EFFECTS OF FOOT PLACEMENT ON RESULTANT JOINT MOMENTS IN THE LOWER EXTREMITY JOINTS DURING THE SQUAT

A DISSERTATION

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BY

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To the Dean of the Graduate School:

I am submitting herewith a dissertation written by Sangwoo Lee entitled "Effects of Foot Placement on Resultant Joint Moments in the Lower Extremity Joints during the Squat." I have examined this dissertation for form and content and recommend that it be accepted in partial fulfillment of the requirements for the degree of Doctor of Philosophy with a major in Kinesiology.

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DEDICATION

I dedicate this dissertation to my family for their never-ending love and blessings.

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ABSTRACT

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EFFECTS OF FOOT PLACEMENT ON RESULTANT JOINT MOMENTS IN THE LOWER EXTREMITY JOINTS DURING THE SQUAT

MAY 2015

The purpose of this study was to investigate the effects of foot placement on normalized resultant joint moments of the lower extremity joints in sagittal and frontal planes during the squat. A total of 42 skilled participants (21 male and 21 female) performed squat trials (five trials per condition) with a weighted barbell of 75% of 1 repetition maximum in six different conditions: 3 stance widths (narrow, medium, and wide) × 2 toe angles (toe forward and toe out). Normalized resultant joint moments of the lower extremities were extracted from each trial using three-dimensional motion analysis. A three-way (3 × 2 x 2) mixed designs MANOVA was conducted to compare the moments with ‘stance width’ (within), ‘toe angle’ (within), and ‘gender’ (between) being factors. Ensemble-average normalized resultant joint moment patterns were also analyzed to identify the changes in the moment patterns. There was a significant interaction between stance width and toe angle in the normalized sagittal-plane resultant joint moments ($p = .017$). The toe out condition showed significantly larger hip extensor moment (HEX) and knee extensor moment (KEX) in the medium and wide stances and significantly larger ankle plantar-flexor moment (AP) in the narrow stance than the toe forward condition. Significant differences in HEX, KEX, and AP among the stance widths were observed during the toe out condition. Significant main effects of the factors (width: $p < .001$; angle: $p < .001$) were observed in the normalized frontal-plane resultant joint moments. In terms of stance width, the largest hip abductor moment (HAB), knee abductor moment (KAB), and ankle everter moment (AE) were produced in the narrow stance, followed by the medium stance, and the wide stance revealed the smallest values. In terms of toe angle, the toe out condition was characterized by significantly larger KAB and AE but smaller HAB than the toe forward condition. The maximum values of HEX, KEX, and AP were observed at the beginning of the upward phase. The maximum values of HAB and KAB occurred approximately in the middle of the downward and upward phases while the smallest HAB and KAB occurred approximately at the bottom of squat. The peak AE was observed around the bottom of squat. Based on the results of this study, an optimal foot placement during the squat would be toes out with wide stance condition.

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CHAPTER I

INTRODUCTION

The squat is one of the most popular exercises in the field of strength and conditioning due to its wide range of applications. It is often included as a core exercise in many workout routines designed to improve athletic performance (Escamilla et al., 2001; Schoenfeld, 2010). It is also an essential component of weightlifting as well as powerlifting competitions (Escamilla, Fleisig, Lowry, Barrentine, & Andrews, 2001). Since a squatting motion is often involved in many daily activities, such as sit-to-stand, lifting boxes, and picking up children, the squat is also considered one of the best exercises for enhancing quality of life (Schoenfeld, 2010). In the clinical setting, it is often used as a means to strengthen lower extremity muscles and connective tissues after joint-related injuries, such as anterior cruciate ligament (ACL) injuries (Stuart, Meglan, Lutz, Growney, & An, 1996; Tagesson, Oberg, Good, & Kvist, 2008; Yack, Collins, & Whieldon, 1993).

In spite of its benefits, the squat also has injury concerns. Poor technique or improper exercise prescription can lead to conditions/injuries such as patellofemoral pain (PFP), ACL tear, menisci/articular cartilage injury, ruptured intervertebral discs, and spondylolysis (Cappozzo, Felici, Figura, & Gazzani, 1985; Escamilla, 2001; Escamilla et al., 2001; Fry, Smith, & Schilling, 2003; Miller, Sedory, & Croce, 1997; Vakos, Nitz, Threlkeld, Shapiro, & Horn, 1994; Yack et al., 1993). Most of the squat-related injury studies have been primarily focused on the lower extremity and lower back due to their vulnerability to injury during the squat. For example, biomechanical (both kinematic and kinetic) risk factors on the lower extremity and lower back during squat exercises have been investigated in terms of the effect of poor squat technique and improper squat exercise prescription on the stress to the lower extremity and lower back (Escamilla, 2001; Escamilla et al., 2001; Lee, Moon, & Eun, 2011; Vakos et al., 1994). Electromyographic (EMG) risk factors, such as activation ratios of vastus lateralis (VL) to vastus medialis oblique (VMO) and biceps femoris (BF) to rectus femoris (RF), have also been examined with respect to their relationship with poor squat technique (Lim et al., 2009; Peng, Kernozek, & Song, 2013; Powers, 2000; Souza & Gross, 1991).

