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To the Graduate Council:

I am submitting herewith a thesis written by Jacob D. Wilbert entitled "Effects of Small and Normalized Q-Factor Changes and Knee Alignment on Knee Biomechanics During Stationary Cycling." I have examined the final electronic copy of this thesis for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Master of Science, with a major in Kinesiology.

Songning Zhang, Major Professor

We have read this thesis and recommend its acceptance:

Songning Zhang, Joshua Weinhandl, Rachel Tatarski

Accepted for the Council:
Dixie L. Thompson

Vice Provost and Dean of the Graduate School

(Original signatures are on file with official student records.)

Effects of Small and Normalized Q-Factor Changes and Knee Alignment on Knee Biomechanics During Stationary Cycling

A Thesis Presented for the
Master of Science
Degree
The University of Tennessee, Knoxville

Jacob Daniel Wilbert
August 2022

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ABSTRACT

Increasing inter-pedal distance (Q-Factor; QF) in cycling increases peak internal knee abduction moments (KAbM). The effect of smaller, normalized changes in QF has not been investigated, and the effect of static knee alignment at varying QFs is unknown. **Purpose:** The primary purpose of this study was to see if significant changes in KAbM were detectable with normalized increases in QF that are smaller than what has previously been investigated. The secondary purpose of this study was to investigate whether static knee alignment accounts for any changes in knee biomechanics while cycling at different QFs. **Methods:** Fifteen healthy participants were included in this study (7 Males, 8 Females, age: 22.7±2.5 years, BMI: 23.95±3.21 kg/m^2; Mean±STD). Motion capture and instrumented pedals were used to collect kinematic (240 Hz) and pedal reaction force (PRF, 1200 Hz) data, respectively, while cycling at five different QFs. The participant's mechanical axis angle (MAA) was determined using motion capture. Each participant's QFs were normalized by starting at 160 mm and increasing by 2% of the participant's trochanteric leg length (L) where the five QF conditions were (in mm): Q1 (160), Q2 (160 + 0.02*L), Q3 (160 + 0.04*L), Q4 (160 + 0.06*L), and Q5 (160 + 0.08*L). A mixed model analysis of variance was performed to detect differences between QF conditions (α = 0.05). Correlation was calculated between MAA and select variables. **Results:** KAbM was increased by at least 30% in Q5 from Q1, Q2, Q3, and Q4. Medial PRF was increased by at least 20% in Q5 from Q1, Q2, and Q3. There were no significant changes seen in peak vertical PRF, sagittal-plane moments and angles, or peak abduction angle that were concurrent with significant changes in KAbM. MAA had varying degrees of correlation with the variables of interest. **Conclusions:** These results suggest that KAbM is more sensitive to changes in QF at greater QF increases. The effect of MAA on frontal-plane knee biomechanics requires further investigation.

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CHAPTER I: INTRODUCTION

BACKGROUND

Osteoarthritis (OA) affected more than 300 million individuals worldwide in 2020, and it has been a severe burden on healthcare systems globally (Peat and Thomas, 2021). OA occurs most often in the knee (Zhang and Jordan, 2010), affecting primarily the medial knee compartment (Felson et al., 2002). It is characterized by the deterioration of articular cartilage, osteophyte growth, and joint space narrowing as well as debilitating pain and stiffness in symptomatic individuals (Hunter and Felson, 2006).

Currently there is no cure for OA, so the objective of treatment is mostly management of symptoms and risk factors. Known risk factors of knee OA include, but are not limited to, malalignment (Sharma, 2001) and obesity (Felson et al., 1988). Severe varus malalignment is associated with an approximately four-fold increase in the progression rate of medial compartment knee OA (Sharma, 2001; Sharma et al., 2010). Severe valgus malalignment is associated with a nearly five-fold increase in the progression rate of lateral compartment knee OA (Sharma, 2001; Sharma et al., 2010). Obesity [body mass index (BMI) ≥ 30 kg/m^2] is recognized as a feasibly modifiable risk factor of knee OA, as increased body mass inherently increases loading on the knee during weight bearing (Felson et al., 1988; Jiang et al., 2012). The American College of Rheumatology (ACR) and Osteoarthritis Research Society International (OARSI) strongly recommend that basic treatment for knee OA include aerobic exercise, strength training, and weight loss (Kolasinski et al., 2020; Zhang et al., 2008). Aerobic exercise is a known facilitator of weight loss; however, due to the symptoms that many individuals with knee OA experience, it may be difficult just to walk or navigate stairs (Stamm et al., 2016).

Stationary cycling is a low-impact form of aerobic exercise that has been shown to reduce symptoms of knee OA (Luan et al., 2021; Mangione et al., 1999; Salacinski et al., 2012). By placing most of the body weight on the seat of the bike, the knee is relatively unloaded (Kutzner et al., 2012). Peak tibiofemoral contact forces during stationary cycling have been shown to be 1 - 1.5 times bodyweight (BW), compared to 1.8 - 2.5 BW during level walking and 4.2 BW during jogging (D'Lima et al., 2008). By reducing knee loads, cycling may offer a less painful form of exercise and reduce the amount of long-term mechanical damage in the joint relative to walking or jogging.

Internal knee extension moment (KEM) and internal knee abduction moment (KAbM) are two variables of interest that are frequently considered in knee OA and clinical cycling literature. Generally, the KEM is reflective of total contact force (TCF) in the knee while the frontal plane moment influences the mediolateral distribution of the TCF (D'Lima et al., 2006; Zhao et al., 2007). Greater KEM and KAbM have been shown and estimated to increase TCF and medial compartment contact force (MCF) respectively, in gait (Richards et al., 2018) and cycling (Thorsen et al., 2021). Greater contact forces increase the risk for development and progression of knee OA in the corresponding area, as is the case with obesity (Felson et al., 1988; Jiang et al., 2012). The objective of many OA-related studies is to reduce KAbM, thereby potentially reducing harmful loading on the medial compartment of the knee.

Increased step width, often expressed as a percent of an individual's leg length, is a gait modification that has been investigated as a means of reducing KAbM, thereby potentially reducing MCF (Fregly et al., 2008; Paquette et al., 2015, 2014). Analogous to step width in gait, the Q-Factor (QF) of a bicycle or cycle ergometer is the horizontal width between the pedals. This measurement is typically taken from the outermost surface of the crank arms. QF influences the mediolateral positioning of the feet and, subsequently, the frontal plane angles and moments of the lower limbs (Thorsen et al., 2020). However, in contrast to increased step width gait modifications, greater QFs increased KAbM (Thorsen et al., 2020) and MCF (Thorsen et al., 2021). In this study (Thorsen et al., 2020), QF was incrementally increased by 42 mm, the width of one commercial pedal extender on each pedal, from 150 mm. It has not yet been shown whether smaller and normalized increases in QF would also induce significant increases in KAbMs. Small changes in loading may not be influential in singularity, but, due to the repetitive nature of cycling, small changes can accumulate over time to result in substantial differences in cumulative loading (Gatti and Maly, 2019; Kumar, 1990).

Static lower limb alignment may influence the mediolateral position of the knee relative to the placement of the foot, which affects knee frontal plane angles and moments. This alignment is determined by the orientations of the weight bearing mechanical axes of the femur and tibia, known as the mechanical axis angle (MAA). If the medial angle formed by the intersection of these two axes is ≤178°, between 178° and 182°, or ≥182°, the alignments are considered varus, neutral, and valgus, respectively (Sharma et al., 2010). These axes and angles are most accurately determined using full limb standing radiographs, but some clinical measurements (Hinman et al., 2006; Kraus et al., 2005; Magee, 2014; Navali et al., 2012) and a motion capture method (Vanwanseele et al., 2009) have been investigated and used as alternatives to the “gold standard” radiographic method. Individuals with different alignments can display differing biomechanics during dynamic tasks.

Greater KAbM and peak knee adduction angles (Bennett et al., 2017a) as well as peak TCF (Heller et al., 2003) have been seen in individuals with varus alignment during walking, compared to those with neutral and valgus alignments. This effect was also recently investigated during cycling. It was found that peak knee adduction angle, but not KAbM, was significantly greater in varus participants, compared to neutral and valgus participants (Shen et al., 2018). It was also noted in this study and a study by Fang et al. (2016) that participants exhibited either a peak KAbM or a peak internal knee adduction moment during the power phase of cycling. Shen et al. (2018) suggest that knee alignment may be responsible in part for this observation, as 90.9% of their varus participants, 72.7% of the neutral participants, and only 50% of the valgus participants exhibited a peak KAbM. It is possible that knee alignment may account for some variation in KAbM during cycling.

STATEMENT OF THE PROBLEM AND PURPOSE

To our knowledge, there have been no previous studies that investigate the effect of QF changes less than 42 mm on KAbM, and the sensitivity of KAbM to smaller and normalized changes (relative to leg length) in QF is presently unknown. Additionally, there have been no studies that investigate static knee alignment as a covariate in the effects of QF on frontal-plane knee moments during ergometer cycling. It has been shown that varus knee alignment alone causes greater KAbM during walking, but the same main effect of alignment was not seen during cycling. It is unknown whether static knee alignment will account for any variation in KAbM at different QFs. Therefore, the primary purpose of this study was to investigate whether significant changes in KAbM are detectable with smaller and normalized changes of QF. The secondary purpose of this study was to investigate the relationship between static knee alignment and KAbM.

RESEARCH HYPOTHESES

1. It was hypothesized that KAbM would be greater with each normalized increase in QF.

2. It was hypothesized that the increases in KAbM would be even greater as MAA decreases, indicating a relationship between static knee alignment and frontal plane knee moments.

DELIMITATIONS

Participants were excluded from this study if:

- They had ever sustained any major lower limb injury requiring surgical intervention.
- They had sustained any lower limb (diagnosed sprains, strains, or fractures) injuries in the past 6 months.
- Their BMI was classified as underweight (BMI < 18.5 kg/m^2) or obese (BMI ≥ 30 kg/m^2).
- They had any preexisting medical condition that would prevent them from riding a cycle ergometer, as determined by the physical activity readiness questionnaire (PAR-Q).

Participants were included in this study if:

- They were between the ages of 18 and 35.
- They were physically active, defined as participating in at least moderate intensity activity 30 min per day and 3 days per week.
- Their BMI was between 18.5 and 29.9 kg/m^2.

LIMITATIONS

- This investigation was performed in a laboratory setting.
- The population studied were young (< 35 years old) and healthy, so conclusions may not be fully generalized to older individuals with osteoarthritis.

- Foot tracking markers were placed on the shoe rather than on the foot, so any free motion of the foot inside the shoe was not tracked.
- The crank arms and pedals used are much heavier and bulkier than typical parts found on a bicycle or ergometer.
- The QF settings were limited to a precision of one millimeter.
- Mechanical axis angles were estimated using non-radiographic methods based on regression equations from previous literature.

SIGNIFICANCE

Many aspects of a bicycle are adjusted when properly fitting it to its rider. Recent literature shows that QF has a significant effect on frontal plane knee biomechanics, yet QF is not commonly included as a parameter of proper bike fit. This investigation should contribute further evidence as to what extent QF should be considered during proper bike fitting. Furthermore, the relationship between lower limb frontal plane alignment and KAbM during cycling is not well understood, so the results of this study may provide evidence that static knee alignment should be considered during bike fit as well. These findings may practically influence personal and clinical decision making with regards to recreational and therapeutic cycling. For example, it is known that wider QFs cause significant increases in KAbM, so individuals with medial compartment knee OA should avoid wide pedals so they do not potentially increase loading on the medial compartment. Additionally, if a meaningful relationship between MAA and KAbM is found, that would suggest that individuals with greater varus knee alignment may have compounded increases in KAbM when pedaling at wider QFs and should take the same precautions.

CHAPTER II: LITERATURE REVIEW

INTRODUCTION

The purpose of this study was to investigate the effect of knee alignment on frontal plane knee biomechanics at different Q-Factors. This literature review will provide background information on knee osteoarthritis, treatments for knee osteoarthritis, gait modification to unload the knee joint, cycling biomechanics, and methods of determining lower limb static alignment. The review of cycling biomechanics will include background information on cycling, knee biomechanics during cycling, and the effects of cycling modifications on knee biomechanics.

KNEE OSTEOARTHRITIS

Background

Epidemiology and risk factors

Osteoarthritis (OA) is a leading cause of disability in older populations and can affect most joint complexes in the body. It has been increasingly recognized as a global healthcare burden with more than 300 million cases worldwide in 2020 and a rising annual cost of $80 billion to the United States alone in 2016 (Peat and Thomas, 2021). There are many modifiable and non-modifiable risk factors associated with the development and progression of OA. Non-modifiable risk factors such as age (Loeser, 2011), sex (Srikanth et al., 2005), genetics (Felson et al., 1998; Spector and MacGregor, 2004), and joint alignment (Sharma et al., 2001) have all been shown to predispose an individual to OA development (Johnson and Hunter, 2014). Modifiable systemic risk factors such as obesity (Felson et al., 1988; Jiang et al., 2012) and diet (McAlindon et al., 1996) have also been shown to predispose an individual to developing OA, and modifiable local risk factors such as muscle strength (Øiestad et al., 2015), occupation (Croft et al., 1992), and injury (Friel and Chu, 2013) are associated with a specific joint's susceptibility to the disease (Johnson and Hunter, 2014).

Etiology, Pathology, and Diagnosis

Osteoarthritis manifests most commonly in the knees, hands, and hips (Zhang and Jordan, 2010); this review will focus on knee OA. OA is generally characterized by structural and functional deterioration of numerous tissues within the joint, most notably the articular cartilage, and it is associated with osteophyte growth on the subchondral bone and narrowing of the joint space (Hunter and Felson, 2006; Lawrence et al., 2008). This damage and bone remodeling may occur over many years of repetitive and excessive loading on the joint, which is considered colloquially as wear-and-tear, and can be exacerbated by increasing joint loads as is the case with obesity and knee OA (Felson et al., 1988). As a standard practice, the progression of knee OA can be classified, or graded, based on these observable changes using radiography such as X-ray and, more recently, magnetic resonance imaging (Johnson and Hunter, 2014). Radiographic knee OA is classified most commonly using the Kellgren-Lawrence scale, a scale from zero to four with higher numbers indicating more severe damage and osteophyte growth as well as greater narrowing of the joint space (Kellgren and Lawrence, 1957). Knee OA can also be classified clinically through physical examination and questionnaires of signs and symptoms, such as the Western Ontario and McMaster Universities Osteoarthritis Index (WOMAC) (Bellamy et al., 1988; Lawrence et al., 2008; Michael et al., 2010).

Treatment

As there is no cure for osteoarthritis, the objective of treatment falls to pain management and functional maintenance of the affected joints (Hunter and Felson, 2006). Depending on the severity of symptoms, namely joint pain and stiffness, treatment can range from simply choosing appropriate footwear to more drastic measures like total joint replacement. Treatment regimens can also be non-pharmacologic, pharmacologic, or a combination of the two (Hunter and Felson, 2006). Several organizations have published recommended clinical practice guidelines (CPGs) for the treatment of OA such as the American Academy of Orthopaedic Surgeons (AAOS), American College of Rheumatology (ACR), Osteoarthritis Research Society International (OARSI), European League Against Rheumatism (EULAR), the Ottawa Panel, and the National Institute for Health and Clinical Excellence (NICE). A general consensus between all parties is to begin basic treatment with patient education, strength training, aerobic exercise, and weight loss, as there is strong evidence that these programs are beneficial for improving joint pain and function without adverse effects (Brosseau et al., 2017a, 2017b; Conaghan et al., 2008; Fernandes et al., 2013; Jevsevar, 2013; Kolasinski et al., 2020; Zhang et al., 2008). These CPGs work hand-in-hand as non-pharmacologic therapies to alleviate joint pain and stiffness, and some expert opinions also recommend supplementing these modalities with pharmacologic remedies such as topical analgesics or non-steroidal anti-inflammatory drugs (Zhang et al., 2008).

Clinical Practice Guidelines

Educating a patient about osteoarthritis, its misconceptions, and potential treatment options is commonly one of the first non-pharmacological CPGs recommended. Such education also includes ensuring a patient’s access to information about knee OA and encouraging selfhelp and self-efficacy, such as adherence to therapy (Conaghan et al., 2008; Fernandes et al., 2013; Kolasinski et al., 2020; Zhang et al., 2008). Improving a patient's knowledge about their condition and what factors put them at risk for OA progression will ensure they know how to avoid certain modifiable risk factors and engage in endorsed positive behaviors, such as weight loss, strength training, and aerobic exercise. Obesity is logically and empirically pinned as one of the most, if not the most, strongly associated and modifiable risk factors of osteoarthritis development and progression in the knee (Jiang et al., 2012). As an individual's weight increases, the load on weight bearing joints increases, and this increased joint contact force can expedite wear-and-tear on the joint surface. Weight loss is strongly recommended to combat obesity and its effects on knee OA (Hunter and Eckstein, 2009; Jiang et al., 2012). Strength training and aerobic exercise are often prescribed for weight loss, but they also have more direct effects on knee OA. Knee extensor muscle weakness is an established modifiable risk factor of the development of knee OA. So, training the quadriceps for activities of daily living by performing exercises that utilize resistance through a range of motion is strongly recommended to combat the incidence and progression of knee OA (Brosseau et al., 2017a; Felson, 2006; Øiestad et al., 2015). Regular (≥30 min/day; ≥5 days/week) moderate-intensity aerobic exercise (3-5.9 METs) is strongly recommended to reduce pain and increase physical function in OA-affected joints, particularly the knee (Brosseau et al., 2017b; Conaghan et al., 2008; Fernandes et al., 2013; Jevsevar, 2013; Kolasinski et al., 2020; Zhang et al., 2008).

Gait Modification

Because of the nature of the disease and its symptoms, most forms of exercise, and many activities in general, involving the affected joint can be painful and otherwise difficult for individuals with knee OA (Stamm et al., 2016). Greater tibiofemoral contact force increases the friction of the articulating surface, potentially making joint motion more painful. Many people with knee OA even find great difficulty in simply walking long distances and climbing stairs under their own body weight (Stamm et al., 2016). It is crucial that patients with knee OA can walk and negotiate stairs with minimal discomfort, as these are prevalent activities of daily living. Knee contact forces can be reduced during these activities through gait modifications (Fregly, 2012).

Gait modification is an antecedent short-term therapy to total knee arthroplasty for individuals with knee osteoarthritis (Fregly, 2012). OA affects the medial compartment of the knee more prevalently than the lateral compartment, as the majority of the contact force in the knee is distributed on the medial side (Felson et al., 2002). Consequently, much of the attention of researchers and clinicians is focused on unloading the medial compartment specifically. Increased medial compartment loading is correlated with a greater internal abduction moment at the knee (Zhao et al., 2007). As it is difficult to measure knee contact forces *in vivo*, internal knee abduction moment (KAbM) has been used as a surrogate for medial compartment loading and is therefore targeted for reduction in gait modification studies (Fregly, 2012; Zhao et al., 2007). Walking gait modifications such as decreased walking speed (Robbins and Maly, 2009), increased step width (Fregly et al., 2008), increased mediolateral trunk sway (Mündermann et al., 2008), and medial knee thrust (Fregly et al., 2007) have all been effective in reducing KAbM. Combinations of gait modifications, such as toe-in with wider step width, have been shown to reduce KAbM more than the same modifications do individually (Bennett et al., 2017b, 2017a).

Step width modifications

Increasing step widths from the preferred step width has been used as a gait modification in level walking and stair ambulation. Absolute ranges of preferred step width in healthy individuals are 7-12 cm in level walking (Helbostad and Moe-Nilssen, 2003; Wert et al., 2010), 13-14 cm in stair ascent (Paquette et al., 2015; Yocum et al., 2018), and 15-17 cm in stair descent (Paquette et al., 2014; Yocum et al., 2018), on average. In studies that have normalized preferred step width to the individual's leg length, these values were 13% in level walking (Donelan et al., 2001), 15.4% in stair ascent (Paquette et al., 2015), and 19.8% in stair descent (Paquette et al., 2014). It is worth considering the influence of individuals' body heights and leg lengths on their absolute preferred step width when comparing to individuals of different leg lengths, as taller and larger individuals may have wider absolute step widths. So, absolute increases in step width may affect KAbM values disproportionally for individuals with different leg lengths. While walking at the preferred step width has been shown to minimize metabolic cost (Donelan et al., 2001), walking at a wider step width may be more comfortable in individuals with medial knee OA

During walking and stair ambulation, two peak KAbMs occur in the frontal plane knee moment: one during weight acceptance and one during push-off (Fregly, 2012; Yocum et al., 2018). The peak at weight acceptance is typically the greater of the two and may contribute more to the onset and progression of OA (Fregly, 2012; Yocum et al., 2018). Using a musculoskeletal model, Fregly et al. (2008) predicted a 9% decrease in peak external knee adduction moment with increased step width during level walking. This prediction was confirmed by a few studies that also found decreases in the peak external knee adduction moment during walking (Yocum et al., 2018; Zhao et al., 2007), stair ascent (Bennett et al., 2017b; Paquette et al., 2015; Yocum et al., 2018), and stair decent (Paquette et al., 2014; Yocum et al., 2018). While walking gait has been researched extensively, cycling has gained increased attention in biomechanical studies for its potential benefits over walking in individuals with obesity and osteoarthritis. Knee loading during cycling will be reviewed in later sections.

