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Effects of Varus Knee Alignment and Using Toe-cages on Frontal Plane Knee Biomechanics in Stationary Cycling

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I am submitting herewith a thesis written by Guangping Shen entitled "Effects of Varus Knee Alignment and Using Toe-cages on Frontal Plane Knee Biomechanics in 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.

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ABSTRACT

Effects of varus knee alignment on the internal knee abduction moment (KAM) in walking has been widely studied. KAM has been shown to be closely associated with the development of medial knee osteoarthritis (OA). Despite the importance of the knee alignment, no studies have explored its effects on knee frontal plane biomechanics during stationary cycling. The purpose of this study was to examine the effects of varus knee alignment and using a toe-cage on the knee frontal plane biomechanics during stationary cycling. Eleven participants in each of the varus and neutral groups participated in the study. The participants performed in six stationary cycling conditions: pedaling at 80 rpm at 0.5 kg (40 Watts), 1.0 kg (78 Watts), and 1.5 kg (117 Watts) with and without a toe-cage. A motion analysis system and a custom instrumented pedal were used to collect kinematic and kinetic data. A varus knee alignment and using toe-cage did not result in greater peak knee adduction angle and peak KAM. These findings suggest stationary cycling may be a safe exercise prescription for people with varus knee alignment, including patients with medial knee compartment OA. In addition, using toe-cage may not have any negative effects on knee joints in stationary cycling. Future studies may be needed to explore the tibiofemoral contact force in subjects with a varus knee alignment during stationary cycling.

Keywords: knee alignment, knee abduction moment, knee OA, cycling, toe-cage

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CHAPTER I

INTRODUCTION

Osteoarthritis (OA) is one of the most prevalent musculoskeletal disorders, that affects approximately 27 million people (Lawrence et al., 2008) and by 2030 it is projected to increase to 67 million in the United States (Hootman & Helmick, 2006). OA mostly affects weight bearing joints of the lower extremity such as knees and hips, which leads to pain, loss of function and restriction in daily activity and disability (van der Waal, Terwee, van der Windt, Bouter, & Dekker, 2005).

Although cycling is not directly included as a non-surgical treatment for OA in Osteoarthritis Research Society International (OARSI) guidelines, it is frequently prescribed by health professionals to improve physical fitness, muscle strength, and function in rehabilitation programs (Gail D Deyle et al., 2005; G. D. Deyle et al., 2000; Naal et al., 2007; Salacinski et al., 2012; Walker, Gotterbarm, Bruckner, Merle, & Streit, 2014). Studies have shown that the varus alignment of knee joints can increase the internal knee abduction moment (KAM) (Barrios, Davis, Higginson, & Royer, 2009; Mundermann, Dyrby, & Andriacchi, 2005; Stief, Bohm, Schwirtz, Dussa, & Doderlein, 2011) and be responsible for the incident and progression of knee OA (Sharma et al., 2010). Although cycling creates relatively small overall loading to the knee, how varus alignment of knee joints affect the external KAM during cycling deserves additional attention.

The measurement of knee alignment is an important component of the diagnosis of musculoskeletal diseases in knee joints (Moreland, Bassett, & Hanker, 1987). Many methods have been developed to investigate knee alignment including direct and indirect measurement methods (Mündermann, 2012). The method using radiographic measurement is identified as the gold standard for the knee alignment measurement, owing to its ability to accurately locate the bony landmarks (Moreland et al., 1987; Sharma et al., 2001). Many studies have applied this method for the alignment measurement (Hinman, May, & Crossley, 2006; Hsu, Himeno, Coventry, & Chao, 1990; Kraus, Vail, Worrell, & McDaniel, 2005; Moreland et al., 1987; Mündermann, 2012; Navali, Bahari, & Nazari, 2012; Sharma et al., 2001; Vanwanseele, Parker, & Coolican, 2009). Knee alignment can be categorized as varus, valgus and neutral (Mündermann, 2012). Biomechanically, a neutral alignment is defined as 180° of mechanical axis angle. A varus alignment is defined as the medial (inside measurement) mechanical axis angle smaller than 180° whereas a valgus alignment was defined as the medial mechanical axis angle greater than 180° (Moreland et al., 1987). The mechanical axis angle of -2.0° to 2.0° seems to be the most frequently used range of neutral alignment in different studies (Felson et al., 2013; Hunter, Sharma, & Skaife, 2009; Issa et al., 2007; Kraus et al., 2005; Leitch, Birmingham, Dunning, & Giffin, 2013; Messier et al., 2014; Sharma et al., 2010; Stief et al., 2014).

In level walking, knee OA patients with a varus alignment have less knee flexion angle at heel strike, less knee range of knee flexion (ROM), greater knee adduction angle and greater external KAM (Foroughi et al., 2010; Hurwitz, Ryals, Case, Block, & Andriacchi, 2002; Messier et al., 2014; Turcot et al., 2013). For healthy subjects with the varus alignment, greater knee adduction angle, increased KAM and increased stress on medial compartment have been observed during level walking (Barrios, Davis, et al., 2009; Barrios & Strotman, 2014; Stief et al., 2011; Yang, Nayeb-Hashemi, Canavan, & Vaziri, 2010). Comparing the knee biomechanics of varus alignment between knee OA patients and healthy subjects, both cohorts showed the increased KAM during level walking. The discrepancy was found in the knee flexion angle and the knee flexion moment as knee OA patients had a smaller knee flexion angle and flexion moment than the normal subjects. Therefore, the decreased knee flexion ROM or the decreased knee flexion moment in knee OA patients was likely due to the presence of the knee OA rather than the knee malalignment.

In cycling, the knee sagittal plane ROM has been found between 65° to 77° and the knee frontal plane ROM between 6° of adduction to 4° of abduction (R. R. Bini, Tamborindeguy, & Mota, 2010; Ercison, Nisell, & Nemeth, 1988; Ericson, 1986; Y. Fang, Fitzhugh, Crouter, Gardner, & Zhang, 2014; Gardner et al., 2015). The kinetic variables in cycling are more sensitive to cycling posture and workload than the kinematic variables (Y. Fang et al., 2014). As a result, there are tremendous discrepancies in the kinetic results, and it entails relating the kinetic variables to the workload and cycling posture when explaining the results from different studies. Regarding cadence and workload, knee kinematics tend to be more influenced by the cadence compared to the workload (R. R. Bini et al., 2010). A knee flexion of 25° – 30°, when the pedal is at the bottom dead center and the cyclist is seated on the saddle, is thought to be as an optimal saddle height for performance and injury prevention (R. Bini, Hume, & Croft, 2011).

STATEMENT OF PROBLEM

To our knowledge, no studies have explored the effects of varus knee alignment on knee frontal plane biomechanics during stationary cycling. Knee malalignment can result in abnormal load on the knee joints in gait (Barrios, Davis, et al., 2009; Stief et al., 2011) and it can cause the incident and progression of knee OA (Sharma et al., 2010; Sharma et al., 2001). Whether varus knee malalignment can lead to excessive load during cycling and increase the risk of knee OA has not been investigated. Furthermore, it remains unclear whether using a toe-cage would negatively influence frontal plane kinetics during cycling. Therefore, the purpose of this study was to examine the effects of varus knee alignment and using toe-cages on the knee frontal plane biomechanics during stationary cycling.

HYPOTHESIS

1. Participants with knee varus alignment will have greater internal knee abduction moment compared to the participants with normal alignment during cycling and walking.
2. Internal knee abduction moment will not differ with or without a toe-cage during cycling.

DELIMITATIONS

1. Subjects should be free from lower extremity injuries during the past six months.
2. Knee alignment obtained from radiographic method is the actual alignment of the subjects.

3. Subjects should be able to ride a stationary bike without any assistance for sixteen minutes.

LIMITATIONS

1. All tests were conducted in a laboratory setting.
2. Pedal reaction forces were collected on the left pedal only.
3. The accuracy of the results was limited by the accuracy of the instruments used in the study; and the accuracy of estimating joint centers was limited by the accuracy of placements of the anatomical markers.

CHAPTER II

LITERATURE REVIEW

The purpose of this study was to examine effects of knee alignment and using toe-cages on knee frontal plane biomechanics during stationary cycling. This chapter was to review existing literature on background, measurement of knee alignment, the biomechanics of knee malalignment and biomechanics of cycling.

BACKGROUND

Osteoarthritis (OA) is one of the most prevalent musculoskeletal disorders, that affects approximately 27 million people (Lawrence et al., 2008) and by 2030 it is projected to increase to 67 million in the United States (Hootman & Helmick, 2006). OA mostly affects weight bearing joints of the lower extremity such as knees and hips, which leads to pain, loss of function and restriction in daily activity and disability (van der Waal et al., 2005). Although the exact cause of OA remains unclear, several risk factors have been identified including modifiable risk factors such as obesity (Lee et al., 2013), occupational activity (Palmer, 2012), and previous injury (Ajuied et al., 2013). In addition, there are non-modifiable risk factors, namely malalignment (Felson et al., 2013; Sharma et al., 2010), ethnicity, age, and female gender (Blagojevic, Jinks, Jeffery, & Jordan, 2010; Wright, Riggs, Lisse, Chen, & Women's Health, 2008). Risk factors contribute to abnormal load-bearing in the joint and subsequent cartilage damage, subsequently leading to incidence and progression of OA.

In addition to the studies on risk factors of OA, there have been a number of studies on treatments of OA including surgical and non-surgical treatments. Surgical treatments using joint replacement surgery are mostly performed in patients with severe OA, which is effective in relieving pain and disability. However, surgical procedures are costly and have risks, complications and long process of rehabilitation. Consequently, it is only considered when non-surgical treatments have been exhausted (Van Manen, Nace, & Mont, 2012).

Osteoarthritis Research Society International (OARSI) has made 29 evidence-based non-surgical recommendations for the management of knee OA, based on studies conducted between 2002 to 2013 (McAlindon et al., 2014; Zhang et al., 2010). Among the 29 non-surgical recommendations, core treatments appropriate for all individuals are identified as land-based exercises, weight management, strength training, water-based exercise, self-management and education (McAlindon et al., 2014). These treatments were found to be effective in reducing pain and improving physical functions because they reduce excessive load on knee joints during exercise while still involving the muscles around the joints.

Although cycling is not directly included as a non-surgical treatment for OA in OARSI guidelines, it is frequently prescribed by health professionals to improve physical fitness, muscle strength, and function in rehabilitation programs (Gail D Deyle et al., 2005; G. D. Deyle et al., 2000; Naal et al., 2007; Salacinski et al., 2012; Walker et al., 2014). Cycling has advantages in reducing the knee joint loads (D'Lima, Steklov, Patil, & Colwell, 2008; Kutzner et al., 2012) and promoting oxidative metabolism for weight loss. During cycling workout, most of the body weight is on the saddle, which allows weight-bearing joints such as knees and hips to experience relatively lower compressive force than during walking or jogging (D'Lima et al., 2008; Kutzner et al., 2012).

The external knee adduction moment (KAM) in walking has been shown to be closely associated with the development of medial knee OA (Mundermann, Dyrby, Hurwitz, Sharma, & Andriacchi, 2004; Sharma et al., 1998; Zhao et al., 2007). Studies have shown that the varus alignment of knee joints can increase the external KAM (Barrios, Davis, et al., 2009; Mundermann et al., 2005; Stief et al., 2011) and be responsible for the progression of knee OA (Sharma et al., 2010). Although cycling creates overall loading in the knee of small magnitude, how knee joints with varus alignment behave during cycling deserves additional attention.

MEASUREMENT OF KNEE ALIGNMENT

The measurement of knee alignment is an important component for the diagnosis of musculoskeletal diseases in knee joints (Mündermann, 2012). Different types of knee alignment measures and measurement methods have been developed during the past few decades.

Common measures of knee alignment

Common measures regarding knee alignment are anatomical axis angle and mechanical axis angle (Mündermann, 2012). Both measures can be performed in the sagittal plane, frontal plane and transverse plane. However, for knee alignment in OA population, the frontal plane knee axis angle is most important, which will be the focus of this project.

The mechanical axis of the femur is formed by a line drawn from the center of the femoral head to the center of the femoral intercondylar notch and the mechanical axis of the tibia is from the center of the intercondylar eminences to the center of the ankle talus (Moreland et al., 1987; Sharma et al., 2001). The mechanical axis angle of the knee joint is the angle between the mechanical axes of femur and tibia.

Using mechanical axis to determine the knee alignment is cumbersome and requires specialized testing protocols such as a full-length lower limb radiograph (Kraus et al., 2005); it has been identified as the gold standard for knee alignment measurements as they have been broadly used in biomechanical studies. Consequently, the rest of the reviews on knee alignment is based on studies using the mechanical axis angles of the knee joint.

The anatomical axis of the femur is formed by a line from the center of the intercondylar eminences to a point 10 cm above the intercondylar eminences midway between the medial and lateral femoral surfaces (Moreland et al., 1987). The anatomical axis of the tibia is formed by a line drawn from the center of the intercondylar eminences to a point 10 cm below the intercondylar eminences, midway between the medial and lateral tibial surfaces (Moreland et al., 1987). Then the anatomical axis angle of the knee joint can be described as the angle between the anatomical axes of femur and tibia.

The anatomical axis angle is relatively easy and convenient to measure because it only requires a regular radiographic facility compared to the measurement of mechanical axis angle. However, the anatomical axis can be easily influenced by the deformity of the tibia or femur because of its dependence on bone morphology (Mündermann, 2012). More importantly, it is not capable of accurately reflecting the mechanical path of the force through the femur as it does not include measurements related to the neck and head of the femur. Hence, there have been few studies utilizing the anatomical axis as an indicator of knee malalignment (Brouwer et al., 2007; Lim, Hinman, Wrigley, Sharma, & Bennell, 2008).

Several studies have correlated the anatomical axis angle with the mechanical axis angle, which is the gold standard for measuring knee alignment, attempting to create alternative approaches. Kraus et al. (2005) compared the measurement of knee alignment by mechanical axis angle and anatomical axis angle in both knees of 57 participants who had knee OA. A significant correlation ($r = 0.65$, $p < 0.0001$) was found between the mechanical axis angle and anatomical axis angle. Hinman et al. (2006) found a significant correlation ($r = 0.88$, $p < 0.001$) between the mechanical axis angle and anatomical axis angle in 40 participants with symptomatic medial knee OA. Issa et al. (2007) also found a significant correlation ($r = 0.86$; 95% CI 0.81, 0.90) between the mechanical and anatomical axis angles in 146 knee OA patients. In addition, Navali et al. (2012) found a significant correlation ($r = 0.93$, $p < 0.001$) between the two angles in 100 knees of 50 participants with frontal knee malalignment.

Common methods of knee alignment measurements

Many methods have been developed to investigate knee alignments including direct and indirect measurement methods (Mündermann, 2012). Direct measurement methods utilize imaging techniques to locate the exact bony landmarks and obtain the mechanical axis angles (Mündermann, 2012). Indirect measurement methods usually need to estimate the location of anatomical landmarks and use a regression equation to estimate the mechanical axis angles (Andriacchi & Strickland, 1985).

Radiographic measurements

The method using radiographic measurement is identified as the gold standard for the knee alignment measurements, owing to its ability of accurately locating the bony landmarks (Moreland et al., 1987; Sharma et al., 2001). Therefore, many studies have applied this method for the alignment measurement (Hinman et al., 2006; Hsu et al., 1990; Kraus et al., 2005; Moreland et al., 1987; Mündermann, 2012; Navali et al., 2012; Sharma et al., 2001; Vanwanseele et al., 2009). Although there are some minor variations in x-ray settings among different studies, the primary protocols are similar and are mainly based on the methods described by Moreland et al. (1987) and Sharma et al. (2001).

In general, the anteroposterior view of a weight-bearing radiograph of the lower extremity including hip, knee and ankle is captured with the graduated-grid x-ray cassette (Moreland et al., 1987; Sharma et al., 2001). The cassette height varied from 91.4 cm to 136.0 cm depending on the body height. The cassette width is usually from 35.5 cm to 36.0 cm. A subject stands barefoot with knees in full extension and the tibial tubercles facing the x-ray beam. The focal distance of 2.4 m is most frequently applied. For the x-ray power settings, the voltage ranges from 77 kilovolts to 95 kilovolts and the electric current commonly has a range of 100 mA/s to 300 mA/s, depending on the limb size and tissue characteristics.

Conventionally, the mechanical axis angles were manually obtained by drawing the lines on hard copy radiographs, which potentially introduces errors to the results. In order to minimize the errors introduced by testers, digital radiographs are now commonly used to calculate the angles between mechanical axes automatically (Sailer et al., 2005). Hankemeier et al. (2006) analyzed lower extremity geometry of 59 long leg radiographs by both conventional and the computer-assisted method, where they found that the computer-assisted method significantly reduced the standard deviation of variables ($p < 0.05$) and reduced the time needed for analysis ($p < 0.001$). Marx et al. (2011) tested the mechanical axis angles in 42 subjects and found the computer-assisted method had both greater intra-rater reliability (ICC: 0.93 – 0.99) and inter-rater reliability (ICC: 0.93 – 0.97) compared to the conventional method (intra-rater reliability: 0.86 – 0.96; inter-rater reliability: 0.88 – 0.94). However, there have been some studies showing no difference in reliability between the two methods (Fakhrai et al., 2010; Sailer et al., 2005; Sled et al., 2011).

Magnetic resonance imaging measurements

The magnetic resonance imaging (MRI) method is another direct measurement method for knee alignments. The MRI method is advantageous in terms of safety, compared to the radiographic method, where the pelvic region of participants is exposed to the radiation. However, the MRI method also has disadvantages. For example, only an open-bore MRI system allows a participant to be positioned in an upright weight-bearing position and entails a cumbersome setup to align the patient's limb in the magnet (Mündermann, 2012). Moreover, when compared to the radiographic methods, the leg length and mechanical axis deviation are underestimated using the MRI method (Hinterwimmer, Graichen, Vogl, &

Abolmaali, 2008; Liodakis et al., 2011). In addition, the magnitude of underestimation of valgus alignment was -3.6 ± 2.8° ($p < 0.05$) compared to the radiographic method (Hinterwimmer et al., 2008). Additionally, the cost of MRI measurement is another prohibiting factor because the MRI usually costs much more than the radiographic method. As a result of the disadvantages described above, few studies have utilized the MRI method for knee alignment testing (Hovinga & Lerner, 2009).

Measurements using 3D motion capture system

3D motion capture systems are commonly used in biomechanics for investigating human movements. Recently, several studies have attempted to measure knee alignment using a 3D motion capture system (Michael A Hunt, Trevor B Birmingham, Thomas R Jenkyn, J Robert Giffin, & Ian C Jones, 2008; Kornaropoulos et al., 2010; Mündermann, Dyrby, & Andriacchi, 2008; Vanwanseele et al., 2009).

In general, reflective markers placed on the medial and lateral malleoli are used to determine the ankle joint centers and markers on medial and lateral epicondyles to determine knee joint centers. Hip joint centers are determined by markers around the pelvis and proximal thigh. Eventually, the mechanical axis angle of the thigh and shank could be obtained by connecting the joint centers. Although the accuracy of locating ankle and knee joint centers based on skin makers has been widely accepted in biomechanics, there are still some discrepancies in the results from different methods of hip joint center estimation that have been developed (Andriacchi & Strickland, 1985; Bell, Brand, & Pedersen, 1989; Seidel, Marchinda, Dijkers, & Soutas-Little, 1995). The accuracy of mechanical axis angle determined by 3D motion capture system depends highly on the accuracy of hip joint center location as the hip joint center is one of the components forming the mechanical axis.

Hunt et al. (2008) correlated the mechanical axis angles estimated using a motion capture system and the radiographic method with the center of hip determined using the Bell method (Bell et al., 1989) and found that two angles were highly correlated ($r = 0.84$, $p < 0.05$). Mündermann et al. (2008) utilized the method proposed by Andriacchi et al. (1985) to determine the hip joint center and found a significant correlation ($r = 0.738$, $p < 0.001$) between the two mechanical axis angles measured from a motion capture method and a radiographic method. Kornaropoulos et al. (2010) found a significant correlation ($r = 0.91$, $p < 0.0001$) between the mechanical axis angles from the motion capture method and a radiographic method using a functional hip center method (Cappozzo, 1984).

Although the measurement method using a 3D motion capture system provides great convenience, safety and a relatively low cost compared to the radiographic method, the accuracy of measurements needs to be improved in order to be acceptable for use in research.

Other estimation methods

Many indirect clinical methods have been studied as alternative ways for determining knee joint alignment (Gibson, Sayers, & Minor, 2008; Hinman et al., 2006; Kraus et al., 2005; Navali et al., 2012; Vanwanseele et al., 2009). Indirect clinical methods include the caliper method, inclinometer method and goniometer method.

With the caliper measurement method, subjects are asked to adduct both of their lower limbs slowly until either the knees or ankles touch (Hinman et al., 2006; Navali et al., 2012). When the knees and ankles touch simultaneously, the alignment is recorded as neutral. If the knees touch first, the subject is classified as valgus malalignment and the distance between the medial malleoli is measured with a caliper. If the ankles touch first, the subject is classified as varus malalignment and the distance between medial knee joint lines (Hinman et al., 2006) or medial epicondyles of femur (Cibere et al., 2004; Navali et al., 2012) is measured. Moderate ($r = 0.76$, $p < 0.001$) to high ($r = 0.90$, $p < 0.0001$) correlations between the radiographic measurement and the caliper method have been found by Hinman et al. (2006) and Navali et al. (2012).

Given the caliper method can be affected by the alignment of both knee joints, a plumb-line can be positioned between the lower limbs and the distance between the plumb-line and the medial knee or ankle is measured to target the alignment of a particular limb. A moderate correlation ($r = 0.71$, $p < 0.001$) was observed between the radiographic and plumb line methods (Hinman et al., 2006).

Errors in the caliper and plumb-line methods can be introduced when the excessive soft tissue on the medial knee joints takes up too much space between knees (Navali et al., 2012). For example, excessive soft tissue in the medial knees can deceptively suggest that participant's knees touch earlier than the ankles, resulting in the wrong judgment and measurement of knee alignments. In addition, another important factor affecting the caliper method is the length of the lower limbs. Given the same mechanical axis angle, a taller subject would have a greater distance between their knees and ankles than shorter subjects. Therefore, some potential exists to improve the accuracy of caliper and plumb-line methods as normalizing the distance between knees and ankles by the length of lower limbs.