The lower extremity joints, including the hip, knee, and ankle joints, biomechanically function as a linked chain during the squat so it is likely that the positions of each of the joints affect loads on the other joints. For example, a squat technique for reducing loads on the knee, such as restricted forward movement of the knee, can place more loads on the hip and lower back (Escamilla, 2001; Fry et al., 2003; Klein, 1961; Schoenfeld, 2010; Swinton, Lloyd, Keogh, Agouris, & Stewart, 2012). While restricted forward movement of the knee serves to reduce loads on the knee, it results in a greater backward movement of the hip and more anterior lean of the trunk, placing greater loads on these joints. If the knee is permitted to move slightly past the toe during the squat, a more appropriate distribution of the loads on the lower extremity joints and lower back may be achieved. In order to properly perform the squat, therefore, interactions among the lower extremity joints need to be carefully considered.

Foot placement has been a key determinant of the safety of the squat. The squat can be classified into different styles based on stance width (narrow, medium, and wide) and toe angle (forward and out). Several studies have focused on the effects of foot placement on the kinetics, kinematics, and muscle activity in the lower extremity joints (Almosnino, Kingston, & Graham, 2013; Escamilla et al., 2001b; McCaw & Melrose, 1999; Miller et al., 1997; Ninos, Irrgang, Burdett, & Weiss, 1997). Despite potential interactions among the hip, knee, and ankle joints in response to the change in foot placement during the squat, only the kinetics and kinematics of the knee have thus far been analyzed in the previous studies. There were two studies analyzing resultant joint moments of the hip, knee, and ankle among various foot placements during the squat (Escamilla et al., 2001a; Swinton et al., 2012). However, the potential interactions among different stance widths and toe angles were not considered in these studies, which only assessed the effects of different stance widths. Moreover, all of the previous studies on the effects of various foot placements during the squat were conducted using only one gender (i.e., male). Therefore, more in-depth scientific investigation on the effects of foot position (stance widths and toe angles) on lower extremity kinetics during the squat and its potential interaction to gender need to be verified.

Purpose of the Study

To investigate the effects of foot placement (stance width and foot angle) on kinetic factors (normalized resultant joint moments in the sagittal and frontal planes) in the lower extremity joints during the squat using three-dimensional (3D) motion analysis.

Research Hypotheses

The peak normalized lower extremity resultant joint moments would differ among stance width conditions and toe angle conditions across both genders.

Significance of the Study

The squat has been regarded as an integral part of various strength and conditioning programs and physical rehabilitation prescriptions. Poor squat technique and inappropriate squat exercise prescription, however, may result in detrimental effects on the lower extremity joints and lower back. The safety of the squat has been one of the biggest concerns and a controversy concerning the proper squat technique still exists.

Foot placement is a key determinant of the safety of the knee joint during the squat. However, the optimum squat foot placement is still unclear among practitioners and researchers. Using three-dimensional motion analysis, this study investigated how the variations in foot placement during the squat affect biomechanical risk factors in the lower extremity joints. The results of this study would help practitioners and researchers better understand the lower extremity injury mechanism during the squat and provide a significant way for optimizing squat performance, while at the same time minimizing the likelihood of lower extremity injury.

Assumptions

1. The body is a linked segment system with frictionless pin joints connecting the segments.

2. The mass, length, and principal moments of inertia of each segment about its center of mass are constant during the squat motion.

3. The weighted bar is fixed to the upper body.

4. The squat motion is symmetrical between the two sides of the body.

5. There is no change in ability to perform the squat between the day of 1RM testing and the day of actual squat testing.

Delimitations

1. Participants performed the squat barefoot to minimize the effects of heel height and shoe cushioning (Ninos et al., 1997).

2. Participants were required to gaze straight ahead while performing the squat to minimize the effects of direction of gaze (Donnelly, Berg, & Fiske, 2006).

3. Participants were required to perform a weighted squat by positioning the thighs parallel to the floor.

4. Participants were not allowed to use a weight belt due to its effects on biomechanical factors (Zink, Whiting, Vincent, & McLaine, 2001).

5. Participants were not allowed to bounce during the transition from descent to ascent (McCaw & Melrose, 1999).

6. Participants were asked to refrain from an overuse of lower extremity, such as lifting a heavy box, for 48 hours prior to the data collection.

7. Participants were asked to place the weighted bar in the same position (top of trapezius on the upper back) for each session.

8. Participants were asked to have a comfortable hold on the weighted bar.

9. Participants performed the squat with their preferred tempo.

10. Testing was conducted in a laboratory setting.

Limitations

Despite the required prior squat experience of at least 3 years, techniques varied with each participant because of their preferred squat exercise style.

Definition of Terms

Global reference frame: the laboratory coordinate system in which body marker coordinates are calculated.

Ground reaction force (GRF): the force exerted by the ground in reaction to a force applied to the ground.

Inverse dynamics: an indirect method of determining resultant joint forces and moments based on kinematics, inertial parameters, and ground reaction forces and moments.

Joint coordinate system: a coordinate system that provides a simple geometric description of the three-dimensional rotational and translational motions between two rigid bodies using two local reference frames and a mutually orthogonal floating.

Kinematics: an area of mechanics that deals with the description of motion, including body and joint positions, angles, velocities, and accelerations.

Kinetics: an area of mechanics that examines the causes of motion, including resultant joint forces and moments.

Local reference frame: the reference frames attached to a segment and expressed relative to the global reference frame.

Orientation angles: the angles about the coordinate axes, formed by a segment with respect to its proximal segment.

Resultant joint moment: a mathematical concept that represents the net effect of all the structures, such as muscles, producing moments of force across a joint.