Osteoarthritis summary

OA is a degenerative joint disease that causes joint pain and functional deficits that debilitates millions of people globally and most commonly affects the knee. There are various risk factors for the onset and progression of OA, some modifiable and others not, that are targets for treatment. Obesity is a highly associated and modifiable risk factor of knee OA, so weight loss is ubiquitously recommended for treatment. Exercise and activities of daily living, such as walking and stair negotiation, can be difficult and painful for individuals with knee OA due to increased joint loading. Increasing step width from the preferred has been shown to reduce medial knee loading, potentially making these tasks less painful to perform.

CYCLING

Since its advent, the bicycle has become a ubiquitous tool for transportation, exercise, recreation, competition, and even rehabilitation, and quite a variety of bicycles exist to accommodate each of these activities. However, as with other forms of human locomotion like walking and running, cycling may yield unintended consequences over time. Given that cycling is a daily activity for many individuals, so too is the risk of developing cycling-related overuse injuries. Somewhere between 14.8% and 33% of cyclists have experienced knee pain or injury associated with long-duration pedaling (Bini and Flores-Bini, 2018). In an effort to reduce the prevalence of these injuries, it is important to understand the effects of the various aspects of bike fit and cycling intensity on performance and health. Furthermore, cycling is widely recommended for knee rehabilitation after injury and/or surgery because the stresses on the structures of the knee are relatively small, depending on the resistance of the bicycle or ergometer (McLeod and Blackburn, 1980). Compared to walking, cycling applies resistance over a greater range of motion at the knee (Bini and Carpes, 2014; Mann and Hagy, 1980) while inducing knee contact forces less than those during level walking (D'Lima et al., 2008). Therefore, cycling may be an effective and preferable alternative to walking as a form of rehabilitation and exercise for individuals with obesity and osteoarthritis.

Background

Bike fit

The rider is attached to a bicycle or cycle ergometer at three points: the saddle (seat), handlebars, and pedals. Therefore, adjustments to these interfaces will affect the riding position, posture, and biomechanics of the lower and upper body and trunk on the bicycle (Bini and Carpes, 2014). Saddle height refers to the vertical position of the saddle, and it affects peak sagittal plane angles of the hip, knee, and ankle (Bini et al., 2011; Bini and Carpes, 2014). Saddle fore-aft position refers to the anterior-posterior position of the seat, and changes to this will affect the sagittal plane angle of the knee as well as the anterior-posterior position of the knee with respect to the pedal (Bini et al., 2013). Saddle tilt is the inclination of the saddle in the sagittal plane and affects the angle between the pelvis and spine segments (Salai et al., 1999; Wadsworth and Weinrauch, 2019). The vertical distance between the handlebars and the saddle affects the inclination of the trunk and distribution of body weight between the saddle and handlebars (Bini and Carpes, 2014; Wadsworth and Weinrauch, 2019). Adjustments to the width between the bicycle pedals influence the frontal plane positioning and biomechanics of the rider's lower limbs (Thorsen et al., 2020). Some competitive cyclists wear cleats that attach to the pedals which are either entirely fixed or allow up to 15° of internal/external (toe-in/toe-out) rotation and 10mm of mediolateral translation called "float" (Wheeler et al., 1995).. The frontal-plane angulation of the foot can be adjusted by attaching wedges of varying degree to the pedals (Gardner et al., 2016).

Crank cycle

Cycling motion is typically described within the 360° of angular motion of the crank arms about the crank axis, commonly known as the crank cycle. During the crank cycle, two main phases occur in a forward cycling motion (Asplund and St Pierre, 2004). The downstroke is from the top dead center (TDC; 0°) position to the bottom dead center (BDC; 180°) position and considered the power phase. It is during this phase that the hip extensors, knee extensors, and ankle plantarflexors work to accelerate the pedal forward and downward to propel the bicycle (Asplund and St Pierre, 2004; So et al., 2005). Peak internal knee extension moment (KEM) and knee power output are typically seen near the 90° mark, when the pedal is at its most forward position and the crank arm is horizontal (Bini and Carpes, 2014; Ettema et al., 2009). Following the BDC point (180°) until the pedal returns to TDC (360°) is the recovery phase (Asplund and St Pierre, 2004). During this phase, knee and hip flexor muscles act to pull the pedal backward and upward, assisting the opposite leg through its power phase (Bini and Carpes, 2014; So et al., 2005).

Instrumented pedals

Similar to a ground reaction force, cyclists experience a reactionary force exerted by the pedals, called pedal reaction force (PRF), which can be measured by pedals instrumented with force sensors. The design of the force-instrumented pedal has evolved over the years since the idea was first incited in 1896 (Sharp, 1977). Beginning with a single uniaxial strain gauge mounted within the pedal (Hoes et al., 1968), the design shifted to single bi- and triaxial strain gauges (Dal Monte et al., 1973; Hull and Davis, 1981), then to a single triaxial piezoelectric transducer (Ericson et al., 1984), and most recently to dual triaxial piezoelectric transducers within the pedal (Broker and Gregor, 1990). The earlier uniaxial and biaxial strain gauges were capable of detecting forces only in one and two dimensions, respectively, where triaxial sensors can detect forces in three dimensions. Piezoelectric sensors allow for a wider range of force detection, and dual sensors are able to determine the center of pressure and magnitude of free moments, whereas single sensor pedals cannot (Broker and Gregor, 1990). Current studies still utilize pedals instrumented with dual triaxial piezoelectric force transducers that are capable of measuring three dimensional forces as well as the central location of force application on the pedal surface (Gardner et al., 2016, 2015; Hummer et al., 2021; Martin and Brown, 2009; Shen et al., 2018; Thorsen et al., 2020).

Knee biomechanics in cycling

Most of the research investigating knee cycling biomechanics has focused on the sagittal plane, and while the values vary from study to study, a range of normal kinematic and kinetic values emerges (Bini and Carpes, 2014; Wozniak Timmer, 1991). The variations may potentially be derived from intentional manipulations aimed at achieving specific results, differences in practice regarding bike fit and measurement, or unintentional differences in the equipment and methods used. Some current research has been more focused on the frontal plane biomechanics of the knee. Recent studies suggest that understanding the frontal plane biomechanics of the knee is important to characterize unilateral overuse pathologies such as medial compartment knee OA (Gardner et al., 2016, 2015; Shen et al., 2018; Thorsen et al., 2020). Transverse plane biomechanics have received even less attention, and as such, only sagittal and frontal plane biomechanics will be considered in this review.

Sagittal plane knee biomechanics

The knee typically goes through a flexion-extension range of motion (ROM) of approximately about 75° throughout the crank cycle (Asplund and St Pierre, 2004; Bini et al., 2011). Starting around 100-110° of flexion at TDC the knee extends to approximately 25-40° at BDC then flexes again as the pedal returns to TDC (Asplund and St Pierre, 2004; Bailey et al., 2003; Bini et al., 2011). Variation in sagittal plane ROM can depend on factors of bike fit, such as saddle height and fore-aft position (Bini et al., 2011, 2013).

As the pedal moves through the crank cycle and knee moves through its range of motion, the knee experiences changing moments. The knee experiences peak internal moments and power near 90° of the crank cycle during the power phase (Bini and Carpes, 2014; Ettema et al., 2009). At workrates around 80W, peak KEM has been reported to be 20-35 Newton-meters (Nm) on average, and KEM depends on factors such as work load and saddle height (Bini et al., 2011; Ericson and Nisell, 1986; Ettema et al., 2009; Fang et al., 2016; Gardner et al., 2016; Thorsen et al., 2020). Soon after experiencing the peak KEM, the moment at the knee becomes an internal flexion moment near 130-140° of the crank cycle, when the knee extensors deactivate and the hamstrings remain active. The knee moment remains as an internal flexion moment until late in the recovery phase (300-315°), when the knee extensor muscles reactivate (da Silva et al., 2016; Ericson and Nisell, 1986; So et al., 2005; Wozniak Timmer, 1991).

Frontal plane knee biomechanics

The frontal plane biomechanics of the knee during cycling have not been examined nearly to the same extent as sagittal plane biomechanics. Previous research shows adduction-abduction ROMs of about 6-10° through the crank cycle, and this range is typically situated around a neutral (0°) knee angle (Fang et al., 2016; Fife et al., 2020; Gardner et al., 2015; Shen et al., 2018; Thorsen et al., 2020). The knee begins near its most adducted position at TDC and approaches its peak adduction angle near a crank angle of 45° degrees, and as the cycle continues past 90°, the knee progressively becomes more abducted until it reaches its peak abduction angle around BDC (Fife et al., 2020; Gardner et al., 2016; Shen et al., 2018). Baseline knee frontal plane angles are influenced by the static alignment of the hip, knee, and ankle joints. If the medial/interior angle formed at the knee by the hip-knee and knee-ankle segments is ≤178°, between 178° and 182°, or ≥182°, the alignments are considered varus, neutral, and valgus, respectively (Sharma et al., 2010). Individuals with varus alignment displayed peak knee adduction angles of about 10°, compared to 5° in those with neutral alignments and -2° in those with valgus alignments (Shen et al., 2018). Peak knee abduction angles were also significantly different being approximately 0°, -5°, and -10° in individuals with varus, neutral, and valgus alignments, respectively. It is evident that the natural frontal plane knee alignment (neutral, varus, or valgus) of an individual may influence peak adduction and abduction angles as well as their ROM (Shen et al., 2018). These results are corroborated by studies of alignment in level walking (Bennett et al., 2017a) and stair ascent (Bennett et al., 2017b) that found more abducted peak knee angles in valgus aligned individuals, compared to neutral and varus, and more adducted peak knee angles in varus aligned individuals, compared to neutral and valgus.

In recent studies of frontal plane knee kinetics, two patterns have emerged in the literature. One pattern that has been found is that peak frontal plane knee moments were experienced only as KAbMs (Gardner et al., 2016; Thorsen et al., 2020). Alternatively, a different study found that eleven of eighteen participants experienced a peak internal knee adduction moment rather than a peak KAbM while the rest did experience peak KAbM (Fang et al., 2016). The natural lower limb alignment of these participants was not reported. In another study, the participants' natural lower limb alignment was assessed, and they were organized into varus, neutral, and valgus groups based on these assessments (Shen et al., 2018). Ten out of 11 varus, eight out of 11 neutral, and 5 out of 10 valgus-aligned individuals experienced a peak KAbM in the knee, and the rest of the participants experienced a peak internal adduction moment (Shen et al., 2018). The commonalities between these findings, though, are the magnitudes and temporal alignment of the peak abduction and adduction moments. In all of these studies, the peak frontal plane moments occur between 90° and 180°, and the peak KAbM tended to be between -7 and -10 Nm during submaximal cycling (Fang et al., 2016; Gardner et al., 2016; Shen et al., 2018; Thorsen et al., 2020). In the studies that reported participants with peak internal knee adduction moments, these values ranged from 4 to 9 Nm at work loads of 0.5-1.5 kg (Fang et al., 2016; Shen et al., 2018). Shen et al. (2018) suggests that the fact that some individuals exhibit knee adduction moments while others exhibit knee abduction moments could be related to their natural knee alignment, but further investigation is needed to determine the cause of this phenomenon in cycling. Previous studies have found an effect of alignment on frontal plane moments and joint contact forces in other tasks. Using musculoskeletal models of patients with instrumented knee arthroses, Heller et al. (2003) found that greater varus malalignment caused increased KAbM and greater varus and valgus malalignments caused increased tibiofemoral compressive force in walking and stair climbing. Alignment generally dictates the mediolateral distribution of knee contact force and, consequently, the development of compartmental osteoarthritis (Sharma et al., 2001). Varus alignment causes increased pressure on the medial compartment of the knee (Bruns et al., 1993), and has been associated with a greater risk of incidence and progression of medial compartment OA (Sharma et al., 2010). Valgus alignment causes increased pressure on the lateral compartment (Bruns et al., 1993), and has been associated with a greater risk of incidence and progression of lateral compartment OA (Felson et al., 2013; Sharma et al., 2010).

Pedal reaction forces and knee loading

The PRF exerted by the pedal on the foot creates compressive forces and external moments in the ankle, knee, and hip. Using the PRFs measured by an instrumented pedal, joint moments can be estimated via three dimensional (3D) inverse dynamics (Ericson and Nisell, 1986). Joint contact forces can be measured *in vivo* using force instrumented joint implants or estimated by musculoskeletal modelling based on the experimentally collected PRF and kinematic data (D’Lima et al., 2008; Thorsen et al., 2021). The direction of the PRF vector is directly related to the direction of the moments at the knee. In the sagittal plane, when the PRF vector is aimed anteriorly or posteriorly to the knee joint, it will create an external extension or flexion moment, respectively (Shen et al., 2018). In the frontal plane, when the PRF vector is oriented medially or laterally to the knee, it creates an external adduction or abduction moment, respectively (Thorsen et al., 2020). The maximum vertical and medial (vertical = 200-230N; medial = 20-30N) pedal reaction forces are seen during the power phase at a work rate of about 80W, concurrent with peak power output, peak internal knee extension moment, and peak KAbM (Bini and Carpes, 2014; Broker and Gregor, 1990; Ericson and Nisell, 1986; Fang et al., 2016; Gardner et al., 2016; Kutzner et al., 2012; Shen et al., 2018; Thorsen et al., 2020). This is also where the knee experiences the greatest amount of total contact force (TCF) and medial compartment contact force (MCF) (Thorsen et al., 2021). A joint's TCF is directly influenced by both PRF and muscle forces. As both PRF and muscle force become greater, tibiofemoral contact force increases proportionally (Zajac and Gordon, 1989). During stationary cycling, the body's weight is supported by the saddle, which unloads the weight of the body from the knees. This results in tibiofemoral contact forces of 1-1.5 BW, which are due to mainly from contributions of muscle forces, compared to 1.8-2.5 BW during level walking and 4.2 BW during jogging (D'Lima et al., 2008).

Sagittal plane moments of the knee joint are reflective of the overall loading to the knee (TCF), while the frontal plane moment influences the mediolateral distribution of the contact force (D'Lima et al., 2006; Zhao et al., 2007). Generally, with increased sagittal plane moments comes greater TCF. In the frontal plane, increases in KAbM typically correspond to increased compression of the medial compartment of the knee, though that is not always the case (Thorsen et al., 2021). In individuals with knee osteoarthritis, it is important to lessen the loads applied to the affected compartment. Many studies have focused on the effects of different cycling modifications on knee motion and loading. The findings of these studies of cycling biomechanics may guide recommendations for healthier cycling and rehabilitation in the future.

Effects of cycling modifications on knee biomechanics

Cadence

Cadence in cycling refers to the speed at which the crank revolves about its axis and is expressed as the number of revolutions per minute (RPM). The literature has shown mixed results regarding the effects of cadence on the kinematics and kinetics of the knee. Fang et al. (2016) found that cadences increased from 60 rpm (70, 80, and 90 rpm) caused greater peak anterior PRF and internal knee flexion moment at a fixed resistance of 1 kg, indicating more backwards pulling on the pedal. They did not find any significant effects of cadence on sagittal plane ROM, peak extension moment, or frontal plane ROM or moments of the knee (Fang et al., 2016). Bini et al. (2010) found that higher cadences [increases from (preferred cadence – 20%) to preferred cadence, to (preferred cadence + 20%)] caused decreases in sagittal knee ROM and absolute knee mechanical work (the integration of power over the entire crank cycle) (Bini et al., 2010). Knee moments were not investigated in this study, so joint kinetics cannot be compared between these two studies.

Ericson et al. (1988) found no effect of cadence (40-100 rpm) on knee ROM or peak angles and they did not report kinetics. In a different study, Barratt et al. (2016) investigated the effects of pedaling speed, not pedaling cadence, on lower limb biomechanics. They reported that higher pedal speeds (lowest = 1.41 m/s; highest = 1.79 m/s) caused greater average knee extension velocity, flexion velocity, and ROM and lesser average KEM. Pedal speed was manipulated by increasing crank arm length at a constant cadence (Barratt et al., 2016). Their choice of experimental variable makes comparisons between these studies difficult because cadence was never isolated as an experimental variable. This raises the question of whether tangential pedal speed or pedaling cadence is more influential on knee angles and moments.

An early study of tibiofemoral compressive forces held resistance constant at 2 kg while cadence was increased from 40 rpm to 60, 80, and 100 rpm. Knee contact forces were calculated with an inverse dynamics approach, and they found no changes as cadence increased (Ericson and Nisell, 1986). A more recent study showed that at set work rates, increasing cadence from 40 to 60 rpm resulted in decreases of 20-40% in tibiofemoral compressive forces, which were measured *in vivo* using an instrumented knee arthrosis (Kutzner et al., 2012). Because the work rate was constant, as cadence increased then resistance had to decrease. This may explain the contrary findings between Ericson & Nisell (1986) and Kutzner et al. (2012). Additionally, the different methods of inverse-dynamics calculation (Ericson and Nisell, 1986) versus instrumented knee arthrosis measurement (Kutzner et al., 2012) of tibiofemoral contact forces may have contributed to the different findings. These results suggest that cadence has little to no effect on knee loading.

Resistance

The effect of resistance, or work load, on knee kinetics is more agreed upon than that of cycling cadence. Logically, increases in resistance require greater force to be placed on the pedal to cause the crank to turn. This, in turn, exerts greater PRFs on the foot and up the kinetic chain, causing greater extension moments and KAbM at the knee (Fang et al., 2016; Shen et al., 2018; Thorsen et al., 2020). Fang et al. (2016) showed that each 0.5 kg increase in resistance from 0.5 kg to 2 kg caused significantly greater KEM: 11.61 Nm at 0.5 kg to 20.23, 26.04, and 34.23 Nm at 1 kg, 1.5 kg, and 2 kg. The same study also showed that KAbM significantly increased from -5.82 Nm at 0.5 kg of resistance to -10.18 Nm at 1.5 kg, and significant increases in KAbM were seen at 2.5 kg when compared to 0.5, 1.5, and 1.5 kg (Fang et al., 2016). Shen et al. (2018) also found significantly greater KEM and KAbM with 0.5 kg increases in resistance from 0.5 kg to 1.5 kg. Thorsen et al. (2020) reported significant increases in KEM and KAbM between work rate conditions of 80 and 120 Watts as well as 120 and 160 Watts at a cadence of 80 rpm. Tibiofemoral compression was also shown to increase with greater workloads (Ericson and Nisell, 1986; Kutzner et al., 2012). These increases in knee loading variables, suggest that cycling with greater resistance (workload or workrate) puts greater mechanical demand on the knee, which may, at some point, be excessive and increase the likelihood of overuse-related damage and pain.

Unlike the effects on knee moments and compressive forces, there are mixed results regarding the effect of resistance on knee kinematics. In a study of lower limb motions during cycling, Ericson et al. (1988) found a small effect of work load on knee extension angle. At a cadence of 60 rpm, increasing the workload from 0 kg to 2 kg to 4 kg only significantly decreased the maximum knee extension angle from 49° to 42° and had no effect on maximum flexion angle or range of motion. No frontal plane kinematics were reported in the study (Ericson et al., 1988). Fang et al. (2016) found a >4° increase in knee extension ROM over a range of workloads of 0.5, 1, 1.5, 2, and 2.5 kg. Shen et al. (2018) found a small, but significant, increase in knee extension ROM of 0.54° between 0.5 and 1kg workloads, which was only seen in individuals with varus lower limb alignments. Thorsen et al. (2020) and Bini and Diefenthaeler (2010) found no significant changes in knee extension ROM at higher work rates and constant cadences. Of the studies that included frontal plane kinematics (Fang et al., 2016; Shen et al., 2018; Thorsen et al., 2020), only Shen et al. (2018) found a significant, albeit small, effect of workload on peak knee adduction angle in individuals with varus lower limb alignments. All other effects of resistance on knee frontal plane kinematics were nonsignificant (Shen et al., 2018).

Q-Factor

In cycling, the horizontal width between the medial aspect of each pedal, known as Q-factor (QF), is analogous to step width in walking. This measurement is taken either between the outermost aspect of the crank arms or the medial aspect of the pedals (Thorsen et al., 2020). Furthermore, the notion of QF being a potential target for modification in cycling, just as step width is in gait, is relatively new (Disley and Li, 2014a; Thorsen et al., 2020). The standard QF for road bicycles is typically near 150 mm and 180 mm for mountain bicycles. Some bicycles have been specially manufactured for world-class cyclists with QF <130 mm, which is more in-line with preferred step width in walking, for a more efficient performance (Disley and Li, 2014a). Although bicycles and cycle ergometers are manufactured with set QF, it is possible to change the QF by adding pedal spacers between the crank arm and the pedal (Thorsen et al., 2020). It is also possible for a pedal to have freedom to translate along a mediolateral axis (Disley and Li, 2014b). Just as an individual can have a preferred step width in gait, they can have a preferred Q-factor (PQF) in cycling (Disley and Li, 2014b).

A study of PQF found a mean (± standard deviation) PQF among ten trained cyclists of 142 mm (± 12 mm). Although nonsignificant, the participants showed the best knee variability and gross mechanical efficiency at their PQF, when compared to QF of 150 mm, 30 mm below their PQF, and 30 mm above their PQF (Disley and Li, 2014b). Additionally, in an effort to predict an individual's PQF based on their anthropometry, they found a strong correlation (R^2 = 0.794, $p < 0.002$) between a participant's PQF and the distance between the medial malleoli during a hanging task. The PQF's ranged approximately from 123 mm to 158 mm with an approximate range of hanging ankle distances of 22-72 mm (Disley and Li, 2014b). This study also attempted to predict PQF using the inter-malleoli distance during a step-up walking task. These self-selected step widths ranged from 37 mm to 139 mm and showed no correlation with PQF (R^2 = 0.091) (Disley and Li, 2014b). However, neither Disley and Li (2014a) nor Disley and Li (2014b) investigated the effects of QF on knee biomechanics.