The inclinometer method was first developed by Hinman et al. (2006) as an alternative clinical measurement of knee alignment. A subject is asked to stand with both feet apart according to foot maps used in previous x-ray tests. The angle of the tibia identified by the tibial tuberosity and neck of talus is measured by a gravity inclinometer. A moderate correlation ($r = 0.80$, $p < 0.001$) with the radiographic method was observed (Hinman et al., 2006). Vanwanseele et al. (2009) also found a significant correlation between the radiographic method and the inclinometer method ($r = 0.83$, $p < 0.001$).

One main concern about the inclinometer method is that it only takes into account the shank angle with respect to the vertical direction (Vanwanseele et al., 2009). However, a shank angle can be easily influenced by factors such as step width and pelvis width. Accordingly, the knee alignment measured by the inclinometer method tends to be valgus with a greater step width and to be varus with a greater pelvis width. Therefore, in order to take comparable measurement by the inclinometer method, the step width according to pelvis width needs to be normalized.

Several studies (Gibson et al., 2008; Hinman et al., 2006; Kraus et al., 2005; Navali et al., 2012) have used an extended goniometer to investigate knee alignment. The arms of the goniometer should be aligned with the thigh to the anterior superior iliac spine and along the axis of the shank to the neck of talus. The axis of the goniometer should be placed on the center of the patella. Moderate correlation coefficients of 0.74 ($p < 0.001$) and 0.70 ($p < 0.001$) were found between the goniometer and radiographic methods by Gibson et al. (2008)

and Kraus et al. (2005) , respectively. However, no significant correlation was found in the study by Hinman et al. (2006) and weak correlation ($r = 0.67$, $p < 0.0001$) was found by Navali et al. (Navali et al., 2012). Hinman et al. (2006) stated that the center of the patella can be an issue for the goniometer method because in subjects with patellar subluxation it is not the center of the knee joint, where the axis of the goniometer is supposed to be placed.

In summary, the advantages of indirect clinical methods (caliper, inclinometer and goniometer) for determining frontal-plane knee alignment are their simplicity, quickness and low-priced process of operation compared to the radiographic and MRI methods. However these methods suffer from relatively low validity. Selection of an appropriate method to measure knee alignment depends on the purpose of the study. A study that needs a high level of accuracy of knee alignment may need to choose the radiographic or MRI method. Alternatively, indirect clinical methods for knee alignment measurements may be more suitable in applications where the priority of measurement is the simplicity or quickness.

Classification of knee alignments

Knee alignment can be categorized as varus, valgus and neutral (Mündermann, 2012). Biomechanically, a neutral alignment is defined as 180° of the mechanical axis angle. A varus alignment is defined as the medial (inside measurement) mechanical axis angle smaller than 180°, whereas a valgus alignment is defined as the medial mechanical axis angle greater than 180° (Moreland et al., 1987).

Several studies attempted to find the range of mechanical axis angle for the normal population (Chao, Neluheni, Hsu, & Paley, 1994; Cooke et al., 1997; Hsu et al., 1990;

Moreland et al., 1987), however, there have not been established ranges of normal knee alignments. All of these studies utilized the standardized radiograph of the entire lower extremity using the method proposed by Moreland et al. (1987) to calculate the mechanical axis angle. The average for normal knee alignment is around 1.0° – 1.3° varus with standard deviations between 2.0° – 2.8° (Table 1).

Table 1. Mean mechanical axis angles of healthy subjects in different studies.

Study	Subject	Mechanical axis angle (°)*
Moreland et al., 1987	25 normal males	-1.3 ± 2.0°
Hsu et al., 1990	120 normal subjects	-1.2 ± 2.2°
Chao et al., 1994	127 healthy Caucasians	-1.2 ± 2.2°
Cooke et al., 1997	119 healthy adults	-0.97 ± 2.86°

*: Deviation from 180°; negative values refer as varus alignment.

In addition, there have been numerous studies regarding the knee alignment, where different ranges of normal alignments were defined by authors based on the design of studies or so called “conventional protocols” (Felson et al., 2013; Hunter et al., 2009; Issa et al., 2007; Kraus et al., 2005; Leitch et al., 2013; Messier et al., 2014; Sharma et al., 2010; Stief et al., 2014). The mechanical axis angle of -2.0° to 2.0° is the most frequently used range of neutral alignment in different studies (Table 2).

In summary, it seems appropriate to propose the mechanical axis angle of -2.0° to 2.0° as the neutral alignment. This range is most widely used and covers the average mechanical axis angle of the normal population.

Table 2. Ranges of neutral alignment in different studies.

Study	Subject	Ranges of neutral alignment*
Kraus et al., 2005	114 knees of 57 knee OA patients	-1.5° – 0°
Issa et al., 2007	146 knee OA patients	-2.0° – 2.0°
Hunter et al., 2009	Literature Review	-2.0° – 0°
Sharma et al., 2010	MOST (2958 knees)	-2.0° – 2.0°
Felson et al., 2013	MOST (5053 knees) & OAI (5953 knees)	-1.0° – 1.0°
Leitch et al., 2013	26 knee OA patients & 13 asymptomatic subjects	-2.0° – 2.0°
Messier et al., 2014	157 knee OA patients	-2.0° – 0°
Stief et al., 2014	18 subjects with knee varus alignment	-1.3° – #

*: Deviation from 180°; negative values refer as varus alignment; positive values refer as valgus alignment.

#: The value was not presented.

BIOMECHANICS OF KNEE MALAIGNMENT IN GAIT

It is necessary to understand biomechanical characteristics of people with knee malalignment during gait because it is possible that the changes of knee biomechanics in gait due to knee malalignment may translate into cycling. Furthermore, there has been no study regarding the effect of malalignment on the knee biomechanics during stationary cycling. Therefore, literature in gait studies is the only source that can be utilized to provide literature support for the present study in cycling.

Two cohorts of subjects have been examined for the biomechanics of knee malalignment, including knee OA patients and healthy subjects who have knee malalignment. Varus alignment was the most common type of knee malalignment in both cohorts of subjects with altered gait patterns compared to a healthy population (Barrios, Davis, et al., 2009; Hurwitz et al., 2002; Miyazaki et al., 2002; Stief et al., 2011; Turcot et al., 2013). While studying knee OA patients is necessary to understand the progressive nature of the disease, evaluating populations at risk for developing OA is also critical for identifying preventative measures (Barrios & Strotman, 2014).

Gait biomechanics of knee malalignment of healthy population

Barrios et al. (2009) studied the biomechanics of varus alignment in healthy subjects. An inclinometer was used to measure the mechanical axis angle of tibia. Subjects with a mechanical axis angle >10° varus were included in the varus group and 7° - 9° varus for the control group. Seventeen healthy subjects with varus alignment and 17 controls with normal alignment were asked to walk at 1.46 m/s. The peak KAM was 42% greater ($p < 0.001$) in the varus group (0.4 ± 0.06 Nm/kg/m) compared to the normal alignment group (0.28 ± 0.05 Nm/kg/m). The peak knee adduction angle was 5.5° greater ($p < 0.001$). The peak knee flexion angle during early stance was approximately 4° greater in the varus group ($p < 0.05$). No difference was observed in the external peak knee flexion moment between the varus group (0.39 ± 0.11 Nm/kg/m) and the control group (0.33 ± 0.09 Nm/kg/m). With the same inclusion criteria, Barrios et al. (2014) conducted another study where 30 varus subjects and 30 normal subjects walked at 1.46 m/s. The peak knee adduction moment was 35% greater ($p < 0.001$) for the varus subjects (0.39 ± 0.07 Nm/kg/m) compared to the control group (0.289 ± 0.0465 Nm/kg/m). The peak knee adduction angle was 5° greater in the varus group compared to the control group ($p < 0.05$). Moreover, the varus group had 3° greater peak knee flexion angle at mid-stance ($p < 0.001$).

As stated in the previous section, the gold standard for knee alignment measurement is the radiographic method (Moreland et al., 1987; Sharma et al., 2001). Although it was clarified that the inclusion criteria was based on a normative database of 30 heathy individuals measured by inclinometer (Barrios, Davis, et al., 2009), the specific process of building the normative database was not clearly articulated. As a result, it is possible that the subjects included in these two studies (Barrios, Davis, et al., 2009; Barrios & Strotman, 2014) might not be the varus alignment or the normal alignment as they were supposed to be. Hence, the results of these two studies may not be representative of gait biomechanics of a specific knee alignment.

Similarly, Stief et al. (2011) studied gait biomechanics of varus knee alignments in youth. Fourteen youth with a varus alignment were recruited for the study and 15 age-matched subjects with a normal knee alignment were included as the control group. The varus alignment was identified by the standard full-limb radiographic method and the mean mechanical axis angle was 8.86 ± 7.38° varus. The neutral alignment was identified as 0° to 1.3° varus. The subjects walked at a self-selected speed and no difference in walking speed was observed between the two groups ($p > 0.05$). Peak knee adduction angle was approximately 7° greater ($p < 0.001$), and the peak external KAM was 32% greater in the varus group compared to the control group ($p < 0.01$). There was no difference in the peak knee flexion angle and the peak knee flexion excursion ($p > 0.05$). However, the peak knee extension angle and the peak external knee extension moment were greater in the control group compared to the varus group ($p < 0.05$). The study concluded that the kinematics and kinetics of varus alignment in healthy participants was somewhat different than the knee OA patients (Stief et al., 2011). In fact, this finding is not useful to the knee OA study because the varus participants were youth instead of the adults, who are the primary population suffering from knee OA (Lawrence et al., 2008). Some studies have shown that youth, who have less developed motor control, could exhibit different gait patterns compared to adults (Ganley & Powers, 2005; Sutherland, 1997). Therefore, it is not applicable to compare the gait pattern of youth with adult knee OA patients. It is important to recruit the participants of ages comparable to knee OA patients so that the findings of study can be generalizable to the mechanism of knee OA development.

There are several studies that have investigated the loading distribution on the knee cartilage in the different alignments. Yang et al. (2010) analyzed loading distribution of three participants with one in a varus, one in a valgus and one in a neutral alignment utilizing participant-specific three-dimensional finite element methods. The study showed the peak compressive load occurred at approximately 25% of the stance phase. Moreover, varus alignment had the largest stress at the medial compartment of the knee compared to the participants with normal alignment and valgus alignment. Werner et al. (2005) investigated the loading distribution of neutral, 3° and 5° in varus, and valgus alignment in cadaver legs with a pressure sensor inserted in the knees. A physiological knee simulator was used to simulate gait trials. The study showed either a 3° or 5° varus or valgus angulation caused a statistically significant change in the load distribution compared to the neutral alignment ($p < 0.05$). More specifically, the varus alignment led to a greater load in the medial compartment, whereas the valgus alignment resulted in a greater load in the lateral compartment.

Several studies have also investigated the correlation between knee malalignment and external KAM during level walking (Andrews, Noyes, Hewett, & Andriacchi, 1996; Barrios, Higginson, Royer, & Davis, 2009). A moderate correlation ($r = 0.69$, $p < 0.001$) was found between the varus alignment (mechanical axis angle) and the peak KAM during level walking in 11 healthy participants (Andrews et al., 1996). Barrios et al. (2009) found there was a significant correlation ($r = 0.74$, $p < 0.001$) between the tibial mechanical axis angle and the external KAM among 37 young asymptomatic knees that varied from normal to varus-aligned. Additionally, dynamic knee alignment appeared to be another good predictor for KAM ($r = 0.68$, $p < 0.001$).

In summary, the mechanical axis angle of knee joints is correlated moderately with the external KAM in healthy participants. Varus alignment of knee joints contributes to an increased external KAM and increased medial compartment load compared to a neutral alignment, suggesting individuals with a varus knee alignment might be more susceptible to developing medial compartment OA.

Gait biomechanics of knee malalignment in population with knee OA

Hurwitz et al. (2002) investigated the knee biomechanics in 62 knee OA participants ($5 \pm 5°$ varus) and 49 asymptomatic controls who walked at approximately 1 m/s. The varus alignment was defined as the mechanical axis angle $> 0°$, whereas the valgus alignment was $< 0°$. The first peak KAM during early stance was significantly greater in the varus group ($p < 0.05$) and the mechanical axis angle was the best predictor of the peak KAM ($r = 0.74$, $p < 0.001$).

Forty-six knee OA patients with varus alignment ($8.2 \pm 5.2°$ varus), 14 knee OA patients with valgus alignment ($3.5 \pm 4.1°$ valgus) and 26 healthy controls were recruited in a study by Turcot et al. (2013). Varus was defined when the knee angle was less than 0° and valgus when the knee angle was greater than 0°. Participants walked at a self-selected speed and there was no difference in gait velocity among the three groups ($p > 0.05$). The knee flexion ROM during stance phase was approximately 10° lower in varus group ($p < 0.001$) and valgus group ($p < 0.05$) compared to the control group, while there was no difference in the knee flexion ROM between malalignment groups, suggesting independent of knee alignment the knee OA patients walked with more extended knee than healthy participants.

The external knee flexion moment was around 38% lower in the varus group ($p < 0.001$) and valgus group ($p < 0.05$). In addition, the varus group had the largest peak knee adduction angle ($p < 0.05$) and the greatest peak KAM ($p < 0.05$) among the three groups.

Foroughi et al. (2010) compared the frontal plane knee biomechanics in 17 knee OA patients and 17 healthy controls. The tibial mechanical axis angle determined by an inclinometer was 0.2 ± 5.1° varus in the OA group and 1.2 ± 3.5° varus in the control group ($p > 0.05$). Participants walked at a self-selected speed and maximum speed. There were no differences for either walking speed between the groups ($p > 0.05$). The peak adduction angle at 30% stance was approximately 2.1° greater in the OA group, but there was no significant difference in the peak external KAM ($p > 0.05$). There was a significant but weak correlation between the peak knee adduction angle and the peak KAM ($r = 0.39$, $p < 0.05$). It is possible that the knee adduction angle at 30% stance instead of the peak adduction angle may give rise to the weak correlation with the KAM.

Comparing the study by Foroughi et al. (2010) and the previous two studies (Hurwitz et al., 2002; Turcot et al., 2013), the mechanical axis angles of the knee OA patients were different. The knee OA patients and the healthy participants in Foroughi et al. (2010) had similar knee alignments ($p > 0.05$), whereas the OA patients in the other studies had markedly greater varus alignment than the healthy participants. Consequently, difference in the alignment led to the discrepancy in the result of the peak external KAM. It is likely that the mechanical axis angle rather than the severity of OA that determined the external KAM in knee OA patients as stated by Hurwitz et al. (2002).

Messier et al. (2014) studied the effects of knee malalignment among 157 knee OA patients with K/L grade 2-3. Participants were divided into three groups: varus group (5.6 ± 3.4° varus; n = 76), neutral group (1.2 ± 0.6° varus; n = 42) and valgus group (2.7 ± 2.3° valgus; n = 39). Varus was defined as the mechanical axis angle > 2° in varus direction, while valgus was defined as < 0° in valgus direction. The neutral alignment was defined as 2° varus to 0°. Participants walked at a self-selected speed and there was no significant difference among groups. The results showed that the peak KAM was approximately 27% greater in the varus group (0.398 Nm/kg) compared to the neutral group (0.290 Nm/kg) after controlling for gender and walking speed ($p < 0.0001$). There was no difference in the external knee flexion moment among groups ($p > 0.05$). This result is similar to the study of Stief et al. (2011) that compared the knee biomechanics among heathy participants with different knee malalignment. These two studies showed that as long as the participants had varus alignment, either knee OA patients or healthy adults showed a greater peak KAM compared to the control during level walking.

Numerous studies have found a significant correlation between the mechanical axis angle and the external KAM or the medial load in knee OA patients and healthy participants. Wada et al. (2001) reported a moderate correlation ($r = 0.62$, $p < 0.001$) between the mechanical axis angle and the external KAM in 69 patients with medial compartment knee OA. Miyazaki et al. (2002) found a weak but significant correlation between KAM and the mechanical axis angle ($r = 0.23$, $p < 0.001$) using the radiographic method in 74 patients with medial compartment knee OA after adjusting for age and pain. After a six year follow-up, they also found the risk of progression of knee OA increased 6.5 times with a 1% increase in KAM.

There are some studies suggesting dynamic knee alignment in the stance phase would be a better predictor for the external KAM (Barrios, Higginson, et al., 2009; Barrios, Royer, & Davis, 2012; Hurwitz et al., 2002; Schmitz & Noehren, 2014). Barrios et al. (2012) analyzed the dynamic versus radiographic alignment in relation to the external KAM in 55 knee OA patients. The results showed the mechanical axis angle was a weak predictor of the peak KAM ($r = 0.096$, $p > 0.05$), but the peak knee adduction angle remained a strong predictor ($r = 0.659$, $p < 0.001$). Schmitz et al. (2014) also found a significant correlation ($r = 0.762$, $p < 0.001$) between the knee adduction angle and the external KAM in 30 healthy participants.

In summary, the primary effect of varus malalignment for knee OA patients seems to be related to an increased external KAM. A decreased knee flexion ROM or a decreased knee flexion moment is probably due to the presence of the knee OA rather than the knee malalignment. The external KAM is correlated well with the mechanical axis angle and the knee adduction angle during level walking in both healthy and knee OA participants.

Gait Biomechanics Summary

During level walking, knee OA patients with a varus alignment had a smaller knee flexion angle at heel strike, smaller knee ROM, increased knee adduction angle and increased external KAM (Foroughi et al., 2010; Hurwitz et al., 2002; Messier et al., 2014; Turcot et al., 2013). For healthy participants with the varus alignment, a greater knee adduction angle, an increased KAM and increased stress on medial compartment were observed during level walking (Barrios, Davis, et al., 2009; Barrios & Strotman, 2014; Stief et al., 2011; Yang et al., 2010). Comparing the knee biomechanics of varus alignment between knee OA patients and healthy participants, both cohorts showed an increased KAM during level walking. The discrepancy was found in the knee flexion angle and the knee flexion moment as knee OA patients had a smaller knee flexion angle and flexion moment than the normal subjects. Therefore, a decreased knee flexion ROM or a decreased knee flexion moment in knee OA patients was likely due to the presence of knee OA rather than knee malalignment.

CYCLING BIOMECHANICS

Studies have shown that cycling has many health benefits for youth, middle-aged and elderly populations (Andersen, Lawlor, Cooper, Froberg, & Anderssen, 2009; Bassett Jr, Pucher, Buehler, Thompson, & Crouter, 2008; Hoevenaar-Blom, Wendel-Vos, Spijkerman, Kromhout, & Verschuren, 2011; Huy, Becker, Gomolinsky, Klein, & Thiel, 2008). Besides health benefits, cycling is also frequently prescribed as a rehabilitation exercise by many health professionals (Gail D Deyle et al., 2005; G. D. Deyle et al., 2000; Naal et al., 2007; Salacinski et al., 2012; Walker et al., 2014).

Injuries occur in cycling without the correct mechanics on the joints. Among the joints of the lower limb, the knee is thought to be the most affected site with prevalence of 42% – 65% (Conti-Wyneken, 1999; Dannenberg, Needle, Mullady, & Kolodner, 1996; Wilber, Holland, Madison, & Loy, 1995). Furthermore, the development of knee OA is also associated with external KAM (Mundermann et al., 2004; Sharma et al., 1998; Zhao et al., 2007). Therefore, it is important to examine the knee biomechanics during cycling to assess whether cycling can be an appropriate exercise for knee OA patients and those with knee varus alignment.

Knee biomechanics of cycling

The basic components of a bicycle include the frame, seat (saddle), handlebar, crank, and pedals. During pedaling, the top most position of the crank and pedal is referred to as top dead center, while the bottom most position is referred to as bottom dead center. Top dead center is defined as 0° or 360°, and the bottom dead center is 180°. Generally, a pedaling cycle can be divided into two phases, the power phase from 0° to 180° and the recovery phase from 180° to 360° (Asplund & St Pierre, 2004).

Kinematics

Ericson et al. (1988; 1986) studied the kinematics of cycling at a power output of 120 W and a cadence of 60 revolution per minute (rpm) with a saddle height of 113% of the distance between the ischial tuberosity and the medial malleolus. The mean ROMs during cycling were 66° (32° – 70°) for the knee, 38° (32° – 70°) for the hip and 24° (2° plantarflexion to 22° dorsiflexion) for ankle. Similar mean knee ROM (65°) was also found in a study by Bini et al. (2010), which investigated the kinematics of cycling at 80% of the participants' maximum power output, a freely chosen cadence and with saddle height of 100% greater trochanter height.

Some studies have also found a different mean knee ROM in cycling. Too et al.

(2000) reported a mean knee ROM of 74 ± 6.0° during a 30-second Wingate test with the saddle height at 109% of the distance from the symphysis pubis to the floor. Gardner et al. (2015) found a peak knee flexion angle of 44.9 ± 7.8° in 11 healthy participants at a power output of 80 W and a cadence of 60 rpm with a neutral foot alignment. The saddle height was set so that the participants' knee angle was 30° when the crank was at bottom dead center. With the same riding position, Fang et al. (2014) showed that the mean knee ROM was approximately 77.4° and there were no significant differences among different cadences (60, 70, 80 and 90 rpm) when cycling at a workload of 1 kg.

There are limited number of studies that have reported on frontal plane kinematics. Bailey et al. (2003) investigated knee kinematics at a power output of 200 W and a cadence of 90 rpm without any modification to the normal riding position. They found the mean knee ROM was 67.5° (41.5° – 109°) and the ROM in frontal plane was around 1° of adduction to 2° of abduction. Umberger et al. (2001) tested knee kinematics of cycling at a power output of 225 W and a cadence of 90 rpm with a self-selected saddle height. The study showed that the mean sagittal plane knee ROM was 73.9° (40.1° – 114 °) and the knee ROM in frontal plane was 5° of adduction to 4° of abduction. Gardner et al. (2015) found the peak knee adduction angle of 2.2 ± 5.3° at a power output of 80 W and a cadence of 60 rpm. In the study by Fang et al. (2014), the knee ROM in the frontal plane was 6.0° of adduction and 3.9° of abduction at a workload of 1 kg and a cadence of 90 rpm.