CHAPTER II

LITERATURE REVIEW

This chapter includes the following sections: description of squat motion, hip complex during the squat, knee complex during the squat, ankle complex during the squat, knee injury studies during the squat, bilateral asymmetry during the squat, and comparison of lower extremity biomechanics between genders.

Description of Squat Motion

The squat is performed by placing a weight bar across the posterior deltoids at the base of Trapezius. The squat begins with the lifter in an upright position, knees and hips fully extended. In general, the squat motion has been often divided into two phases: the descent and ascent phases (Figure 1). During the descent phase, the lifter simultaneously flexes at the hip and knee joints and dorsiflexes the ankle until the desired squat depth is achieved (i.e. the thighs are parallel to the floor). The lifter then reverses direction and ascends back to the upright position. In the ascent phase, the lifter extends the hip and knee joints and plantar flexes the ankle joint to return to the upright position (McCaw & Melrose, 1999; Schoenfeld, 2010).

The squat motion recruits most of the lower extremity musculature, including the quadriceps, hamstrings, hip extensors, hip adductors, hip abductors, and triceps surae. In addition to the lower extremity musculature, significant isometric activities of the upper extremity are required by wide range of supporting muscles, including the abdominals, erector spine, trapezius, rhomboids, and many others to facilitate postural stabilization of the trunk.

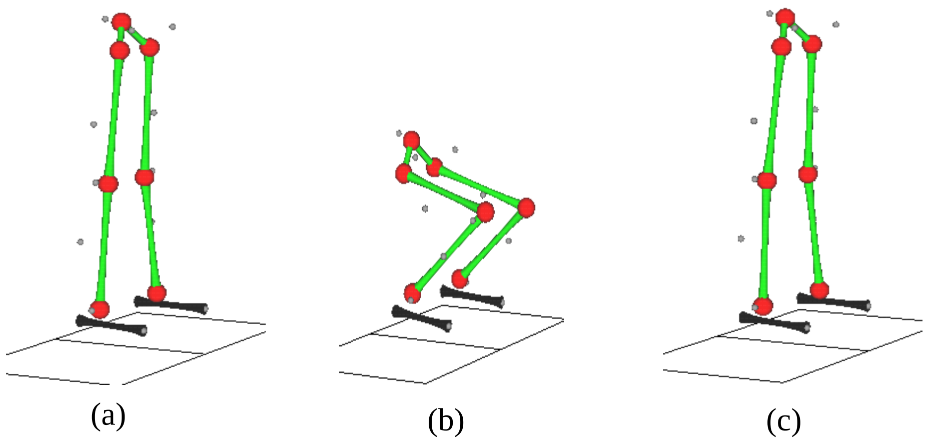


Figure 1: Typical events and phases setting during the squat. (a): starting position (upright position), (b): parallel position (when the thighs are parallel to the floor), and (c): ending position (upright position). The descent and ascent phases are defined between (a) and (b) and between (b) and (c), respectively.

Hip Complex during the Squat

Structure of the Hip

The hip joint is a ball and socket joint. The ball is the head of the femur and socket is the concave acetabulum. Joint cartilage covers both articulating surfaces for the stability of the hip joint. Outside the joint capsule, there are three major ligaments which also contribute to the stability of the hip, including the iliofemoral, pubofemoral, and ischiofemoral ligaments. Inside the joint capsule, the ligamentum teres provides a direct attachment from the rim of the acetabulum to the head of the femur. The iliofemoral and pubofemoral ligaments act to strengthen the joint capsule anteriorly, whereas the ischiofemoral ligament acts to reinforce the joint capsule posteriorly. Several bursae also exist in the hip joint to provide the hip with lubrication. The iliopsoas bursa and deep trochanteric bursa are the major bursae in the hip. The iliopsoas bursa is located between the iliopsoas and articular capsule and act to reduce the friction between these structures. The deep trochanteric bursa is positioned between the greater trochanter of the femur and gluteus maximus at the site of its attachment to the iliotibial and provides a cushion (Hall, 2011).

Movements at the Hip and Muscles Corresponding to the Movements

Flexion/Extension

Six muscles primarily responsible for the hip flexion are those muscles crossing the hip joint anteriorly; the iliacus, psoas major, pectineus, rectus femoris, Sartorius, and tensor fascia latae. The iliacus and psoas major are the major hip flexors. The rectus femoris is a two-joint muscle active during both hip flexion and knee extension. The Sartorius is also a two-joint muscle and the longest muscle in the human body. The hip extensors are the gluteus maximus and three hamstrings; biceps femoris, semitendinosus, and semimembranosus. The gluteus maximus usually acts only when the hip is flexed. The hamstrings are two-joint muscles active during both hip extension and knee flexion (Hall, 2011).

Abduction/Adduction

The gluteus medius is the major hip abductor with the gluteus minimus assisting. The hip adductors are those muscles crossing the hip joint medially, including the adductor longus, adductor brevis, adductor magnus, and gracilis. The gracilis also contributes to the knee flexion and the other three adductor muscles also contribute hip flexion and lateral rotation.

Medial (Internal)/Lateral (External) Rotation

Six muscles function only as lateral rotators at the hip: piriformis, gemellus superior, gemellus inferior, obturator internus, obturator externus, and quadratus femoris. The major medial rotator of the femur is the gluteus minimus with the tensor fascia latae, semitendinosus, semimembranosus, and gluteus medius assisting.