A recent study has shown that wider QF on a cycling ergometer increase knee extension and abduction ROM, peak knee frontal-plane angle, peak medial PRF, and KAbM (Thorsen et al., 2020). Starting from 150 mm, 21 mm pedal spacers were added to either side in 3 increments, creating QF conditions of 192, 234, and 276 mm. They hypothesized that as QF increased KAbM would decrease, as it does with an increased step width gait modification. However, the results showed that at all higher QF, the medial ground reaction force and KAbM actually became significantly larger (Thorsen et al., 2020). In a follow-up investigation, musculoskeletal modelling was used to associate the increased KAbM with increased medial compartment knee contact force. The authors suggest that QF modulations can be used to control medial compartment loading in such a way that would promote a positive physiological adaptation to increased loading on the medial compartment (Thorsen et al., 2021).

In another recent study, the effect of QF modulation on the ankle, knee, and hip frontal plane kinematics was investigated, but no knee moments were reported (Fife et al., 2020). Beginning at a QF of 150mm and increasing to 190 mm and 210 mm, the knees and hips became systematically more abducted at higher QFs, but there were no observable affects at the ankle (Fife et al., 2020). Although this study yielded a stronger effect of QF, these results align with what was found by Thorsen et al. (2020) that the knee became significantly more abducted at a QF of 276 mm compared to 150 mm.

Saddle height and fore-aft position

Saddle height has been considered as the most controversial aspect of bike fit, and, therefore, it has been the emphasis of numerous studies (Bini et al., 2011). There are several different methods of determining saddle height based on anthropometrics and knee kinematics. Three methods identify the saddle height as a percentage of leg length, as measured from the greater trochanter, ischial tuberosity, or inseam heights to the floor while standing (Bini et al., 2011; Hamley and Thomas, 1967; Nordeen-Snyder, 1977; Shennum and deVries, 1976). Another method, the Holmes method (Holmes et al., 1994), suggests that the knee should be between 25° and 30° of flexion when the pedal is at BDC during a static fitting. This method is commonly used and is recommended to reduce knee joint loading and improve cycling efficiency (Bini et al., 2011; Holmes et al., 1994).

Many manifestations of overuse injuries of the knees from cycling are attributed to improper saddle position. Higher saddle heights cause the knees to be more extended throughout the crank cycle and may irritate the iliotibial band, put more stress on the biceps femoris tendon, or increase patellofemoral loading (Asplund and St Pierre, 2004). Lower saddle heights cause the knees to be more flexed and put more stress on the patellar and quadriceps tendons, increasing risk of overuse at those sites (Asplund and St Pierre, 2004). A lower saddle height, where the knee is flexed to 40° at BDC during a static fitting, caused an increase in KEM. Saddle height has not been shown to effect the frontal plane angles and moments of the knee (Hummer et al., 2021).

Saddle fore-aft position, otherwise known as saddle depth, dictates the rider's anteroposterior position on the bike. While this aspect of bike fit does not receive as much attention as saddle height, there is a well-established consensus on the proper fore-aft position, known as the "knee over pedal spindle" method (Burke, 2002; Wadsworth and Weinrauch, 2019). Combined with the bicycle's seat tube angle which is typically 72-74°, the fore-aft position of the saddle should place the anterior aspect of the rider's knee directly above the pedal spindle (axis) when the pedal is at the 90° position. This is typically achieved by dropping a plumb line from the lateral femoral condyle and aligning it vertically with the center of the pedal spindle (Burke, 2002; Wadsworth and Weinrauch, 2019). Another variation of this measurement places the patella directly over the pedal spindle by dropping the plumb line from the patella, rather than from the lateral condyle of the femur (Bini and Carpes, 2014). Bini et al. (2013) found that a maximally forward position on the saddle caused a 7° increase in knee flexion and a maximally backward saddle position caused a 5° decrease in knee flexion angle at 90° (3 o'clock) of the crank cycle, when compared to the preferred fore-aft position on the saddle. The preferred position placed the riders' sacrum 0.32 m horizontally behind the bottom bracket (crank axis) of the bicycle, and the riders were moved 0.06 m forward and 0.03 m backward at the maximal positions. They found that patellofemoral and tibiofemoral compressive forces were not substantially affected by saddle fore-aft position, but tibiofemoral anterior shear force was significantly increased with more backwards saddle positions (Bini et al., 2013).

Foot position

There are a few aspects of foot position on the pedal whose effects on knee biomechanics have been researched: anterior-posterior position, inversion-eversion angle of the ankle, and toe-in angle. When comparing the anterior position of the foot (“ball” of foot placed on pedal) to the posterior position of the foot (pedal contact is 10 cm posterior of the “ball”), knee joint anterior shear force was significantly increased when cycling in the posterior position (Ericson and Nisell, 1986). Additionally, cycling in the anterior position resulted in a 7° increase in peak knee flexion, a 10° decrease in peak knee extension, and a 3° decrease in knee ROM (Ericson et al., 1988). Frontal plane biomechanics were not reported.

When the frontal plane angle of the ankle (inversion/eversion) was modified using lateral or medial wedges on the pedals, significant changes in knee kinematics and kinetics were found. Gregerson et al. (2006) reported peak frontal plane angles and moments of the knee at 5 conditions of the ankle frontal plane angle: neutral (no wedge) and 5° & 10° each of inversion and eversion. Greater angles of ankle inversion caused greater KAbM at the knee. KAbM was -3.55 Nm at 10° of eversion, -7.84 Nm at neutral, and -11.53 Nm at 10° of inversion (Gregersen et al., 2006). Ten degrees of eversion caused a significantly greater KAbM of -10.07, compared to -8.11 Nm at the neutral condition. This study found no significant differences in frontal plane knee kinematics (Gregersen et al., 2006). In a similar study by Gardner et al. (2016), 5° and 10° wedges were used to further investigate the effects of ankle eversion on frontal plane knee biomechanics in individuals with osteoarthritis and healthy controls. Peak KAbM was significantly reduced in both groups with a 10° wedge, compared to no wedge, and no significant differences in KEM were found between wedge conditions or groups. They reported significantly increased vertical PRFs with both the 5° and 10° wedges in both the healthy and osteoarthritic individuals, although the osteoarthritic individuals did not report increased pain in the wedged conditions. Additionally, they found significant increases in peak knee flexion angle at 5° and 10° (Gardner et al., 2016).

In an investigation of the effects of 5° and 10° toe-in angles on knee biomechanics in healthy and osteoarthritic participants, there was a 3.3° decrease in peak flexion angle and a 2.4° decrease in peak adduction angle at a 10° toe-in angle in the healthy group (Gardner et al., 2015). Greater decreases were seen in the OA group (4.3° decrease in peak flexion angle; 3.2° decrease in peak adduction angle), and the differences at a 5° were smaller in both groups, but still significant. Increased toe-in angles had no significant effect on KEM or KAbM, but the peak vertical PRF significantly increased by 9.8 N in the healthy group and 14.7 N in the OA group at a 5° toe-in angle (Gardner et al., 2015).

Overall, some changes in the position of the foot relative to the knee have been shown to affect biomechanics at the knee. The position of the foot relative to the knee may also be changed by moving the pedal itself, as is the case with QF. Foot position is a potential modifiable target for reducing frontal plan knee loading.

Cycling summary

Bicycles are popular tools for transportation, competition, exercise, and rehabilitation. Because stationary cycling is inherently low-impact, it is recommended as a form of exercise and therapy for individuals who are overweight/obese or suffer from knee OA. There are various aspects of a bicycle that can be manipulated to fit the rider optimally. Some of these components have been researched extensively, while others are only beginning to be investigated. Because OA predominantly affects the medial compartment, frontal plane knee biomechanics have seen increased interest in the literature. Increased KEM, increased KAbM, and varus limb alignment, are correlated with greater compressive force on the medial compartment of the knee. Q-factor has been shown to have a great amount of influence on the medial pedal reaction force and frontal plane knee angles, moments, and contact forces. Consequently, Q-factor is a potential target for manipulation in cycling exercises for people who are susceptible to or are suffering from knee OA.

METHODS OF DETERMINING LOWER LIMB STATIC ALIGNMENT

Knowing an individual's static lower-limb alignment is particularly important in investigations of frontal plane knee biomechanics. Static alignment influences the baseline adduction or abduction angle of the knee, which also influences the frontal plane moment arm lengths and moments at the knee (Weidenhielm et al., 1995). The "gold standard" method of determining static knee alignment is by measuring the medial angle of the intersection of the mechanical axes of the femur and tibia from an anterior full-limb radiograph (Sharma et al., 2001). The mechanical axis of the femur is formed by the line connecting the center of the femoral head to the midpoint of the femoral intercondylar notch. The mechanical axis of the tibia is formed by the line connecting the center of the talus to the midpoint of the tibial spines (Chao et al., 1994). If the medial angle formed by the intersection of the femoral and tibial mechanical axes is ≤178°, 179-181°, or ≥182°, the alignments are considered varus, neutral, and valgus, respectively (Sharma et al., 2010). Although it is the "gold standard," this method exposes the pelvis and leg to radiation, making it costly in terms of both health and finance (Hinman et al., 2006). As a result, several studies have investigated alternative methods of determining static alignment using 3D motion capture (Vanwanseele et al., 2009) and physical examinations (Bennett, 2016; Cibere et al., 2004; Hinman et al., 2006; Kraus et al., 2005; Navali et al., 2012).

Of these non-radiographic methods, prediction of joint centers and mechanical axis angles (MAA) using 3D motion capture has been shown to have the highest correlation

(Pearson's correlation coefficient (r) = 0.934) with the radiographic MAA (Navali et al., 2012; Vanwanseele et al., 2009). However, motion capture technology is likely not available in most clinical settings, so physical examination is a more realistic method, though less accurate. Furthermore, physical examination can be used to prescreen individuals for alignment, and motion capture or radiography can be used later to determine and confirm the true alignment. The simplest physical examination is called the Magee method (Magee, 2014), and it has been shown to have a moderate (Spearman's rho = -0.54; $P < 0.001$) relationship with the radiographic MAA (Hinman et al., 2006). This method has the patient adduct their lower limbs until either the knees or ankles contact. If they contact simultaneously, the patient is neutral; if the knees contact first, the patient is valgus; and if the ankles contact first, the patient is varus (Magee, 2014). This method requires no tools, but it only provides a rough qualitative estimation of alignment and is subject to the amount of soft tissue surrounding the joints. For a quantitative clinical measurement of alignment, several other methods may be used: caliper, plumb-line, goniometer, and inclinometer (Hinman et al., 2006).

The caliper method determines the remaining intercondylar or intermalleolar distance of the joint that was left uncontacted by the Magee method, and it is strongly correlated with the radiographic MAA ($r = 0.76$, $P < 0.001$) (Cibere et al., 2004; Hinman et al., 2006). The plumb-line method is similar to the caliper method except that the distance between the uncontacted joints are measured to a centralized plumb-line, rather than to the contralateral joint (Jonson and Gross, 1997). This method is also highly reliable and strongly correlated ($r = 0.71$, $P < 0.001$) with the radiographic MAA, and it considers the alignment of each limb separately (Hinman et al., 2006). The goniometer method takes the tibiofemoral angle by positioning the axis of a long-arm goniometer over the center of patella and one arm along the center of the thigh and the other along center of the patella tendon. This method has been shown to have both no relationship ($r = 0.32$, $P = 0.12$) (Hinman et al., 2006) and a strong relationship ($r = 0.70$, $P < 0.001$) (Kraus et al., 2005) with the radiographic MAA. The final quantitative clinical method and strongest relationship with the MAA according to Hinman et al. (2006), is the tibial inclination method ($r = 0.80$, $P < 0.001$). The tibial inclination method uses a gravity inclinometer attached to calipers to measure the frontal plane inclination of the tibia. The points of the caliper arms are placed on the neck of the talus and the tibial tuberosity and the gravity inclinometer is read to measure the tibia's frontal plane angle relative to the vertical (Hinman et al., 2006). A later study supports this strong relationship ($r = 0.831$, $P < 0.001$) between tibial inclination and the radiographic MAA, although this method only accounts for tibial alignment and does not consider the orientation of the femur (Vanwanseele et al., 2009). This method may also be subject to error due to variations in stance width.

CHAPTER III: MATERIALS AND METHODS

Participants

Fifteen healthy adults (7 Males, 8 Females, age: 22.73±2.55 years, height: 1.718±0.081 meters, body mass: 70.929±12.602 kilograms, BMI: 23.953±3.207; Mean±SD) were included in this study. These participants were physically active, engaging in moderate intensity physical activity for at least 30 minutes per day and 3 days per week, had no lower limb injuries within the past 6 months, and no history of severe lower limb musculoskeletal injuries (any muscle, ligament, or bone injury requiring surgical intervention) or diseases (e.g. osteoarthritis). Participants were recruited among a university population through posted flyers, emails, and in-person advertisement. A minimum sample size was calculated *a priori* using a QF main effect on peak internal knee abduction moment (KAbM) (η^2 = 0.721) from previous literature (Thorsen et al., 2020) with an α level of 0.05, β level (power) of 0.80, and calculated Cohen's f effect size of 1.607. The results of this power analysis (G*Power 3.1.9.7; Heinrich Heine University Dusseldorf, Dusseldorf, Germany) estimated a minimum sample size of 2. A post hoc power analysis confirmed that this study was sufficiently powered, and another a priori power analysis using the η^2 of the current study estimated a minimum sample size of 4. A written document of informed consent, approved by the University of Tennessee Institutional Review Board, was reviewed and signed by each participant prior to testing.

Instrumentation

Motion capture

A 13-camera motion capture system (240 Hz; Vicon Motion Analysis Inc., Oxford, UK) captured 3D kinematic data of the participants. During data collection, participants wore compression spandex shorts, a tight shirt and/or sports bra, and standardized lab shoes (Air

Pegasus, Nike, Beaverton, OR, USA). Reflective markers were placed bilaterally on the following anatomic landmarks for joint center and segment demarcation: acromion process, iliac crest, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, heads of the 1st and 5th metatarsals, and tip of the 2nd toe. Rigid shells with four fixed, non-colinear reflective markers were fixed to the trunk, pelvis, thighs, shanks, and shoes for dynamic segment tracking.

Reflective markers were also placed on the front of the cycle ergometer and bilaterally on the crank axes and pedals. Specifically, on each pedal, a rigid three-marker cluster was attached to the lateral aspect and a single wand marker was mounted pointing inferiorly and laterally to the anterior aspect of each pedal for dynamic tracking of the pedals. An additional marker was placed centrally on the anterior surface of each pedal to define anterior orientation of pedal.

Cycle Ergometer and Instrumented Pedal

The participants rode a stationary bike (Excalibur Sport, Lode, Groningen, Netherlands) with custom adjustable crank arms and a pair of custom instrumented pedals for experimental testing (Figure 1). The customized adjustable crank arms allowed for continuous changes in QF using three different sized blocks for mounting the pedals on the crank arms (small, medium, large; Figure 2). Any conditions with a QF at or below 172 mm was achieved using the small block, a QF between 173mm and 232 mm was achieved using the medium block, and a QF at or above 233 mm was achieved using the large block. The bike was aligned with the global lab coordinate system using a custom jig fixed to a floor-mounted force platform. The vertical position of the bike saddle was set so that the participant's right knee was flexed to between 25° and 30° when the pedal was at bottom dead center (BDC) (Holmes et al., 1994). The fore-aft

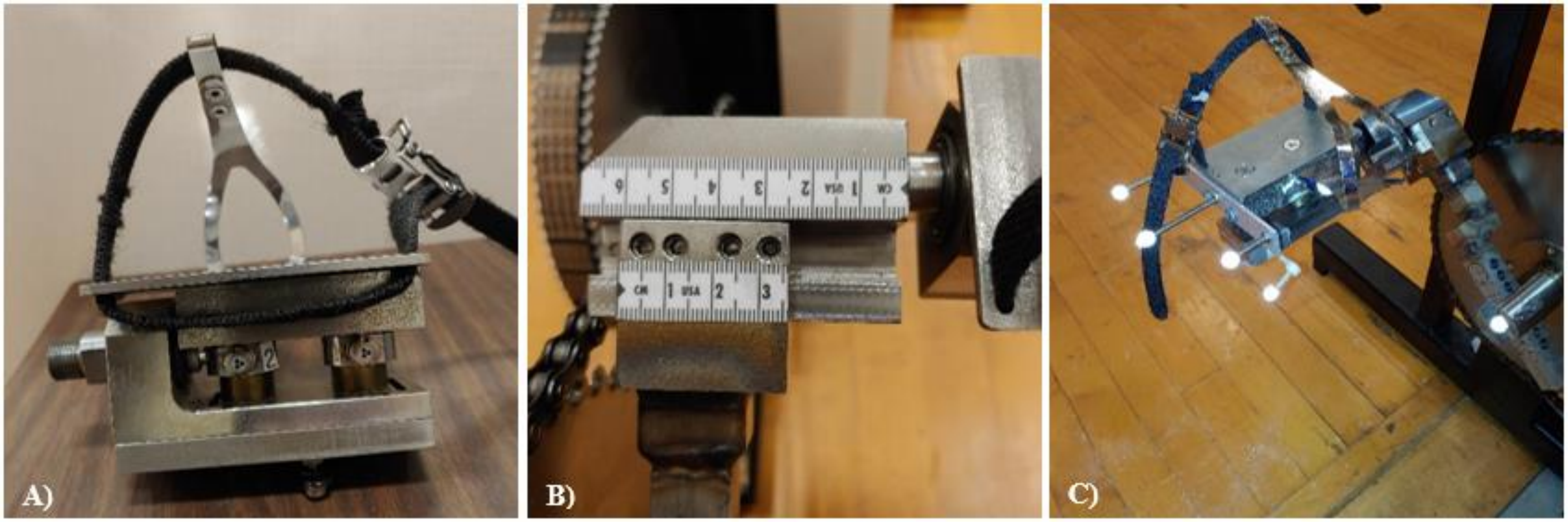

Figure 1: Photographs of the custom crank arm and pedal assembly. A) Posterior view of the right instrumented pedal. B) Adjustable pedal mount for changing Q-Factor with large mounting block. C) Complete assembly of pedal and crank arms on cycle ergometer.

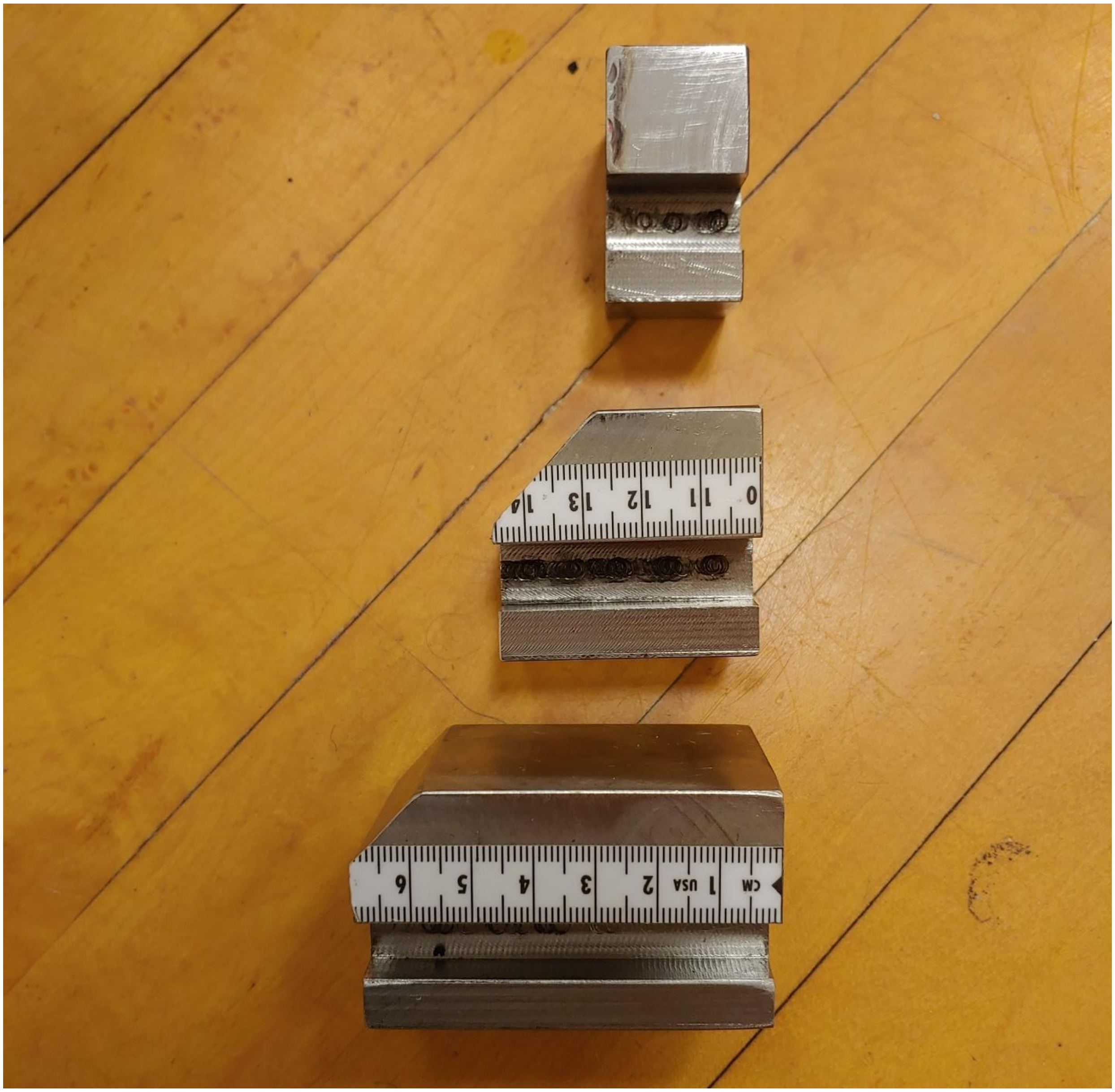

Figure 2: Image of the right small (top), medium (middle), and large (bottom) pedal mounting blocks used with the custom crank arms. The right pedal is screwed into the right face of each block.

position of the saddle was set where the participant's knee was aligned vertically with the pedal spindle, confirmed using a plumbline, when the crank arm was at the 90° position (Burke, 2002). The position of the handlebar was set where the angle between the trunk and thigh segments was 90° when the crank arm was at 90° (Thorsen et al., 2020). Angular measurements of the hip and knee during bike fitting were confirmed with a standard goniometer, and the crank position was determined visually when the pedal and foot were at their lowest and most forward positions for BDC and 90°, respectively.