Overall, the kinematic results from the literature are variable, since studies used different protocols of cycling posture, power output, workload and cadence. The knee sagittal plane ROM during cycling varied from 65° to 77° and the knee ROM in frontal plane was approximately 6° adduction to 4° abduction.

Kinetics

Gregor et al. (1985) investigated the sagittal knee kinetics in five recreational cyclists at a power output of 160 W and a cadence of 60 rpm. The results showed the peak knee extension moment was 53 Nm and the peak knee flexion moment was 34 Nm. Ericson et al. (1986) studied the sagittal plane knee moment during "standardized ergometer cycling" at a power output of 120 W, a cadence of 60 rpm with a mid-saddle height (113% of distance between the ischial tuberosity and the medial malleolus) and anterior foot position. The results showed that the knee extension moment was 28.8 Nm and flexion moment was 11.9 Nm. It has been shown that the knee sagittal-plane kinetic variables are easily influenced by the workload (R. R. Bini et al., 2010). It is obvious that the discrepancy in sagittal plane knee moment between the two studies (Ericson et al., 1986; Gregor et al., 1985) was from the different workload.

Frontal plane knee load is closely associated with the development of knee OA (Mundermann et al., 2004; Sharma et al., 1998; Zhao et al., 2007). Accordingly, the magnitude of frontal plane knee load during exercise is most critical to the knee OA patients and it dictates whether an exercise can be a non-surgical treatment for knee OA (Zhang et al., 2010). Moreover, a common injury during cycling, patellofemoral pain syndrome, is thought to be caused by the external KAM during the power phase (Boyd, Neptune, & Hull, 1997; Wolchok, Hull, & Howell, 1998). Given that both knee OA and patellofemoral pain syndrome are related to the KAM, it is essential to review the literature about the frontal plane knee load during cycling.

Ericson et al. (1984) studied frontal plane knee load during cycling at 120 W and 60 rpm. They found the external peak KAM was 24.5 Nm and the external peak knee abduction moment was 2.9 Nm. Gardner et al. (2015) examined the knee load in 11 healthy participants at a power output of 80 W and a cadence of 60 rpm with a neutral foot alignment. The results showed the peak extensor moment was 26.27 ± 9.60 Nm and the internal peak knee abduction moment was 9.00 ± 4.74 Nm. In the study by Fang et al. (2015), the mean internal knee adduction moment was 7 Nm and the mean internal knee abduction moment was around 7.78 Nm during cycling with workload of 1 kg at cadences of 60, 70, 80, 90 rpm. Ruby et al. (1992) analyzed the frontal plane knee load at a power output of 225 W and a cadence of 90 rpm. The external peak KAM was 15.3 Nm and the external peak knee abduction moment was 11.2 Nm. Gregersen et al. (2003) investigated the frontal plane knee load at a power output of 225 W and a cadence of 90 rpm. The external peak KAM was 7.8 Nm and the external peak knee abduction moment was 8.1 Nm.

The kinetic results from different studies varied, which can be attributable to several factors. First, the difference in cadence and particularly workload among studies can lead to the various kinetic results. Second, even if the workload and cadence were identical (Gregersen & Hull, 2003; Ruby et al., 1992), differences still existed in the frontal plane knee load during cycling and it is probably due to the difference in the saddle height or depth used in the studies. Third, most of early studies utilized a one-sensor instrumented pedals (Ericson et al., 1984; Gregersen & Hull, 2003; Ruby et al., 1992), whereas the recent studies used pedals with two sensors (Y. Fang et al., 2014; Gardner et al., 2015). Pedals with one sensor are not capable of measuring the medial-lateral center of pressure (COP) displacement and it makes the kinetic data such as knee moment less accurate than pedals with two sensors.

In summary, the kinetic variables in cycling are more sensitive to cycling posture and workload than kinematic variables. Moreover, studies utilized different instrumented pedals to measure kinetic variables. Therefore, there were discrepancies in the results among studies. It is important to relate the kinetic variables to the workload and cycling posture when interpreting the results from cycling studies.

Cadence and workload

Cadence and workload are the primary variables that can be manipulated during cycling exercises (Asplund & St Pierre, 2004). Many studies have investigated the effects of cadence and workload on the knee biomechanics during cycling.

Ericson et al. (1988) studied the effect of cadence and workload on the knee kinematics during cycling. Participants were asked to cycle at a workload of 2 kg and cadences of 40, 60, 80, 100 rpm or at a cadence of 60 rpm and workloads of 0, 2, 4 kg. When the workload increased from 0 to 4 kg, the maximum knee extension angle was significantly decreased from 49° to 42 ° ($p < 0.05$), yet the knee ROM was not influenced by change of workload.

In another study by Ericson et al. (Ericson et al., 1986), the participants were asked to cycle at power outputs of 0, 120, 240 W and cadences of 40, 60, 80, 100 rpm. Both knee flexion and extension moments increased with an increase in power output. Particularly, the external knee flexion moment was influenced mostly with an increase from 9 to 50 Nm when the power output changed from 0 W to 240 W ($p < 0.05$). When the cadence changed from 40 rpm to 100 rpm with a constant workload of 2 kg, there was a significant but very small increase in the external knee flexion moment from 28 Nm to 32 Nm ($p < 0.05$).

Bini et al. (2010) asked the participants to cycle at cadences of 40 rpm or 70 rpm and workloads of 0, 5, 10 N. The results showed no difference in the mean knee angle, knee ROM and knee mechanical work when the cadence changed from 40 to 70 rpm. When the workload increased from 0 N to 10 N, the kinematic variables in the knee joint were not influenced but the knee mechanical work and the total mechanical work of the lower extremity joints were significantly increased ($p < 0.05$).

Fang et al. (2015) had participants cycling in eight testing conditions that included five workload conditions (0.5, 1.0, 1.5, 2.0, and 2.5 kg) at 60 rpm, and four cadence conditions (60, 70, 80, and 90 rpm) with 1 kg workload. The results showed that the cadence had a significant effect on the knee abduction ROM and knee flexion moment ($p < 0.05$). The workload had a significant effect on the knee extension ROM ($p < 0.01$), knee abduction ROM ($p < 0.01$). Moreover, the knee extension moment increased from 11.6 Nm to 37.2 Nm ($p < 0.001$) and the knee abduction moment increased from 5.8 Nm to 14.4 Nm ($p < 0.05$) when the workload increased from 0.5 kg to 2.5 kg.

Overall, most studies have found that knee kinematics was hardly influenced by cadence and workload and the knee kinetics was mostly affected by the workload.

Saddle height and depth

Saddle height and depth are the variables related to cycling posture. There are many studies that have investigated the effects of saddle height and depth on performance, but only few studies have focused on the biomechanics of cycling (R. Bini et al., 2011; Ercison et al., 1988).

Ericson et al. (1987) studied the effect of three different saddle heights (102, 113, 120% of the distance between the ischial tuberosity and medial malleolus) on the kinetics of knee joints in the sagittal plane. A kinetic model was used to estimate the patellofemoral compressive forces during cycling. The results showed that the patellofemoral compressive force was inversely related to the saddle height. In addition, the external knee flexion moment was decreased from 32 Nm to 20 Nm and the external knee extension moment was increased from 11 Nm to 19 Nm when the saddle height increased from 102 % to 120%.

Bini et al. (2011) reviewed the literature related to the effect of saddle height on the knee injury and the performance during cycling. It was stated in the review that there was limited number of articles and controversial results regarding the optimal saddle height for injury prevention. Considering effects of saddle height on both performance and injury prevention, the range of 25° – 30° of knee flexion, when the pedal is at the bottom dead center and the cyclist is seated on the saddle, can be the optimal saddle height.

Saddle depth is the same as seat tube angle in terms of their function in cycling. The seat tube angle is the angle formed between the rear of the seat tube (posterior direction) and level ground. The more forward a cyclist sits on the saddle, the deeper the saddle is and the larger the seat tube angle is. Studies have shown that the increased saddle depth can increase the hip extension angle and ankle ROM, whereas the knee kinematics was not influenced by saddle depth (Price & Donne, 1997; B. Umberger, Scheuchenzuber, & Manos, 1998). No kinetic results were reported regarding the effect of saddle depth in cycling. Based on the practical experience in cycling, it is advocated that the saddle depth should be set as the knee in line with the pedal spindle when the crank is in the forward horizontal position (90°) (Burke, 2003).

In summary, a knee flexion between 25° – 30° can be a good choice for the saddle height during cycling. The saddle depth had little effect on the knee kinematics and it can be set as the knee in line with the pedal spindle when the crank is in 90° position.

Cycling biomechanics summary

During cycling, the knee sagittal plane ROM varied from 65° to 77° and the knee frontal plane ROM was around 6° of adduction to 4° of abduction. The kinetic variables in cycling are more sensitive to cycling posture and workload than the kinematic variables. As a result, there were tremendous discrepancies in the kinetic results and it entails relating the kinetic variables to the workload and cycling posture when explaining the results from different studies.

Regarding the cadence and workload, cycling cadence did not influence the knee kinematics. The knee kinetics were mostly associated with workload. A knee flexion of 25° – 30°, when the pedal is at the bottom dead center and the cyclist is seated on the saddle, is thought to be as an optimal saddle height for performance and injury prevention. The saddle depth had little effect on the knee kinematics. Based on the practical experience, it can be set as the knee in line with the pedal spindle when the crank is in the forward horizontal position.

CHAPTER III

METHODS

PARTICIPANTS

Eleven participants in each of varus group (age: 24.4±2.8 years, height: 1.78±0.08 m, weight: 75.1±16.5 kg, BMI: 23.6±4.6 kg/m^2) and neutral group (age: 24.0±4.1 years, height: 1.76±0.10 m, weight: 73.1±15.3 kg, BMI: 23.4±2.9 kg/m^2) were recruited to participate in the study. Participants were recruited from the UT student population by flyers and announcement in Kinesiology and Physical Education and Activity Program classes. To be included, varus participants had a knee alignment of a minimum of 2° deviation from neutral in varus direction. Potential participants were asked to attend a preliminary screening session during which alignment screening was performed using a previously validated clinical method (Hinman et al., 2006). The method measures the distance between either the medial epicondyles of the knees (varus alignment) or the medial malleoli of the ankles (valgus alignment) when the participants are standing and moving their feet together until either their ankles or knees touch. If the participant fit the criteria using the clinical method, then he or she was asked to attend a full-limb radiographic measurement session to confirm the alignment type. The exclusion criteria for the study included the following: a body mass index (BMI) greater than 35 kg/m^2, major lower extremity injury or surgery, any injury within the past three months, any chronic disease, diagnosis of arthritis in the lower extremity, and inability to ride a stationary bike for about 15 minutes or unable to see, hear, or follow instructions.

During the full-limb radiographic measurement session, the anteroposterior view of a weight-bearing radiograph of the lower extremity including hip, knee and ankle was captured with the graduated-grid x-ray cassette. The cassette height was 130.0 cm and the width was 36.0 cm. A participant stood barefoot with knees in full extension and the tibial tubercles facing the x-ray beam. The x-ray tube was placed at a distance of 1.83 m from the cassette. For the x-ray power settings, 95 kilovolts and 300 mA/s – 500 mA/s were applied, depending on the limb size and tissue characteristics.

Participants who met the alignment requirements attended one testing session. The testing session lasted 90 minutes. Participants gave their written informed consent approved by the University of Tennessee Institutional Review Board, prior to the x-ray testing session.

An effect size of 1.09 was calculated using the values of knee adduction moment in the study by Barrios et al. (2009). A sample size of 17 was estimated in a power analysis with an effect size of 1.09, a β level of 0.95 and α level of 0.05 (G*Power 3.1.3, National Instruments Corporation).

INSTRUMENTATION

A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to obtain the three dimensional (3D) kinematics during the test. Reflective anatomical and tracking markers were placed on both sides of feet, ankles, legs, knees, thighs, and hips during testing. For the thighs and legs, the tracking markers were attached to the respective segment via a semi-rigid thermoplastic shell. The tracking markers for the feet were placed directly to the posterior heel of the shoes.

A Monark cycle ergometer (828E, Monark, Sweden) equipped with a weighted brake system was used in the study. The saddle height and depth can be adjusted by moving the seat upward, downward, forward and backward. The location of handlebars can move up and down and rotate forward and backward.

A customized instrumented bike pedal was used on the cycle ergometer, which allows recordings of three dimensional pedal reaction forces (PRF) and moments. The assembly contained two 3D force sensors (Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A, Kistler, Switzerland). The charge amplifiers were necessary to convert the charge measured by the force sensors to a voltage value used by the Vicon Nexus software. The sensors were placed in the left pedal and a dummy pedal of the same mass and design was used on the opposite limb to minimize asymmetries during the testing.

PROCEDURES

The participants were asked to wear spandex shorts and running shoes (Noveto, Adidas). For the biomechanical testing, the reflective anatomical markers were placed bilaterally on the acromion process, iliac crests, anterior superior iliac spine, posterior superior iliac spine, greater trochanters, medial and lateral epicondyles, medial and lateral malleoli, 1^{st} and 5^{th} metatarsal heads, tip of the second toe, and midpoint of the front edge of both pedals. A cluster of four tracking markers on a thermoplastic shell was attached to the shanks, thighs, pelvis and trunk. Three markers were also attached to the lateral, superior and inferior heel counters of the shoes. The three lateral pedal markers were also used as tracking markers for both pedals. A crank tracking marker was placed on the axes of both cranks, and an additional tracking marker was placed on the front body of the bike. After the static and dynamic calibrations of the cameras, the participants performed a cycling assessment on the instrumented cycling ergometer, and gait assessment. The saddle height on the cycle ergometer was set so that the angle of the knee joint was approximately 30° when the crank was at bottom dead center (R. Bini et al., 2011). The horizontal saddle depth was set so that the knee was in line with the pedal spindle when the crank was in the forward horizontal position (Burke, 2003). The position of the handlebars was set so that the angle between the participant's trunk and thigh was 90° when the crank was in the forward horizontal position.

Participants performed a 2-min warm-up on the cycle ergometer before data collection. A 2-min rest was provided between the warm-up and subsequent cycling testing. The participants pedaled for 2-min in each of 6 cycling conditions: pedaling at 80 rpm at 0.5 kg (40 Watts), 1.0 kg (78 Watts), and 1.5 kg (117 Watts) with and without toe-cage. Simultaneous recordings of kinematic (240 Hz) and kinetic (1200 HZ) data were performed on five consecutive pedaling cycles for each condition which began during the last 30 seconds of each test condition. Participants were given two minutes of rest between conditions and continued to the next condition when they felt ready to proceed.

DATA AND STATISTICAL ANALYSES

The obtained radiographs were analyzed using InteleViewer software (Intelerad, Montreal, Quebec, Canada). A 2.54 cm diameter sphere was used to calibrate each participant's radiograph. The mechanical axis of each limb was then determined using the following standard procedures (Moreland et al., 1987). The mechanical axis of the femur was measured by a line drawn from the center of the femoral head to the center of the tibial intercondylar eminence and the mechanical axis of the tibia was from the center of the intercondylar eminence to the center of talus. The mechanical axis angle of the knee joint was measured by the angle between the mechanical axes of femur and tibia. Two investigators independently performed the same measurements on each radiograph. Inter-rater reliability, as measured by intra-class correlation, showed an excellent agreement between investigators ($r = 0.995$).

Pedal reaction force (PRF), moments of force, and center of pressure (COP) on the left pedal were computed in Visual 3D (C-Motion Inc.) using the method described by Gardner et al. (2015). A right-hand rule was used to determine the polarity of the joint angles and moments and an x-y-z Cardan rotation sequence was used to compute joint angles. In cycling conditions, the movement cycle of a trial was defined from the top dead center (0°) to the following top dead center (360°). Both kinematic and kinetic data were filtered using a low-pass 4th order Butterworth filter with zero lag at a cutoff frequency of 6 Hz (Gardner et al., 2015). A customized program (VB_V3D, MS VisualBASIC 6.0) was utilized to identify peak angles, velocities, moments and powers. The variables of interest were organized and reported using another customized program (VB_Table, MS VisualBASIC 6.0). The pedal reaction force and joint moment in cycling were not normalized by the participant's body weight as the participant placed most of weight on the seat and handlebars (Y. Fang et al., 2014; Gardner et al., 2015).

A 2 × 2 × 3 (group × toe-cage × workload) mixed design analysis of variance (ANOVA) was used to examine the effect of knee alignment, foot alignment and workload on selected biomechanical variables (IBM SPSS Statistics 22, Chicago, IL). When a three way interaction was found, two-way ANOVAs were followed. When a two-way ANOVA was significant, a post hoc analysis with Bonferroni adjustments was performed to detect specific differences. An alpha level of 0.05 was set a priori.

CHAPTER IV

EFFECTS OF VARUS KNEE ALIGNMENT AND USING TOE-CAGES ON FRONTAL PLANE KNEE BIOMECHANICS IN STATIONARY CYCLING

ABSTRACT

Effects of varus knee alignment on the internal knee abduction moment (KAM) in walking has been widely studied. KAM has been shown to be closely associated with the development of medial knee osteoarthritis (OA). Despite the importance of knee alignment, no studies have explored its effect on knee frontal plane biomechanics during stationary cycling. The purpose of this study was to examine the effects of varus knee alignment and using a toe-cage on the knee frontal plane biomechanics during stationary cycling. Eleven participants in each of the varus and neutral groups participated in the study. The participants cycled for 2-min in each of six stationary cycling conditions: pedaling at 80 rpm at 0.5 kg (40 Watts), 1.0 kg (78 Watts), and 1.5 kg (117 Watts) with and without a toe-cage. A motion analysis system and a custom instrumented pedal were used to collect kinematic and kinetic data. A varus knee alignment and using a toe-cage did not result in greater peak knee adduction angle and peak KAM. These findings suggest stationary cycling may be a safe exercise prescription for people with varus knee alignment, including patients with medial knee compartment OA. In addition, using a toe-cage may not have negative effects on knee joints in stationary cycling. Future studies may be needed to explore the tibiofemoral contact force in participants with a varus knee alignment during stationary cycling.

Keywords: knee alignment, knee abduction moment, knee OA, cycling, toe-cage

INTRODUCTION

Cycling is frequently prescribed as a rehabilitation exercise by many health professionals (Gail D Deyle et al., 2005; G. D. Deyle et al., 2000; Naal et al., 2007; Salacinski et al., 2012; Walker et al., 2014), given that cycling has advantages in reducing the knee joint loads (D'Lima et al., 2008; Kutzner et al., 2012). Despite the relatively lower joint load during cycling, the prevalence of chronic bicycle injuries can be as high as 85% due to its highly repetitive nature (Dettori & Norvell, 2006; Wanich, Hodgkins, Columbier, Muraski, & Kennedy, 2007). Among the joints of the lower limb, the knee is thought to be the most affected site with injury prevalence of 42% – 65% (Conti-Wyneken, 1999; Dannenberg et al., 1996; Wilber et al., 1995).

The internal knee abduction moment (KAM) is a surrogate measure for loading to the medial compartment of the knee joint in walking and has been shown to be closely associated with the development of medial knee osteoarthritis (OA) (Mundermann et al., 2004; Sharma et al., 1998; Zhao et al., 2007). Studies have shown that the frontal plane knee malalignment, mostly varus alignment, can lead to a significant increase of KAM during walking in both healthy population (Barrios, Davis, et al., 2009; Stief et al., 2011) and knee OA patients (Messier et al., 2014; Turcot et al., 2013). Furthermore, longitudinal studies have shown that varus alignment was associated with incident and progression of medial knee OA (Sharma et al., 2010; Sharma et al., 2001).

Although many studies have investigated the effects of knee alignment during walking, there are limited number of studies in cycling. Recently, Gardner et al. (2015)

compared the KAM in patients with medial knee OA and healthy controls during stationary cycling and no significant difference was found between groups. The knee alignment of the participants was not measured in the study, and it is likely that the knee alignment data may help explain their results on KAM.

Many stationary bikes have toe-cages available and they are used by cyclists to constrict their feet on the pedals during cycling. However, previous studies have suggested allowing some freedom between the foot and pedal may be beneficial for reducing overuse knee injuries (Boyd et al., 1997). It is still unclear whether a toe-cage would have any negative effects on knee biomechanics during stationary cycling.

Despite the importance of the knee alignment, no studies have explored its effects on knee frontal plane biomechanics during cycling. Considering the significant effects of knee alignment on gait biomechanics, it is reasonable to hypothesize that the knee alignment may have a similar influence during cycling. Furthermore, it remains unclear whether using a toe-cage would negatively influence the frontal plane loading in the knee joints. Therefore, the purpose of this study was to examine the effect of varus knee alignment and using a toe-cage on the knee frontal plane biomechanics during stationary cycling. It was hypothesized that participants with a knee varus alignment will have a greater KAM compared to participants with a neutral alignment during cycling, and KAM will not differ when using or not using a toe-cage.

METHODS

Participants

Eleven participants in each of varus group (age: 24.4±2.8 years, height: 1.78±0.08 m, weight: 75.1±16.5 kg, BMI: 23.6±4.6 kg/m^2) and neutral group (age: 24.0±4.1 years, height: 1.76±0.10 m, weight: 73.1±15.3 kg, BMI: 23.4±2.9 kg/m^2) were recruited to participate in the study. An anteroposterior full limb radiograph was obtained to measure the knee mechanical axis angle (MAA). To be included in the varus group, participants had a mechanical axis angle less than 178° (Sharma et al., 2010). For the neutral group, participants had a MAA between 178°and 182° (Sharma et al., 2010). The exclusion criteria included a body mass index (BMI) greater than 35 kg/m^2, any injury within the past three months, and inability to ride a stationary bike for about 15 minutes. Using an effect size of 1.09 calculated from the knee adduction moment in a study by Barrios et al. (2009), a sample size of 17 was estimated with a β level of 0.95 and α level of 0.05 (G*Power 3.1.3). Participants were asked to read and sign an informed consent approved by the University Institutional Review Board prior to the radiographic measurement session.

Instrumentation

All potential participants attended a full-limb radiographic measurement session. The anteroposterior view of a full-length lower extremity weight-bearing radiograph was captured with the graduated-grid x-ray cassette (Moreland et al., 1987; Sharma et al., 2001). The cassette size was 130.0 cm (height) by 36.0 cm (width). The participant stood barefoot with knees in full extension and the tibial tubercles facing the x-ray beam. The x-ray tube was placed at a distance of 1.83 m from the cassette. An x-ray power settings of 95 kilovolts and 300 mA/s – 500 mA/s were applied, depending on the limb size and tissue characteristics.