Horizontal Abduction/Adduction

Horizontal abduction and adduction of the femur occur when the hip is flexed at 90° while the femur is either abducted or adducted. The coordinated actions of several muscles are required for these actions. For example, the hip flexors need to be produced to elevate the femur. The hip abductors can then produce horizontal abduction and, from a horizontally abducted position, the hip adductors can produce horizontal adduction. The muscles located on the posterior portion of the hip are more effective as horizontal abductors and adductors than those on the anterior portion because the former muscles are stretched when the femur is flexed at 90°, whereas tension in the later muscles is reduced when the femur is in this position.

Knee Complex during the Squat

Structure of the Knee

The knee is a synovial joint, including three articulations in the joint capsule. The two articulations of the tibiofemoral joint act as weight-bearing joints with the third articulation being the patellofemoral joint. The medial and lateral condyles of the tibia and femur articulate to form two condyloid joints. These joints function together as a hinge joint with slight lateral and rotational movements allowed. The tibiofemoral joint is a principal joint at the knee. This joint is formed by a femur (convex femoral condyles) as a proximal segment and a tibia (concave superior surface of the tibial plateau) as a distal segment. It carries out movements that occur in sagittal plane throughout a range of motion of 0° to approximately 150° of knee flexion, such as knee flexion/extension with posterior/anterior glide of the tibial plateau on the femoral condyles (Li et al., 2004). In addition, a small amount of axial rotation also occurs at the tibiofemoral joint while performing downward motion, such as the squat, with the femur rotating externally during flexion and internally during extension with respect to the tibia.

The patellofemoral joint is also one of the knee joints at the knee complex. This joint is formed by the femur as a proximal segment and a patella as a distal segment. The patella is the largest sesamoid bone in the body and embedded in quadriceps tendon. Its primary functions are to increase mechanical leverage of the quadriceps and to minimize the friction between femur and quadriceps tendon. This joint can be classified as a gliding joint in which the patella slides over the trochlear surfaces of the femur during flexion and extension of the knee. This joint coexists with the tibiofemoral joint in the knee joint by providing additional mechanical leverage in extension due to a greater force arm, as well as reducing wear on the quadriceps and patellar tendons from friction against the intercondylar groove (Schoenfeld, 2010).

Many ligaments cross the knee joint and they serve as main static stabilizers of the knee joint during the squat. The knee is supported by an array of ligaments and cartilage. There are four major ligaments at the knee complex: anterior cruciate ligament (ACL), posterior cruciate ligament (PCL), lateral collateral ligament (LCL), and medial collateral ligament (MCL). The ACL is often considered as the most important stabilizer of the knee joint among the ligaments and cartilage due to its primary role. Its primary role is to prevent anterior tibial translation at the knee, particularly at lower flexion angles (Yack et al., 1993). It also plays an important role in limiting internal/external rotation of the knee and deterring varus/valgus movement. The PCL serves as a counterpart to the ACL. Its primary role is to restrain posterior tibial translation at knee flexion angles larger than 30° (Fox, Harner, Sakane, Carlin, & Woo, 1998; Li et al., 2004). The LCL and MCL act as stabilizers of the knee during the squat movements that occur in the frontal plane by helping to provide resistance to knee abduction/adduction torques (Schoenfeld, 2010). The LCL goes from the top portion of the fibula (the bone on the outside of the lower leg) to the lower and outside portion of the femur and helps keep the outer side of the knee joint stable. The MCL goes from the inside portion of the tibia to the inner surface of the femur and helps keep the inside portion of the knee joint stable.

The menisci are cartilaginous discs located between the tibial and femoral condyles. They are also joined together by the transverse ligament at the knee. The menisci distribute the load at the knee over a larger surface area by deepening the articulating depressions of the tibial plateaus and also help absorb shock at the knee. Many bursae are located in and around the knee joint capsule to reduce friction during the knee motions. The suprapatellar bursa located between the femur and quadriceps femoris tendon is the largest bursa in the body. The subpopliteal bursa located between the lateral condyle of the femur and popliteal muscle and the semimembranosus bursa located between the medial head of the gastrocnemius and semimembranosus tendons also play an important role in the reduction of friction in the knee joint.

Movements at the Knee and Muscles Corresponding to the Movements

Flexion/Extension

Knee musculature serves as dynamic joint stabilizers of the knee joint during the squat whereas the ligaments are main static stabilizers of the knee joint (Schoenfeld, 2010). Flexion and extension are the major movements occurring at the tibiofemoral joint. Three hamstring muscles primarily responsible for the knee flexion are biceps femoris, semitendinosus, and semimembranosus. As discussed above, the hamstrings are two-joint muscles active during both hip extension and knee flexion. Muscles including the gracilis, Sartorius, popliteus, and gastrocnemius assist with the knee flexion. The hamstrings muscles technically act as antagonists of the quadriceps muscles, opposing knee extensor moments (Schoenfeld, 2010). The hamstrings produce a counter-regulatory force on the tibia by pulling it posteriorly, thereby attenuating anterior tibial translation. The hamstrings also help to neutralize the anterior tibiofemoral shear caused by the quadriceps pulling and thus alleviating stress on the ACL (Escamilla, 2001). Several studies have shown that greater hamstrings activity was observed during the ascent phase as compared with the descent phase of the squat (Escamilla et al., 1998; Lee et al, 2011; Wilk et al., 1996). The studies reported that the peak activity occurred near 70° of knee flexion during the ascent phase. The hamstrings do not change length too much throughout the squat because the hamstrings are biarticular muscles crossing two joints so that they work nearly isometrically during the descent and ascent phases (Escamilla, 2001). The hamstrings are concurrently shortening at the knee and lengthening at the hip during the descent phase, and lengthening at the knee and shortening at the hip during the ascent phase.