Two custom force-instrumented bicycle pedals were used to measure pedal reaction forces (PRF) during experimental testing. Each pedal contained two triaxial piezoelectric force transducers (Type 9027C, Kistler, Winterthur, Switzerland), and analog signals from these sensors were amplified (Type 5073A, Kistler, Winterthur, Switzerland). These signals were temporally aligned and sampled simultaneously at 1200 Hz with the motion capture data using Vicon Nexus (Version 2.12, Vicon Motion Analysis Inc., Oxford, UK) of the motion capture system. Each pedal was affixed with a toe cage and strap to minimize relative motion between foot and pedal. The Q-Factor was measured as the horizontal distance between the medial edge of the pedals.

Experimental Protocol

Participants were screened for inclusion prior to data collection using questionnaires. Upon arrival, participants changed clothes and donned a tight shirt and/or sports bra, standardized compression shorts, and standardized lab shoes. Their height and weight were taken on a standard stadiometer, their standing leg length (L) was taken between the greater trochanter and the floor, while shod, using a meterstick (Johnson Level & Tool Mfg. Co. Inc.; Milwaukee, WI, USA) (Donelan et al., 2001), and the width between their anterior superior iliac spine (ASIS) was taken using analog linear calipers (Anthropometer Model 01291; Lafayette Instrument Company; Lafayette, IN, USA). These leg lengths were later used to prescribe normalized QF conditions, and the ASIS width was used later to position the participant's feet during the static capture for motion capture calibration. Participants were initially screened for static knee alignments using the Magee method (Magee, 2014), where the alignment was determined visually by observing whether the ankles or knees contacted first when one leg was eccentrically adducted from a slightly abducted hip position while standing. If the ankles contacted first, the person was deemed having a varus alignment, and if the knees contacted first, the person was deemed having valgus alignment. The individual whose ankles and knees contacted simultaneously was deemed neutrally aligned. This initial assessment of lower limb alignment was further quantified using a caliper method (Navali et al., 2012). A digital spring caliper (Fred V. Fowler Company, Inc., Newton, MA, USA) was used to measure the remaining distance between the medial malleoli for valgus participants or femoral condyles for varus participants to the nearest tenth of a millimeter during the Magee test. This distance was recorded as a positive number for the inter-malleolus distance and negative for the inter-femoral epicondyle distance. Based on the regression equation reported by Navali et al. (2012), the MAA was estimated using Equation 1:

$$MAA = 0.125 * (\text{caliper reading}) + 177.333 \qquad (1)$$

Since these methods are less accurate and cannot account for the alignment of each limb individually, alignments were ultimately determined using the standardized static trial during motion capture (Vanwanseele et al., 2009). Furthermore, these two clinical assessments were only used in comparison with the motion capture estimates outside of the final study.

Anatomic and tracking reflective markers were affixed to the participant and the cycle ergometer, and a static motion capture trial was taken which was used to calibrate the anatomic and tracking markers as well as to later analyze static frontal plane knee alignment. The participants stood on a single force platform with their feet parallel at approximately the width of their ASIS width with their arms folded across their chest. A large piece of paper with a series of parallel lines was placed underneath of both feet to guide the stance width and ensure proper anteroposterior alignment of feet. The feet were placed on the two lines that were closest to the ASIS width and aligned so that the lines ran from the middle of the heel through the 2nd toe.

Subsequent to the static trial, the participant's anatomical markers were removed, leaving only the cluster markers for the dynamic trials. They began with a two-minute warm-up ride at 80 W and 160 mm QF, and they were given at least two minutes of rest after the warm-up before testing began (Shen et al., 2018; Thorsen et al., 2020). Each participant completed five tests at five different experimental QF conditions: Q1 (160 mm), Q2 (160 mm + 0.02L), Q3 (160 mm + 0.04L), Q4 (160 mm + 0.06L), and Q5 (160 mm + 0.08L), where L is each participant's leg length in millimeters. The QF conditions were randomized in two different steps. The order of pedal mounting block sizes was first randomized, then the order of the QF conditions performed within each mounting block were randomized. Q1 was always set with the small block. Depending on leg length, Q2 through Q5 were achieved using the medium block for some participants, while Q2 through Q4 were achieved using the medium block and Q5 using the large block for taller participants ($L \geq 925$ mm). Each condition was changed by 2% of L (to the nearest millimeter) from its surrounding conditions, so an individual with a 1-meter trochanteric leg length would have QF conditions of: 160, 180, 200, 220, and 240 mm. The average QF change between conditions for all participants was 18 mm. These increments are smaller than what has been shown to have a significant effect on KAbM in previous literature (Thorsen et al., 2020). These small, normalized changes in QF tested if KAbM is sensitive to smaller changes in QF than 42 mm (Thorsen et al., 2020). Each experimental QF condition was tested while cycling at 120 W and 80 RPM for two minutes, and participants were given a two-minute rest period between conditions (Thorsen et al., 2020). Kinematics and kinetics during the final 10 seconds of each condition were recorded, and data from five consecutive crank cycles were truncated from the 10 seconds of collected data for further analysis (Shen et al., 2018; Thorsen et al., 2020). Upon completion of each condition, participants were asked to give a rating of perceived effort (RPE; 6-20 Borg scale) (Borg, 1998).

Data Processing and Analysis

Static and dynamic trial marker coordinates were manually labelled with a custom marker set in Vicon Nexus 2.12 (Vicon Motion Analysis Inc., Oxford, UK). The 10 seconds of dynamic trial data were cropped to five consecutive crank cycles of good quality data. Good quality data were considered when there were no visible abnormalities in the participant's movement, minimal gaps in kinematic tracking, and PRF data was present and continuous. Dynamic trial marker labels were confirmed and any gaps in marker coordinate data were filled using a rigid body fill or pattern fill. One trial extended from when the left crank arm was at 1 o'clock through a full cycle until the right crank arm reached its 1 o'clock, so each trial contained 1 full revolution of each crank arm beginning and ending at the 12 o'clock position with a time buffer at the beginning at end.

The data were exported as C3D files to Visual3D (Version 6, C-Motion Inc., Germantown, MD, USA) for kinematic and kinetic computation and analysis. Marker coordinate data and analog pedal reaction force data were filtered using a fourth-order, zero-lag, low-pass Butterworth filter with a cutoff frequency of 6 Hz (Gardner et al., 2015; Thorsen et al., 2020). A crank cycle of pedaling movement was defined as a full revolution of the crank arm with the beginning (0°) and end (360°) at the top dead center (TDC) position. The forwardmost position, BDC and backmost position were defined as 90°, 180°, and 270°, respectively. Sagittal and frontal plane ranges of motion (ROM), peak angles, and peak moments were calculated for the right knee over a full crank cycle. Angular kinematics and kinetics were computed using the joint coordinate system (Grood and Suntay, 1983) and expressed using an XYZ Cardan sequence and right-hand-rule, such that right knee extension (x-axis), adduction (y-axis), and internal rotation (z-axis) were positive. Joint moments were expressed as the internal moments. The PRF was expressed as the resultant and as each of its 3D components, such that positive X, Y, and Z vectors were respectively directed laterally, anteriorly, and superiorly, from the right pedal. All variables of interest were time-normalized to a full crank cycle (0-360°) to account for any variations in pedaling timing and cadence.

The medial hip-knee-ankle (HKA) angle was found as the deviation from 180° of the frontal plane knee joint angle in Visual 3D. The hip joint center location was offset 23.4% medially and 4.7% superiorly of the intertrochanteric distance from the ipsilateral greater trochanter marker location, corrected for the radius of the marker and thickness of marker base (Bennett et al., 2016). The knee joint center was defined as the midpoint between the medial and lateral femoral epicondyle markers, and the ankle joint center was defined as the midpoint between the medial and lateral malleolus markers. The calculated HKA angle was expressed as its deviation from 180° (HKA deviation = HKA – 180; adduction = negative; abduction =

positive) to estimate the MAA deviation from 180 using Equation 2, based on a regression analysis from previous literature (Vanwanseele et al., 2009).

$$\text{MAA deviation} = -4.05 + 1.05 * \text{HKA deviation} \qquad (2)$$

Finally, the MAA deviation was added to 180° to obtain the full MAA.

The selected variables of interest included sagittal, frontal, and transverse knee peak angles, ROMs, and peak moments as well as the sagittal and frontal ankle and hip peak moments.

Statistical Analysis

A Shapiro-Wilks test was used to test the data for normal distribution. A mixed model analysis of variance (ANOVA) was used to investigate differences in the variables of interest between QF conditions (IBM SPSS 28, Chicago, IL). A mixed model analysis of covariance (ANCOVA) was used to investigate differences in the variables of interest, using MAA as the covariate, and Pearson correlation coefficients were found between MAA and variables of interest. The α level was set at 0.05 *a priori*. If a main effect of QF was detected, pairwise comparisons were made *post hoc* with Bonferroni adjustments. The results of the ANOVA and ANCOVA were then compared to determine the effect of MAA. Since MAA did not have any meaningful effects on the key loading variables, only the ANOVA results were reported hereafter.

CHAPTER IV:

Effects of Small and Normalized Q-Factor Changes and Knee Alignment on Knee Biomechanics During Stationary Cycling

Abstract

Increasing inter-pedal distance (Q-Factor; QF) in cycling increases peak internal knee abduction moments (KAbM). The effect of smaller, normalized changes in QF has not been investigated, and the effect of static knee alignment at varying QFs is unknown. **Purpose:** The primary purpose of this study was to see if significant changes in KAbM were detectable with normalized increases in QF that are smaller than what has previously been investigated. The secondary purpose of this study was to investigate whether static knee alignment accounts for any changes in knee biomechanics while cycling at different QFs. **Methods:** Fifteen healthy participants were included in this study (7 Males, 8 Females, age: 22.7±2.5 years, BMI: 23.95±3.21 kg/m^2; Mean±STD). Motion capture and instrumented pedals were used to collect kinematic (240 Hz) and pedal reaction force (PRF, 1200 Hz) data, respectively, while cycling at five different QFs. The participant's mechanical axis angle (MAA) was determined using motion capture. Each participant's QFs were normalized by starting at 160 mm and increasing by 2% of the participant's trochanteric leg length (L) where the five QF conditions were (in mm): Q1 (160), Q2 (160 + 0.02*L), Q3 (160 + 0.04*L), Q4 (160 + 0.06*L), and Q5 (160 + 0.08*L). A mixed model analysis of variance was performed to detect differences between QF conditions ($\alpha = 0.05$). Correlation was calculated between MAA and select variables. **Results:** KAbM was increased by at least 30% in Q5 from Q1, Q2, Q3, and Q4. Medial PRF was increased by at least 20% in Q5 from Q1, Q2, and Q3. There were no significant changes seen in peak vertical PRF, sagittal-plane moments and angles, or peak abduction angle that were concurrent with significant changes in KAbM. MAA had varying degrees of correlation with the variables of interest. **Conclusions:** These results suggest that KAbM is more sensitive to changes in QF at greater QF increases. The effect of MAA on frontal-plane knee biomechanics requires further investigation.

Introduction

Stationary cycling is a low-impact form of aerobic exercise that has been shown to reduce the symptoms of knee OA (Luan et al., 2021; Mangione et al., 1999; Salacinski et al., 2012). Peak tibiofemoral contact forces during stationary cycling have been shown to be 1 - 1.5 times bodyweight (BW), compared to 1.8 - 2.5 BW during level walking and 4.2 BW during jogging (D'Lima et al., 2008). The seat of the bike supports most of the body's weight, so the knee is relatively unloaded when compared to walking and jogging (Kutzner et al., 2012). Therefore, individuals with knee OA may find stationary cycling to be a more feasible form of aerobic exercise than its higher impact alternatives. Obesity is a leading risk factor of knee OA (Felson et al., 1988; Jiang et al., 2012), so being able to exercise is important to this population because aerobic exercise is an effective prescription for weight loss.

Although cycling is low impact, there is still inherent knee loading involved, so it is important to minimize harmful knee contact forces where possible. As *in vivo* measurements of tibiofemoral contact forces are quite difficult to obtain, internal knee extension moment (KEM) and internal knee abduction moment (KAbM) are two commonly used surrogate variables. Generally, the KEM is reflective of total contact force (TCF) in the knee while the frontal plane moment is thought to influence the mediolateral distribution of the TCF (D'Lima et al., 2006; Zhao et al., 2007). Greater KEM and KAbM have been shown to increase TCF and medial compartment contact force (MCF) respectively, in gait (Richards et al., 2018) and cycling (Thorsen et al., 2021). Greater contact forces increase the risk for development and progression of knee OA in the corresponding area, as is the case with obesity (Felson et al., 1988; Jiang et al., 2012).

Increased step width, often evaluated as a percent of an individual's leg length, is a gait modification that has been investigated as a means of reducing KAbM, thereby potentially reducing MCF and the risk of medial compartment knee OA (Fregly et al., 2008; Paquette et al., 2015, 2014). Analogous to step width in gait, the Q-Factor (QF) of a bicycle or cycle ergometer is the horizontal distance between the lateral surface of each crank arm. Therefore, QF influences the mediolateral positioning of the pedals and feet and, subsequently, the frontal plane angles and moments of the lower limbs (Thorsen et al., 2020). However, in contrast to increased step width gait modifications, greater QFs increased KAbM (Thorsen et al., 2020) and MCF (Thorsen et al., 2021), indicating less preferable loading for individuals with medial compartment knee OA. In this study (Thorsen et al., 2020), QF was incrementally increased by 42 mm, the width of one commercial pedal extender on each pedal, from 150 mm. It has not yet been shown whether smaller and normalized increases in QF would also induce significant increases in KAbMs. Small changes in loading may not be influential in singularity, but, due to the repetitive nature of cycling, small changes can accumulate over time to result in substantial differences in cumulative loading (Gatti and Maly, 2019; Kumar, 1990). So, it is important to understand the extent to which QF affects KAbM as to avoid unnecessary knee joint loading.

Static lower limb alignment influences the mediolateral position of the knee relative to the placement of the foot, which may also affect knee frontal plane angles and moments. This alignment is determined by the orientations of the weight bearing mechanical axes of the femur and tibia. If the medial angle formed by the intersection of these two axes, known as the mechanical axis angle (MAA), is ≤178°, between 178° and 182°, or ≥182°, the alignments are considered to be varus, neutral, and valgus, respectively (Sharma et al., 2010). These axes and angles are most accurately determined using full-limb, standing radiography, but hip-knee-ankle angles obtained with 3D motion capture were shown to have a high correlation (Pearson's correlation coefficient = 0.934) with radiographic MAAs and can be used to predict such MAAs through a regression model (Vanwanseele et al., 2009).

Greater KAbM and peak knee adduction angles (Bennett et al., 2017a) as well as peak TCF (Heller et al., 2003) have been seen in individuals with varus alignment during walking, compared to those with neutral and valgus alignments. This effect was also recently investigated during cycling. It was found that peak knee adduction angle, but not KAbM, was significantly greater in varus participants, compared to neutral and valgus participants (Shen et al., 2018). It was also noted in this study and a study by Fang et al. (2016) that participants exhibited either a KAbM or an internal knee adduction moment during the power phase of cycling. Shen et al. (2018) suggest that knee alignment may be responsible in part for this observation, as 90.9% of their varus participants, 72.7% of the neutral participants, and only 50% of the valgus participants exhibited KAbM.

To our knowledge, there have been no previous studies that investigate the effect of QF changes less than 42 mm on KAbM, and the sensitivity of KAbM to smaller but normalized changes (relative to leg length) in QF is presently unknown. Additionally, there have been no studies that investigate static knee alignment as a covariate in the effects of QF on frontal-plane knee moments during ergometer cycling. It is unknown whether static knee alignment will account for any variation in KAbM at different QFs. Therefore, the primary purpose of this study was to investigate whether significant changes in KAbM are detectable with smaller and normalized changes in QF. The secondary purpose of this study was to investigate the relationship between static knee alignment and KAbM. It was hypothesized that KAbM would be greater with each normalized increase in QF. It was also hypothesized that the increases in

KAbM would be even greater as MAA decreases, indicating a relationship between static knee alignment and frontal plane knee moments.

Materials and Methods

Participants

Fifteen adults between 18 and 35 years of age (7 Males, 8 Females, age: 22.7±2.5 years, height: 1.71±0.08 m, body mass: 70.93±12.60 kg, BMI: 23.95±3.21 kg/m^2; Mean±SD) participated in this study. All participants were physically active, engaging in at least 30 minutes of moderate intensity exercise three days per week. All participants were free from lower extremity injury within 6 months of their inclusion of the study, and they were free from any history of musculoskeletal disease (e.g. Osteoarthritis) and severe lower extremity injury requiring surgical intervention. A minimum sample size was calculated *a priori* using a QF main effect on peak KAbM ($\eta^2 = 0.721$) from previous literature (Thorsen et al., 2020) with an α level of 0.05, β level (power) of 0.80, and calculated Cohen's f effect size of 1.607. The results of this power analysis (G*Power 3.1.9.7; Heinrich Heine University Dusseldorf, Dusseldorf, Germany) estimated a minimum sample size of 2. Post hoc power analysis confirmed that this study was sufficiently powered. A written document of informed consent, approved by the University of Tennessee Institutional Review Board, was reviewed and signed by each participant prior to testing.

Instrumentation

A 13-camera motion capture system (240 Hz; Vicon Motion Analysis Inc., Oxford, UK) captured three-dimensional (3D) kinematic data of the participants. During data collection, participants wore compression spandex shorts, a tight shirt and/or sports bra, and standardized lab shoes (Air Pegasus, Nike, Beaverton, OR, USA). Reflective markers were placed bilaterally over clothing and shoes at the following anatomical landmarks for segment demarcation: acromion process, iliac crest, greater trochanter, medial and lateral femoral epicondyles, medial and lateral malleoli, heads of the 1st and 5th metatarsals, and tip of the 2nd toe. Rigid shells with four fixed, non-colinear reflective markers were fixed to the trunk, pelvis, thighs, shanks, and shoes for dynamic segment tracking. Reflective markers were also placed on the front of the cycle ergometer, the crank axis, and, to each pedal, a rigid three-marker cluster was attached to the lateral aspect and a single wand marker was mounted to the anterolateral aspect. An additional marker was placed centrally on the anterior surface of both pedals to define their anterior orientation.

During testing, the participants rode a Lode cycle ergometer (Excalibur Sport, Lode, Groningen, Netherlands) with custom adjustable crank arms and a pair of custom instrumented pedals (Figure 3). The customized adjustable crank arms allowed for continuous changes in QF using three different sized blocks for mounting the pedals on the crank arms (small, medium, large; Figure 2). Any conditions with a QF at or below 172 mm was achieved using the small block, a QF between 173mm to 232 mm was achieved using the medium block, and a QF at or above 233 mm was achieved using the large block. Each pedal contained two triaxial piezoelectric force transducers (Type 9027C, Kistler, Winterthur, Switzerland) for measurement of the pedal reaction force (PRF; 1200 Hz). The bike was aligned with the global lab coordinate system using a custom jig fixed to a floor-mounted force platform so that the ergometer's crank axis was parallel to the lab's mediolateral axis. The vertical position of the saddle was adjusted so that the participant's knee was flexed to between 25° and 30° when the pedal was at bottom dead center (BDC) (Holmes et al., 1994). The fore-aft position of the saddle was set where the

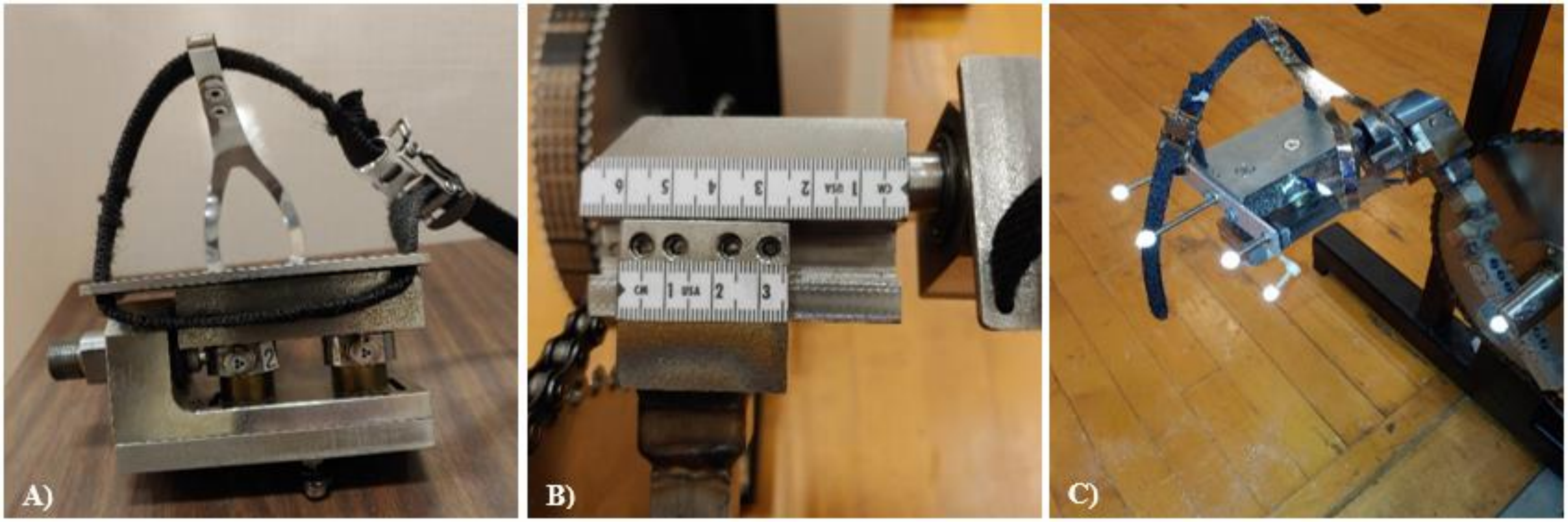

Figure 3: Images of the custom pedal and crank arm assembly. A) Right custom force-instrumented pedal, B) Custom crank arm pedal mount with large mounting block, C) Custom crank arm with reflective markers on Lode ergometer.

participant's knee was aligned vertically with the pedal spindle, confirmed using a plumbline, when the crank arm was at the 90° position (Burke, 2002). The position of the handlebar was set where the angle between the trunk and thigh segments was 90° when the crank arm was at 90° (Thorsen et al., 2020). Angular measurements of the hip and knee during bike fitting were confirmed with a standard goniometer, and the crank position was determined visually when the pedal and foot were at their lowest and most forward positions for BDC and 90°, respectively.