A nine-camera motion analysis system (240 Hz, Vicon Motion Analysis Inc., UK) was used to obtain the three dimensional (3D) kinematics during the test. The reflective anatomical markers were placed bilaterally on acromion process, iliac crests, anterior superior iliac spine, posterior superior iliac spine, greater trochanters, medial and lateral epicondyles, medial and lateral malleoli, 1^{st} and 5^{th} metatarsal heads, tip of the second toe, and midpoint of the front edge of both pedals. A cluster of four tracking markers on a thermoplastic shell was attached to the shanks, thighs, pelvis and trunk. Three markers were also attached to the lateral, superior and inferior heel counters of the standard lab shoes (Noveto, Adidas). Three lateral pedal markers were also used as tracking markers for both pedals. A crank tracking marker was placed on the axes of both cranks, and an additional tracking marker was placed on the front body of the bike.

A Monark cycle ergometer (818E, Monark, Sweden) was used in the study. The saddle height on the bike was set so that the angle of the knee joint was approximately 30° when the crank was at bottom dead center (R. Bini et al., 2011). The saddle depth was set so that the knee was in line with the pedal spindle when the crank was in the forward horizontal position (Burke, 2003). The position of the handlebars was set so that the angle between the participant's trunk and thigh was 90 ° when the crank was in the forward horizontal position.

A customized instrumented bike pedal was used on the cycle ergometer, which allows recordings of three dimensional pedal reaction forces (PRF) and moments (Y Fang, 2014;

Gardner et al., 2015). The assembly contained two 3D force sensors (Type 9027C, Kistler, Switzerland) coupled with two industrial charge amplifiers (Type 5073A, Kistler, Switzerland). The charge amplifiers were necessary to convert the charge measured by the force sensors to a voltage value used by the Vicon Nexus software. The sensors were placed in the left pedal and a dummy pedal of the same mass and design was used on the opposite limb to minimize asymmetries during the testing.

Experimental protocol

Participants performed a 2-min warm-up on the cycle ergometer before data collection. A 2-minute rest was provided between the warm-up and subsequent cycling testing. The participants pedaled for 2-min in each of six cycling conditions: pedaling at 80 rpm at 0.5 kg (40 Watts), 1.0 kg (78 Watts), and 1.5 kg (117 Watts) with and without toe-cage. All the conditions were randomized by toe-cage conditions first, and followed by workload conditions. Simultaneous recordings of kinematic (240 Hz) and kinetic (1200 HZ) data were performed on five consecutive pedal cycles which began during the last 30 seconds of each test condition.

Data and Statistical Analyses

The obtained radiographs were analyzed using InteleViewer software (Intelerad, Montreal, Quebec, Canada). A 2.54 cm diameter sphere was used to calibrate each participant's radiograph. The mechanical axis of each limb was then determined using the following standard procedures (Moreland et al., 1987). The mechanical axis of the femur was measured by a line drawn from the center of the femoral head to the center of the tibial intercondylar eminence and the mechanical axis of the tibia was from the center of the intercondylar eminence to the center of talus. The mechanical axis angle of the knee joint was measured by the angle between the mechanical axes of femur and tibia. Two investigators independently performed the same measurements on each radiograph. Inter-rater reliability, as measured by intra-class correlation, showed an excellent agreement between investigators ($r = 0.995$).

Pedal reaction force (PRF), moments of force, and center of pressure (COP) on the left pedal were computed in Visual 3D (C-Motion Inc.) using the method described by Gardner et al. (2015). A right-hand rule was used to determine the polarity of the joint angles and moments and an x-y-z Cardan rotation sequence was used to compute joint angles. In cycling conditions, the movement cycle of a trial was defined from the top dead center (0°) to the following top dead center (360°). Both kinematic and kinetic data were filtered using a low-pass 4^{th} order Butterworth filter with zero lag at a cutoff frequency of 6 Hz (Gardner et al., 2015). A customized program (VB_V3D, MS VisualBASIC 6.0) was utilized to identify peak angles, velocities, moments and powers. The variables of interest were organized and reported using another customized program (VB_Table, MS VisualBASIC 6.0). The pedal reaction force and joint moment during cycling were not normalized by the participant’s body weight as the participant placed most of their weight on the seat and handlebars (Y. Fang et al., 2014; Gardner et al., 2015).

A 2 × 2 × 3 (group × toe-cage × workload) mixed design analysis of variance (ANOVA) was used to examine the effect of knee alignment, foot alignment and workload on selected biomechanical variables (IBM SPSS Statistics 22, Chicago, IL). When a three way interaction was found, two-way ANOVAs were followed. When a two-way ANOVA was significant, a post hoc analysis with Bonferroni adjustments was performed to detect specific differences. An alpha level of 0.05 was set a priori.

RESULTS

No significant differences were found for age, height, weight or BMI between the groups (Table 3). The mechanical axis angle (MAA) for the varus group was smaller than that of the neutral group with a mean difference of 4.9±0.5° ($p<0.001$).

Pedal Reaction Force

Workload was significant for peak medial PRF and peak vertical PRF ($p<0.001$): it was increased from 0.5 to 1.0 kg ($p<0.001$), 1.0 to 1.5 kg ($p=0.001$), and 0.5 to 1.5 kg ($p<0.001$, Table 4). The ANOVA also revealed a significant interaction of group and toe-cage for peak vertical PRF ($p=0.026$). However, the post hoc comparisons showed no additional differences.

Knee Joint Moment and Angle

Workload main effect was significant for peak knee extension moment ($p<0.001$); it was increased from 0.5 to 1.0 kg ($p<0.001$), 1.0 to 1.5 kg ($p=0.006$), and 0.5 to 1.5 kg ($p<0.001$, Table 4). Workload was also significant for peak knee abduction moment ($p<0.001$): it was increased from 0.5 to 1.0 kg ($p=0.001$), 1.0 to 1.5 kg ($p<0.001$), and 0.5 to 1.5 kg ($p<0.001$).

A three-way interaction was found for peak knee internal rotation moment ($p=0.004$,

Table 4). Post hoc results showed a group × toe-cage interaction ($p=0.004$) at workload of 0.5 kg. Further analysis showed that toe-cage increased peak knee internal rotation moment in the varus group ($p=0.019$) but decreased it in the neutral group ($p=0.047$). The greater workload resulted in a greater knee internal rotation moment in both conditions with and without toe-cage ($p<0.001$). In addition, a toe-cage × workload interaction ($p=0.034$) was also found in the neutral group. Post hoc comparisons showed the greater workload led to a greater internal rotation moment without toe-cage ($p<0.001$). The same trend was also found for the varus group ($p=0.003$).

A group × workload interaction for knee extension ROM ($p=0.048$) and toe-cage main effect were significant ($p=0.024$, Table 5). Post hoc comparisons showed that knee extension ROM was greater at 1.5 kg compared to 1.0 kg ($p=0.02$) for neutral group. The peak knee abduction angle in the neutral group was greater compared to the varus group ($p=0.026$), and the toe-cage reduced the magnitude of peak knee abduction angle ($p=0.015$). The peak knee adduction angle occurred earlier as the workload increased from 0.5 to 1.0 kg ($p=0.013$) and 0.5 to 1.5 kg ($p=0.001$).

Ankle Joint Moment and Angle

The toe-cage main effect was significant for peak ankle inversion moment ($p=0.008$, Table 4) and it was greater in the toe-cage condition. A significant three-way interaction existed in peak ankle external rotation moment ($p=0.022$). Post hoc results showed a group × toe-cage interaction ($p=0.01$) at a workload of 1.5 kg. Further analysis revealed that the toe-cage decreased the ankle external rotation moment in the varus group ($p=0.001$); varus group had a greater external rotation moment with toe-cage (p=0.043) and without toe-cage (p<0.001). For the varus group, it was increased from 0.5 to 1.0 kg (p=0.006), 1.0 to 1.5 kg (p=0.019), and 0.5 to 1.5 kg (p<0.001); for neutral group, it was greater at 1.5 kg compared to 0.5 kg (p=0.018).

A three-way interaction was found for peak ankle eversion angle (p=0.035, Table 5), yet no further significant results were revealed in the post hoc analysis. Additionally, the ankle inversion ROM was greater at 1.5 kg compared to 0.5 kg (p=0.012).

Foot Angle

Toe-cage significantly increased the peak foot eversion angle (p<0.001) and the mean external rotation angle (p=0.001, Table 5)

DISCUSSION

The purpose of this study was to examine effects of varus knee alignment and using a toe-cage on knee frontal plane biomechanics during stationary cycling. Our first hypothesis that the participants with a varus alignment would have greater KAM compared to those with a neutral alignment was not supported by the results as no difference in the peak KAM was observed between the varus and neutral groups.

Effect of knee alignments on frontal-plane knee biomechanics

The KAM in the varus group did not differ from that of the neutral group. The magnitude of the KAM depends on the resultant PRF and its frontal plane moment arm with respect to knee joint center. The peak medial and vertical PRF did not differ between groups in the current study. It has been suggested that the frontal plane moment arm of the ground reaction force (GRF) in walking is dependent on lower limb alignment and more highly associated with the magnitude of the adduction moment (Hunt, Birmingham, Giffin, & Jenkyn, 2006). In the current study, the static knee alignment measured by the MAA for the varus group (174.3±1.4°) was significantly more varus than that of the neutral group (179.2±1.0°). Although the moment arm of PRF was not examined in this study, our results showed that the peak knee adduction angle for the varus group (10.2±2.0°) did not differ from that of the neutral group (8.6±2.1°). As the peak knee adduction angle is more representative of the actual skeletal geometry of lower extremity during movement (Barrios et al., 2012; M. A. Hunt, T. B. Birmingham, T. R. Jenkyn, J. R. Giffin, & I. C. Jones, 2008), no difference found for the peak knee adduction angle may be partially responsible for the lack of difference of the KAM between the two alignment groups in the current study. Another contributing factor may be related to the temporal difference of the peak knee adduction angle and the peak vertical PRF during cycling. The results showed that the peak knee adduction angle occurred at 23.8° of crank angle (13.2% power phase) and the peak vertical PRF at 86.1° of crank angle (47.8% power phase), whereas the peak KAM occurred at 75.2° of crank angle (41.8% power phase). It is likely that the large temporal separation between the peak knee adduction angle and the peak vertical PRF diminished the effect of knee alignment on the magnitude of KAM. To further investigate the effect of knee dynamic alignment, we also examined the knee frontal plane angle at the time of peak KAM for 1.0 kg workload condition and found no difference between the varus (5.1±1.7°) and neutral (1.7±1.8°) groups, although the mean difference was more than 3°.

The result of KAM in cycling is not supported by findings from previous gait studies regarding the effect of knee alignment, which have shown that a static varus alignment is associated with a greater peak knee adduction angle and an increased KAM during walking in both healthy (Barrios, Davis, et al., 2009; Barrios & Strotman, 2014; Stief et al., 2011) and knee OA populations (Hurwitz et al., 2002; Messier et al., 2014; Turcot et al., 2013). In addition, the peak knee adduction angle has been shown to occur at about the same time (22% of stance) as the peak KAM (23% of stance) during walking (Barrios et al., 2012). It is possible that temporal alignment of the peak adduction angle and GRF is one of the reasons for the varus group having a greater KAM during walking. These results suggest that stationary cycling introduces a less “harmful” frontal-plane movement and loading to the knee joint compared to walking for people with a varus knee alignment. Additionally, the peak knee extension moment in the current study did not differ between two alignment groups, indicating that the overall knee joint loading during cycling was similar. Therefore, it seems that the stationary cycling may be a safe aerobic exercise prescription for people with neutral or varus knee alignment, including patients with medial knee OA who have high incidence of knee varus alignment (Sharma et al., 2010). However, further investigations are warranted to examine tibiofemoral contact force in participants with knee malalignment to confirm these findings.

The peak knee abduction angle in the varus group (0.02±1.6°) was smaller than that of the neutral group (-5.5±1.6°). During stationary cycling, the knee was in an adducted position at the beginning, changed to abduction early in the power phase, and reached its peak at

183.8° of crank angle. Considering the peak knee abduction angle occurred much later than the peak KAM (75.2° of crank angle), the peak KAM was unlikely to be positively correlated with the peak abduction angle. However, the varus participants still had smaller peak knee abduction angles which placed them in relatively more adducted knee position during the majority of the power phase compared to the neutral participants. Although the peak KAM did not differ between the two alignment groups and the actual magnitudes were much smaller than fully weight bearing exercises, e.g. walking (Y. Fang et al., 2014; Gardner et al., 2015), effects of this more adducted knee alignment for people with a varus alignment deserve more attention in future research, especially in the context of long term effects of engaging in stationary cycling as exercises.

The peak KAM increased by 51.3 % and 31.7 % when the workload changed from 0.5 to 1.0 kg and 1.0 to 1.5 kg, respectively. The average peak KAM for both neutral and varus groups in our study was -7.2±3.6 Nm (0.6 %BW×Height) at a workload of 1.0 kg and a cadence of 80 rpm, which is similar to -7.0±4.3 Nm in the same condition by Fang et al. (2014). These values are also much lower compared to walking, where the peak KAM was 2.23 - 5.10 %BW×Height for knee OA patients and 2.60 - 3.16 %BW×Height for healthy controls (Foroughi, Smith, & Vanwanseele, 2009).

Effect of using a toe-cage on knee biomechanics

The motivation for examining the effects of toe-cage was to assess if it would have any negative effects on knee biomechanics during stationary cycling as a toe-cage is often available and used on many stationary bikes. Our second hypothesis was supported by the results as no difference in the peak KAM was observed with or without toe-cage during the stationary cycling. It is likely that the small magnitude of change in the foot external rotation angle and eversion angle induced by the toe-cage did not produce any change in the KAM during cycling. Furthermore, our result showed there was no difference in the peak knee extension moment. Based on these results, using a toe-cage may not appear to be harmful to the knee mechanics during cycling.

The usage of the toe-cage increased the peak foot eversion angle by 2.3° and the mean foot external rotation angle by 1.1°. The toe-cage was applied to the foot by tightening the straps between the toe-cage and pedal. A tight toe-cage might minimize the foot inversion, which might have caused the slight increased foot eversion angle in the toe-cage conditions. However, the ankle eversion angle was not affected by the toe-cage, considering no difference was shown by post hoc analysis although a group × toe-cage × workload interaction for the peak ankle eversion angle was found. The peak ankle inversion moment in the toe-cage condition was 0.27 Nm greater compared to the condition without a toe-cage. It appears that the restricted foot position caused by the toe-cage may require a greater inversion moment during the power phase of pedaling. However, the increase in the peak inversion moment was small and may not have any clinical significance.

Another interesting finding was that the use of toe-cage reduced the peak knee abduction angle during cycling. Perhaps a more everted foot position caused a decreased peak knee abduction angle in the toe-cage condition. One previous study in cycling reported that cyclists with a greater knee abduction angle may be at increased risk of overuse knee injuries

(Bailey et al., 2003). This finding seems to imply that cycling with a toe-cage use can be beneficial for preventing overuse knee injuries. However, it should be noted that the absolute change caused by toe-cage in the current study was less than 1°. Therefore, caution should be used when interpreting the results of the current study.

There are a few limitations of this study. Seventeen out of 22 participants showed a knee adduction moment instead of KAM. This finding has been reported in a previous study by Fang et al. (2014). The decreased sample size may have reduced statistical power for KAM. The participants might have a different experience in cycling, which may have introduced some variability to pedaling techniques and the results.

CONCLUSION

The findings of this study indicate that a varus knee alignment did not result in greater peak knee adduction angle and peak KAM, suggesting stationary cycling may be a safe exercise prescription for people with varus knee alignment, including patients with medial knee compartment OA. Using a toe-cage did not led to a greater peak KAM, suggesting it may not have any negative effects on knee joints during stationary cycling. This is the first study that examined the effects of varus knee alignment and using a toe-cage on knee frontal-plane biomechanics during stationary cycling. Future studies should explore the tibiofemoral contact force in participants with a varus knee alignment during stationary cycling.

LIST OF REFERENCES

Ajuied, A., Wong, F., Smith, C., Norris, M., Earnshaw, P., Back, D., & Davies, A. (2013). Anterior Cruciate Ligament Injury and Radiologic Progression of Knee Osteoarthritis: A Systematic Review and Meta-analysis. *Am J Sports Med*. doi:10.1177/0363546513508376

Andersen, L. B., Lawlor, D. A., Cooper, A. R., Froberg, K., & Anderssen, S. A. (2009). Physical fitness in relation to transport to school in adolescents: the Danish youth and sports study. *Scand J Med Sci Sports, 19*(3), 406-411.

Andrews, M., Noyes, F. R., Hewett, T. E., & Andriacchi, T. P. (1996). Lower limb alignment and foot angle are related to stance phase knee adduction in normal subjects: a critical analysis of the reliability of gait analysis data. *J Orthop Res, 14*(2), 289-295. doi:10.1002/jor.1100140218

Andriacchi, T., & Strickland, A. (1985). Gait analysis as a tool to assess joint kinetics *Biomechanics of normal and pathological human articulating joints* (pp. 83-102): Springer.

Asplund, C., & St Pierre, P. (2004). Knee pain and bicycling: fitting concepts for clinicians. *Phys Sportsmed, 32*(4), 23-30. doi:10.3810/psm.2004.04.201

Bailey, M. P., Maillardet, F. J., & Messenger, N. (2003). Kinematics of cycling in relation to anterior knee pain and patellar tendinitis. *J Sports Sci, 21*(8), 649-657. doi:10.1080/0264041031000102015

Barrios, J. A., Davis, I. S., Higginson, J. S., & Royer, T. D. (2009). Lower extremity walking mechanics of young individuals with asymptomatic varus knee alignment. *J Orthop Res, 27*(11), 1414-1419. doi:10.1002/jor.20904

Barrios, J. A., Higginson, J. S., Royer, T. D., & Davis, I. S. (2009). Static and dynamic correlates of the knee adduction moment in healthy knees ranging from normal to varus-aligned. *Clin Biomech (Bristol, Avon), 24*(10), 850-854. doi:10.1016/j.clinbiomech.2009.07.016

Barrios, J. A., Royer, T. D., & Davis, I. S. (2012). Dynamic versus radiographic alignment in relation to medial knee loading in symptomatic osteoarthritis. *Journal of Applied Biomechanics, 28*(5), 551-559.

Barrios, J. A., & Strotman, D. E. (2014). A sex comparison of ambulatory mechanics relevant to osteoarthritis in individuals with and without asymptomatic varus knee alignment. *Journal of Applied Biomechanics, 30*(5), 632-636. doi:10.1123/jab.2014-0039

Bassett Jr, D. R., Pucher, J., Buehler, R., Thompson, D. L., & Crouter, S. E. (2008). Walking, cycling, and obesity rates in Europe, North America, and Australia. *Journal of Physical Activity and Health, 5*(6), 795-814.

Bell, A. L., Brand, R. A., & Pedersen, D. R. (1989). Prediction of hip joint centre location from external landmarks. *Human Movement Science, 8*(1), 3-16.

Bini, R., Hume, P. A., & Croft, J. L. (2011). Effects of bicycle saddle height on knee injury risk and cycling performance. *Sports Med, 41*(6), 463-476. doi:10.2165/11588740-000000000-00000

Bini, R. R., Tamborindeguy, A. C., & Mota, C. B. (2010). Effects of saddle height, pedaling cadence, and workload on joint kinetics and kinematics during cycling. *J Sport Rehabil, 19*(3), 301-314.

Blagojevic, M., Jinks, C., Jeffery, A., & Jordan, K. P. (2010). Risk factors for onset of osteoarthritis of the knee in older adults: a systematic review and meta-analysis. *Osteoarthritis Cartilage, 18*(1), 24-33. doi:10.1016/j.joca.2009.08.010

Boyd, T. F., Neptune, R. R., & Hull, M. L. (1997). Pedal and knee loads using a multi-degree-of-freedom pedal platform in cycling. *J Biomech, 30*(5), 505-511.

Brouwer, G., Van Tol, A., Bergink, A., Belo, J., Bernsen, R., Reijman, M., . . . Bierma‐Zeinstra, S. (2007). Association between valgus and varus alignment and the development and progression of radiographic osteoarthritis of the knee. *Arthritis & Rheumatism, 56*(4), 1204-1211.

Burke, E. (2003). *High-tech cycling*: Human Kinetics.

Cappozzo, A. (1984). Gait analysis methodology. *Human Movement Science, 3*(1), 27-50.

Chao, E. Y., Neluheni, E. V., Hsu, R. W., & Paley, D. (1994). Biomechanics of malalignment. *Orthop Clin North Am, 25*(3), 379-386.

Cibere, J., Bellamy, N., Thorne, A., Esdaile, J. M., McGorm, K. J., Chalmers, A., . . . Kopec, J. (2004). Reliability of the knee examination in osteoarthritis: effect of standardization. *Arthritis Rheum, 50*(2), 458-468. doi:10.1002/art.20025

Conti-Wyneken, A. (1999). Bicycling injuries. *Phys Med Rehabil Clin N Am, 10*(1), 67-76.

Cooke, D., Scudamore, A., Li, J., Wyss, U., Bryant, T., & Costigan, P. (1997). Axial lower-limb alignment: comparison of knee geometry in normal volunteers and osteoarthritis patients. *Osteoarthritis Cartilage, 5*(1), 39-47.

D'Lima, D. D., Steklov, N., Patil, S., & Colwell, C. W., Jr. (2008). The Mark Coventry Award: in vivo knee forces during recreation and exercise after knee arthroplasty. *Clin Orthop Relat Res, 466*(11), 2605-2611. doi:10.1007/s11999-008-0345-x

Dannenberg, A. L., Needle, S., Mullady, D., & Kolodner, K. B. (1996). Predictors of injury among 1638 riders in a recreational long-distance bicycle tour: Cycle Across Maryland. *Am J Sports Med, 24*(6), 747-753.

Dettori, N. J., & Norvell, D. C. (2006). Non-traumatic bicycle injuries : a review of the literature. *Sports Med, 36*(1), 7-18.