Four quadriceps muscles primarily responsible for the knee extension are rectus femoris, vastus lateralis, vastus medialis, and vastus intermedius which carry out concentric contraction of knee extension as well as eccentric contraction of knee flexion. Muscle activities of the quadriceps are likely to peak at approximately 90° of knee flexion (Escamilla et al., 1998; Lee et al., 2011), remaining relatively constant thereafter. It could be suggested that squatting at knee flexion angles greater than 90° might not lead to enhancements in quadriceps development.

Abduction/Adduction

The primary contribution to knee abduction/adduction comes from co-contraction of the hamstrings and quadriceps with secondary contributions resulting from the gracilis and tensor fascia latae.

Medial (Internal)/Lateral (External) Rotation

The semimembranosus, semitendinosus, and popliteus are responsible for medial rotation of the tibia with the gracilis and Sartorius assisting and the biceps femoris is responsible for lateral rotation of the tibia.

Patellofemoral Joint Motion

During flexion and extension at the tibiofemoral joint, the patella glides inferiorly and superiorly against the distal end of the femur. The vastus lateralis is primarily responsible for pulling the patella laterally, whereas the vastus medialis oblique counteract the lateral pull of the vastus lateralis.

Ankle Complex during the Squat

Structure of the Ankle

The ankle consists of three joints including the distal tibiofibular, tibiotalar, and fibulotalar joints. The distal tibiofibular joint is a syndesmosis joint supported by the anterior and posterior tibiofibular ligaments as well as the crural interosseous tibiofibular ligament. Motion at the ankle joint mostly occurs at the tibiotalar joint, where the convex surface of the superior talus articulates with the concave surface of the distal tibia. Three ligaments, including the anterior and posterior talofibular and the calcaneofibular ligaments, reinforce the joint capsule laterally and the four bands of the deltoid ligament reinforce the joint capsule medially.

Movements at the Ankle and Muscles Corresponding to the Movements

Dorsiflexion/Plantarflexion

Although dorsiflexion and plantarflexion occur with slight external and internal rotations, respectively, it is believed that motion at the ankle occurs primarily in the sagittal plane with the ankle functioning as a hinge joint. The tibialis anterior, extensor digitorum longus, and peroneus tertius are the major dorsiflexors of the foot. The extensor halluces longus assists in dorsiflexion. Two joint muscles, including the gastrocnemius and soleus, are the major plantarflexors of the foot. The tibialis posterior, peroneus longus, peroneus brevis, plantaris, flexor halluces longus, and flexor digitorum logus assist in plantarflexion.

Knee Injury Studies during the Squat

Internal forces acting on the knee joint during the squat were estimated from modeling techniques and not measured directly. There have been several studies using the modeling techniques to identify the squat-related knee injuries (Table 1).

A mathematical model was developed to estimate internal forces of the knee during the squat (Zheng, Fleisig, Escamilla, & Barrentine, 1998). Tibiofemoral compressive force and forces acting on the quadriceps muscle and ACL/PCL ligaments were estimated using the mathematical model. They compared the calculated internal forces among different types of exercises (seated knee extension, leg press, and squat) using the mathematical model. Surface EMG data, MVIC (Maximum Voluntary Isometric Contraction) data, PCSA (physiological cross-sectional area), muscle fiber length, and maximum muscle force were used as components needed to estimate the internal forces in the model. There are several features and limitations in the model. The muscle fiber length was considered. The force-velocity relation was, however, not included in the model and the quadriceps muscle force was calculated as a single muscle. In addition, since the sagittal plane was only considered for the computation of the internal forces, the collateral ligaments were not included in the model. The participant's body was scaled using body weight only.

Another study conducted by Escamilla et al. (2009) estimated patellofemoral joint contact force during the wall squat and one-leg squat using a modified mathematical model (Table 1). The original mathematical model was modified by adding the force-velocity relation to the model with consideration for eccentric and concentric muscle actions in the study. The modified mathematical model enabled researchers to obtain more accurate and in-depth data as compared to the original mathematical model. However, there are limitations in the modified mathematical model regarding the computations of kinematic parameters because the model was developed with focus on the computations of the internal forces.

Table 1

A *Comparison of Knee Joint Models Used in the Calculation of Internal Forces (Joint, Muscle, and Ligament)*