Experimental Protocol

Prior to experimental testing, the participants' standing leg length (L) was taken between the greater trochanter and the floor, while shod, using a meter stick (Johnson Level & Tool Mfg. Co. Inc.; Milwaukee, WI, USA) (Donelan et al., 2001), and the width between their anterior superior iliac spine (ASIS) was taken using analog linear calipers (Anthropometer Model 01291; Lafayette Instrument Company; Lafayette, IN, USA). A ruled foot position template was placed underneath the participant during a static calibration capture so that the lines closest to the width of their ASIS ran from the center of either heel through the 2nd toe of the same foot. This placed each persons' feet parallel and approximately beneath the weight bearing axis of the hips.

Participants began the experimental protocol with a two-minute warm-up ride, and they were given at least two minutes after the warm-up before testing began (Shen et al., 2018; Thorsen et al., 2020). Each participant completed five tests at five different experimental QF conditions: Q1 (160 mm), Q2 (160 mm + 0.02L), Q3 (160 mm + 0.04L), Q4 (160 mm + 0.06L), and Q5 (160 mm + 0.08L), where L is each participant's leg length in millimeters. The QF conditions were randomized in two different steps. The order of pedal mounting block sizes was first randomized, then the order of the QF conditions performed within each mounting block were randomized. Q1 was always set with the small block. Depending on leg length, Q2 through

Q5 were achieved using the medium block for some participants, while Q2 through Q4 were achieved using the medium block and Q5 using the large block for taller participants (L ≥ 925 mm). Each of the conditions was different by 2% of L (rounded to the nearest millimeter), so an individual with a 1-meter trochanteric leg length would have QF conditions of: 160, 180, 200, 220, and 240 mm. The average QF change between conditions for all participants was 18 mm. These increments are smaller than what has been shown to have a significant effect on KAbM in previous literature (42 mm) (Thorsen et al., 2020). Each condition was performed cycling at 120 W and 80 RPM for two minutes. Kinematics and kinetics during the final 10 seconds of each condition were recorded, and data from five consecutive crank cycles were truncated from the 10 second's collected data for further analysis (Shen et al., 2018; Thorsen et al., 2020). Participants were given at least two-minutes of rest between conditions (Thorsen et al., 2020). During this period, participants were asked for their rating of perceived exertion (RPE; 6-20 Borg scale) (Borg, 1998).

Data Processing and Analysis

Static and dynamic trial marker coordinates were manually labelled with a custom marker set in Vicon Nexus 2.12 (Vicon Motion Analysis Inc., Oxford, UK). The 10 seconds of dynamic trial data were cropped to five consecutive crank cycles of good quality data. Good quality data were considered when there were no visible abnormalities in the participant's movement, minimal gaps in kinematic tracking, and PRF data was present and continuous. Dynamic trial marker labels were confirmed and any gaps in marker coordinate data were filled using a rigid body fill or pattern fill.

Kinematic and kinetic computation was performed in Visual3D (Version 6, C-Motion Inc., Germantown, MD, USA). Marker coordinate data and analog pedal reaction force data were filtered using a fourth-order, zero-lag, low-pass Butterworth filter with a cutoff frequency of 6 Hz (Gardner et al., 2015; Thorsen et al., 2020). A crank cycle was defined as a full revolution of the crank arm with the beginning (0°) and end (360°) at the top dead center (TDC) position. The forwardmost position, BDC and backmost position were defined as 90°, 180°, and 270°, respectively. Sagittal, frontal, and transverse plane ranges of motion (ROM), peak angles, and peak moments were calculated for each participants' dominant side knee, ankle, and hip over a full crank cycle. Angular kinematics and kinetics of each participant's dominant leg, determined as the leg with which they would kick a soccer ball, were computed using the joint coordinate system (Grood and Suntay, 1983) and expressed using an XYZ Cardan sequence and right-hand-rule, such that right knee extension (x-axis), adduction (y-axis), and internal rotation (z-axis) were positive. Joint moments were expressed as the internal moments. The PRF was expressed as the resultant and as each of its 3D components, such that positive X, Y, and Z vectors were respectively directed laterally, anteriorly, and superiorly, from the right pedal. All variables of interest were time-normalized to a full crank cycle (0-360°) to account for any variations in pedaling timing and cadence.

The hip joint center location was offset 23.4% medially then 4.7% superiorly of the intertrochanteric distance from the ipsilateral greater trochanter marker location, corrected for the radius of the marker and thickness of marker base (Bennett et al., 2016). The knee joint center was defined as the midpoint between the medial and lateral femoral epicondyle markers, and the ankle joint center was defined as the midpoint between the medial and lateral malleolus markers. The calculated hip-knee-ankle (HKA) angle was expressed as its deviation from 180° (HKA deviation = HKA – 180°; adduction = negative; abduction = positive) to estimate the mechanical axis angle (MAA) deviation from 180° using Equation 1, based on a regression analysis from previous literature (Vanwanseele et al., 2009).

$$\text{MAA deviation} = -4.05 + 1.05 * \text{HKA deviation} \qquad (1)$$

Finally, the MAA deviation was added to 180° to obtain the MAA.

The selected variables of interest included sagittal, frontal, and transverse knee peak angles, ROMs, and peak moments as well as the sagittal and frontal ankle and hip peak moments.

Statistical Analysis

A Shapiro-Wilks test was used to test the data for normal distribution. A mixed model analysis of variance (ANOVA) was used to investigate differences in the variables of interest between QF conditions (IBM SPSS 28, Chicago, IL). A mixed model analysis of covariance (ANCOVA) was used to investigate differences in the variables of interest, using MAA as the covariate, and Pearson correlation coefficients were found between MAA and variables of interest. The α level was set at 0.05 *a priori*. If a main effect of QF was detected, pairwise comparisons were made *post hoc* with Bonferroni adjustments. The results of the ANOVA and ANCOVA were then compared to determine the effect of MAA. Since MAA did not have any meaningful effects on the key loading variables, only the ANOVA results were reported hereafter.

Results

The RPE responses were not significantly different across the QF conditions. The Q1 condition was the same for all participants at 160 mm. The mean QF values were 178±1 mm (mean±STD) for Q2, 196±2 mm for Q3, 214±3 mm for Q4, and 232±4 mm for Q5. Each of these QFs was significantly different from the others ($p < 0.001$ for all pairwise comparisons).

The ANOVA results found a significant QF effect on peak vertical PRF ($p = 0.01$, Table 1), and post hoc comparisons showed that vertical PRF was significantly greater for Q1 than Q2 and Q4 ($p \leq 0.017$ for both comparisons). The QF main effect was also significant for peak medial PRF ($p < 0.001$, Table 1). The post hoc comparisons found that it was significantly greater (more negative) in Q5 than Q1, Q2, and Q3 ($p < 0.001$ for all comparisons).

There was a significant QF main effect on knee abduction ROM ($p <0.001$, Table 1). The post hoc analyses found that it was significantly smaller for Q4 than Q1 ($p = 0.009$) and smaller for Q5 than Q1 and Q3 ($p \leq 0.008$ for both comparisons). A significant main effect of QF was found for knee external rotation ROM ($p = 0.007$, Table 1). Post hoc comparisons reveal that external rotation ROM was significantly less for Q5 than for Q1 and Q2 ($p \leq 0.046$ for both comparisons). Representative mean waveforms of these variables are shown in Figure 4.

There was a significant main effect of QF on peak KAbM ($p < 0.001$, Table 1), and the post hoc comparisons found the KAbM was significantly greater for Q5 than Q1, Q2, Q3, and Q4 ($p \leq 0.004$ for all comparisons). Additionally, a main effect of QF was found significant on peak knee internal rotation moment ($p < 0.001$, Table 1). The post hoc comparisons found that it was significantly greater for Q4 than Q1 ($p < 0.011$) and significantly greater for Q5 than Q1, Q2, and Q3 ($p \leq 0.001$ for all comparisons). Representative mean waveforms for these variables are shown in Figure 4.

The main effect of QF was significant for peak ankle inversion moment ($p < 0.001$, Table 2). Post hoc tests found that it was significantly smaller for Q3 than Q1 ($p = 0.008$), significantly smaller for Q4 than for Q1, Q2, and Q3 ($p \leq 0.023$ for all comparisons), and significantly less for

Table 1: Peak pedal reaction forces (N), knee angles and ROMs (°), and knee moments (Nm): Mean ± STD.

Variables	Q1	Q2	Q3	Q4	Q5	F	P
Vertical PRF	233.5±34.8	217.4±32.6 [a]	222.2±29.2	217.2±27.0 [a]	224.8±27.9	3.708	**0.010**
Medial PRF	-45.3±10.5	-45.7±7.8	-46.1±9.5	-50.1±10.2	-55.4±11.2 [a,b,c]	8.732	**<0.001**
Extension Angle	-33.1±6.4	-34.0±6.7	-33.2±6.5	-32.6±7.0	-32.6±6.3	1.767	0.148
Extension ROM	77.4±7.5	76.9±7.4	76.8±7.4	77.7±7.6	77.9±7.3	1.603	0.186
Abduction Angle*	0.83±4.17	1.19±3.99	0.76±4.00	0.48±4.19	0.15±3.93	1.799	0.145
Abduction ROM*	-5.7±2.6	-4.9±2.0	-5.2±2.5	-4.3±2.7 [a]	-3.8±2.5 [a,c]	7.364	**<0.001**
External Rotation Angle	-6.7±4.6	-7.4±3.9	-6.8±3.9	-6.7±3.7	-6.5±4.4	1.245	0.303
External Rotation ROM	-11.4±5.9	-11.1±5.5	-10.3±6.0	-10.1±5.6	-9.6±6.2 [a,b]	3.981	**0.007**
Extension Moment	34.7±7.2	35.2±4.8	34.9±5.6	35.1±5.9	38.1±6.8	2.421	0.590
Abduction Moment	-9.8±4.5	-9.6±4.1	-9.5±4.5	-10.6±4.0	-12.9±4.6 [a,b,c,d]	10.121	**<0.001**
Internal Rotation Moment	7.9±3.7	8.8±2.8	8.7±3.2	9.9±2.8 [a]	11.1±3.3 [a,b,c]	9.811	**<0.001**

[a]: Significantly different from Q1
[b]: Significantly different from Q2
[c]: Significantly different from Q3
[d]: Significantly different from Q4
*13 out of 15 participants displayed this pattern. The remaining participants are not included in the analysis of the variable. Values for peak knee abduction angle refer to the minimum angle, closest to an abducted position.

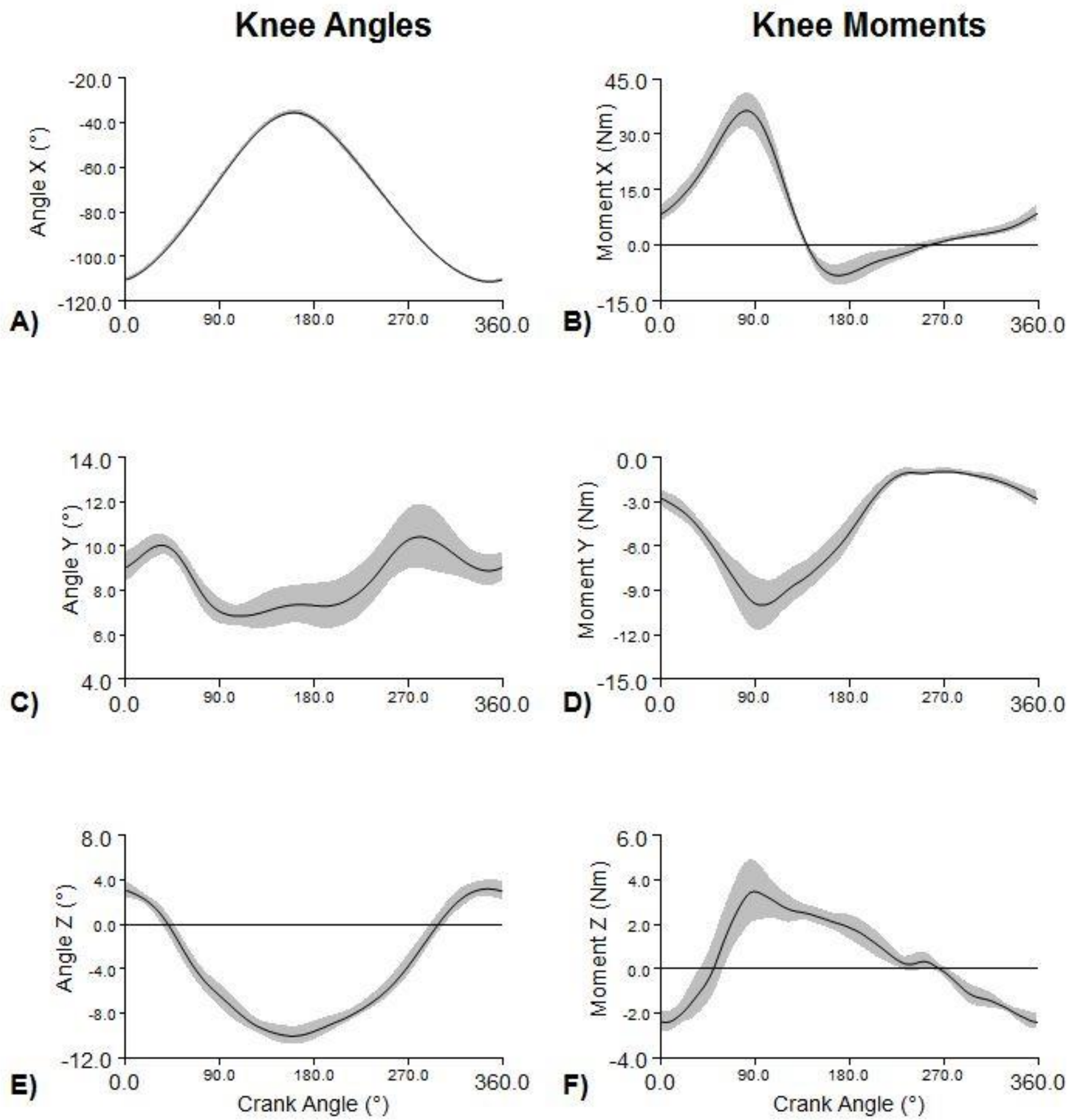

Figure 4: Representative mean knee joint angles and moments for A) sagittal plane knee angle, B) sagittal plane knee moment, C) frontal plane knee angle, D) frontal plane knee moment, E) transverse plane knee angle, and F) transverse plane knee moment. The bold line represents the mean value, and the shaded region represents 1 standard deviation. Positive values correspond to extension, adduction, and internal rotation angles and moments. X, Y, and Z refer to the sagittal, frontal, and transverse planes, respectively.

Table 2: Peak ankle and hip moments (Nm): Mean±STD.

Variables	Q1	Q2	Q3	Q4	Q5	F	P
Ankle Plantarflexion Moment	-15.8±4.2	-14.4±3.6	-14.3±3.0	-14.3±2.5	-14.5±3.6	1.642	0.177
Ankle Inversion Moment	1.25±0.95	0.95±0.57	0.75±0.49 [a]	0.21±0.62 [a,b,c]	0.27±0.46 [a,b]	16.313	**<0.001**
Hip Extension Moment	-18.1±9.2	-12.3±7.1	-16.0±9.3	-13.9±7.8	-12.5±6.1	2.786	**0.036**
Hip Abduction Moment	-18.2±6.8	-18.5±5.5	-18.3±6.7	-20.3±6.4	-24.2±7.7 [a,b,c,d]	8.519	**<0.001**

Negative values correspond to ankle plantarflexion and eversion moments and hip extension and abduction moments. Positive values indicate ankle dorsiflexion and inversion moments and hip flexion and adduction moments.

[a]: Significantly different from Q1

[b]: Significantly different from Q2

[c]: Significantly different from Q3

[d]: Significantly different from Q4

Q5 than for Q1 and Q2 ($p < 0.001$ for both comparisons). There was a significant main effect of QF on peak hip extension moment ($p = 0.036$, Table 2), but post hoc comparisons did not detect any specific differences between conditions. A significant main effect of QF was also found for peak hip abduction moment ($p < 0.001$, Table 2), and post hoc comparisons showed that the moment was significantly greater for Q5 than for Q1, Q2, Q3, and Q4 ($p \leq 0.017$ for all comparisons). Representative mean waveforms for these variables are shown in Figure 5.

There was significant correlation found between MAA and vertical PRF at Q4 only ($r = -0.561$, $p = 0.03$; Table 3) as well as between MAA and knee abduction ROM at Q5 only ($r = -0.580$, $p = 0.038$; Table 3). All correlations in other conditions and between other variables were nonsignificant, and they varied in degree from moderate to low correlation.

Discussion

The main purpose of this study was to investigate whether significant changes in KAbM are detectable with smaller and normalized changes of QF. The primary hypothesis that KAbM would become greater with increased QFs was partially supported by the results of this study. Although each increase from one QF to the next were equal in magnitude, not all changes in QF resulted in significantly increased KAbM. Interestingly, the peak KAbM in Q1 through Q4 were not statistically different, but the peak KAbM for Q5 was statistically different from all other conditions ($p \leq 0.004$ for all comparisons). With a mean QF difference (2% of leg length) of 18 mm between conditions across all participants, these comparisons equated to average differences in QF of 18 mm (Q4-Q5; range: 16-20 mm), 36 mm (Q3-Q5; range 32-40 mm), 54 mm (Q2-Q5; range 48-60 mm), and 72 mm (Q1-Q5; range 64-80 mm). The most relevant of these comparisons is the significant difference between Q4 and Q5, where peak KAbM increased by

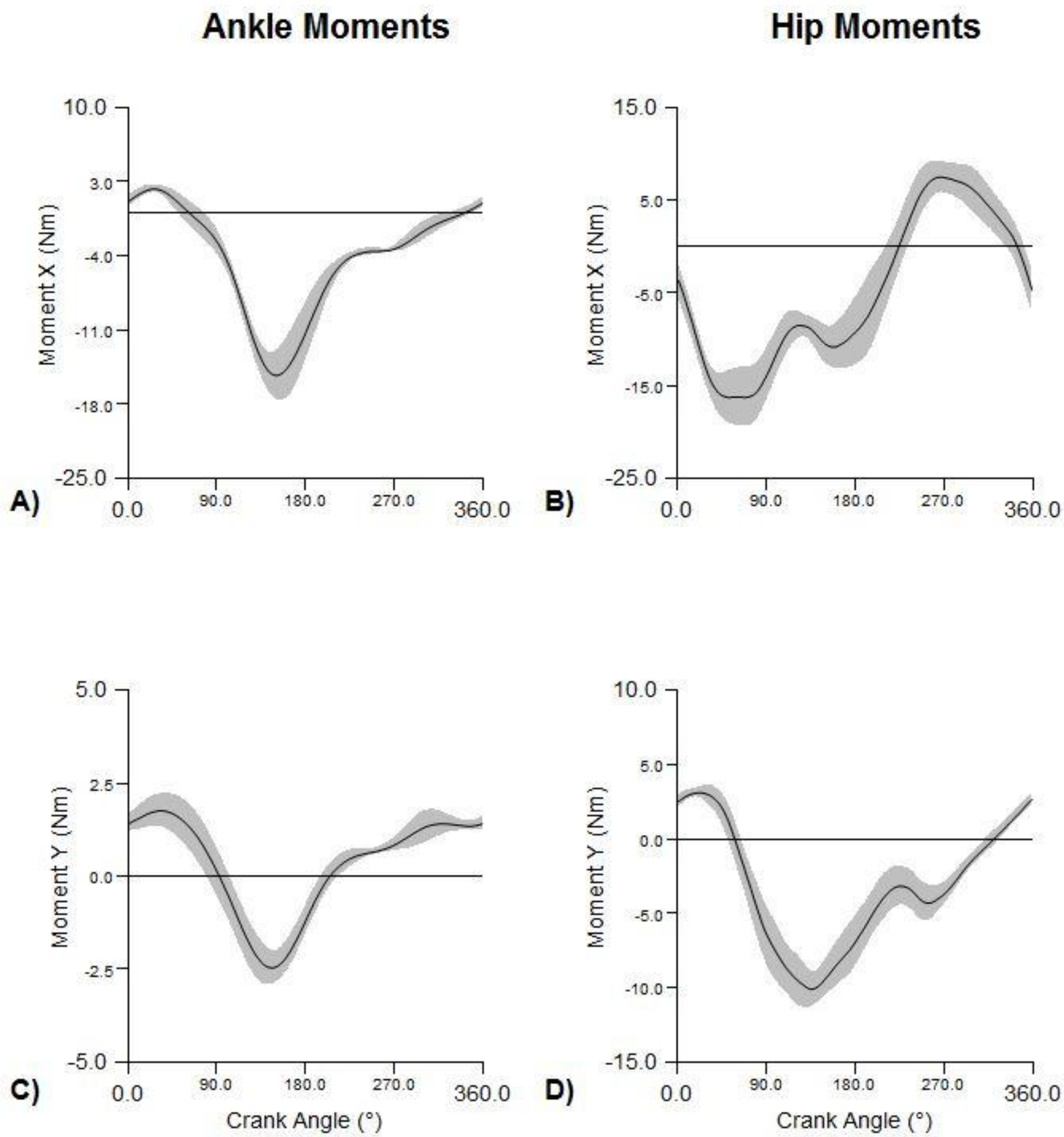

Figure 5: Representative mean ankle and hip joint moments for A) sagittal plane ankle moment, B) sagittal plane hip moment, C) frontal plane ankle moment, and D) frontal plane hip moment. The bold line represents the mean value, and the shaded region represents 1 standard deviation. Positive values correspond to dorsiflexion/flexion and inversion/adduction moments, and X and Y refer to sagittal and frontal planes, respectively.