Deyle, G. D., Allison, S. C., Matekel, R. L., Ryder, M. G., Stang, J. M., Gohdes, D. D., . . . Garber, M. B. (2005). Physical therapy treatment effectiveness for osteoarthritis of the knee: a randomized comparison of supervised clinical exercise and manual therapy procedures versus a home exercise program. *Phys Ther, 85*(12), 1301-1317.

Deyle, G. D., Henderson, N. E., Matekel, R. L., Ryder, M. G., Garber, M. B., & Allison, S. C. (2000). Effectiveness of manual physical therapy and exercise in osteoarthritis of the knee. A randomized, controlled trial. *Ann Intern Med, 132*(3), 173-181. doi:200002010-00002 [pii]

Ercison, M., Nisell, R., & Nemeth, G. (1988). Joint Motions of the Lower Limb during Ergometer Cycling. *J Orthop Sports Phys Ther, 9*(8), 273-278.

Ericson, M. (1986). On the biomechanics of cycling. A study of joint and muscle load during exercise on the bicycle ergometer. *Scand J Rehabil Med Suppl, 16*, 1-43.

Ericson, M., Bratt, A., Nisell, R., Nemeth, G., & Ekholm, J. (1986). Load moments about the hip and knee joints during ergometer cycling. *Scand J Rehabil Med, 18*(4), 165-172.

Ericson, M., & Nisell, R. (1987). Patellofemoral joint forces during ergometric cycling. *Phys Ther, 67*(9), 1365-1369.

Ericson, M., Nisell, R., & Ekholm, J. (1984). Varus and valgus loads on the knee joint during ergometer cycling. *Scand J Sports Sci, 6*, 39-45.

Fakhrai, N., Widhalm, P., Chiari, C., Weber, M., Langs, G., Donner, R., . . . Peloschek, P. (2010). Automatic assessment of the knee alignment angle on full-limb radiographs. *Eur J Radiol, 74*(1), 236-240. doi:10.1016/j.ejrad.2009.02.004

Fang, Y. (2014). *Effects of Cycling Workload and Cadence on Frontal Plane Knee Load.* (Master of Science), The University of Tennessee, Knoxville.

Fang, Y., Fitzhugh, E. C., Crouter, S. E., Gardner, J. K., & Zhang, S. (2014). Effects of Workloads and Cadences on Frontal Plane Knee Biomechanics in Cycling. *Med Sci Sports Exerc*. doi:10.1249/mss.0000000000000759

Felson, D. T., Niu, J., Gross, K. D., Englund, M., Sharma, L., Cooke, T. D., . . . Nevitt, M. C. (2013). Valgus malalignment is a risk factor for lateral knee osteoarthritis incidence and progression: findings from the Multicenter Osteoarthritis Study and the Osteoarthritis Initiative. *Arthritis Rheum, 65*(2), 355-362. doi:10.1002/art.37726

Foroughi, N., Smith, R., & Vanwanseele, B. (2009). The association of external knee adduction moment with biomechanical variables in osteoarthritis: a systematic review. *Knee, 16*(5), 303-309. doi:10.1016/j.knee.2008.12.007

Foroughi, N., Smith, R. M., Lange, A. K., Baker, M. K., Fiatarone Singh, M. A., & Vanwanseele, B. (2010). Dynamic alignment and its association with knee adduction moment in medial knee osteoarthritis. *Knee, 17*(3), 210-216. doi:10.1016/j.knee.2009.09.006

Ganley, K. J., & Powers, C. M. (2005). Gait kinematics and kinetics of 7-year-old children: a comparison to adults using age-specific anthropometric data. *Gait Posture, 21*(2), 141-145. doi:10.1016/j.gaitpost.2004.01.007

Gardner, J. K., Zhang, S., Liu, H., Klipple, G., Stewart, C., Milner, C. E., & Asif, I. M. (2015). Effects of toe-in angles on knee biomechanics in cycling of patients with medial knee osteoarthritis. *Clinical Biomechanics, 30*(3), 276-282.

Gibson, K., Sayers, S., & Minor, M. (2008). Accuracy of a non-radiographic method of measuring varus/valgus alignment in knees with osteoarthritis. *Missouri medicine, 106*(2), 132-135.

Gregersen, C. S., & Hull, M. L. (2003). Non-driving intersegmental knee moments in cycling computed using a model that includes three-dimensional kinematics of the shank/foot and the effect of simplifying assumptions. *J Biomech, 36*(6), 803-813.

Gregor, R. J., Cavanagh, P. R., & LaFortune, M. (1985). Knee flexor moments during propulsion in cycling–a creative solution to Lombard's Paradox. *J Biomech, 18*(5), 307-316.

Hankemeier, S., Gosling, T., Richter, M., Hufner, T., Hochhausen, C., & Krettek, C. (2006). Computer-assisted analysis of lower limb geometry: higher intraobserver reliability compared to conventional method. *Computer Aided Surgery, 11*(2), 81-86.

Hinman, R. S., May, R. L., & Crossley, K. M. (2006). Is there an alternative to the full‐leg radiograph for determining knee joint alignment in osteoarthritis? *Arthritis Care & Research, 55*(2), 306-313.

Hinterwimmer, S., Graichen, H., Vogl, T. J., & Abolmaali, N. (2008). An MRI-based technique for assessment of lower extremity deformities—reproducibility, accuracy, and clinical application. *European radiology, 18*(7), 1497-1505.

Hoevenaar-Blom, M. P., Wendel-Vos, G. W., Spijkerman, A. M., Kromhout, D., & Verschuren, W. M. (2011). Cycling and sports, but not walking, are associated with 10-year cardiovascular disease incidence: the MORGEN Study. *European Journal of Cardiovascular Prevention & Rehabilitation, 18*(1), 41-47.

Hootman, J. M., & Helmick, C. G. (2006). Projections of US prevalence of arthritis and associated activity limitations. *Arthritis Rheum, 54*(1), 226-229. doi:10.1002/art.21562

Hovinga, K. R., & Lerner, A. L. (2009). Anatomic variations between Japanese and Caucasian populations in the healthy young adult knee joint. *Journal of Orthopaedic Research, 27*(9), 1191-1196.

Hsu, R. W., Himeno, S., Coventry, M. B., & Chao, E. Y. (1990). Normal axial alignment of the lower extremity and load-bearing distribution at the knee. *Clin Orthop Relat Res*(255), 215-227.

Hunt, M. A., Birmingham, T. B., Giffin, J. R., & Jenkyn, T. R. (2006). Associations among knee adduction moment, frontal plane ground reaction force, and lever arm during walking in patients with knee osteoarthritis. *J Biomech, 39*(12), 2213-2220.

Hunt, M. A., Birmingham, T. B., Jenkyn, T. R., Giffin, J. R., & Jones, I. C. (2008). Measures of frontal plane lower limb alignment obtained from static radiographs and dynamic gait analysis. *Gait Posture, 27*(4), 635-640. doi:10.1016/j.gaitpost.2007.08.011

Hunt, M. A., Birmingham, T. B., Jenkyn, T. R., Giffin, J. R., & Jones, I. C. (2008). Measures of frontal plane lower limb alignment obtained from static radiographs and dynamic gait analysis. *Gait Posture, 27*(4), 635-640.

Hunter, D. J., Sharma, L., & Skaife, T. (2009). Alignment and osteoarthritis of the knee. *J Bone Joint Surg Am, 91 Suppl 1*, 85-89. doi:10.2106/jbjs.h.01409

Hurwitz, D. E., Ryals, A. B., Case, J. P., Block, J. A., & Andriacchi, T. P. (2002). The knee adduction moment during gait in subjects with knee osteoarthritis is more closely correlated with static alignment than radiographic disease severity, toe out angle and pain. *J Orthop Res, 20*(1), 101-107. doi:10.1016/S0736-0266(01)00081-X

Huy, C., Becker, S., Gomolinsky, U., Klein, T., & Thiel, A. (2008). Health, medical risk factors, and bicycle use in everyday life in the over-50 population. *J Aging Phys Act, 16*(4), 454-464.

Issa, S. N., Dunlop, D., Chang, A., Song, J., Prasad, P. V., Guermazi, A., . . . Sharma, L. (2007). Full-limb and knee radiography assessments of varus-valgus alignment and their relationship to osteoarthritis disease features by magnetic resonance imaging. *Arthritis Rheum, 57*(3), 398-406. doi:10.1002/art.22618

Kornaropoulos, E. I., Taylor, W. R., Duda, G. N., Ehrig, R. M., Matziolis, G., Müller, M., . . . Heller, M. O. (2010). Frontal plane alignment: An imageless method to predict the mechanical femoral–tibial angle (mFTA) based on functional determination of joint centres and axes. *Gait Posture, 31*(2), 204-208.

Kraus, V. B., Vail, T. P., Worrell, T., & McDaniel, G. (2005). A comparative assessment of alignment angle of the knee by radiographic and physical examination methods. *Arthritis Rheum, 52*(6), 1730-1735. doi:10.1002/art.21100

Kutzner, I., Heinlein, B., Graichen, F., Rohlmann, A., Halder, A. M., Beier, A., & Bergmann, G. (2012). Loading of the knee joint during ergometer cycling: telemetric in vivo data. *J Orthop Sports Phys Ther, 42*(12), 1032-1038. doi:10.2519/jospt.2012.4001

Lawrence, R. C., Felson, D. T., Helmick, C. G., Arnold, L. M., Choi, H., Deyo, R. A., . . . National Arthritis Data, W. (2008). Estimates of the prevalence of arthritis and other rheumatic conditions in the United States. Part II. *Arthritis Rheum, 58*(1), 26-35. doi:10.1002/art.23176

Lee, J., Song, J., Hootman, J. M., Semanik, P. A., Chang, R. W., Sharma, L., . . . Dunlop, D. D. (2013). Obesity and other modifiable factors for physical inactivity measured by accelerometer in adults with knee osteoarthritis. *Arthritis Care Res (Hoboken), 65*(1), 53-61. doi:10.1002/acr.21754

Leitch, K. M., Birmingham, T. B., Dunning, C. E., & Giffin, J. R. (2013). Changes in valgus and varus alignment neutralize aberrant frontal plane knee moments in patients with unicompartmental knee osteoarthritis. *J Biomech, 46*(7), 1408-1412. doi:10.1016/j.jbiomech.2013.01.024

Lim, B. W., Hinman, R. S., Wrigley, T. V., Sharma, L., & Bennell, K. L. (2008). Does knee malalignment mediate the effects of quadriceps strengthening on knee adduction moment, pain, and function in medial knee osteoarthritis? A randomized controlled trial. *Arthritis Rheum, 59*(7), 943-951. doi:10.1002/art.23823

Liodakis, E., Kenawey, M., Doxastaki, I., Krettek, C., Haasper, C., & Hankemeier, S. (2011). Upright MRI measurement of mechanical axis and frontal plane alignment as a new technique: a comparative study with weight bearing full length radiographs. *Skeletal radiology, 40*(7), 885-889.

Marx, R., Grimm, P., Lillemoe, K., Robertson, C., Ayeni, O., Lyman, S., . . . Pavlov, H. (2011). Reliability of lower extremity alignment measurement using radiographs and PACS. *Knee Surgery, Sports Traumatology, Arthroscopy, 19*(10), 1693-1698.

McAlindon, T. E., Bannuru, R. R., Sullivan, M. C., Arden, N. K., Berenbaum, F., Bierma-Zeinstra, S. M., . . . Underwood, M. (2014). OARSI guidelines for the non-surgical management of knee osteoarthritis. *Osteoarthritis Cartilage, 22*(3), 363-388. doi:10.1016/j.joca.2014.01.003

Messier, S. P., Pater, M., Beavers, D. P., Legault, C., Loeser, R. F., Hunter, D. J., & DeVita, P. (2014). Influences of alignment and obesity on knee joint loading in osteoarthritic gait. *Osteoarthritis Cartilage, 22*(7), 912-917. doi:10.1016/j.joca.2014.05.013

Miyazaki, T., Wada, M., Kawahara, H., Sato, M., Baba, H., & Shimada, S. (2002). Dynamic load at baseline can predict radiographic disease progression in medial compartment knee osteoarthritis. *Annals of the Rheumatic Diseases, 61*(7), 617-622. doi:Doi 10.1136/Ard.61.7.617

Moreland, J. R., Bassett, L. W., & Hanker, G. J. (1987). Radiographic analysis of the axial alignment of the lower extremity. *J Bone Joint Surg Am, 69*(5), 745-749.

Mündermann, A. (2012). Use of Anthropometry for the Measurement of Lower Extremity Alignment *Handbook of Anthropometry* (pp. 2951-2970): Springer.

Mundermann, A., Dyrby, C. O., & Andriacchi, T. P. (2005). Secondary gait changes in patients with medial compartment knee osteoarthritis: increased load at the ankle, knee, and hip during walking. *Arthritis Rheum, 52*(9), 2835-2844. doi:10.1002/art.21262

Mündermann, A., Dyrby, C. O., & Andriacchi, T. P. (2008). A comparison of measuring mechanical axis alignment using three-dimensional position capture with skin markers and radiographic measurements in patients with bilateral medial compartment knee osteoarthritis. *Knee, 15*(6), 480-485.

Mundermann, A., Dyrby, C. O., Hurwitz, D. E., Sharma, L., & Andriacchi, T. P. (2004). Potential strategies to reduce medial compartment loading in patients with knee osteoarthritis of varying severity: reduced walking speed. *Arthritis Rheum, 50*(4), 1172-1178. doi:10.1002/art.20132

Naal, F. D., Fischer, M., Preuss, A., Goldhahn, J., von Knoch, F., Preiss, S., . . . Drobny, T. (2007). Return to sports and recreational activity after unicompartmental knee arthroplasty. *Am J Sports Med, 35*(10), 1688-1695. doi:10.1177/0363546507303562

Navali, A. M., Bahari, L. A., & Nazari, B. (2012). A comparative assessment of alternatives to the full-leg radiograph for determining knee joint alignment. *Sports Med Arthrosc Rehabil Ther Technol, 4*(1), 40. doi:10.1186/1758-2555-4-40

Palmer, K. T. (2012). Occupational activities and osteoarthritis of the knee. *Br Med Bull, 102*, 147-170. doi:10.1093/bmb/lds012

Price, D., & Donne, B. (1997). Effect of variation in seat tube angle at different seat heights on submaximal cycling performance in man. *J Sports Sci, 15*(4), 395-402. doi:10.1080/026404197367182

Ruby, P., Hull, M. L., & Hawkins, D. (1992). Three-dimensional knee joint loading during seated cycling. *J Biomech, 25*(1), 41-53.

Sailer, J., Scharitzer, M., Peloschek, P., Giurea, A., Imhof, H., & Grampp, S. (2005). Quantification of axial alignment of the lower extremity on conventional and digital total leg radiographs. *European radiology, 15*(1), 170-173.

Salacinski, A. J., Krohn, K., Lewis, S. F., Holland, M. L., Ireland, K., & Marchetti, G. (2012). The effects of group cycling on gait and pain-related disability in individuals with mild-to-moderate knee osteoarthritis: a randomized controlled trial. *J Orthop Sports Phys Ther, 42*(12), 985-995. doi:10.2519/jospt.2012.3813

Schmitz, A., & Noehren, B. (2014). What predicts the first peak of the knee adduction moment? *Knee*. doi:10.1016/j.knee.2014.07.016

Seidel, G. K., Marchinda, D. M., Dijkers, M., & Soutas-Little, R. W. (1995). Hip joint center location from palpable bony landmarks–a cadaver study. *J Biomech, 28*(8), 995-998.

Sharma, L., Hurwitz, D. E., Thonar, E. J., Sum, J. A., Lenz, M. E., Dunlop, D. D., . . . Andriacchi, T. P. (1998). Knee adduction moment, serum hyaluronan level, and disease severity in medial tibiofemoral osteoarthritis. *Arthritis Rheum, 41*(7), 1233-1240.

doi:10.1002/1529-0131(199807)41:7<1233::AID-ART14>3.0.CO;2-L

Sharma, L., Song, J., Dunlop, D., Felson, D., Lewis, C. E., Segal, N., . . . Nevitt, M. (2010). Varus and valgus alignment and incident and progressive knee osteoarthritis. *Annals of the Rheumatic Diseases, 69*(11), 1940-1945. doi:10.1136/ard.2010.129742

Sharma, L., Song, J., Felson, D. T., Cahue, S., Shamiyeh, E., & Dunlop, D. D. (2001). The role of knee alignment in disease progression and functional decline in knee osteoarthritis. *JAMA, 286*(2), 188-195.

Sled, E. A., Sheehy, L. M., Felson, D. T., Costigan, P. A., Lam, M., & Cooke, T. D. (2011). Reliability of lower limb alignment measures using an established landmark-based method with a customized computer software program. *Rheumatol Int, 31*(1), 71-77. doi:10.1007/s00296-009-1236-5

Stief, F., Bohm, H., Dussa, C. U., Multerer, C., Schwirtz, A., Imhoff, A. B., & Doderlein, L. (2014). Effect of lower limb malalignment in the frontal plane on transverse plane mechanics during gait in young individuals with varus knee alignment. *Knee, 21*(3), 688-693. doi:10.1016/j.knee.2014.03.004

Stief, F., Bohm, H., Schwirtz, A., Dussa, C. U., & Doderlein, L. (2011). Dynamic loading of the knee and hip joint and compensatory strategies in children and adolescents with varus malalignment. *Gait Posture, 33*(3), 490-495. doi:10.1016/j.gaitpost.2011.01.001

Sutherland, D. (1997). The development of mature gait. *Gait Posture, 6*(2), 163-170. doi:Doi 10.1016/S0966-6362(97)00029-5

Too, D., & Landwer, G. E. (2000). The effect of pedal crank arm length on joint angle and power production in upright cycle ergometry. *J Sports Sci, 18*(3), 153-161. doi:10.1080/026404100365054

Turcot, K., Armand, S., Lubbeke, A., Fritschy, D., Hoffmeyer, P., & Suva, D. (2013). Does knee alignment influence gait in patients with severe knee osteoarthritis? *Clin Biomech (Bristol, Avon), 28*(1), 34-39. doi:10.1016/j.clinbiomech.2012.09.004

Umberger, B., Scheuchenzuber, H., & Manos, T. (1998). Differences in power output during cycling at different seat tube angles. *Journal of Human Movement Studies, 35*(1), 21-36.

Umberger, B. R., & Martin, P. E. (2001). Testing the planar assumption during ergometer cycling. *Journal of Applied Biomechanics, 17*(1), 55-62. Retrieved from <Go to ISI>://000167377900005

van der Waal, J. M., Terwee, C. B., van der Windt, D. A., Bouter, L. M., & Dekker, J. (2005). Health-related and overall quality of life of patients with chronic hip and knee complaints in general practice. *Quality of life research, 14*(3), 795-803.

Van Manen, M. D., Nace, J., & Mont, M. A. (2012). Management of primary knee osteoarthritis and indications for total knee arthroplasty for general practitioners. *The Journal of the American Osteopathic Association, 112*(11), 709-715.

Vanwanseele, B., Parker, D., & Coolican, M. (2009). Frontal knee alignment: three-dimensional marker positions and clinical assessment. *Clin Orthop Relat Res, 467*(2), 504-509. doi:10.1007/s11999-008-0545-4

Wada, M., Maezawa, Y., Baba, H., Shimada, S., Sasaki, S., & Nose, Y. (2001).

Relationships among bone mineral densities, static alignment and dynamic load in patients with medial compartment knee osteoarthritis. *Rheumatology (Oxford), 40*(5), 499-505.

Walker, T., Gotterbarm, T., Bruckner, T., Merle, C., & Streit, M. R. (2014). Return to sports, recreational activity and patient-reported outcomes after lateral unicompartmental knee arthroplasty. *Knee Surg Sports Traumatol Arthrosc*. doi:10.1007/s00167-014-3111-5

Wanich, T., Hodgkins, C., Columbier, J. A., Muraski, E., & Kennedy, J. G. (2007). Cycling injuries of the lower extremity. *J Am Acad Orthop Surg, 15*(12), 748-756.

Werner, F. W., Ayers, D. C., Maletsky, L. P., & Rullkoetter, P. J. (2005). The effect of valgus/varus malalignment on load distribution in total knee replacements. *J Biomech, 38*(2), 349-355. doi:10.1016/j.jbiomech.2004.02.024

Wilber, C., Holland, G., Madison, R., & Loy, S. (1995). An epidemiological analysis of overuse injuries among recreational cyclists. *Int J Sports Med, 16*(03), 201-206.

Wolchok, J. C., Hull, M. L., & Howell, S. M. (1998). The effect of intersegmental knee moments on patellofemoral contact mechanics in cycling. *J Biomech, 31*(8), 677-683.

Wright, N. C., Riggs, G. K., Lisse, J. R., Chen, Z., & Women's Health, I. (2008). Self-reported osteoarthritis, ethnicity, body mass index, and other associated risk factors in postmenopausal women-results from the Women's Health Initiative. *J Am Geriatr Soc, 56*(9), 1736-1743. doi:10.1111/j.1532-5415.2008.01812.x

Yang, N. H., Nayeb-Hashemi, H., Canavan, P. K., & Vaziri, A. (2010). Effect of frontal plane tibiofemoral angle on the stress and strain at the knee cartilage during the stance phase of gait. *J Orthop Res, 28*(12), 1539-1547. doi:10.1002/jor.21174

Zhang, W., Nuki, G., Moskowitz, R. W., Abramson, S., Altman, R. D., Arden, N. K., . . . Tugwell, P. (2010). OARSI recommendations for the management of hip and knee osteoarthritis: part III: Changes in evidence following systematic cumulative update of research published through January 2009. *Osteoarthritis Cartilage, 18*(4), 476-499. doi:10.1016/j.joca.2010.01.013

Zhao, D., Banks, S. A., Mitchell, K. H., D'Lima, D. D., Colwell, C. W., Jr., & Fregly, B. J. (2007). Correlation between the knee adduction torque and medial contact force for a variety of gait patterns. *J Orthop Res, 25*(6), 789-797. doi:10.1002/jor.20379

APPENDICES

APPENDIX A: TABLES IN CHAPTER V

Table 3. Demographic characteristics of participants (mean ± SD).