Anatomical planes and variables	Sagittal plane only. Quadriceps muscle force and ACL/PCL ligament forces, and tibiofemoral compressive force were calculated.	Sagittal plane only. Patellofemoral compressive force and ACL/PCL ligament forces were computed.	Main sagittal plane, but, A/P and M/L translations, and I/E and var/val rotations are calculated using equations used in a previous study (Walker et al., 1988)
Main features	EMG-driven model (estimated using surface EMG data). The weight factor was used to minimize the errors between muscle force estimation (Dennis & Schnabel, 1983). The force-velocity relationship was NOT included. The quadriceps muscle force was calculated as a single muscle. The collateral ligaments were NOT included. The body was scaled using a body weight only.	Same features as the mathematical model except for the inclusion of the force-velocity relationship. Mainly used to calculate the patellofemoral compressive force and ligament forces (ACL & PCL). The accuracy of the data highly depends on the accuracy surface EMG and MVIC data.	Data obtained from all 3 anatomical axes. Accurate examination of the fiber length operating range in relation to the joint angle. Tendon lengths are set using an experimentally measured relationship between fiber length and joint angle. Unlike the mathematical methods, the body model was scaled using both mass and measurement data (length of segments) which enables us to create more accurate body model.
Accuracy	Least accurate due to the relatively small number of subjects used for the calculation of the muscle modeling parameters and methodological limitation.	More accurate than the mathematical model (Zheng et al., 1998) due to the improvement in the methodology.	Most accurate due to the more detailed methodology (e.g., scaling method and determination of the muscle geometry) Furthermore, the number of subjects in the dataset was also larger.
Comprehensiveness	Focused on the computations of internal forces (e.g., joint contact, muscle, and ligament forces) and limited to the computation of kinematic parameters		Highly comprehensive (e.g., kinematic, kinetic, and internal forces are computed)

(continued)

Author	Zheng et al., 1998	Escamilla et al., 2009, 2010, & 2012	Arnold et al., 2010
Model	A mathematical model	A modified mathematical model	A lower limb model used in OpenSim
Components	6 muscle-tendon compartments	6 muscle-tendon compartments	15 muscle-tendon compartments (7 segments, 5 joints, and 44 muscle-tendon compartments for the entire mode)
Muscle force generation	Cross-bridge model	A Hill-type muscle model was used to characterize muscle force generation (Zajac, 1989).	A Hill-type muscle model was used to characterize muscle force generation (Zajac, 1989).
Muscle modeling parameters	Muscle fiber length considering sarcomere length (Herzog et al., 1990), pennation angle, PCSA obtained from the direct measurement made in 3 cadaver subject (Wickiewicz et al., 1983), and maximum isometric force, surface EMG and MVIC data obtained from participants, and a weight factor	The mathematical model (Zheng et al., 1998) was modified by adding a force-velocity relationship to this model for eccentric and concentric muscle actions.	PCSA (cm), maximum isometric force (N), optimal fiber length considering sarcomere length (cm), tendon slack length (cm), pennation angle (°) obtained from the direct measurements made in 21 cadaver subjects (Ward et al., 2009)
Muscle-tendon geometry	Geometry of the lower extremity (Pierrynowski, 1991)	Geometry of the lower extremity (Pierrynowski, 1991)	Line segment paths constrained to origin and insertion points, via points, and wrapping surfaces (e.g., a cylinder wrapping surface in the knee joint considering the muscle-tendon geometry when the knee is fully flexed)
Degree of freedom	1	1	1
Number of the knee joint	2 (tibiofemoral and patellofemoral joints)	2 (tibiofemoral and patellofemoral joints)	2 (tibiofemoral and patellofemoral joints)

Note. MVIC and PCSA represent maximum voluntary isometric contraction and physiological cross-sectional area, respectively.

Bilateral Asymmetry during the Squat

The squat is often considered as a symmetrical exercise between the two lower limbs. According to previous squat studies, it has been assumed that there was no bilateral asymmetry in kinematics and kinetics between the two lower limbs during the squat (Almosnino et al,, 2013; Escamilla et al,, 2001; Lee et al., 2011; McCaw & Melrose, 1999; Miller et al., 1997; Ninos et al., 1997). In the previous studies, one side of the lower extremity was only selected for data analysis based on leg dominance or personal preference under an assumption that results obtained from the both sides of the lower extremity are identical. Moreover, one side was just selected and analyzed without describing clear justification in some of the previous studies. Although the squat is relatively a symmetrical movement, as a human body is originally designed to be asymmetric, unlike a machine, the bilateral asymmetry in biomechanics of the lower extremity must be present during the squat.

For a better understanding of the bilateral asymmetry during the squat, there have been studies on the bilateral asymmetry of the lower extremity during the squat under various conditions (Flanagan & Salem, 2007; Hodges, Patrick, & Reiser, 2011; Kobayashi et al., 2010; Newton et al., 2006; Salem, Salinas, & Harding, 2003; Sato & Heise, 2012; Webster, Austin, Feller, Clark, & McClelland, 2014). There were two studies on the effects of different squat loads on the bilateral asymmetry in vertical ground reaction force (VGRF), knee flexion angle, and resultant extension moments at the hip, knee, and ankle during the squat (Flanagan & Salem, 2007; Kobayashi et al., 2010). These studies indicated that there were significant bilateral asymmetries in the kinematics and kinetics of the lower extremity across different squat loads conditions. There were also two studies on the influences of fatigue on the bilateral asymmetry in VGRF and resultant extension moments acting on the lower extremity joints during the squat (Hodges et al., 2011; Webster et al., 2014). These studies suggested symmetrical patterns were achieved when bilaterally fatigued for both patients with anterior cruciate ligament reconstruction (ACLR) and healthy people groups.

There was also a study on the influences of ACLR on bilateral asymmetry in kinematics and kinetics of the lower extremity during the squat (Salem et al., 2003). In this study, patients with ACLR surgery exhibited the bilateral asymmetry in the hip and knee joints which showed the patients with ACLR surgery equally distributed the muscular effort between the hip and knee extension moments in the unaffected side, whereas increased the muscular effort at the hip and decreased the muscular effort at the knee in the affected side. The results of all the previous studies indicated repetitive squat trials with excessive bilateral asymmetry of the lower extremity may result in detrimental effects on performance and increase the risk of the lower extremity injuries. Therefore, it is important to minimize bilateral asymmetry as much as possible during the squat.