Table 3: Pearson correlation coefficients between MAA and select variables.

Variables	Q1	Q2	Q3	Q4	Q5
Vertical PRF	-0.358	-0.279	-0.455	-0.561*	-0.340
Medial PRF	0.064	0.228	0.419	0.432	0.212
Peak Knee Abduction Angle	-0.274	-0.348	-0.390	-0.391	-0.418
Knee Abduction ROM	-0.474	-0.374	-0.304	-0.262	-0.580*
Peak Knee Abduction Moment	-0.055	-0.100	-0.038	-0.197	-0.086
Peak Knee Extension Moment	-0.090	0.073	0.346	0.377	0.210
Peak Knee Internal Rotation Moment	0.194	0.309	0.204	0.350	0.325
Peak Ankle Inversion Moment	-0.424	-0.430	0.007	-0.222	0.003
Peak Hip Abduction Moment	-0.212	-0.350	-0.477	-0.502	-0.340

*: $p < 0.05$

21.7%. In terms of absolute QF change, the difference between Q4 and Q5 is a considerably smaller difference in QF than has previously been shown to cause significant increases in KAbM. Thorsen et al. (2020) used 42 mm increments in QF at the same workrate (120W) and cadence (80 rpm) as the present study and found significant increases in KAbM with each incremental increase in QF from 150 mm to 276 mm. The results of the present study indicate that it is possible to see a significant increase in KAbM with as small of a change as 2% of leg length, which ranged from 16 to 20 mm for the participants in this study. However, the fact that other QF changes of equal and greater magnitude did not cause significant changes in KAbM should not be ignored.

A similar phenomenon was observed by Thorsen et al. (2020) at workrates of 160W and 80W, but not at 120W which was employed in the present study. When comparing between QF conditions within the workrate of 160W, they found that KAbM significantly increased when QF changed from 150 mm to 192 mm, but neither the step from 192 mm to 234 mm nor the step from 234 mm to 276 mm resulted in significantly different KAbMs. Additionally, in the 80W workrate condition, there were no significant increases in KAbM between incremental 42 mm changes. KAbM was only significantly greater at a QF of 234 mm compared to 150 mm and at 276 mm compared to 150 and 192 mm. However, when comparing between QF conditions when workrates were combined, there was a significant increase in KAbM between all QF condition comparisons (Thorsen et al., 2020). It was expected that if a single incremental increase in QF caused an increase in KAbM, then the other changes of the same and greater magnitude would also cause increases in KAbM. A potential explanation for why this was not observed may be that there is an interaction effect where KAbM is more sensitive to changes in QF at higher QFs. This is supported by the present findings that KAbM at Q5 was significantly higher than at Q4 and the other three QFs while there were no significant differences between Q1, Q2, Q3, and Q4 themselves.

The PRF can influence KAbM in a couple of different ways. The medial PRF is more influential in modulating the length of the frontal-plane moment arm of the knee while the vertical PRF has a greater influence on the magnitude of the resultant PRF vector. The medial PRF was significantly increased in Q5 compared to Q1, Q2, and Q3 in the present study. A general increase in medial PRF with increased QFs was expected and agrees with previous literature (Thorsen et al., 2020). There were a couple significant differences found in vertical PRF, but they did not likely have meaningful influence on the observed differences in KAbMs. The differences in vertical GRF were not present in the comparisons among the QFs with observed significant changes in KAbM. Additionally, there was no consistent pattern in these changes, which is supported by previous research (Thorsen et al., 2020). Given a constant vertical PRF between Q5 and the other conditions, it appears that the increases in KAbM were primarily caused by increases in the PRF moment arm and to lesser extent by the PRF itself. In most comparisons, this is supported by either concurrent increases or lack thereof in both medial PRF and KAbM. The only comparison where this was not upheld was between Q4 and Q5 where there was a significant increase in KAbM but not in medial PRF. This could be explained by the fact that there are other variables that can influence KAbM, such as the frontal-plane knee angle, that may have contributed to this change.

The frontal-plane knee angle can also influence KAbM by changing the position of the knee relative to the PRF, thereby changing the length of the moment arm. The most common pattern of knee frontal-plane angle during early power phase was knee abduction among our participants. Peak knee abduction angle occurred almost simultaneously with the peak KAbM, but no differences in peak knee abduction angles between conditions were found. Thorsen et al. (2020) and Fife et al. (2020) also investigated the effect of QF on peak knee abduction angle, and they both found that, to some extent, increases in QF caused the knee to become more abducted. These findings disagree with the current study, but the greatest change in knee abduction angle in either of these studies, found by Thorsen et al. (2020), was only about 2° between QFs of 150 and 276 mm. We also observed that the knee abduction ROMs in the present study were significantly smaller in Q4 than in Q1 and in Q5 than Q1 and Q3. Without significant changes in peak knee abduction angles, these changes indicate that at the onset of the crank cycle the knee was less adducted in the higher QFs. Therefore, given the results of the current study and these studies, it appears that changes in peak frontal-plane knee angles and their ROMs during power phase may not have meaningful contribution to changes in peak KAbM by themselves. It is possible that subtle changes in frontal-plane knee angle in combination with subtle changes in medial PRF may result in more notable changes in KAbM. This may explain why there was a significant increase in KAbM, but neither peak knee abduction angle nor medial PRF were significantly different in the same comparison. Additionally, variability in the temporal overlap between the peak medial PRF, vertical PRF, and knee abduction angle may explain the current observations. In order to determine how these variables contribute to changes in peak KAbM, further investigation would need to be performed to determine each variable's effect on the moment arm at the instance of peak KAbM.

Interestingly, changes of QF seem to have similar effects on the knee transverse-plane kinetics and kinematics as it did on the knee frontal-plane variables. The peak internal rotation moment was significantly greater at Q4 than Q1 and at Q5 than Q1, Q2, and Q3. The peak external rotation angle was not significantly different between conditions, but the external rotation ROM was significantly smaller at Q5 than Q1 and Q2. These observations suggest that, to some extent, there may be a coupled response of the frontal- and transverse-plane biomechanics of the knee to changes in QF. Further investigation is warranted to determine the nature and extent of this relationship during cycling.

The knee was not the only joint that saw changes in its peak frontal-plane moment. Both the ankle and hip had significant differences in some comparisons as well, although the values and changes in the ankle were less substantial than those at the knee and hip. The peak ankle inversion moment was shown to be decreased at higher QFs, but they were never greater than 1.25 Nm or less than 0.21 Nm. The hip abduction moment was significantly increased in Q5 compared to Q1, Q2, Q3, and Q4, concurrent with the changes in KAbM. This finding indicates that there are potentially important accommodations occurring in the hip that contribute to the whole lower limb's adjustment when pedaling at wider QFs. This may explain why a significant difference was seen in KAbM but not medial PRF between Q4 and Q5. It is possible that the responses of the knee and hip to QF changes are coupled, and further investigation of this effect is needed to better understand how the rider adapts wholistically to wider QFs.

The secondary aim of this study was to examine the relationship between an individual's MAA and their KAbM. The hypothesis that MAA would account for some variance in KAbM was not supported by the findings of the current study. MAA did not account for any significant portion of the variance in KAbM based on the initial ANCOVA, nor was a significant correlation between the two variables found ($|r| \leq 0.197$, $p \geq 0.482$ for all QF conditions). In a previous study of the effects of knee alignment on knee biomechanics during cycling, no significant effects were found for knee alignment group on KAbM, mediolateral PRF, or vertical PRF (Shen et al., 2018). These alignment group comparisons cannot be directly related to the current study [MAA range: 172.3° – 179.7°; 12 varus (MAA ≤ 178°) and 3 neutral (MAA > 178° and < 182°) participants], but they lend support to the findings that there was no consistent correlation between MAA and these variables. There was a significant correlation between MAA and vertical PRF in Q4; but considering the ANCOVA results and lack of significant correlation elsewhere, this relationship is likely not meaningful. The same previous study did find that knee alignment significantly affects the peak frontal-plane angles of the knee (Shen et al., 2018). The current study found no relationship between MAA and the peak knee abduction angle, although there was a significant correlation between MAA and knee abduction ROM for Q5. Again, these results cannot be directly compared due to differences in participant group and study aims, but they would seem to disagree about the nature of the relationship between MAA and peak frontal-plane knee angles. Static knee alignments are most accurately determined through standing, full-limb radiography, but the present study used 3D motion capture and a regression model from previous literature to estimate MAA (Vanwanseele et al., 2009). Differing methods of determining MAA may explain some of the observed differences between the present study and studies that utilized radiography, such as Shen et al. (2018).

As previously stated, KAbM may be influenced by the vertical PRF, medial PRF, and the frontal-plane knee angle. Because MAA had no consistent relationships with these variables, it is agreeable that the same was found for KAbM. Static frontal-plane knee alignment was expected to influence peak frontal-plane angles of the knee and peak KAbM, as it has previously been shown in level walking (Barrios and Strotman, 2014; Bennett et al., 2017a). While walking and cycling share similarities, ultimately, they have different dynamic processes. Previous research has already shown that KAbM is increased by widening the pedals in cycling (Thorsen et al., 2020) and decreased by widening stance in walking (Bennett et al., 2017a). Additionally, MAA is captured in a standing posture, which is a weight-bearing position similar to walking. Conversely, cycling is performed in a seated, non-weight-bearing position where both the feet (by pedals) and hips (by saddle) are more constrained than in walking. Therefore, it is reasonable to expect that MAA may affect cycling dynamics differently than it affects walking. However, it cannot be concluded entirely that MAA bears no effect on cycling biomechanics, as the participants in this study had relatively homogenous MAA alignments and no participants had valgus alignment. Therefore, it remains possible that individuals with valgus alignment respond differently to changes in QF than individuals with varus or neutral alignment.

One limitation is that the crank arm and adjustable QF assembly were custom built, so it would be difficult for outside research groups to replicate these conditions. The crank arms and pedals were also considerably heavier and bulkier than parts typically found on a bicycle or ergometer. It is possible that the increased inertia of the crank arm and pedal assembly could have altered the rider's biomechanics. This warrants further investigation to examine if the increased inertia contributed to changes in sagittal-plane and frontal-plane lower limb kinetics. Another limitation is that although the crank arms allowed for a continuous change in QF, QF could only be measured with millimeter level precision (Figure 3, panel B). Consequently, the normalized QF changes for each participant were rounded to the nearest millimeter. However, this limitation is unlikely to have had any substantial effect on the KAbM results of the study. A third limitation of this study was that MAA was not determined using radiography but was estimated using a validated 3D motion capture and a regression equation from previous literature (Vanwanseele et al., 2009). Future studies should try to include more participants with valgus and neutral alignments, as the range of MAA values in the present study was relatively small and predominantly categorized as varus.

Conclusion

This study was the first to investigate the effects of small and normalized changes in QF while controlling for static frontal-plane knee alignment. The results of this study show that it is possible to detect a significant increase in KAbM with changes in QF as small as 2% of one's leg length at a high QF. These findings suggest that KAbM becomes more sensitive to changes in QF at greater QFs. Static knee alignment does not seem to be meaningfully related to any of the knee kinetic variables. The results suggest that people with varus and neutral knee alignment may not need to be concerned with alignment associated changes in knee loading when cycling at different pedal widths on stationary bikes and cycle ergometers. More research is required to fully elucidate the significance of QF and knee alignment as parameters of bike fit and their impact on frontal-plane lower limb biomechanics in cycling.

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APPENDICES

Appendix A: Individual Participant Characteristics

Table 4: Individual participant characteristics.

Participant	Sex	Age (years)	Height (m)	Mass (kg)	BMI (kg/m2)
S2	M	19	1.78	70.3	22.2
S3	F	25	1.65	66.7	24.5
S4	M	23	1.82	81.6	24.8
S5	M	20	1.73	56.7	19.0
S6	M	27	1.83	99.8	29.8
S7	F	21	1.70	64.9	22.4
S8	M	28	1.88	89.3	25.3
S9	F	24	1.58	57.6	23.2
S10	F	22	1.68	60.8	21.6
S11	M	24	1.70	71.2	24.6
S12	F	23	1.68	63.5	22.6
S13	F	20	1.68	83.9	29.9
S15	F	22	1.63	72.6	27.4
S16	M	21	1.73	66.7	22.4
S17	F	22	1.73	58.5	19.6
Mean	-	22.7	1.72	70.9	24.0
SD	-	2.5	0.08	12.6	3.2

Abbreviations: BMI – Body Mass Index

Table 5: Individual anthropometrics and prescribed Q-Factor (QF) intervals.

Participant	ASIS (cm)	Leg Length (mm)	MAA (°)	QF Interval (mm)
S2	21.5	915	173.3	18
S3	20.5	897	178.1	18
S4	26.0	921	172.3	18
S5	22.0	885	173.5	18
S6	28.0	982	173.3	20
S7	21.0	895	178.6	18
S8	23.5	1000	176.7	20
S9	20.0	820	176.1	16
S10	20.0	887	175.5	18
S11	25.0	840	173.7	17
S12	25.0	917	177.2	18
S13	22.5	910	179.7	18
S15	25.0	860	177.2	17
S16	23.0	905	174.0	18
S17	25.5	910	173.8	18
Mean	23.2	903	175.5	18
SD	2.4	46	2.3	1

Abbreviations: ASIS – Anterior Superior Iliac Spine Width; MAA – Mechanical Axis Angle

Table 6: Individual Q-Factor conditions (mm).

Participant	Q1 160	Q2 160 + 0.02L	Q3 160 +0.04L	Q4 160 + 0.06L	Q5 160 + 0.08L
S2	160	178	196	214	232
S3	160	178	196	214	232
S4	160	178	196	214	232
S5	160	178	196	214	232
S6	160	180	200	220	240
S7	160	178	196	214	232
S8	160	180	200	220	240
S9	160	176	192	208	224
S10	160	178	196	214	232
S11	160	177	194	211	228
S12	160	178	196	214	232
S13	160	178	196	214	232
S15	160	177	194	211	228
S16	160	178	196	214	232
S17	160	178	196	214	232
Mean	-	178.0	196.0	214.0	232.0
SD	-	1.04	2.08	3.11	4.15

L = Participant trochanteric leg length (mm). Each condition is increased by 2% of L from the previous starting at 160 mm for each participant.

Table 7: Individual responses for RPE.

Participant	RPE				
	Q1	**Q2**	**Q3**	**Q4**	**Q5**
S2	6	8	9	10	11
S3	12	11	13	12	12
S4	12	12	12	12	12
S5	10	15	14	14	12
S6	8	8	8	8	8
S7	11	11	11	11	11
S8	7	7	7	7	7
S9	14	13	13	14	14
S10	17	16	12	12	15
S11	11	11	9	8	8
S12	15	13	13	12	14
S13	11	12	13	12	15
S15	11	13	12	11	12
S16	9	11	10	9	11
S17	14	11	12	14	13
Mean	11.20	11.47	11.20	11.07	11.67
SD	3.00	2.47	2.11	2.25	2.47

Abbreviations: RPE – Rating of Perceived Exertion

Appendix B: Informed Consent Form

Consent for Research Participation

Research Study Title: *The Effects of Knee Alignment and Personalized Q-Factor Changes on Knee Biomechanics During Cycling*

Researcher(s): Jacob Wilbert B.S., University of Tennessee, Knoxville

Sean Brown M.A., University of Tennessee, Knoxville

Songning Zhang Ph.D., University of Tennessee, Knoxville

Why am I being asked to be in this research study?

We are asking you to be in this research study because you meet the following requirements:

- Between 18 and 35 years old
- Participate in moderate intensity physical activity at least 3 times per week for 30 minutes
- Body mass index (BMI) between 18.5 and 29.9 kg/m^2
- No lower limb injuries within the past 6 months
- No history of musculoskeletal disease affecting the lower limbs
- No history of severe injury that required surgery

What is this research study about?

The primary purpose of the research study is to determine whether knee joint malalignments affect knee biomechanics when cycling at various pedal widths.

The secondary purpose of the research study is to determine whether small increases in pedal width that are personalized to the rider will affect knee biomechanics.

How long will I be in the research study?

If you agree to be in the study, your participation will last for up to 1.5 hours during a single session.

What will happen if I say "Yes, I want to be in this research study"?

Eligibility to participate in this study will be determined based on your responses to the Physical Activity Readiness Questionnaire and participant screening questionnaire prior to scheduling an in-person session.

If you are eligible and agree to be in this study, we will ask you to do the following:

- Attend a single, 1.5-hour session at the Biomechanics/Sports Medicine Laboratory in the Health, Physical Education, and Recreation (HPER) building at the University of Tennessee, Knoxville. During this session, we will ask you for personal and contact information and experimental testing will be performed.

- Prior to experimental testing:
  - We will take measurements of your body that will require a member of the research team to make physical contact with you including:
    - Height
    - Weight
    - Leg length
    - Knee alignment
  - Change into appropriate attire either owned by you or provided by the research team. An appropriate changing area will be provided.
  - We will attach compressive sleeves and reflective markers to your body. This will require palpation and physical manipulation of your body by a member of the research team.
  - We will adjust the bike so that you fit properly. This will require some physical contact as we measure hip and knee angles.
  - We will ask you to complete a warm-up ride on the bike for 2 minutes.
- During experimental testing:
  - We will ask you to cycle at a cadence of 80 rotations per minute and a power output of 120 Watts for 2 minutes in each test condition with 2 minutes of rest in between bouts.
  - During each test condition we will ask you to pedal at a predetermined pedal width. We will change the pedal width during your rest period between conditions.
  - The bike we will ask you to ride on is deconstructed and has exposed moving parts such as the gear-mounted chain.

What happens if I say "No, I do not want to be in this research study"?

Being in this study is up to you. You can say no now or leave the study later. Either way, your decision won't affect your grades, your relationship with your instructors, your academic standing, or your employment with The University of Tennessee, Knoxville.

What happens if I say "Yes" but change my mind later?

Even if you decide to be in the study now, you can change your mind and stop at any time. This decision won't affect your grades, your relationship with your instructors, your academic standing, or your employment with The University of Tennessee, Knoxville.

If you decide to stop before the study is completed, you should promptly inform the principal investigator. In this event, all materials, information, and data collected from you will be destroyed/deleted and not used in the study.

Are there any possible risks to me?

There is minimal risk in this study. The duration and intensity of exercise will not exceed a moderate level. We will ensure you pass the PARQ form, which will help ensure you are safe to exercise. The researchers in this study are certified in first aid, CPR, and AED.

Other people may see you participating in the study, but access to the research lab is limited to Faculty, Staff, and Biomechanics graduate personnel. Anyone who may see you participating in the study will not have access to any of your identifiable information. All digital information will be coded and stored with no identification attached. All physical documents, including an identification key, will be kept in a locked cabinet and only the research team will have access to them.

Are there any benefits to being in this research study?

There is a possibility that you may benefit from being in the study, but there is no guarantee that will happen. Completion of the protocol includes estimation of knee joint alignment, so it is possible for you to learn the estimated alignment of your knee. However, the value we calculate is merely an estimate and should not be considered as a legitimate medical evaluation. Even if you don't benefit from being in the study, your participation may help us to learn more about the effects of knee alignment and pedal width on knee biomechanics during cycling. We hope the knowledge gained from this study will benefit others in the future.

Who can see or use the information collected for this research study?