	Varus Group	**Neutral Group**
Age (years)	24.4±2.8	24.0±4.1
Height (m)	1.78±0.08	1.76±0.10
Weight (kg)	75.1±16.5	73.1±15.3
BMI (kg/m^2)	23.6±4.6	23.4±2.9
Knee MAA*	174.3±1.4	179.2±1.0

*: significant group difference.
BMI: Body Mass Index
MAA: Mechanical Axis Angle

Table 4. Peak pedal reaction forces (N) and peak knee, ankle joint moments (Nm) (mean ± SD).

		Varus Group						Neutral Group					
		Without Toe-cage			With Toe-cage			Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
Knee	Extension Mom^{Yabc}	21.4±5.4	33.9±10.0	35.0±16.1	22.8±6.5	31.7±10.3	39.3±10.0	23.4±7.4	35.3±10.9	42.5±11.7	26.6±8.0	34.1±11.0	40.3±13.2
	Abduction Mom^{Yabc}	-5.3±1.9	-7.7±2.4	-10.6±4.3	-5.2±2.1	-7.6±4.0	-9.6±3.8	-4.1±2.2	-7.2±4.2	-8.7±5.2	-4.8±2.7	-6.6±3.9	-8.9±6.4
	Crank Angle at Abduction Mom	92.2±41.8	87.4±42.2	89.8±36.0	85.4±47.0	112.3±50.4	91.2±38.4	75.4±33.6	99.8±19.2	107.5±29.3	95.0±63.8	102.7±36.0	102.2±26.4
	Int. Rotation Mom*	4.7±2.2	5.8±3.0	8.6±4.0	3.5±2.5	6.7±4.8	9.9±4.4	3.6±1.6	5.6±2.0	7.3±2.1	4.6±2.3	5.4±2.6	6.7±4.4
Ankle	Inversion Mom^{Z}	1.5±1.4	1.6±1.5	1.4±1.3	1.8±1.8	2.0±1.7	2.1±1.6	1.4±0.9	1.8±0.9	2.0±1.4	1.6±1.1	2.0±1.1	2.0±1.4
	Ext. Rotation Mom*$^{\&XYZ}$	-2.0±1.6	-3.8±1.3	-5.2±1.5	-2.2±1.3	-3.3±1.9	-4.0±2.0	-1.5±0.5	-2.1±0.9	-2.3±0.7	-1.5±0.7	-1.7±0.7	-2.3±0.9
PRF	Medial PRF^{Yabc}	21.2±8.9	31.8±9.5	42.8±13.1	16.8±9.6	29.5±16.0	38.7±17.6	18.4±7.6	31.4±14.3	41.4±13.4	21.3±9.5	30.1±12.5	37.4±16.2
	Vertical PRF$^{\#Yabc}$	152.8±31.8	180.3±41.8	236.9±51.1	155.9±42.4	193.4±43.7	236.3±40.6	158.2±30.5	199.5±44.7	236.5±44.4	154.1±32.9	192.0±37.2	226.3±49.0

*: significant group x toe-cage x workload interaction
$^{\&}$: significant group x workload interaction
$^{\#}$: significant group x toe-cage interaction
^{X}: significant group main effect
^{Y}: significant workload main effect
^{Z}: significant toe-cage main effect

Post hoc comparisons:
For workload main effect only: ^{a}: significantly different between 0.5 – 1.0 kg, ^{b}: significant difference between 1.0 – 1.5 kg, ^{c}: significant difference between 0.5 – 1.5 kg
PRF: Pedal Reaction Force
Mom: moment
Ext.: External
Int.: Internal

Table 5. Peak knee, ankle and foot angles and ROM (°) (mean ± SD).

		Varus Group						Neutral Group					
		Without Toe-cage			With Toe-cage			Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
Knee	Extension ROM[&¥Z]	74.4±5.6	75.1±6.3	74.2±5.4	74.3±5.7	74.1±5.9	74.0±5.5	75.1±6.1	76.2±6.5	76.8±6.0	75.0±6.2	75.6±5.6	75.7±5.8
	Adduction Angle	10.3±4.8	9.7±5.3	10.1±5.2	11.6±4.1	11.0±3.7	10.4±4.3	5.2±9.3	5.0±8.8	5.2±8.9	6.0±8.4	5.5±8.7	5.4±9.0
	Crank Angle at Adduction Angle[Yac]	29.8±12.5	24.5±5.8	22.1±7.2	27.8±9.1	22.1±11.5	23.5±11.5	26.4±11.5	21.6±8.2	19.2±7.2	26.9±11.5	22.1±9.6	20.2±10.1
	Abduction Angle[XZ]	-0.3±4.6	-0.7±5.1	-0.2±4.3	0.3±4.5	0.9±3.3	0.0±4.0	-5.0±5.1	-5.2±5.5	-5.6±5.4	-4.7±5.0	-5.0±5.3	-4.8±5.9
	Crank Angle at Abduction Angle	199.7±28.3	179.0±44.2	184.3±40.8	192.0±20.6	188.2±27.8	182.4±27.8	184.3±43.7	193.4±32.6	180.5±43.7	177.6±42.7	180.5±43.2	164.2±49.9
	Abduction ROM	-9.0±4.3	-9.3±5.1	-9.5±4.5	-9.9±3.7	-9.1±3.8	-9.7±4.6	-8.8±5.2	-9.4±3.8	-10.2±4.3	-9.1±4.0	-9.7±4.1	-9.9±3.7
Ankle	Eversion Angle*	-1.7±7.5	-2.0±6.6	-2.7±6.5	-2.4±6.7	-2.9±7.9	-1.4±6.8	-2.8±4.1	-1.8±3.8	-1.1±5.2	-2.4±4.1	-1.4±2.0	-1.4±3.7
	Inversion ROM[Yc]	3.4±1.8	3.6±2.1	4.2±2.5	3.8±2.0	3.7±1.5	3.5±1.6	3.2±2.1	3.9±2.2	4.0±1.9	2.6±1.6	3.5±2.1	4.2±2.3
Foot	Eversion Angle[Z]	-3.1±4.2	-2.8±3.7	-2.7±5.5	-5.4±3.7	-5.3±4.4	-4.8±4.6	-4.7±5.3	-2.9±4.9	-2.6±5.2	-7.1±3.9	-5.5±3.5	-4.7±3.3
	Mean Ext. Rotation Angle[%Z]	-15.1±4.6	-14.6±5.0	-14.2±4.6	-15.8±3.9	-15.2±4.5	-15.3±3.9	-12.0±5.0	-11.1±5.5	-11.4±4.6	-13.3±4.1	-12.8±5.1	-12.5±4.7

*: group x toe-cage x workload interaction
[&]: significant group x workload interaction
[#]: significant group x toe-cage interaction
[X]: significant group main effect
[Y]: significant workload main effect
[Z]: significant toe-cage main effect

Post hoc comparisons:
For workload main effect only: [a]: significantly different between 0.5 – 1.0 kg, [b]: significant difference between 1.0 – 1.5 kg, [c]: significant difference between 0.5 – 1.5 kg
For neutral group only: [£]: significantly different between 0.5 – 1.0 kg, [§]: significant difference between 1.0 – 1.5 kg, [¥]: significant difference between 0.5 – 1.5 kg
ROM: Range of Motion

Ext.: External
Int.: Internal
[%]: Mean angle of first 25% of pedal cycle

APPENDIX B: INDIVIDUAL SUBJECT CHARACTERISTICS

Table 6. Individual subject characteristics.

Subject	Group	Gender	Age (years)	Height (m)	Weight (kg)	BMI (kg/m^2)	Alignment (°)
1	Varus	Male	25	1.83	110.16	32.89	173.45
2	Varus	Male	28	1.75	88.71	28.97	171.63
4	Varus	Male	22	1.83	69.18	20.68	174.65
8	Varus	Male	22	1.65	61.68	22.63	176.17
13	Varus	Male	23	1.85	82.49	24.10	175.91
18	Varus	Male	26	1.64	54.74	20.23	173.21
20	Varus	Male	20	1.83	67.82	20.14	174.47
22	Varus	Male	22	1.76	61.22	19.76	174.89
25	Varus	Male	25	1.79	78.85	24.75	175.36
26	Varus	Male	26	1.87	62.44	17.95	174.76
29	Varus	Male	29	1.82	88.88	26.98	172.98
7	Neutral	Male	29	1.90	94.46	26.44	178.00
9	Neutral	Male	29	1.85	84.53	24.70	178.52
10	Neutral	Male	24	1.82	83.18	25.11	178.54
11	Neutral	Male	19	1.74	70.91	23.42	178.04
15	Neutral	Female	20	1.72	58.42	19.86	179.19
16	Neutral	Female	21	1.63	53.67	20.20	179.75
21	Neutral	Male	21	1.89	81.77	23.01	180.08
23	Neutral	Male	20	1.81	94.84	28.95	180.31
24	Neutral	Female	24	1.73	58.49	19.54	180.01
27	Neutral	Male	27	1.63	57.66	21.83	180.34
30	Neutral	Female	30	1.64	65.81	24.47	178.15
Mean±SD	Varus		24.36±2.80	1.78±0.08	75.11±16.46	23.55±4.56	174.32±1.37
Mean±SD	Neutral		24.00±4.12	1.76±0.10	73.07±15.29	23.41±2.93	179.18±0.95

APPENDIX C: INFORMED CONSENT FORM

INFORMED CONSENT FORM

Investigators: Hunter Bennett, Guangping Shen, and Songning Zhang, PhD (faculty advisor)
Address: Biomechanics/Sports Medicine Lab
136 HPER
The University of Tennessee at Knoxville
1914 Andy Holt Avenue
Knoxville, TN 37996
Phone: (865) 974-2091

Introduction
You are invited to participate in a research study entitled, "Effects of foot position modifications in level and stair gait in stationary cycling on lower extremity biomechanics in adults with knee mal-alignments." The purpose of this proposed research is to investigate ground reaction forces and lower extremity biomechanics characteristics of two gait modification strategies, toeing-in and toeing-in with increased step width during level ground walking and stair negotiation, and two foot positions in stationary cycling in adults with knee mal-alignment (varus/valgus deformity) who are otherwise healthy This consent form may contain words that you do not understand. Please ask the study staff to explain any words or information that you do not clearly understand. Before agreeing to be in this study, it is important that you read and understand the following explanation of the procedures, risks, and benefits.

Testing Protocol
As a participant, you may be asked to undergo one full-length leg x-ray, which will cover both lower extremities/legs simultaneously, at the Tennessee Orthopedics Clinic. This will be of no cost to you. All included participants will then perform the following data collection procedures:

The biomechanical testing session will take about 2.5 hours. At the beginning, you will fill out a questionnaire about your current physical activity and overall readiness for physical activity, and a subject information sheet. You will be asked to walk five times in each of nine testing conditions including level ground walking, and stair climb and stair descent of one flight of stairs in your natural gait, with your toes turned inwards, and with your toes turned inward and with a wider step width. You will also be asked to ride a stationary bicycle in six cycling conditions: 40, 78, and 117 watts at 80 RPM with your feet parallel and in your self-selected foot placement on the pedals. For each cycling condition you will cycle for 2 minutes. A minimum of two minutes rest will be provided after each condition. During testing, we will perform a 3-dimensional motion analysis. Reflective markers will be applied to your trunk, legs, and shoes. You will be asked wear t-shirt and tight-fitting short during the test.

No part of the attachment of these reflective markers will impede your ability to engage in normal and effective motions during the test. If you have any further questions, interests or concerns about any instrumentation, please feel free to ask the investigators.

Potential Risks
Risks associated with this study are minimal. The full-length leg x-ray involves a small amount of radiation. The radiation exposure from the x-ray is about 513 microsievert. This exposure is about the same amount of radiation as you would get from living in a high altitude city such as Denver for 4 weeks, or taking 10 airplane flights from New York to Los Angeles.

The walking, stair climb and descent, and cycling to be tested in the study session are no different from what you would do in normal daily movements or moderate exercises. The long standing x-ray provides less exposure than three days of natural background radiation exposure. During the testing sessions, the investigators and/or a qualified research assistant will be stationed close to you and provide assistance in case that you lose balance. Should any injury occur during the course of testing, standard first aid procedures will be administered as necessary at least one researcher with a basic knowledge of athletic training and/or first aid procedures will be present at each test session. The University of Tennessee does not "automatically" reimburse subjects for medical claims, or provide other compensation. If physical injury is suffered in the course of research, please contact Hunter Bennett, Guangping Shen, or Dr. Songning Zhang (974-2091).

Benefits of Participation
You will receive one full-length leg x-ray at no cost to you, which will provide information about your lower extremity alignments and may help understand your risk of developing knee osteoarthritis later in your life. If you are interested in the results of your walking and cycling performance, we can provide video animations of your performance during these tests free of charge. Additionally, the results of this study may provide valuable insights into the gait modifications and foot positions in cycling on knee joint loading of people with mal-alignments of lower limbs which may provide further insights of knee joint loading for individuals with knee osteoarthritis, a degenerative joint disease, which is commonly associated with knee alignment.

Compensation
You will not be compensated for participating in this study.

Voluntary Participation and Withdrawal
Your participation is entirely voluntary and your refusal to participate will involve no penalty or loss of benefits to which you are otherwise entitled. It is your obligation to ask questions regarding any aspect of this study that you do not understand. You may stop participating in this study voluntarily or may be asked to stop if you fail to follow the study procedures or if the Investigator feels that it is in your best interest to stop. Furthermore, participation or non-participation in this study will have no effect on current or future treatments you receive from your physician.

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

Contact Information
If you have any questions at any time about the study you can contact Hunter Bennett, Guangping Shen, or their advisor Dr. Songning Zhang. Questions about your rights as a participant can be addressed to Research Compliance Services in the Office of Research at the University of Tennessee at (865) 974-3466.

Consent Statement
The study has been explained fully to my satisfaction and I agree to participate as described. I have been given the opportunity to discuss all aspects of this study and to ask questions. Answers to such questions, if any, were satisfactory. I am qualified for the study and freely give my informed consent to serve as a subject. By signing this consent form, I have not given up any of my legal rights as a participant.

Subject's Name: Signature: Date:

__________________________ _________________________ _________________

Investigator's Signature: Date:

__________________________ ____________________

APPENDIX D: FLYER

RESEARCH PARTICIPANTS NEEDED FOR A STUDY ON THE EFFECTS OF FOOT POSITIONS IN LEVEL WALKING, STAIR NEGOTIATION, AND STATIONARY 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 two foot positions during level walking, stair ascent and descent, and cycling on the knee joint. Test and control participants will be required to attend one 2.5 hour testing session in the Biomechanics/Sports Medicine lab, as well as obtaining one standing full-length x-ray* of lower extremities (no cost to you) at Tennessee Orthopedic Clinic.

*The exposure is small and includes 513 microsievert (similar to living in Denver, CO for 4 weeks)

Qualifications to participate in the study include:

• Between the ages of 18 and 30 yrs.
• Healthy
• No musculoskeletal injuries in past 3 months
• Able to ascend/descend stairs without the use of handrail
•Able to ride stationary bike for minimum 15 minutes

If you would like to participate or for more information contact Hunter or Steven at the UT Biomechanics/Sports Medicine Lab.
Office: 865-974-2091
Email: hbennet4@utk.edu or gshen1@utk.edu

Contact: Hunter B.
P: 974-2091
E: hbennet4@utk.edu

Contact: Steven S.
P: 974-2091
E: gshen1@utk.edu

Contact: Hunter B.
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Contact: Steven S.
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Contact: Hunter B.
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APPENDIX E: PHYSICAL ACTIVITY READINESS QUESTIONNAIRE (PAR-Q)

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

PAR-Q & YOU

(A Questionnaire for People Aged 15 to 69)

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

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

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

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

If you answered

YES to one or more questions

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

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

NO to all questions

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

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

DELAY BECOMING MUCH MORE ACTIVE:

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

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

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

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

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

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

NAME ____

SIGNATURE ____ DATE ________________________

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

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

CSEP SCPE Supported by: Health Canada Santé Canada

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APPENDIX F: INDIVIDUAL RESULTS FOR SELECTED VARIABLE

Table 7. Peak knee extension moment (Nm).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	26.362±1.273	45.392±3.652	53.626±3.175	35.663±2.403	46.082±6.667	61.226±1.150
9	Neutral	35.758±3.406	41.941±2.257	53.410±3.998	30.284±0.729	44.875±2.527	56.420±4.759
10	Neutral	19.589±3.474	39.371±5.114	41.346±2.315	19.664±2.080	30.682±3.559	46.561±3.009
11	Neutral	10.622±1.613	12.535±1.388	19.457±3.172	11.429±0.762	13.027±2.902	18.607±2.786
15	Neutral	17.855±2.883	19.641±2.477	28.150±1.386	19.546±1.334	18.687±1.531	21.764±2.655
16	Neutral	30.997±3.256	35.790±2.572	39.287±2.937	34.152±2.994	35.732±0.931	43.573±2.103
21	Neutral	21.504±5.418	36.650±7.234	58.688±6.026	31.142±5.393	42.430±5.365	37.585±8.764
23	Neutral	17.714±4.795	47.917±2.808	49.342±11.055	37.001±7.532	38.775±2.864	39.522±4.689
24	Neutral	30.873±2.287	43.085±5.430	44.943±2.531	27.165±2.245	39.639±3.662	45.847±4.498
27	Neutral	27.275±3.546	36.827±2.380	43.607±3.445	25.232±0.901	40.453±2.998	43.239±3.077
30	Neutral	19.092±0.247	28.616±1.587	35.403±3.525	21.269±2.182	24.799±1.158	29.330±3.401
1	Varus	27.657±3.400	48.411±9.182	49.228±4.632	31.995±3.601	33.892±4.054	53.141±9.630
2	Varus	28.116±4.569	36.699±2.621	49.803±2.500	26.710±1.811	40.445±3.559	43.949±2.881
4	Varus	11.888±1.372	13.557±4.115	21.750±3.823	16.112±4.060	11.752±3.611	26.242±1.789
8	Varus	22.295±2.483	43.094±7.614	-3.884±3.979	26.890±2.688	43.736±3.534	45.111±3.122
13	Varus	17.478±4.104	31.752±11.188	43.359±5.041	24.230±5.791	38.945±5.938	49.765±2.962
18	Varus	18.296±2.584	27.835±2.229	24.767±1.003	15.738±2.582	24.537±4.090	26.110±1.674
20	Varus	17.917±4.081	28.552±0.877	40.951±8.091	19.703±3.074	23.225±3.783	33.513±3.768
22	Varus	16.581±0.897	26.657±2.335	34.809±11.143	10.418±2.141	21.940±2.973	28.303±8.525
25	Varus	23.133±3.645	31.253±4.895	32.518±5.825	23.370±3.208	29.489±2.306	32.924±4.494
26	Varus	27.635±3.066	42.780±3.855	40.751±3.755	28.023±4.526	38.039±5.447	44.737±4.415
29	Varus	24.663±5.982	42.333±3.189	50.459±8.868	27.792±3.472	42.951±7.939	48.337±5.480
Mean±SD	Neutral	23.422±7.437	35.251±10.931	42.478±11.652	26.595±8.001	34.107±10.975	40.334±13.151
Mean±SD	Varus	21.424±5.371	33.902±9.981	34.955±16.116	22.816±6.547	31.723±10.328	39.285±10.036

Table 8. Peak knee abduction moment (Nm).

		Without Toe-cage			With Toe-cage		
Subject	Group	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	-8.136±0.411	-15.710±2.053	-19.130±1.823	-10.441±1.259	-15.247±1.206	-20.963±1.728
9	Neutral	-4.299±0.053	-4.265±0.824	-6.347±0.778	-4.182±0.287	-4.470±0.862	-3.532±1.198
10	Neutral	-5.957±0.841	-11.603±1.689	-14.362±1.947	-5.891±0.488	-8.436±0.868	-15.727±0.957
11	Neutral	–	–	–	–	–	–
15	Neutral	–	–	–	–	–	–
16	Neutral	-2.000±0.272	-4.283±0.416	-4.069±0.644	-1.885±0.207	-2.519±0.575	-3.917±0.441
21	Neutral	–	–	–	–	–	–
23	Neutral	-3.800±0.872	-5.590±0.975	-7.008±1.663	-4.827±0.999	-6.693±0.434	-9.227±1.898
24	Neutral	-1.395±0.581	-3.850±0.842	-4.738±1.175	-2.032±0.968	-4.382±0.593	-3.341±0.481
27	Neutral	-2.935±1.391	-5.435±0.523	-7.671±0.969	-3.638±0.756	-5.431±0.503	-8.345±1.189
30	Neutral	-4.067±0.468	-6.598±0.953	-6.500±0.811	-5.465±0.692	-5.706±0.510	-5.991±0.456
1	Varus	-4.881±1.196	-7.537±0.709	-9.314±1.232	-5.284±0.967	-5.567±1.019	-6.936±0.864
2	Varus	-8.741±1.491	-9.078±0.868	-14.495±1.778	-7.836±0.628	-10.415±1.297	-12.760±1.029
4	Varus	-4.982±1.121	-6.137±1.170	-10.615±1.575	-6.616±1.592	-3.411±0.329	-9.621±0.660
8	Varus	-4.466±0.543	–	-8.226±1.175	-5.352±0.201	-3.654±0.469	-7.296±0.939
13	Varus	-5.716±0.781	-6.233±1.299	-6.632±1.866	-3.497±0.330	-7.138±1.019	-7.207±0.821
18	Varus	-3.572±0.855	-3.772±0.344	-6.648±0.227	-3.144±0.392	-4.368±0.478	-5.084±0.504
20	Varus	-3.331±1.549	-6.334±1.140	-9.374±1.210	-2.224±0.726	-4.967±1.314	-6.145±1.476
22	Varus	–	-7.665±0.737	-5.068±1.980	–	–	–
25	Varus	-5.724±1.067	-7.618±0.392	-10.666±1.560	-4.772±0.924	-9.085±2.424	-10.639±0.931
26	Varus	-8.433±1.334	-11.513±0.973	-16.555±2.106	-9.161±1.800	-15.682±2.969	-16.986±4.172
29	Varus	-3.506±0.520	-11.530±1.331	-18.810±2.953	-4.433±0.891	-11.396±2.638	-13.007±3.392
Mean±SD	Neutral	-4.073±2.169	-7.167±4.246	-8.728±5.241	-4.795±2.709	-6.610±3.899	-8.881±6.387
Mean±SD	Varus	-5.335±1.917	-7.742±2.427	-10.582±4.345	-5.232±2.145	-7.568±4.010	-9.568±3.766

–: Subject did not show the peak knee abduction moment.