Comparison of Lower Extremity Biomechanics between Genders

Gender differences in lower extremity biomechanics during landing, cutting, and jumping have been reported (Decker et al., 2003; Ford, Myer, & Hewett, 2003; Russell, Palmieri, Zinder, & Ingersoll, 2006). These studies showed that there were significant differences in biomechanical risk factors in lower extremity between genders. Females were more prone to knee injuries such as ACL tear and PFP than males in the occurrence of a sudden deceleration of movement. The gender differences have also been observed during closed kinetic chain (CKC) exercises such as squat and lunge (Dwyer, Boudreau, Mattacola, Uhl, & Lattermann, 2010; Willson, Ireland, & Davis, 2006). The results of the studies exhibited that there were significant differences in biomechanical and neuromuscular risk factors between genders during the CKC exercises such as valgus angle, flexion/extension angles, joint moment, muscular imbalance, and muscle activity in the knee joint. These studies also showed that females have been shown to exhibit greater risk of knee injuries while squatting when compared with males.

The primary causes of the gender differences can be categorized into three different factors (Decker et al., 2003; Dwyer et al., 2010; Ford et al., 2003; Russell et al., 2006; Willson et al., 2006). First, there are anatomical differences between genders. Females are characterized by increased Q-angle and narrower notch. The degree of the Q-angle is determined by the width of the pelvis. Since females have a wider pelvis than males, the Q-angle is greater in females and it may increase the risk for the ACL tear. In addition, the ACL moves within the notch which lies between femoral condyles. Since females have a narrower notch than males, the space for ACL movement is more limited in females. Second, there is a hormonal difference between genders. A previous study showed that hormone levels vary in females during the menstrual cycle and it may affect knee stability (Ford et al., 2003). The change in hormones may be a possible contributor to decreased ligament and muscle strength. Third, there are neuromuscular differences between genders. Females have less muscle strength in proportion to the size of the bone than males so that females rely less on the muscles and more on the ACL and PCL to hold the knee. It may increase the chance of the knee injuries. Females also show decreased activity of the knee flexors (hamstrings) while performing landing and squatting. The decreased muscle activity can result in muscular imbalance between quadriceps and hamstrings muscles. The muscular imbalance is one of the knee injury risk factors.

CHAPTER III

METHODS

This chapter is divided into the following sections: participants, trial conditions, experimental setup, data processing, data analysis, and statistical analysis.

Participants

Following a statistical power analysis using G*Power 3 (Faul, Erdfelder, Lang, & Buchner, 2007), a total of forty-two healthy college students were recruited for this study (21 male and 21 female). Anthropometric information is presented in Table 2.

Table 2

Participant Characteristics

	Age (year)	Mass (kg)	Height (cm)	Leg dominance
Female (n=21)	23.3 ± 3.3	62.0 ± 9.1	164.5 ± 2.7	right (20), both (1)
Male (n=21)	26.4 ± 4.4	79.4 ± 10.5	178.1 ± 6.8	right (21)

(mean ± SD)

The inclusion criteria were:

1. Prior squat experience of at least 3 years

2. No knee or lower-back injury history at least 12 months prior to this study

3. Ability to perform a squat with appropriate traditional squat posture

4. No powerlifting experience

Prior to the initiation of the study, the participants were informed of the purposes of this study and were asked to sign an informed consent form approved by Texas Woman's University Institutional Review Board.

Trial Conditions

Six different squat conditions were used based on three stance widths and two toe angles. Stance widths and toe angles were determined by the shoulder width of each participant (distance between both acromions) and the direction of the foot (straight line drawn from the middle of calcaneus to the second toe), respectively. Stance width conditions used were narrow stance (NS; 75% of shoulder width), medium stance (MS; 100% of shoulder width), and wide stance (WS; 140% of shoulder width). Toe angle conditions used were forward (TF; toes pointing directly forward) and out (TO; toes pointing 30° out from the forward direction). Markings were displayed on the floor to standardize the foot placements. Each participant performed 30 squat trials (3 stance widths × 2 toe angles x 5 repetitions per condition) in a randomized order (just conditions) to prevent any order effects.

Squat trials were performed with a weighted barbell placed on the upper trapezius. The weight of the barbell was determined based on 1 repetition maximum (1RM) and 75% of 1RM was used in this study (Caruso et al., 2012; McCaw, & Melrose, 1999). Each participant's 1RM was determined one week before the day of data collection using 1RM testing protocol developed by the National Strength & Conditioning Association (Baechle & Earle, 2000). A minimum rest period of 2-3 minutes between each of the 30 squat trials was allowed to minimize fatigue. Water was also allowed for participants during the resting period to prevent dehydration.

Each participant performed the squat barefoot to eliminate the effects of shoes. While the speed of squat and hand-position on the barbell was not controlled, participants were asked to use a specified squat depth (until the thigh became parallel to the floor). Each participant was asked to perform the squat as consistently as possible in terms of speed, depth, and hand position across all trials as inconsistent speed, depth, and hand position can result in a large variability in data across squat trials.