We will protect the confidentiality of your information by deidentifying the data such that only a subject number will be assigned to it. If information from this study is published or presented at scientific meetings, your name and other personal information will not be used.

We will make every effort to prevent anyone who is not on the research team from knowing that you gave us information or what information came from you. Although it is unlikely, there are times when others may need to see the information, we collect about you. These include:

- People at the University of Tennessee, Knoxville who oversee research to make sure it is conducted properly.
- Government agencies (such as the Office for Human Research Protections in the U.S. Department of Health and Human Services), and others responsible for watching over the safety, effectiveness, and conduct of the research.
- If a law or court requires us to share the information, we would have to follow that law or final court ruling.

What will happen to my information after this study is over?

We will not keep your identifying information to use for future data analysis, presentations, and/or publications. Your name and other information that can directly identify you will be deleted from your research data collected as part of the study.

We may share your research data with other researchers without asking for your consent again, but it will not contain information that could directly identify you.

Will it cost me anything to be in this research study?

It will not cost you anything to be in this study.

What else do I need to know?

We use procedures to lower the possibility of these risks happening. Even so, you may still experience problems or injury, even when we are careful to avoid them. Please tell the researcher in charge, Jacob Wilbert (Email: jacdwil1@vols.utk.edu | Phone: (865) 974-2091), about any injuries or other problems that you have during this study.

The University of Tennessee does not automatically pay for medical claims or give other compensation for injuries or other problems.

Who can answer my questions about this research study?

If you have questions or concerns about this study, or have experienced a research related problem or injury, contact the researchers, Jacob Wilbert (Email: jacdwil1@vols.utk.edu | Phone: (865) 974-2091) or Dr. Songning Zhang (Email: szhang@utk.edu).

For questions or concerns about your rights or to speak with someone other than the research team about the study, please contact:

Institutional Review Board
The University of Tennessee, Knoxville
1534 White Avenue
Blount Hall, Room 408
Knoxville, TN 37996-1529
Phone: 865-974-7697
Email: utkirb@utk.edu

STATEMENT OF CONSENT

I have read this form and the research study has been explained to me. I have been given the chance to ask questions and my questions have been answered. If I have more questions, I have been told who to contact. By signing this document, I am agreeing to be in this study. I will receive a copy of this document after I sign it.

______________________	______________________	__________
Name of Adult Participant	Signature of Adult Participant	Date

Researcher Signature (to be completed at time of informed consent)

I have explained the study to the participant and answered all of his/her questions. I believe that he/she understands the information described in this consent form and freely consents to be in the study.

______________________	______________________	__________
Name of Research Team Member	Signature of Research Team Member	Date

Appendix C: Recruitment Flyer

RESEARCH PARTICIPANTS NEEDED FOR A STUDY ON KNEE ALIGNMENT IN CYCLING

A team of researchers from the Department of Kinesiology Recreation and Sports Studies at UT are conducting a research study to understand the effects of knee alignment at different pedal widths on lower limb biomechanics during cycling. Participants will be required to attend one testing session that will last up to 1.5 hours in the UT Biomechanics/Sports Medicine lab.

Qualifications to participate in the study include:

- Between the ages of 18 and 35 yrs.
- Physically active
- No lower limb injuries in past 6 months
- No history of lower limb joint disease or injury requiring surgical intervention

If you would like to participate or for more information, contact Jacob Wilbert at the UT Biomechanics/Sports Medicine Lab.
Phone: 865-974-2091
Email: jacdwil1@vols.utk.edu

Contact: Jacob Wilbert
P: 865-974-2091
E: jacdwil1@vols.utk.edu

Contact: Jacob Wilbert
P: 865-974-2091
E: jacdwil1@vols.utk.edu

Contact: Jacob Wilbert
P: 865-974-2091
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Contact: Jacob Wilbert
P: 865-974-2091
E: jacdwil1@vols.utk.edu

Appendix D: Participant Questionnaires

PARTICIPANT SCREENING QUESTIONNAIRE

All answers to the following questions are confidential and will not be shared outside of the research team. All identifying information will be removed from any information that is used for research purposes. The following questionnaire will be used to determine your eligibility to participate in the study entitled: *The Effects of Knee Alignment and Personalized Q-Factor Changes on Knee Biomechanics during Cycling*. Please complete the following form to the best of your knowledge and return it to the principal investigator. If you are eligible to participate in this study, you will be contacted via email to schedule a session. If you are ineligible, the PI will notify you and this form will be destroyed.

Name:		**Email:**	
Sex: Male Female	**Age:**	**Height:**	**Weight:**

1) Do you participate in 30 minutes of moderate intensity exercise at least 3 times per week?	YES NO
2) Have you sustained any lower limb musculoskeletal injuries within the past 6 months?	YES NO
a. If YES, please list or describe:	
3) Have you been diagnosed with a musculoskeletal disease affecting the lower extremities?	YES NO
a. If YES, please list or describe:	
4) Have you ever sustained a lower extremity injury requiring surgical intervention?	YES NO
a. If YES, please list or describe:	
5) Have you ever been assessed by x-ray for lower limb joint malalignment?	YES NO
a. If YES, please list or describe the results:	

Please sign and return this form or direct any questions or concerns to the principal investigator.

Principal investigator: **Jacob Wilbert**

Email: jacdwil1@vols.utk.edu

By signing this document below, I am agreeing to the use of this information by the research team for the study entitled *The Effects of Knee Alignment and Personalized Q-Factor Changes on Knee Biomechanics During Cycling*.

Signature: Date:

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The Physical Activity Readiness Questionnaire for Everyone

The health benefits of regular physical activity are clear; more people should engage in physical activity every day of the week. Participating in physical activity is very safe for MOST people. This questionnaire will tell you whether it is necessary for you to seek further advice from your doctor OR a qualified exercise professional before becoming more physically active.

GENERAL HEALTH QUESTIONS

Please read the 7 questions below carefully and answer each one honestly: check YES or NO.	YES	NO
1) Has your doctor ever said that you have a heart condition ☐ **OR** high blood pressure ☐?	☐	☐
2) Do you feel pain in your chest at rest, during your daily activities of living, **OR** when you do physical activity?	☐	☐
3) Do you lose balance because of dizziness **OR** have you lost consciousness in the last 12 months? Please answer **NO** if your dizziness was associated with over-breathing (including during vigorous exercise).	☐	☐
4) Have you ever been diagnosed with another chronic medical condition (other than heart disease or high blood pressure)? **PLEASE LIST CONDITION(S) HERE:** ____________	☐	☐
5) Are you currently taking prescribed medications for a chronic medical condition? **PLEASE LIST CONDITION(S) AND MEDICATIONS HERE:** ____________	☐	☐
6) Do you currently have (or have had within the past 12 months) a bone, joint, or soft tissue (muscle, ligament, or tendon) problem that could be made worse by becoming more physically active? Please answer **NO** if you had a problem in the past, but it **does not limit your current ability** to be physically active. **PLEASE LIST CONDITION(S) HERE:** ____________	☐	☐
7) Has your doctor ever said that you should only do medically supervised physical activity?	☐	☐

☑ **If you answered NO to all of the questions above, you are cleared for physical activity. Please sign the PARTICIPANT DECLARATION. You do not need to complete Pages 2 and 3.**

- Start becoming much more physically active – start slowly and build up gradually.
- Follow Global Physical Activity Guidelines for your age (https://www.who.int/publications/i/item/9789240015128).
- You may take part in a health and fitness appraisal.
- If you are over the age of 45 yr and NOT accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.
- If you have any further questions, contact a qualified exercise professional.

PARTICIPANT DECLARATION
If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for its records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME ______________________________ DATE ______________

SIGNATURE ______________________________ WITNESS ______________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ______________________________

If you answered YES to one or more of the questions above, COMPLETE PAGES 2 AND 3.

⚠ **Delay becoming more active if:**

- ✓ You have a temporary illness such as a cold or fever; it is best to wait until you feel better.
- ✓ You are pregnant - talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the ePARmed-X+ at www.eparmedx.com before becoming more physically active.
- ✓ Your health changes - answer the questions on Pages 2 and 3 of this document and/or talk to your doctor or a qualified exercise professional before continuing with any physical activity program.

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FOLLOW-UP QUESTIONS ABOUT YOUR MEDICAL CONDITION(S)

1. Do you have Arthritis, Osteoporosis, or Back Problems?

If the above condition(s) is/are present, answer questions 1a-1c — If **NO** ☐ go to question 2

1a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
1b.	Do you have joint problems causing pain, a recent fracture or fracture caused by osteoporosis or cancer, displaced vertebra (e.g., spondylolisthesis), and/or spondylolysis/pars defect (a crack in the bony ring on the back of the spinal column)?	YES ☐ NO ☐
1c.	Have you had steroid injections or taken steroid tablets regularly for more than 3 months?	YES ☐ NO ☐

2. Do you currently have Cancer of any kind?

If the above condition(s) is/are present, answer questions 2a-2b — If **NO** ☐ go to question 3

2a.	Does your cancer diagnosis include any of the following types: lung/bronchogenic, multiple myeloma (cancer of plasma cells), head, and/or neck?	YES ☐ NO ☐
2b.	Are you currently receiving cancer therapy (such as chemotheraphy or radiotherapy)?	YES ☐ NO ☐

3. Do you have a Heart or Cardiovascular Condition? This Includes Coronary Artery Disease, Heart Failure, Diagnosed Abnormality of Heart Rhythm

If the above condition(s) is/are present, answer questions 3a-3d — If **NO** ☐ go to question 4

3a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
3b.	Do you have an irregular heart beat that requires medical management? (e.g., atrial fibrillation, premature ventricular contraction)	YES ☐ NO ☐
3c.	Do you have chronic heart failure?	YES ☐ NO ☐
3d.	Do you have diagnosed coronary artery (cardiovascular) disease and have not participated in regular physical activity in the last 2 months?	YES ☐ NO ☐

4. Do you currently have High Blood Pressure?

If the above condition(s) is/are present, answer questions 4a-4b — If **NO** ☐ go to question 5

4a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
4b.	Do you have a resting blood pressure equal to or greater than 160/90 mmHg with or without medication? (Answer **YES** if you do not know your resting blood pressure)	YES ☐ NO ☐

5. Do you have any Metabolic Conditions? This Includes Type 1 Diabetes, Type 2 Diabetes, Pre-Diabetes

If the above condition(s) is/are present, answer questions 5a-5e — If **NO** ☐ go to question 6

5a.	Do you often have difficulty controlling your blood sugar levels with foods, medications, or other physician-prescribed therapies?	YES ☐ NO ☐
5b.	Do you often suffer from signs and symptoms of low blood sugar (hypoglycemia) following exercise and/or during activities of daily living? Signs of hypoglycemia may include shakiness, nervousness, unusual irritability, abnormal sweating, dizziness or light-headedness, mental confusion, difficulty speaking, weakness, or sleepiness.	YES ☐ NO ☐
5c.	Do you have any signs or symptoms of diabetes complications such as heart or vascular disease and/or complications affecting your eyes, kidneys, **OR** the sensation in your toes and feet?	YES ☐ NO ☐
5d.	Do you have other metabolic conditions (such as current pregnancy-related diabetes, chronic kidney disease, or liver problems)?	YES ☐ NO ☐
5e.	Are you planning to engage in what for you is unusually high (or vigorous) intensity exercise in the near future?	YES ☐ NO ☐

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6. Do you have any Mental Health Problems or Learning Difficulties? This includes Alzheimer's, Dementia, Depression, Anxiety Disorder, Eating Disorder, Psychotic Disorder, Intellectual Disability, Down Syndrome

If the above condition(s) is/are present, answer questions 6a-6b — If **NO** ☐ go to question 7

6a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
6b.	Do you have Down Syndrome **AND** back problems affecting nerves or muscles?	YES ☐ NO ☐

7. Do you have a Respiratory Disease? *This includes Chronic Obstructive Pulmonary Disease, Asthma, Pulmonary High Blood Pressure*

If the above condition(s) is/are present, answer questions 7a-7d — If **NO** ☐ go to question 8

7a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
7b.	Has your doctor ever said your blood oxygen level is low at rest or during exercise and/or that you require supplemental oxygen therapy?	YES ☐ NO ☐
7c.	If asthmatic, do you currently have symptoms of chest tightness, wheezing, laboured breathing, consistent cough (more than 2 days/week), or have you used your rescue medication more than twice in the last week?	YES ☐ NO ☐
7d.	Has your doctor ever said you have high blood pressure in the blood vessels of your lungs?	YES ☐ NO ☐

8. Do you have a Spinal Cord Injury? *This includes Tetraplegia and Paraplegia*

If the above condition(s) is/are present, answer questions 8a-8c — If **NO** ☐ go to question 9

8a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
8b.	Do you commonly exhibit low resting blood pressure significant enough to cause dizziness, light-headedness, and/or fainting?	YES ☐ NO ☐
8c.	Has your physician indicated that you exhibit sudden bouts of high blood pressure (known as Autonomic Dysreflexia)?	YES ☐ NO ☐

9. Have you had a Stroke? *This includes Transient Ischemic Attack (TIA) or Cerebrovascular Event*

If the above condition(s) is/are present, answer questions 9a-9c — If **NO** ☐ go to question 10

9a.	Do you have difficulty controlling your condition with medications or other physician-prescribed therapies? (Answer **NO** if you are not currently taking medications or other treatments)	YES ☐ NO ☐
9b.	Do you have any impairment in walking or mobility?	YES ☐ NO ☐
9c.	Have you experienced a stroke or impairment in nerves or muscles in the past 6 months?	YES ☐ NO ☐

10. Do you have any other medical condition not listed above or do you have two or more medical conditions?

If you have other medical conditions, answer questions 10a-10c — If **NO** ☐ read the Page 4 recommendations

10a.	Have you experienced a blackout, fainted, or lost consciousness as a result of a head injury within the last 12 months **OR** have you had a diagnosed concussion within the last 12 months?	YES ☐ NO ☐
10b.	Do you have a medical condition that is not listed (such as epilepsy, neurological conditions, kidney problems)?	YES ☐ NO ☐
10c.	Do you currently live with two or more medical conditions?	YES ☐ NO ☐

PLEASE LIST YOUR MEDICAL CONDITION(S) AND ANY RELATED MEDICATIONS HERE: ______________________________

GO to Page 4 for recommendations about your current medical condition(s) and sign the PARTICIPANT DECLARATION.

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If you answered NO to all of the FOLLOW-UP questions (pgs. 2-3) about your medical condition, you are ready to become more physically active - sign the PARTICIPANT DECLARATION below:

- It is advised that you consult a qualified exercise professional to help you develop a safe and effective physical activity plan to meet your health needs.
- You are encouraged to start slowly and build up gradually - 20 to 60 minutes of low to moderate intensity exercise, 3-5 days per week including aerobic and muscle strengthening exercises.
- As you progress, you should aim to accumulate 150 minutes or more of moderate intensity physical activity per week.
- If you are over the age of 45 yr and **NOT** accustomed to regular vigorous to maximal effort exercise, consult a qualified exercise professional before engaging in this intensity of exercise.

If you answered **YES** to **one or more of the follow-up questions** about your medical condition:

You should seek further information before becoming more physically active or engaging in a fitness appraisal. You should complete the specially designed online screening and exercise recommendations program - the **ePARmed-X+ at www.eparmedx.com** and/or visit a qualified exercise professional to work through the ePARmed-X+ and for further information.

Delay becoming more active if:

- You have a temporary illness such as a cold or fever; it is best to wait until you feel better.
- You are pregnant - talk to your health care practitioner, your physician, a qualified exercise professional, and/or complete the ePARmed-X+ **at www.eparmedx.com** before becoming more physically active.
- Your health changes - talk to your doctor or qualified exercise professional before continuing with any physical activity program.

- You are encouraged to photocopy the PAR-Q+. You must use the entire questionnaire and NO changes are permitted.
- The authors, the PAR-Q+ Collaboration, partner organizations, and their agents assume no liability for persons who undertake physical activity and/or make use of the PAR-Q+ or ePARmed-X+. If in doubt after completing the questionnaire, consult your doctor prior to physical activity.

PARTICIPANT DECLARATION

- All persons who have completed the PAR-Q+ please read and sign the declaration below.
- If you are less than the legal age required for consent or require the assent of a care provider, your parent, guardian or care provider must also sign this form.

I, the undersigned, have read, understood to my full satisfaction and completed this questionnaire. I acknowledge that this physical activity clearance is valid for a maximum of 12 months from the date it is completed and becomes invalid if my condition changes. I also acknowledge that the community/fitness center may retain a copy of this form for records. In these instances, it will maintain the confidentiality of the same, complying with applicable law.

NAME ____________________ DATE ____________________

SIGNATURE ____________________ WITNESS ____________________

SIGNATURE OF PARENT/GUARDIAN/CARE PROVIDER ____________________

For more information, please contact

www.eparmedx.com

Email: eparmedx@gmail.com

Citation for PAR-Q+

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PRINT FORM RESET FORM

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Appendix E: Post-trial survey question

1. **How would you rate your physical exertion during this bout of cycling?**

Borg's Scale of Perceived Exertion (RPE)

Scale	Perceived Exertion
6	
7	Very, very light
8	
9	Very Light
10	
11	Fairly Light
12	
13	Somewhat Hard
14	
15	Hard
16	
17	Very Hard
18	
19	Very, very hard
20	Maximum exertion

Appendix F: Individual Results for Selected Variables

Table 8: Individual mean peak vertical pedal reaction forces (N).

Participant	Q1	Q2	Q3	Q4	Q5
S2	235.992±30.347	233.998±14.990	221.378±16.282	216.593±15.274	226.975±28.809
S3	221.264±5.824	210.538±18.078	212.099±18.341	222.457±4.991	218.398±12.525
S4	174.925±20.285	178.399±16.824	163.321±11.314	158.687±10.691	159.426±12.618
S5	224.627±43.064	215.775±19.695	215.306±33.829	214.240±28.615	223.779±30.547
S6	281.713±19.888	251.916±8.588	256.828±20.890	225.246±20.272	233.684±19.938
S7	217.324±20.905	202.234±16.088	238.126±18.653	225.312±32.322	203.806±10.216
S8	240.622±29.466	262.489±29.205	245.528±22.470	228.441±42.804	241.480±17.241
S9	262.230±12.595	214.245±50.425	234.374±25.837	252.682±24.817	229.606±23.463
S10	205.827±8.155	181.392±7.003	199.258±9.748	195.758±21.566	215.040±17.625
S11	259.770±18.728	241.957±19.629	240.343±16.866	228.492±21.231	272.986±20.796
S12	-	204.455±23.746	231.592±27.087	219.398±16.221	234.167±38.357
S13	303.077±20.670	268.520±6.633	257.267±8.028	274.532±36.167	263.976±23.067
S15	231.947±20.552	229.707±16.243	-	192.942±16.623	235.784±14.418
S16	222.659±13.452	216.275±13.878	229.260±32.637	212.775±14.931	228.062±13.722
S17	186.818±30.305	148.518±31.223	166.675±7.018	190.563±22.652	185.311±17.797
Mean±STD	233.485±34.779	217.361±32.611	222.240±29.179	217.208±26.956	224.832±27.867

Positive values indicate peak vertically directed pedal reaction forces during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 9: Individual mean peak medial pedal reaction forces (N).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-55.331±10.345	-48.160±5.701	-52.833±5.721	-53.331±7.099	-61.905±11.746
S3	-30.851±3.566	-43.185±3.328	-39.106±6.523	-46.243±2.352	-45.132±5.188
S4	-28.124±6.102	-37.180±6.663	-30.689±3.174	-29.047±2.435	-30.405±4.675
S5	-40.347±10.950	-41.629±7.058	-39.389±8.485	-46.788±8.011	-50.872±11.118
S6	-48.591±10.992	-45.715±6.754	-49.519±9.290	-48.283±11.837	-56.041±8.310
S7	-33.414±7.024	-37.671±4.764	-46.919±6.200	-55.880±13.439	-57.672±4.243
S8	-46.934±14.352	-53.719±15.711	-57.296±5.813	-57.256±16.398	-59.616±9.657
S9	-60.324±4.919	-60.325±13.817	-53.020±7.057	-68.955±10.101	-61.739±10.794
S10	-54.820±2.991	-45.040±1.084	-45.643±4.895	-44.333±4.814	-59.141±24.812
S11	-56.562±7.966	-54.996±6.117	-53.592±3.056	-55.917±4.061	-72.082±5.084
S12	-	-51.096±10.244	-55.799±13.042	-63.342±10.519	-70.295±18.336
S13	-53.177±2.730	-47.234±6.136	-54.129±4.025	-56.981±15.904	-47.955±12.569
S15	-50.792±13.060	-49.343±9.007	-	-38.117±9.929	-65.620±10.311
S16	-36.186±6.814	-40.356±5.527	-40.625±12.957	-41.975±4.786	-50.806±9.883
S17	-38.828±7.588	-29.908±8.849	-26.761±2.945	-44.649±6.909	-41.087±5.787
Mean±STD	-45.306±10.540	-45.704±7.809	-46.094±9.532	-50.073±10.184	-55.358±11.201

Negative values indicate a peak medially directed pedal reaction forces during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 10: Individual mean peak knee extension moments (Nm).