Table 9. Peak knee internal rotation moment (Nm).

		Without Toe-cage			With Toe-cage		
Subject	Group	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	5.259±0.506	7.519±0.252	7.870±1.671	6.239±0.651	6.289±1.739	9.441±1.403
9	Neutral	0.878±0.438	2.975±0.555	4.680±1.103	2.381±0.808	3.766±0.727	-3.339±1.712
10	Neutral	3.740±0.788	5.539±0.922	7.691±1.011	6.613±0.680	7.653±0.911	8.236±1.164
11	Neutral	–	–	–	–	–	–
15	Neutral	1.343±0.581	2.337±1.515	4.041±0.403	1.733±0.766	-0.303±0.320	2.932±1.339
16	Neutral	3.497±0.539	6.087±0.631	7.897±0.705	3.694±0.330	4.517±0.456	7.483±0.549
21	Neutral	–	–	–	–	–	–
23	Neutral	5.095±0.728	4.508±0.842	7.149±1.792	8.659±1.672	6.322±1.106	10.448±1.430
24	Neutral	3.898±1.273	6.056±1.334	6.428±0.955	3.022±0.841	5.196±0.622	6.815±1.538
27	Neutral	3.456±0.809	7.974±0.732	10.392±0.886	3.197±0.747	8.610±1.638	9.099±1.273
30	Neutral	5.191±0.260	7.004±0.700	9.699±1.019	6.140±0.761	6.293±0.873	9.215±1.233
1	Varus	0.974±0.273	3.955±0.928	6.204±1.630	-0.148±0.344	2.729±1.628	9.749±3.881
2	Varus	4.069±1.219	3.752±0.225	10.463±0.587	3.092±0.289	6.428±2.004	10.495±0.958
4	Varus	5.188±0.238	4.655±1.418	6.961±1.265	3.506±0.465	4.560±0.385	8.165±1.375
8	Varus	3.466±0.242	–	–	–	–	–
13	Varus	4.847±2.067	4.311±1.875	4.955±0.736	3.526±1.807	2.916±0.914	7.292±3.394
18	Varus	–	–	–	–	–	–
20	Varus	5.044±0.787	4.585±1.063	6.354±1.370	1.438±0.723	3.184±1.308	3.786±0.894
22	Varus	–	3.647±0.807	4.097±1.977	–	–	–
25	Varus	7.359±1.079	10.745±1.518	13.437±3.105	8.226±2.313	13.327±3.512	13.884±1.066
26	Varus	8.363±1.645	11.158±2.302	15.729±2.195	5.541±1.162	14.843±3.530	18.118±1.963
29	Varus	2.754±0.658	5.412±0.746	9.529±1.972	2.776±1.104	5.417±1.053	7.566±1.874
Mean±SD	Neutral	3.595±1.587	5.555±1.951	7.316±2.074	4.631±2.345	5.372±2.594	6.703±4.353
Mean±SD	Varus	4.674±2.249	5.802±2.970	8.637±3.962	3.495±2.529	6.675±4.765	9.882±4.415

–: Subject did not show the peak knee internal rotation moment.

Table 10. Peak ankle inversion moment (Nm).

Subject	Group	Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	2.751±0.177	3.265±0.520	5.291±1.476	3.321±0.197	3.582±0.627	3.868±0.486
9	Neutral	1.688±0.061	1.592±0.390	1.598±0.325	2.015±0.196	2.160±0.365	1.773±0.132
10	Neutral	2.770±0.346	3.084±0.219	3.211±0.216	3.250±0.114	3.618±0.356	4.453±0.064
11	Neutral	1.573±0.240	0.858±0.228	2.174±0.532	1.491±0.094	1.673±0.246	2.013±0.225
15	Neutral	0.104±0.178	0.563±0.343	0.091±0.260	-0.362±0.198	0.402±0.204	0.274±0.077
16	Neutral	1.335±0.170	2.210±0.266	1.810±0.243	1.615±0.106	3.092±1.030	2.179±0.434
21	Neutral	1.000±0.182	1.378±0.251	2.403±0.250	0.982±0.366	1.564±0.522	1.349±0.628
23	Neutral	0.566±0.299	1.029±0.189	0.378±0.235	1.019±0.807	0.956±0.499	0.329±0.585
24	Neutral	0.597±0.273	1.507±0.527	0.865±0.302	0.583±0.213	1.285±0.134	0.788±0.177
27	Neutral	1.264±0.141	1.599±0.099	2.102±0.426	1.395±0.265	1.078±0.378	1.138±0.301
30	Neutral	2.055±0.157	2.757±0.242	2.540±0.168	2.292±0.387	2.587±0.231	3.366±0.377
1	Varus	3.592±0.552	3.518±0.518	1.586±0.292	3.812±0.790	3.148±0.345	3.180±1.270
2	Varus	2.125±0.234	2.119±0.067	2.942±0.189	1.868±0.225	2.745±0.149	2.780±0.247
4	Varus	0.481±0.081	0.150±0.194	0.650±0.172	0.067±0.171	0.487±0.038	0.544±0.372
8	Varus	0.446±0.310	3.383±1.462	1.785±1.361	1.000±0.361	4.492±0.899	2.403±0.662
13	Varus	3.266±0.353	2.081±0.444	2.982±1.140	6.013±0.761	3.349±0.518	3.779±0.465
18	Varus	2.635±0.674	3.071±0.560	2.474±0.266	2.451±0.247	2.698±0.327	4.586±0.514
20	Varus	0.127±0.102	0.100±0.182	0.507±0.166	0.492±0.055	0.358±0.226	0.865±0.178
22	Varus	2.345±0.822	1.336±0.219	0.087±0.359	0.972±0.309	1.036±0.321	1.688±0.277
25	Varus	-2.183±0.515	-4.922±0.603	-0.138±0.090	-0.271±0.166	0.065±0.171	0.175±0.110
26	Varus	-0.320±0.310	-2.090±0.515	-0.497±0.212	0.050±0.292	-0.161±0.320	-0.596±0.174
29	Varus	1.705±0.317	3.013±0.429	3.101±0.520	2.254±0.805	3.915±0.825	3.172±0.473
Mean±SD	Neutral	1.428±0.862	1.804±0.908	2.042±1.433	1.600±1.095	2.000±1.095	1.957±1.409
Mean±SD	Varus	1.460±1.438	1.639±1.541	1.407±1.348	1.760±1.843	2.012±1.685	2.052±1.641

Table 11. Peak ankle external rotation moment (Nm).

		Without Toe-cage			With Toe-cage		
Subject	Group	0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	–	–	–	–	–	–
9	Neutral	-2.264±0.328	-2.422±0.415	-3.234±0.441	-1.631±0.207	-1.662±0.383	-2.874±0.508
10	Neutral	-1.843±0.351	-3.174±0.394	-3.080±0.411	-1.521±0.207	-1.578±1.018	-3.265±0.203
11	Neutral	–	–	–	–	–	–
15	Neutral	-1.691±0.285	-1.123±0.265	-1.579±0.349	-0.854±0.161	-0.744±0.163	-0.784±0.554
16	Neutral	-1.015±0.257	-1.281±0.260	-1.387±0.269	-1.261±0.079	-1.544±0.436	-1.436±0.242
21	Neutral	–	–	–	–	–	–
23	Neutral	-1.964±0.494	-3.378±0.765	-2.821±0.913	-2.995±0.333	-2.452±0.166	-3.255±0.650
24	Neutral	-1.247±0.109	-2.257±0.343	-2.290±0.476	-0.626±0.476	-1.147±0.221	-1.911±0.451
27	Neutral	-1.324±0.623	-2.013±0.130	-2.515±0.707	-1.108±0.445	-2.975±0.515	-2.921±0.922
30	Neutral	-0.608±0.235	-1.285±0.273	-1.460±0.325	-1.664±0.437	-1.490±0.239	-1.725±0.361
1	Varus	-3.386±0.625	-4.983±0.287	-6.415±0.772	-3.834±0.578	-3.958±0.643	-4.493±0.318
2	Varus	-4.064±0.828	-5.363±0.497	-6.914±0.521	-3.989±0.443	-5.718±0.644	-5.224±0.284
4	Varus	-1.790±0.350	-2.037±0.468	-3.825±0.606	-2.039±0.438	-1.662±0.177	-3.241±0.488
8	Varus	-1.136±0.117	–	–	–	–	–
13	Varus	–	–	–	–	–	–
18	Varus	–	–	–	–	–	–
20	Varus	-1.945±1.010	-2.694±0.462	-3.657±0.655	-0.870±0.635	-1.856±0.895	-2.394±0.589
22	Varus	1.281±0.411	-3.570±0.199	-2.974±0.841	-0.225±0.832	-0.262±0.563	-0.311±0.329
25	Varus	-2.953±0.750	-4.313±0.439	-5.870±0.952	-1.874±0.411	-3.555±0.682	-4.470±0.550
26	Varus	-2.726±0.379	-4.989±0.752	-6.110±1.093	-3.010±1.161	-5.442±1.828	-6.942±1.365
29	Varus	-1.561±0.653	-2.583±0.221	-6.066±1.078	-1.869±0.505	-3.833±0.663	-5.108±1.353
Mean±SD	Neutral	-1.494±0.545	-2.117±0.863	-2.296±0.742	-1.458±0.723	-1.699±0.707	-2.272±0.932
Mean±SD	Varus	-2.031±1.556	-3.816±1.275	-5.229±1.496	-2.214±1.333	-3.286±1.898	-4.023±2.022

–: Subject did not show the peak ankle external rotation moment.

Table 12. Peak medial pedal reaction force (N).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	23.479±2.113	44.034±4.348	55.909±5.366	29.920±3.980	42.636±3.112	57.105±4.043
9	Neutral	20.423±1.575	26.995±1.972	36.321±3.594	20.816±1.966	30.434±3.110	31.593±4.537
10	Neutral	21.260±2.384	39.208±6.484	45.052±4.748	26.441±1.772	29.781±1.878	47.800±3.757
11	Neutral	4.051±0.961	6.180±0.649	16.825±3.750	5.984±1.337	7.979±1.332	11.918±4.720
15	Neutral	10.631±1.361	14.177±2.206	29.982±2.775	12.931±2.406	11.989±2.014	19.090±2.285
16	Neutral	24.482±2.045	24.837±2.604	41.220±2.326	21.825±1.618	28.708±1.419	41.909±2.035
21	Neutral	6.060±3.831	16.166±3.843	23.284±1.829	8.808±3.479	17.897±7.196	12.643±6.227
23	Neutral	22.444±2.686	43.546±1.561	50.414±11.039	37.020±5.885	42.511±4.841	51.828±5.595
24	Neutral	24.441±3.180	47.059±8.086	48.753±5.562	21.019±2.673	38.159±3.105	42.805±4.811
27	Neutral	21.291±5.839	41.075±2.504	49.413±3.363	18.447±3.018	42.742±6.714	44.450±6.853
30	Neutral	23.826±0.639	42.144±1.877	58.350±5.325	30.710±1.841	38.028±1.360	50.666±5.553
1	Varus	10.906±4.222	35.457±5.299	42.424±4.988	10.925±2.148	19.627±7.644	42.778±17.123
2	Varus	40.316±7.880	45.907±4.137	70.111±5.090	35.781±2.429	55.313±8.137	66.115±3.664
4	Varus	18.900±1.317	21.053±5.278	33.345±6.094	18.978±4.303	16.278±1.809	37.151±3.508
8	Varus	21.242±1.015	–	–	–	–	–
13	Varus	18.092±3.568	24.245±9.138	29.396±1.302	9.391±4.433	24.396±2.423	36.346±4.134
18	Varus	10.548±4.207	17.613±2.782	32.420±2.014	11.596±1.969	19.395±5.834	17.341±4.089
20	Varus	18.561±3.700	34.300±3.403	46.805±4.428	15.367±3.947	26.681±4.926	32.416±3.724
22	Varus	–	25.001±2.127	29.226±7.436	0.568±2.868	4.804±3.282	7.215±4.103
25	Varus	22.799±2.715	31.755±4.761	38.827±8.929	22.401±4.574	36.795±9.680	39.405±3.196
26	Varus	31.256±4.197	41.949±8.102	50.432±6.866	23.911±2.467	49.013±11.413	58.875±5.295
29	Varus	19.229±3.487	40.419±2.384	55.206±3.175	19.301±2.783	42.658±1.573	48.891±6.059
Mean±SD	Neutral	18.399±7.643	31.402±14.316	41.411±13.413	21.266±9.517	30.078±12.519	37.437±16.159
Mean±SD	Varus	21.185±8.909	31.770±9.522	42.819±13.128	16.822±9.633	29.496±15.957	38.653±17.558

–: Subject did not show the peak medial pedal reaction force.

Table 13. Peak vertical pedal reaction force (N).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	196.432±6.829	258.793±14.367	296.628±3.963	199.583±16.830	246.111±21.585	313.087±11.451
9	Neutral	181.880±7.298	230.250±9.996	277.522±15.743	175.470±8.005	213.695±4.816	254.168±12.845
10	Neutral	162.085±15.225	242.314±11.440	269.292±11.139	156.697±4.534	220.850±24.740	288.284±8.831
11	Neutral	89.010±9.568	108.111±5.774	134.596±7.134	76.393±8.402	106.418±10.583	130.396±10.401
15	Neutral	161.663±8.655	165.222±5.676	210.324±14.385	145.166±6.858	183.751±16.555	190.074±7.248
16	Neutral	140.570±10.069	168.196±11.501	192.386±12.163	152.611±11.353	172.622±5.354	207.638±5.748
21	Neutral	196.771±17.135	191.519±23.887	244.254±15.620	168.188±7.783	172.862±9.800	203.463±16.399
23	Neutral	163.715±30.247	250.146±19.204	248.009±45.827	182.920±22.076	226.668±10.231	211.954±24.689
24	Neutral	163.467±6.561	209.938±25.623	246.272±13.736	170.958±4.902	202.094±15.924	240.951±10.882
27	Neutral	148.713±15.555	183.306±4.995	243.354±11.143	129.609±7.800	188.955±14.015	226.880±15.261
30	Neutral	135.661±8.883	186.269±4.911	239.237±16.821	137.235±8.145	178.307±5.063	222.448±20.348
1	Varus	205.740±10.274	254.175±11.388	275.966±15.589	216.834±5.631	208.003±16.396	268.792±27.680
2	Varus	183.815±17.019	154.880±90.048	305.164±11.253	193.719±13.913	238.350±13.049	295.023±8.722
4	Varus	86.452±7.004	92.292±11.503	120.857±13.552	71.315±12.793	82.556±8.198	137.595±7.557
8	Varus	128.094±6.791	173.688±29.382	212.090±28.869	118.409±10.569	205.285±9.705	225.190±19.529
13	Varus	162.390±14.002	168.471±24.442	226.662±16.870	187.230±20.608	209.500±13.326	246.720±10.219
18	Varus	176.781±11.845	209.127±11.234	217.803±6.160	181.732±10.930	197.205±27.100	219.977±13.693
20	Varus	149.777±15.193	194.468±14.326	269.685±13.154	170.683±13.108	202.649±16.228	236.675±8.223
22	Varus	132.175±8.130	151.871±5.405	232.139±13.446	111.076±10.765	158.097±9.308	235.470±30.975
25	Varus	159.066±17.482	176.662±16.779	225.583±25.104	142.252±11.643	183.255±23.106	217.011±17.222
26	Varus	139.720±5.005	188.676±7.390	220.211±21.237	151.995±10.293	196.600±32.449	243.741±21.008
29	Varus	156.404±7.839	218.784±13.585	299.282±42.879	170.097±9.887	245.928±13.682	272.953±20.794
Mean±SD	Neutral	158.179±30.482	199.460±44.695	236.534±44.421	154.076±32.939	192.030±37.237	226.304±48.955
Mean±SD	Varus	152.765±31.784	180.281±41.758	236.858±51.120	155.940±42.407	193.402±43.749	236.286±40.641

Table 14. Peak knee extension ROM (°).

Subject	Group	Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	70.533±0.564	70.551±0.823	71.356±0.532	70.636±1.029	72.245±0.605	71.682±0.507
9	Neutral	75.288±0.508	76.108±0.322	76.499±0.387	74.747±0.196	76.086±0.600	75.526±0.707
10	Neutral	65.665±0.353	68.025±0.724	67.783±0.276	65.203±0.246	66.540±0.516	67.927±1.036
11	Neutral	75.745±0.608	80.014±0.658	79.856±0.513	76.861±1.646	78.656±0.999	78.716±0.899
15	Neutral	72.952±0.770	71.486±0.622	73.325±0.505	74.296±0.920	73.083±0.478	73.357±0.364
16	Neutral	86.700±1.321	87.672±0.734	88.640±1.341	87.054±0.689	85.088±0.831	86.181±1.234
21	Neutral	70.201±1.009	70.018±1.953	73.900±0.777	69.571±0.805	70.269±2.160	69.418±1.259
23	Neutral	72.365±0.472	74.572±0.875	76.356±0.505	71.212±1.692	74.851±0.653	74.637±1.188
24	Neutral	75.022±1.014	74.852±0.609	73.936±0.902	74.563±0.339	73.245±0.347	72.646±1.016
27	Neutral	84.740±1.100	86.214±0.677	85.092±0.803	83.332±0.579	84.136±0.761	84.406±0.571
30	Neutral	76.450±0.998	78.715±0.360	78.205±0.979	77.379±0.729	77.772±1.531	78.678±0.962
1	Varus	67.335±1.147	67.325±0.688	67.268±1.617	65.868±0.505	64.339±1.299	65.716±0.828
2	Varus	82.698±1.098	83.940±1.151	81.535±0.547	82.607±0.812	82.159±0.641	81.748±0.554
4	Varus	69.446±0.746	66.554±0.759	67.990±0.742	67.504±1.434	69.444±0.603	69.147±0.516
8	Varus	73.946±1.124	78.889±0.864	75.852±0.904	75.721±0.820	74.790±0.574	75.227±1.134
13	Varus	71.368±0.645	72.728±0.952	71.633±0.618	73.248±0.638	73.317±0.815	74.403±0.623
18	Varus	83.109±1.361	83.921±0.932	81.313±1.975	82.361±0.626	83.178±0.209	80.296±1.431
20	Varus	80.690±0.874	80.282±0.498	81.619±1.055	78.446±0.988	79.653±0.471	81.494±1.184
22	Varus	76.592±1.523	75.091±1.666	74.181±1.309	75.275±1.404	75.319±1.494	73.393±0.771
25	Varus	68.165±0.282	69.252±1.053	68.756±0.455	67.623±0.514	67.572±1.509	67.631±0.474
26	Varus	73.455±1.426	76.992±0.768	74.151±0.431	76.937±0.323	74.274±1.180	73.011±2.301
29	Varus	72.103±1.425	70.831±0.610	71.425±2.210	71.856±0.412	71.448±0.766	71.539±0.875
Mean±SD	Neutral	75.060±6.129	76.203±6.453	76.814±6.001	74.987±6.192	75.634±5.580	75.743±5.793
Mean±SD	Varus	74.446±5.639	75.073±6.263	74.157±5.410	74.313±5.745	74.136±5.894	73.964±5.456

Table 15. Peak knee adduction angle (°).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	15.775±0.657	16.053±0.221	16.680±0.415	16.686±0.626	14.916±0.389	15.662±0.513
9	Neutral	0.157±0.636	-0.449±0.746	0.370±0.991	1.784±0.183	0.292±0.613	-1.201±0.879
10	Neutral	16.844±0.449	16.886±0.679	18.507±0.553	18.621±0.636	19.928±0.105	20.339±0.778
11	Neutral	4.036±1.511	3.543±0.754	2.386±1.241	5.494±0.348	2.657±0.257	0.849±0.558
15	Neutral	-13.412±0.760	-12.692±0.662	-13.172±0.511	-10.781±0.789	-11.039±0.580	-11.694±0.538
16	Neutral	5.129±0.630	8.017±0.395	9.393±0.895	7.665±0.572	8.607±1.152	8.754±0.994
21	Neutral	-1.361±0.888	-1.609±1.066	-0.047±1.403	-0.714±2.247	-0.918±0.812	0.646±1.492
23	Neutral	10.500±1.059	7.930±1.049	3.934±0.514	5.335±1.121	3.695±0.694	4.871±1.516
24	Neutral	3.864±1.553	4.720±0.497	6.475±1.220	6.635±1.224	7.906±0.622	7.792±1.050
27	Neutral	-0.958±0.222	-0.679±0.345	0.635±0.344	0.799±0.822	0.922±0.330	0.160±0.699
30	Neutral	16.715±0.833	13.672±1.230	12.102±0.632	14.013±0.601	13.496±0.509	13.387±0.687
1	Varus	0.889±1.851	-0.338±0.996	-0.146±1.889	4.612±1.156	4.064±0.743	1.155±2.036
2	Varus	13.585±0.570	11.913±0.657	12.942±0.545	12.890±0.304	11.721±0.405	12.719±0.159
4	Varus	14.917±0.352	14.784±1.479	15.243±0.981	15.847±0.366	13.079±0.845	12.329±0.664
8	Varus	10.732±1.307	11.910±0.971	11.809±0.473	11.109±0.592	11.528±0.859	11.528±0.313
13	Varus	15.429±1.449	9.737±1.679	9.833±0.994	9.296±0.195	9.811±0.639	10.036±0.338
18	Varus	14.647±0.794	12.885±1.163	14.743±0.765	16.217±0.480	12.558±1.654	14.202±3.263
20	Varus	9.773±2.369	10.387±0.575	11.966±1.170	12.320±1.244	13.016±1.027	12.263±0.767
22	Varus	11.413±5.736	13.732±1.461	10.315±1.614	13.964±2.276	14.921±1.122	13.745±0.880
25	Varus	2.920±1.089	0.000±0.491	0.459±0.511	3.919±1.063	4.065±0.957	3.060±1.046
26	Varus	8.126±1.252	8.377±1.309	11.942±1.697	14.004±1.761	13.777±1.771	11.117±0.718
29	Varus	10.651±1.275	13.793±0.960	12.193±0.792	13.915±1.268	12.281±1.454	12.529±0.790
Mean±SD	Neutral	5.208±9.295	5.036±8.816	5.206±8.913	5.958±8.444	5.497±8.662	5.415±9.034
Mean±SD	Varus	10.280±4.763	9.744±5.252	10.118±5.180	11.645±4.143	10.984±3.660	10.426±4.289

Table 16. Peak knee abduction angle (°).