Participant	Q1	Q2	Q3	Q4	Q5
S2	39.052±5.873	37.285±3.199	39.298±3.045	33.715±2.897	37.494±6.390
S3	34.800±4.813	42.160±4.934	37.904±7.253	44.859±1.410	43.864±4.327
S4	23.257±5.508	29.266±4.070	21.653±2.422	25.129±0.759	25.622±6.414
S5	32.184±8.929	35.112±5.093	29.453±6.865	30.451±5.961	34.953±5.615
S6	47.733±9.056	41.879±5.919	36.953±3.925	39.433±10.258	46.023±6.247
S7	36.771±4.494	32.640±4.961	39.889±5.030	39.120±5.292	37.839±3.493
S8	24.614±8.525	36.271±13.325	40.151±6.415	38.023±15.130	39.324±8.514
S9	43.935±2.032	32.347±8.996	41.276±5.756	46.224±3.575	42.915±4.936
S10	29.915±1.788	27.589±1.549	31.706±2.225	31.292±2.322	29.879±2.507
S11	43.411±5.469	43.632±5.078	38.985±3.519	36.632±3.962	46.763±3.750
S12	-	32.166±8.646	27.971±1.665	31.723±3.425	28.991±6.857
S13	31.270±3.216	37.781±1.712	35.362±1.550	35.661±8.673	40.784±11.510
S15	33.988±4.785	35.683±4.652	-	28.012±5.541	44.664±2.938
S16	36.567±4.642	34.333±3.929	35.350±9.111	34.934±3.262	42.827±6.750
S17	28.183±5.067	29.453±8.427	32.131±3.770	30.945±5.603	30.148±3.628
Mean±STD	34.691±7.203	35.173±4.836	34.863±5.642	35.077±5.865	38.139±6.804

Positive values indicate a peak (maximum) knee extension moment during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 11: Individual mean peak knee abduction moments (Nm).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-12.673±2.140	-9.403±1.581	-10.641±1.574	-11.360±1.689	-15.233±3.590
S3	-5.929±1.035	-7.026±1.340	-8.221±0.768	-9.189±0.969	-9.669±1.294
S4	-4.417±1.210	-7.105±1.896	-6.565±0.887	-5.980±0.605	-5.769±0.829
S5	-4.156±0.321	-1.720±0.558	-2.616±1.190	-2.593±1.379	-5.884±1.544
S6	-18.515±2.778	-17.867±1.516	-20.296±3.630	-18.381±3.191	-21.113±2.728
S7	-6.627±1.812	-7.502±0.757	-8.303±1.813	-11.783±3.186	-12.112±0.907
S8	-7.503±2.410	-12.066±3.560	-12.017±1.034	-14.178±3.987	-12.933±1.660
S9	-11.608±0.943	-12.016±2.973	-11.171±1.270	-14.076±2.582	-11.826±1.541
S10	-14.283±1.148	-10.972±0.475	-10.833±1.340	-8.395±1.300	-16.225±2.736
S11	-12.357±2.011	-13.266±1.598	-13.158±0.947	-14.154±0.952	-18.977±1.597
S12	-	-9.621±2.547	-9.877±1.762	-13.308±2.693	-16.253±5.815
S13	-13.856±0.823	-11.013±1.737	-9.896±1.945	-10.633±2.712	-12.565±5.266
S15	-11.745±3.341	-11.896±2.394	-	-9.507±2.604	-15.714±1.968
S16	-10.135±1.574	-9.442±2.145	-8.200±2.721	-9.138±1.051	-12.618±2.266
S17	-3.716±1.533	-2.536±1.058	-1.584±1.019	-5.825±1.932	-6.074±1.855
Mean±STD	-9.823±4.495	-9.563±4.072	-9.527±4.522	-10.567±4.016	-12.864±4.622

Negative values indicate a peak (minimum) knee abduction moment during the power stroke. Values that were missing or removed from the sample are denoted with "-".

Table 12: Individual mean peak knee extension angles (°).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-34.763±0.453	-33.741±0.655	-33.825±1.671	-33.666±0.394	-33.063±1.337
S3	-30.535±0.428	-23.866±0.292	-29.187±0.570	-23.630±0.583	-27.847±1.135
S4	-45.395±0.600	-44.451±0.341	-44.878±0.290	-44.442±0.879	-44.949±0.475
S5	-33.032±2.729	-30.937±3.628	-31.246±2.080	-32.351±1.557	-29.176±1.661
S6	-34.837±0.676	-37.619±0.493	-35.689±0.798	-36.495±1.572	-37.077±0.922
S7	-30.560±1.257	-32.175±1.523	-28.308±0.403	-32.223±1.257	-32.399±0.831
S8	-42.156±0.968	-42.982±1.019	-43.459±0.941	-43.874±0.402	-42.272±1.240
S9	-21.018±1.176	-24.269±2.528	-26.130±1.842	-20.303±1.523	-23.407±0.908
S10	-35.769±0.482	-36.986±0.837	-30.672±0.784	-31.350±0.412	-32.377±0.900
S11	-28.039±0.952	-28.266±0.640	-27.898±1.063	-26.583±0.425	-26.514±0.762
S12	-26.446±0.716	-26.759±1.391	-24.450±1.519	-26.098±1.211	-23.830±1.401
S13	-33.033±0.797	-35.023±0.631	-33.904±0.381	-35.709±0.725	-33.249±1.039
S15	-26.310±1.317	-31.023±1.198	-29.073±0.344	-27.541±0.977	-28.507±1.382
S16	-35.269±1.025	-37.148±0.474	-35.429±0.782	-34.172±0.678	-34.887±0.743
S17	-40.051±1.592	-44.065±0.577	-43.579±0.559	-39.901±0.924	-38.720±0.858
Mean±STD	-33.148±6.404	-33.954±6.700	-33.182±6.455	-32.556±7.003	-32.552±6.319

Full extension = 0°; More negative values = greater knee flexion.

Table 13: Individual mean knee extension ranges of motion (°).

Participant	Q1	Q2	Q3	Q4	Q5
S2	78.953±0.427	79.277±0.765	78.886±1.537	79.119±0.786	78.849±0.834
S3	79.803±0.634	83.998±0.227	81.104±0.647	84.518±0.246	81.197±1.135
S4	66.975±0.585	69.125±0.452	68.378±0.265	68.314±0.834	67.908±0.295
S5	83.005±2.684	83.233±3.470	82.395±1.849	83.546±1.744	85.966±1.262
S6	71.844±0.456	71.114±0.341	71.302±0.740	72.478±1.347	71.590±0.715
S7	75.213±1.578	76.399±1.565	77.274±0.864	77.531±0.910	76.403±1.268
S8	69.280±0.860	67.801±1.468	67.815±0.800	66.641±0.487	68.821±1.068
S9	92.596±0.887	89.594±2.407	87.572±1.844	90.019±1.990	88.450±1.640
S10	80.551±0.757	79.549±1.061	83.974±0.860	84.438±0.287	83.939±0.803
S11	88.665±0.962	87.773±0.674	87.531±1.133	88.058±0.149	88.064±0.700
S12	82.643±0.597	82.123±1.251	84.400±1.285	81.247±0.778	84.927±1.097
S13	66.734±0.772	65.818±0.707	65.495±0.772	66.613±0.987	66.856±1.197
S15	72.066±1.588	72.713±2.211	72.456±0.704	70.843±1.029	75.329±1.735
S16	75.160±0.846	72.152±0.715	73.260±0.584	75.052±0.900	74.521±1.089
S17	77.063±1.622	72.100±0.415	70.727±0.788	76.354±0.883	75.624±0.759
Mean±STD	77.370±7.523	76.851±7.429	76.838±7.429	77.651±7.635	77.896±7.304

Individual knee extension ranges of motion were calculated as the difference between the peak knee extension angle and the initial knee sagittal plane angle. Positive values indicate the degree of knee extension from start to peak extension during the power stroke.

Table 14: Individual mean peak knee abduction angles (°).

Participant	Q1	Q2	Q3	Q4	Q5
S2	0.184±0.378	0.665±1.069	-0.071±0.651	0.216±0.654	0.481±0.926
S3	-2.277±0.144	-3.899±0.481	-1.912±1.190	-4.734±0.943	-3.589±0.495
S4	3.683±0.098	3.972±0.505	5.493±0.254	5.020±0.605	2.827±0.817
S5	-6.481±1.245	-5.329±0.588	-5.310±1.336	-6.342±0.575	-4.794±0.702
S6	4.882±0.775	7.012±1.303	8.481±0.741	7.279±0.738	6.351±1.082
S7	0.577±0.364	-0.150±0.934	-1.469±0.541	-0.675±0.861	-1.167±0.673
S8	4.489±0.965	5.410±1.127	3.570±0.763	4.445±1.191	5.209±0.713
S9	-	-	-	-	-
S10	4.981±0.819	6.138±0.865	3.657±0.937	2.493±1.101	2.587±0.381
S11	-	-	-	-	-
S12	1.004±1.256	0.486±1.076	0.512±0.809	2.696±0.710	-0.917±0.562
S13	-5.460±0.904	-3.110±0.558	-3.836±0.786	-4.687±0.678	-5.844±2.240
S15	1.601±1.665	0.877±0.901	0.704±0.398	-0.822±0.906	0.085±1.557
S16	6.656±0.660	4.607±0.583	3.313±1.194	3.452±1.044	4.277±0.641
S17	-3.095±0.941	-1.234±1.375	-3.297±1.353	-2.098±1.296	-3.527±1.343
Mean±STD	0.826±4.166	1.188±3.990	0.757±4.001	0.480±4.189	0.152±3.927

Peak knee abduction angles were the negative-most values of knee frontal plane angle during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 15: Individual mean knee abduction ranges of motion (°).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-6.180±0.646	-5.847±1.047	-5.848±0.622	-5.186±0.981	-4.454±0.918
S3	-4.127±0.866	-3.819±0.742	-2.492±1.008	-3.718±0.680	-3.380±0.366
S4	-1.149±0.198	-3.219±0.374	-1.520±0.389	-1.800±0.716	-0.288±0.658
S5	-5.356±0.788	-4.453±1.542	-5.500±0.743	-3.848±1.062	-1.913±1.025
S6	-5.474±0.579	-2.309±0.648	-5.115±0.800	-2.940±1.114	-0.679±1.346
S7	-6.458±0.593	-3.577±1.017	-5.267±0.585	-2.987±0.800	-3.236±0.533
S8	-4.888±0.832	-4.229±1.137	-2.844±0.688	-2.544±1.323	-3.380±0.785
S9	-	-	-	-	-
S10	-7.504±1.008	-7.288±0.917	-7.189±0.338	-6.357±1.683	-6.835±0.956
S11	-	-	-	-	-
S12	-10.783±1.664	-8.444±1.796	-7.954±1.034	-6.106±0.644	-7.631±0.525
S13	-5.958±1.378	-5.528±0.638	-5.780±0.639	-3.488±1.026	-5.183±1.626
S15	-9.765±1.365	-7.854±1.549	-10.739±0.599	-11.797±1.355	-7.876±1.890
S16	-2.407±1.100	-3.433±1.150	-2.569±1.302	-2.543±1.360	-1.609±0.690
S17	-4.270±0.850	-3.064±1.005	-4.498±1.341	-2.699±1.291	-2.353±0.825
Mean±STD	-5.717±2.636	-4.851±1.976	-5.178±2.532	-4.309±2.650	-3.755±2.517

Knee abduction range of motion was calculated as the difference between the peak knee abduction angle and the initial knee frontal plane angle. All but two of the participants displayed a peak abduction angle and range of motion during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 16: Individual mean peak knee adduction angles (°).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-	-	-	-	-
S3	-	-	-	-	-
S4	-	-	-	-	-
S5	-	-	-	-	-
S6	-	-	-	-	-
S7	-	-	-	-	-
S8	-	-	-	-	-
S9	5.533±0.544	6.018±1.684	6.572±0.448	5.166±0.791	5.183±1.971
S10	-	-	-	-	-
S11	4.789±0.557	5.694±0.406	5.424±0.718	5.456±0.988	5.421±0.722
S12	-	-	-	-	-
S13	-	-	-	-	-
S15	-	-	-	-	-
S16	-	-	-	-	-
S17	-	-	-	-	-
Mean±STD	5.161±0.526	5.856±0.229	5.998±0.812	5.311±0.205	5.302±0.169

Peak knee abduction angles were the positive-most values of knee frontal plane angle during the power stroke. Values that were missing or removed from the sample are denoted with "-".

Table 17: Individual mean knee adduction ranges of motion (°).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-	-	-	-	-
S3	-	-	-	-	-
S4	-	-	-	-	-
S5	-	-	-	-	-
S6	-	-	-	-	-
S7	-	-	-	-	-
S8	-	-	-	-	-
S9	0.551±0.602	0.595±2.100	0.838±0.649	2.507±0.000	3.137±1.087
S10	-	-	-	-	-
S11	2.388±0.702	2.384±0.977	2.524±0.520	1.813±0.695	4.247±0.765
S12	-	-	-	-	-
S13	-	-	-	-	-
S15	-	-	-	-	-
S16	-	-	-	-	-
S17	-	-	-	-	-
Mean±STD	1.470±1.299	1.489±1.265	1.681±1.192	2.160±0.491	3.692±0.785

Knee adduction range of motion was calculated as the difference between peak adduction angle and the initial knee frontal plane angle. Two participants displayed peak adduction angles and ranges of motion. Values that were missing or removed from the sample are denoted with “-“.

Table 18: Individual mean peak ankle plantarflexion moments (Nm).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-14.856±1.464	-13.938±0.662	-13.600±0.720	-15.658±2.289	-12.242±1.963
S3	-14.211±1.436	-16.114±1.983	-12.703±0.993	-16.569±0.678	-13.672±0.981
S4	-14.460±1.067	-13.000±0.829	-14.001±0.919	-13.583±2.859	-13.637±0.746
S5	-18.147±2.721	-14.937±1.746	-15.624±1.976	-17.628±2.514	-20.742±2.833
S6	-17.934±1.390	-15.293±2.124	-15.815±2.669	-16.878±1.286	-14.511±1.251
S7	-7.119±0.732	-10.492±1.104	-10.721±1.198	-11.406±2.211	-10.452±1.062
S8	-24.187±3.131	-24.142±1.635	-20.120±1.888	-16.609±1.594	-20.932±1.105
S9	-14.424±1.658	-16.015±4.307	-11.754±2.227	-13.758±3.360	-8.835±1.291
S10	-16.522±0.922	-14.172±0.492	-13.692±0.594	-10.717±0.441	-16.007±0.549
S11	-19.322±0.963	-16.912±1.995	-18.246±1.819	-15.820±2.423	-17.155±1.564
S12	-	-12.190±1.638	-15.424±1.421	-12.225±1.562	-16.016±3.158
S13	-20.223±1.592	-12.422±1.178	-11.467±1.412	-12.839±4.764	-14.251±3.054
S15	-10.970±0.625	-10.249±2.584	-	-9.769±0.780	-8.862±2.332
S16	-15.437±2.434	-16.591±0.639	-17.384±1.415	-15.813±1.311	-16.365±0.265
S17	-13.540±1.865	-9.790±0.561	-9.531±1.281	-15.181±1.026	-13.269±0.964
Mean±STD	-15.811±4.150	-14.417±3.560	-14.291±3.011	-14.297±2.456	-14.463±3.623

Negative values indicate a peak ankle plantarflexion moments during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 19: Individual mean peak ankle inversion moments (Nm).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-	0.069±0.000	-0.153±0.153	0.285±0.219	-0.391±0.000
S3	1.062±0.275	1.210±0.183	1.368±0.276	1.199±0.430	0.865±0.184
S4	0.121±0.115	-	-	-	-
S5	3.072±0.699	1.758±0.493	1.306±0.382	0.638±0.437	0.787±0.455
S6	0.820±0.171	1.200±0.263	0.553±0.157	0.034±0.221	0.228±0.479
S7	0.589±0.131	0.357±0.053	0.464±0.151	0.075±0.128	0.214±0.190
S8	0.929±0.143	0.945±0.210	1.198±0.728	0.921±0.194	0.681±0.610
S9	2.215±0.418	0.790±0.318	0.755±0.160	0.741±0.162	0.435±0.411
S10	0.517±0.209	0.453±0.123	-0.027±0.337	-0.648±0.234	-0.281±0.428
S11	2.055±0.355	1.222±0.098	0.937±0.344	0.021±0.119	0.246±0.313
S12	-	1.168±0.560	0.370±0.505	-0.360±0.383	-0.121±0.243
S13	0.242±0.255	0.577±0.202	0.687±0.353	-0.484±0.335	-
S15	0.396±0.891	0.150±0.146	-	-0.710±0.280	-0.244±0.301
S16	1.811±0.430	1.723±0.447	1.039±0.000	0.256±0.424	0.861±0.750
S17	2.350±0.461	1.686±0.329	1.184±0.167	0.999±0.215	-
Mean±STD	1.245±0.946	0.951±0.570	0.745±0.490	0.212±0.623	0.273±0.460

Positive values indicate peak (maximum) ankle inversion moments during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

Table 20: Individual mean peak hip extension moments (Nm).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-5.283±3.457	-8.453±1.783	-3.446±1.189	-2.127±2.013	-5.661±1.539
S3	-21.683±1.111	-13.827±2.389	-13.182±2.160	-16.239±2.158	-15.104±0.895
S4	-25.280±2.282	-12.981±1.699	-17.815±1.772	-11.875±0.625	-21.998±3.445
S5	-4.526±3.074	-6.539±1.551	-13.008±10.447	-6.808±0.000	-7.461±1.845
S6	-24.040±3.429	-17.402±1.377	-25.893±1.779	-13.082±2.441	-11.459±4.468
S7	-21.760±1.554	-3.980±0.885	-17.064±6.573	-4.070±0.676	-1.373±1.550
S8	-24.117±2.833	-19.490±3.042	-19.014±5.218	-14.316±2.856	-22.161±2.725
S9	-8.580±1.039	-4.381±10.899	-1.660±1.409	-7.661±0.954	-9.463±3.281
S10	-17.373±3.666	-14.311±1.029	-12.945±2.047	-17.544±2.297	-15.582±4.048
S11	-8.614±0.910	-8.761±0.531	-9.765±2.002	-10.717±0.753	-8.937±1.334
S12	-	-8.694±5.115	-15.389±4.988	-13.715±1.940	-8.601±4.886
S13	-35.941±2.857	-15.520±4.911	-21.479±3.455	-26.577±8.119	-14.840±7.759
S15	-21.021±3.265	-24.574±1.389	-	-20.719±1.030	-21.719±2.584
S16	-16.768±2.866	-24.019±3.234	-37.451±2.436	-28.816±2.014	-12.668±2.653
S17	-	-1.793±0.000	-	-	-10.436±4.607
Mean±STD	-18.076±9.163	-12.315±7.065	-16.009±9.266	-13.876±7.786	-12.498±6.146

Negative values indicate peak (minimum) knee extension moments during the power stroke. Values that were missing or removed from the sample are denoted with "-".

Table 21: Individual peak mean hip abduction moments (Nm).

Participant	Q1	Q2	Q3	Q4	Q5
S2	-20.733±3.426	-17.642±1.573	-19.243±2.080	-20.129±1.574	-24.383±5.891
S3	-10.889±1.089	-15.359±1.661	-14.922±2.078	-17.860±1.627	-18.903±2.038
S4	-11.276±1.894	-18.743±4.160	-13.960±1.410	-13.256±0.613	-11.892±4.792
S5	-12.741±3.902	-12.415±2.408	-12.964±3.958	-16.426±3.681	-17.852±4.217
S6	-24.191±4.223	-24.007±3.215	-25.357±5.720	-26.501±5.591	-33.302±4.196
S7	-14.761±3.408	-17.293±2.725	-19.300±3.527	-27.128±7.071	-26.206±2.170
S8	-11.926±5.466	-17.167±5.410	-19.914±2.737	-21.234±5.391	-20.755±2.443
S9	-19.052±2.074	-18.553±4.473	-18.059±2.864	-20.028±3.116	-18.868±3.766
S10	-26.727±1.879	-18.777±0.504	-17.316±1.675	-14.877±2.361	-28.076±4.159
S11	-24.501±3.378	-24.702±3.029	-21.928±1.663	-24.491±1.403	-33.199±3.317
S12	-	-22.840±3.854	-27.836±6.164	-27.524±3.982	-37.653±8.914
S13	-31.997±1.847	-28.907±3.120	-30.217±3.147	-32.546±11.694	-29.609±7.490
S15	-19.549±6.906	-21.423±4.188	-	-15.706±3.232	-30.874±3.673
S16	-10.436±1.413	-8.358±1.358	-6.947±2.179	-8.654±0.557	-15.342±1.719
S17	-15.506±3.552	-10.730±3.694	-8.659±0.832	-18.235±2.910	-16.796±2.441
Mean±STD	-18.163±6.780	-18.461±5.469	-18.330±6.691	-20.306±6.363	-24.248±7.705

Negative values indicate peak (minimum) hip abduction moments during the power stroke. Values that were missing or removed from the sample are denoted with “-“.

VITA

Jacob Wilbert lived in Medina, TN until he graduated from South Gibson County High School in Spring 2016 and departed to attend The University of Tennessee, Knoxville in Fall 2016. He received his Bachelor of Science degree majoring in Biological Sciences with a concentration in Biochemistry and Cellular & Molecular Biology in Spring 2020. He was admitted into the MS program in Kinesiology with a concentration in Biomechanics in the Department of Kinesiology, Recreation, and Sport Studies at The University of Tennessee, Knoxville in 2020. He completed his studies in Spring 2022 and will receive his Master of Science degree in Summer 2022. Starting in Fall 2022, Jacob will attend Alabama State University for a Master of Science program in Prosthetics and Orthotics. He will begin his career in clinical prosthetics/orthotics, and in the future, he hopes to conduct biomechanical research on how these devices affect human movement and subsequently how their designs could be improved.