Subject	Group	Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	-0.359±1.166	1.468±0.484	-0.126±0.397	-0.453±0.627	-0.504±0.138	0.261±0.398
9	Neutral	-5.572±0.398	-5.930±0.580	-6.047±0.839	-5.026±0.452	-4.744±0.526	-6.030±0.697
10	Neutral	4.958±0.299	5.987±0.347	6.386±0.264	5.872±0.244	6.706±0.515	7.343±0.386
11	Neutral	-2.492±0.673	-5.598±0.378	-5.542±0.588	-3.590±0.727	-6.937±0.481	-6.146±0.974
15	Neutral	-15.139±0.713	-15.103±1.789	-15.328±0.963	-13.353±1.235	-14.853±0.617	-17.052±1.479
16	Neutral	-5.470±0.369	-5.537±0.199	-5.069±0.315	-4.788±0.940	-4.341±0.269	-5.151±0.384
21	Neutral	-9.670±0.707	-10.341±1.356	-10.370±1.137	-9.687±0.838	-9.658±0.668	-8.927±0.794
23	Neutral	-4.358±0.809	-3.543±0.469	-6.098±1.116	-5.636±1.202	-6.102±0.511	-4.737±0.964
24	Neutral	-6.595±1.412	-6.939±0.489	-5.849±0.246	-4.275±1.074	-3.877±0.645	-3.807±0.916
27	Neutral	-7.347±0.440	-7.328±0.972	-6.858±0.461	-8.075±0.928	-5.387±0.658	-5.936±0.455
30	Neutral	-2.950±0.296	-3.785±0.458	-6.335±0.462	-3.051±1.134	-5.645±0.508	-2.968±2.091
1	Varus	-8.269±1.662	-8.905±1.551	-5.497±1.321	-5.104±1.530	-3.533±0.980	-6.183±1.043
2	Varus	-0.150±0.191	-0.838±0.311	-0.530±0.166	0.038±0.314	-0.289±0.356	1.149±0.449
4	Varus	7.713±0.364	8.632±0.601	8.063±1.147	10.023±1.320	6.101±0.404	7.101±0.518
8	Varus	-4.421±0.535	-4.481±1.122	-3.575±1.152	-4.332±0.882	-3.021±0.718	-4.515±1.093
13	Varus	-0.419±1.425	-1.414±1.767	-3.469±0.730	-2.347±1.181	-0.589±0.907	-2.489±0.630
18	Varus	5.124±1.367	4.113±0.973	4.221±0.743	5.527±0.793	4.394±0.665	4.958±0.653
20	Varus	-1.726±0.812	-0.946±0.535	-0.446±0.556	-0.312±0.980	1.745±0.654	-0.260±0.379
22	Varus	-4.681±1.300	-7.247±0.692	-5.591±0.964	-3.353±0.989	-2.521±0.439	-3.955±0.507
25	Varus	0.975±0.639	-0.549±0.288	0.269±0.595	-0.371±1.001	0.542±2.176	0.767±0.822
26	Varus	-0.790±1.778	-0.174±1.523	1.697±0.811	1.539±0.676	2.147±0.864	1.414±1.639
29	Varus	3.391±1.571	4.445±0.792	3.172±3.048	2.212±1.107	4.889±1.602	1.896±2.394
Mean±SD	Neutral	-5.000±5.146	-5.150±5.542	-5.567±5.421	-4.733±4.960	-5.031±5.323	-4.832±5.910
Mean±SD	Varus	-0.296±4.592	-0.669±5.119	-0.153±4.270	0.320±4.460	0.897±3.285	-0.011±4.038

Table 17. Peak knee abduction ROM (°).

Subject	Group	Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	-14.006±1.049	-13.411±0.699	-15.623±0.608	-15.206±0.969	-14.265±0.608	-14.054±0.716
9	Neutral	-5.698±0.651	-5.807±0.907	-6.458±1.655	-6.631±0.583	-5.334±0.692	-5.253±1.124
10	Neutral	-11.097±0.667	-10.275±0.912	-11.620±0.632	-11.495±0.564	-12.334±0.613	-12.501±0.558
11	Neutral	-4.590±1.188	-7.998±0.628	-7.410±1.771	-4.504±1.035	-7.284±0.803	-6.557±1.419
15	Neutral	-0.299±0.395	-2.110±1.859	-2.540±1.416	-2.401±1.343	-4.744±0.629	-5.648±1.219
16	Neutral	-7.931±0.889	-10.489±0.255	-11.280±1.386	-9.068±1.061	-10.153±1.475	-12.206±1.422
21	Neutral	-6.146±1.126	-8.816±1.705	-10.275±2.237	-8.590±2.622	-8.295±0.684	-9.276±0.704
23	Neutral	-11.512±1.442	-10.396±0.984	-9.542±1.296	-8.283±1.387	-8.964±0.581	-9.011±0.860
24	Neutral	-10.419±1.647	-11.665±0.844	-12.465±1.438	-10.775±0.634	-11.757±0.859	-12.020±0.683
27	Neutral	-5.502±0.531	-6.466±1.093	-7.511±0.366	-7.688±1.735	-5.952±0.451	-5.972±1.110
30	Neutral	-19.088±0.544	-15.950±1.559	-17.716±0.669	-15.299±0.979	-17.866±0.417	-15.968±1.639
1	Varus	-8.743±2.142	-7.599±1.604	-4.813±1.458	-9.427±1.149	-7.470±1.668	-7.093±1.612
2	Varus	-11.901±0.620	-11.325±0.548	-12.170±0.768	-11.627±0.476	-10.545±0.395	-10.288±0.698
4	Varus	-4.719±0.578	-4.935±1.978	-6.160±2.314	-4.421±1.563	-4.812±0.735	-3.099±0.901
8	Varus	-12.169±1.066	-14.777±1.827	-13.627±0.873	-12.627±0.557	-12.471±0.783	-15.243±1.401
13	Varus	-15.175±2.273	-10.184±2.007	-12.685±1.197	-10.902±0.808	-10.241±0.546	-11.864±0.500
18	Varus	-8.245±0.989	-7.021±1.982	-9.829±1.186	-7.758±1.068	-6.246±1.262	-8.492±3.327
20	Varus	-10.545±1.483	-10.102±1.092	-11.909±1.625	-12.214±2.079	-10.397±1.588	-12.395±0.778
22	Varus	-13.034±5.326	-20.122±1.552	-14.993±1.201	-15.959±1.700	-16.655±1.228	-17.182±0.477
25	Varus	0.004±1.276	-0.259±0.646	0.239±0.584	-3.133±0.706	-3.543±2.035	-1.815±1.720
26	Varus	-8.648±1.461	-7.750±1.620	-10.028±1.323	-11.607±1.673	-11.284±1.839	-9.621±1.231
29	Varus	-5.633±1.347	-8.194±0.915	-8.357±3.192	-9.215±2.048	-6.886±1.129	-9.176±1.808
Mean±SD	Neutral	-8.753±5.162	-9.398±3.795	-10.222±4.270	-9.085±3.991	-9.723±4.061	-9.860±3.716
Mean±SD	Varus	-8.982±4.322	-9.297±5.149	-9.485±4.472	-9.899±3.708	-9.141±3.785	-9.661±4.618

Table 18. Peak ankle eversion angle (°).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	-0.152±1.105	2.385±0.743	0.017±0.834	-0.673±0.471	-0.886±0.745	1.234±0.454
9	Neutral	-3.600±0.509	-3.159±0.583	-3.318±0.797	-1.031±0.487	-1.957±0.677	-3.249±0.610
10	Neutral	-4.540±0.500	-2.228±0.453	-3.349±0.289	-4.597±0.755	-4.076±0.440	-4.544±0.557
11	Neutral	-0.907±1.306	-1.251±1.124	5.254±1.362	-1.640±0.806	-0.655±0.540	1.171±1.439
15	Neutral	-3.373±0.799	-1.413±1.942	-0.030±0.922	1.262±0.917	0.524±0.875	2.041±1.076
16	Neutral	-2.163±0.588	-2.153±1.259	-3.087±0.538	-3.377±0.645	-3.562±0.350	-4.708±0.287
21	Neutral	-0.789±1.013	-0.291±0.233	-2.297±0.363	-0.898±1.419	-2.076±0.887	-2.391±0.622
23	Neutral	-0.600±1.560	0.071±1.319	3.676±0.538	1.526±0.878	1.142±0.766	2.723±1.730
24	Neutral	0.694±1.132	2.039±0.854	2.033±0.678	0.162±2.010	0.779±0.608	0.438±0.846
27	Neutral	-14.318±1.697	-12.144±1.036	-13.939±0.812	-13.339±1.002	-4.403±0.449	-9.325±1.291
30	Neutral	-1.543±0.582	-1.523±1.755	3.352±0.909	-4.101±0.755	-0.136±0.670	0.956±0.670
1	Varus	-20.362±3.117	-17.312±0.570	-15.342±1.653	-17.561±2.192	-21.425±1.124	-15.521±1.938
2	Varus	-3.027±0.396	-3.437±0.395	-4.530±0.446	-2.628±0.527	-3.869±0.834	-1.715±0.413
4	Varus	1.503±0.603	-0.953±1.132	0.674±2.560	-0.981±1.294	1.457±0.428	2.055±0.771
8	Varus	6.519±0.525	5.796±0.693	4.745±0.224	6.927±0.381	5.811±0.604	5.781±0.390
13	Varus	6.016±1.329	6.127±0.704	7.719±1.888	6.955±0.177	8.545±0.752	11.065±0.916
18	Varus	-4.144±0.684	-3.633±0.674	-2.629±1.259	-4.064±1.179	-5.453±1.028	-4.407±0.847
20	Varus	4.099±2.800	3.572±0.538	1.510±0.655	1.918±0.780	2.084±1.500	0.506±0.567
22	Varus	-6.681±0.536	-5.174±0.417	-6.530±1.325	-7.104±0.820	-6.915±1.097	-5.226±0.742
25	Varus	-0.652±0.417	-1.797±0.086	-1.128±0.957	-2.011±0.930	-2.353±0.843	-2.568±1.103
26	Varus	0.806±1.031	0.460±1.559	-4.183±0.555	-3.612±0.887	-2.827±2.507	-0.144±2.140
29	Varus	-2.725±2.272	-6.030±1.116	-9.604±3.460	-3.746±1.622	-6.653±4.206	-4.863±1.662
Mean±SD	Neutral	-2.845±4.127	-1.788±3.843	-1.063±5.248	-2.428±4.140	-1.392±1.973	-1.423±3.733
Mean±SD	Varus	-1.695±7.485	-2.035±6.562	-2.663±6.506	-2.355±6.733	-2.872±7.949	-1.367±6.787

Table 19. Peak ankle inversion ROM (°).

Subject	Group	Without Toe-cage			With Toe-cage		
		0.5 kg	1.0 kg	1.5 kg	0.5 kg	1.0 kg	1.5 kg
7	Neutral	-0.341±0.764	0.812±0.429	1.517±0.792	0.237±0.260	1.281±0.717	1.013±0.730
9	Neutral	1.890±0.806	2.488±0.798	2.853±0.683	2.983±0.657	2.559±0.530	3.215±0.894
10	Neutral	2.787±0.360	1.915±0.534	3.131±0.704	1.595±0.448	2.028±0.813	3.322±0.410
11	Neutral	4.038±1.391	3.523±0.963	5.862±1.835	5.433±0.502	4.673±1.045	8.268±1.372
15	Neutral	7.180±1.701	8.059±2.725	5.754±2.375	2.345±1.691	7.676±3.837	8.018±1.107
16	Neutral	4.744±0.578	4.839±1.908	2.746±1.268	1.488±0.759	2.466±0.940	2.682±0.862
21	Neutral	3.340±1.019	6.120±0.559	4.861±0.140	4.666±1.172	3.461±0.289	4.102±0.239
23	Neutral	0.646±0.628	2.407±1.362	4.311±0.512	1.739±1.083	2.811±0.727	3.309±1.394
24	Neutral	4.320±0.857	3.737±1.440	3.332±0.956	2.480±1.152	3.237±0.698	4.089±0.343
27	Neutral	2.523±2.072	3.067±1.192	2.148±0.921	1.891±0.717	1.485±0.527	2.783±0.754
30	Neutral	3.926±0.573	6.391±1.909	7.697±1.102	4.259±1.236	7.150±0.800	5.809±0.563
1	Varus	4.045±3.438	4.210±0.771	5.257±1.730	7.845±1.828	6.294±1.400	7.205±2.661
2	Varus	3.888±0.449	4.544±0.445	4.630±1.450	3.071±1.017	4.907±0.490	4.708±0.651
4	Varus	4.762±1.062	3.230±0.128	4.749±1.733	5.820±1.538	4.065±0.764	3.086±0.950
8	Varus	0.652±0.275	2.530±0.739	1.354±0.385	1.710±0.801	1.567±0.375	1.987±0.512
13	Varus	1.786±1.175	2.182±0.831	3.932±2.196	1.447±0.238	2.842±0.225	2.875±0.661
18	Varus	6.458±2.247	5.409±1.150	10.236±1.582	5.551±0.936	4.280±1.743	3.382±3.038
20	Varus	0.859±2.141	1.739±0.405	3.589±0.805	2.891±0.529	4.113±1.847	1.598±0.679
22	Varus	4.834±1.162	8.845±1.048	5.405±2.183	2.423±1.054	2.181±1.336	2.707±0.546
25	Varus	2.307±0.478	2.143±0.832	2.154±0.988	2.375±0.691	3.155±0.626	2.450±0.796
26	Varus	4.435±1.138	2.212±1.725	1.295±1.527	3.844±1.292	2.020±1.526	3.017±1.157
29	Varus	2.983±2.778	2.813±1.163	3.961±3.657	4.432±2.181	5.085±2.986	5.321±2.082
Mean±SD	Neutral	3.187±2.053	3.942±2.187	4.019±1.864	2.647±1.560	3.530±2.140	4.237±2.252
Mean±SD	Varus	3.364±1.815	3.623±2.085	4.233±2.464	3.764±1.977	3.683±1.465	3.485±1.642

Table 20. Peak foot eversion angle (°).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	-9.678±0.313	-9.050±0.848	-7.897±0.617	-10.401±0.168	-8.023±0.203	-7.253±0.170
9	Neutral	-5.242±0.407	-5.030±0.563	-6.362±0.139	-6.185±0.416	-6.966±0.000	-6.547±0.341
10	Neutral	4.093±0.202	3.252±0.499	1.120±0.219	-1.521±0.072	-3.783±0.185	-3.503±0.171
11	Neutral	-8.807±0.410	-1.603±0.590	4.155±0.683	-10.581±0.381	-4.872±1.737	-4.059±0.963
15	Neutral	-5.169±1.547	-5.360±1.602	-9.336±1.950	-11.116±1.052	-11.010±0.403	-2.561±1.240
16	Neutral	-1.921±0.336	-0.798±0.895	-7.123±3.159	-8.536±2.408	-4.038±1.806	-6.739±0.932
21	Neutral	-2.908±1.541	0.776±1.827	4.057±1.201	-5.331±0.392	-5.466±0.537	-4.271±0.659
23	Neutral	-6.301±0.443	-4.979±0.343	-2.962±1.531	-5.679±0.495	-6.242±0.394	-4.646±0.471
24	Neutral	3.126±0.773	4.939±0.749	4.593±0.372	-0.268±2.137	-0.252±0.433	2.828±0.113
27	Neutral	-5.185±0.174	-3.172±0.151	-4.338±0.206	-6.139±0.449	0.018±0.507	-4.750±0.286
30	Neutral	-13.773±0.719	-11.176±0.406	-4.462±1.524	-12.497±0.452	-10.245±0.891	-10.205±0.670
1	Varus	-7.660±0.878	-4.636±0.688	-4.614±0.407	-7.190±0.176	-10.880±0.205	-9.215±0.599
2	Varus	-9.523±0.246	-9.571±0.082	-11.520±0.114	-9.609±0.151	-10.241±0.321	-10.382±0.181
4	Varus	-3.465±1.759	-5.590±0.431	-2.605±0.459	-3.871±0.327	-5.942±0.551	-4.326±0.349
8	Varus	3.385±0.511	1.847±1.748	4.267±0.550	0.642±0.774	0.518±0.786	0.056±0.589
13	Varus	-1.141±0.843	-1.843±0.962	1.040±2.549	-2.459±0.180	0.371±0.285	1.682±2.345
18	Varus	3.157±1.629	2.811±0.812	8.070±0.609	0.565±0.310	0.378±0.408	0.690±0.835
20	Varus	0.360±3.927	-4.592±2.197	-6.092±0.821	-8.222±0.585	-7.179±0.927	-11.331±0.357
22	Varus	-4.776±1.045	1.688±0.589	-1.810±1.067	-5.349±0.589	-1.719±1.901	-1.700±1.005
25	Varus	-5.726±0.569	-3.117±0.715	-5.236±1.149	-8.061±0.391	-8.376±0.411	-8.045±0.190
26	Varus	-3.959±0.911	-4.542±0.593	-5.463±0.241	-8.866±0.235	-7.091±0.921	-3.799±0.497
29	Varus	-5.079±1.320	-3.269±1.035	-5.994±0.810	-7.430±0.338	-8.663±0.334	-6.297±0.643
Mean±SD	Neutral	-4.706±5.261	-2.927±4.901	-2.596±5.198	-7.114±3.937	-5.534±3.541	-4.701±3.281
Mean±SD	Varus	-3.130±4.166	-2.801±3.711	-2.723±5.451	-5.441±3.676	-5.347±4.410	-4.788±4.611

Table 21. Mean foot external rotation angle (°).

Subject	Group	Without Toe-cage 0.5 kg	Without Toe-cage 1.0 kg	Without Toe-cage 1.5 kg	With Toe-cage 0.5 kg	With Toe-cage 1.0 kg	With Toe-cage 1.5 kg
7	Neutral	-10.413±0.726	-10.702±0.356	-9.249±0.352	-10.877±0.173	-8.550±0.164	-7.794±0.240
9	Neutral	-6.015±0.267	-5.639±0.402	-6.456±0.124	-6.493±0.098	-6.367±0.307	-6.300±0.287
10	Neutral	2.387±0.260	2.193±0.549	0.056±0.268	-1.783±0.078	-4.272±0.221	-3.833±0.174
11	Neutral	-10.198±0.175	-3.883±1.305	0.538±0.480	-11.273±0.184	-7.210±0.925	-4.477±0.953
15	Neutral	-8.482±0.260	-9.149±0.895	-12.090±0.458	-13.518±0.258	-12.321±0.234	-8.254±0.790
16	Neutral	-6.275±0.295	-9.225±0.211	-11.008±0.545	-11.521±0.507	-10.032±0.263	-11.014±0.189
21	Neutral	-3.961±0.206	-1.464±0.703	1.570±0.617	-5.260±0.290	-6.239±0.415	-5.787±0.280
23	Neutral	-7.152±0.334	-6.711±0.699	-5.012±0.153	-6.169±0.139	-7.368±0.288	-5.808±0.493
24	Neutral	0.841±0.473	1.972±0.281	1.225±0.319	-1.380±1.559	-1.553±0.140	0.011±0.245
27	Neutral	-6.595±0.265	-4.737±0.155	-5.477±0.288	-6.961±0.400	-0.620±0.294	-6.181±0.116
30	Neutral	-14.736±0.265	-13.765±0.344	-8.573±0.410	-13.713±0.228	-11.892±0.380	-12.322±0.399
1	Varus	-7.846±0.549	-5.557±0.437	-5.946±0.473	-8.104±0.160	-11.215±0.068	-9.975±0.405
2	Varus	-10.344±0.133	-10.235±0.064	-12.192±0.069	-10.349±0.057	-11.106±0.172	-11.510±0.072
4	Varus	-3.671±1.029	-7.907±0.614	-4.506±1.580	-7.540±0.258	-5.783±0.262	-4.762±0.461
8	Varus	1.005±0.360	-0.290±0.372	0.906±0.219	-1.181±0.161	-1.604±0.161	-1.840±0.178
13	Varus	-2.579±0.510	-2.772±0.531	-1.355±0.855	-3.187±0.147	-0.511±0.289	0.325±1.188
18	Varus	1.366±1.087	0.588±0.533	5.575±0.689	-0.416±0.254	-1.577±0.389	-1.218±0.379
20	Varus	-3.115±2.875	-9.791±1.000	-10.175±0.354	-9.788±0.356	-10.906±0.280	-12.648±0.301
22	Varus	-5.660±0.656	-4.641±1.603	-5.864±1.491	-7.770±0.303	-4.660±0.667	-5.373±0.466
25	Varus	-6.650±0.323	-3.656±0.468	-5.581±0.198	-8.591±0.208	-8.620±0.329	-8.316±0.113
26	Varus	-5.175±1.411	-5.480±0.549	-6.432±0.171	-9.391±0.384	-8.025±0.550	-4.672±0.477
29	Varus	-5.981±0.578	-3.866±0.861	-6.507±0.748	-7.435±0.775	-8.389±0.453	-6.979±0.284
Mean±SD	Neutral	-6.418±4.906	-5.555±5.084	-4.952±5.076	-8.086±4.371	-6.948±3.779	-6.523±3.373
Mean±SD	Varus	-4.423±3.534	-4.873±3.475	-4.734±4.942	-6.705±3.469	-6.581±4.014	-6.088±4.248

VITA

Guangping (Steven) Shen was born in Fushun, China, to the parents of Zaishi Shen and Shengqin Jin. He grew up and attended elementary through high school in his hometown. After he graduated from Fushun Korean Chinese No. 1 high school, he went to Beijing Sport University, and received his Bachelor of Science degree in Kinesiology with a concentration in athletic training. After his graduation, he got accepted by the University of Tennessee and started to pursue a master's degree in exercise science with a concentration in Biomechanics. He graduated with a Master of Science degree in 2015.