Prior to actual squat trials, participants were allowed to make multiple practice trials using only a weighted barbell (20.4 kg) to adjust to the 6 different foot positions. The preferred squat speed and hand position and desired depth of the squat (until the thigh became parallel to the floor) were also determined during the practice trials. The preferred hand position was marked with a tape on the bar for consistent hand positioning. The success of a squat trial was determined by the accuracy of squat motion and participant's input.

Experimental Setup

Motion Capture

A 250-Hz 10-camera Vicon motion capture system (Centennial, CO, USA) was used to capture the 3D coordinates of the retro-reflective markers placed on the participant's body. The 'Squat' marker set (Table 3) was used to create a stick figure of the squat motion. A total of 27 markers were attached on the participant's body. A static posture trial was collected prior to squat trials and used to find the locations of a group of secondary points (Table 3). Participants oriented in the direction of the positive X-axis of the global reference frame (Figure 2). The positive Y-axis pointed leftward perpendicular to the X-axis. The positive Z-axis was upward. Two AMTI forceplates (Advanced Mechanical Technology, Inc., Watertown, MA, USA) were used to collect the ground reaction force (GRF) data.

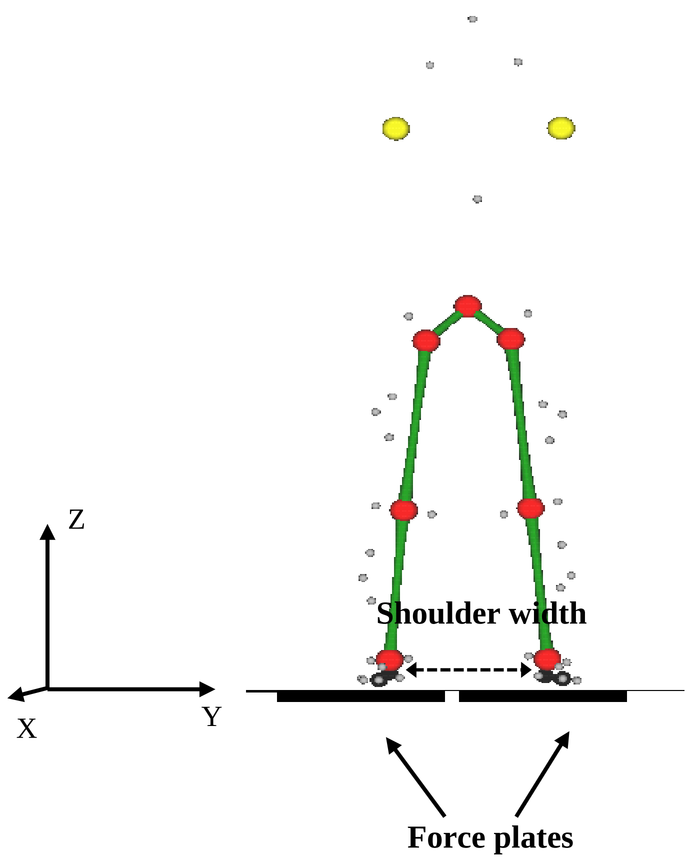


Figure 2: Experimental setting.

Table 3

The 35-Point (27 Markers) 'Squat' Body Model/Marker Set

Segment	Markers/computed points	Description
Pelvis	Markers (3)	Three pelvic markers: right and left anterior superior iliac spines (ASISs) and middle posterior superior iliac spines (PSIS).
	Computed (4)	Hip joints (right and left), L4/L5, and mid-ASIS. The hip joints were calculated using the 'Tylkowski Method' and L4/L5 was computed using the 'MacKinnon Method' (see Table 4). The mid-ASIS is the mid-point of right and left anterior superior iliac spines.
Legs	Markers (12×2)	Five thigh markers each: lateral, upper, and lower thigh and lateral and medial femoral epicondyle. Five shank markers each: lateral, upper, and lower shank and lateral and medial malleoli. Two foot markers each: heel and toe.
	Computed (2×2)	Knee and ankle joints. The knee and ankle joint centers were computed in the motion trials using the 'Mid-Point Method' (see Table 2).

Table 4

Methods Used in Locating the Computed Joints

Method	Description
Tylkowski	In the static posture trial, the hip joint center was computed using the Tylkowski method (Tylkowski, Simon, & Mansour, 1982). Three pelvic markers were used for this method.
MacKinnon	In the static posture trial, the L4/L5 joint was computed using the MacKinnon method (MacKinnon and Winter, 1993) based on the three pelvic markers (right and left ASISs and mid-PSIS). The joint position was then described with respect to the pelvic reference frame defined by the pelvic markers. In the motion trials, this joint position was used to compute the L4/L5 joint.
Mid-point	A joint center was computed using the mid-point of the two markers (lateral and medial femoral epicondyle markers in the case of the knee).

Data Processing

Captured Trial Processing

The captured squat motions and GRF data were initially processed on Vicon Nexus to generate data files in the C3D format. The C3D files were then imported into Kwon3D Motion Analysis Suite (Version XP, Visol Inc., Seoul, Korea) for subsequent data processing and analysis. Biomechanical kinetic variables (normalized sagittal- and frontal-plane resultant joint moments) were computed and extracted using Kwon3D. To eliminate the noise due to experimental errors, the raw coordinates were filtered using a Butterworth zero phase-lag fourth-order low-pass filter with a cut-off frequency of 6 Hz.

In order to facilitate data analysis, the squat motion was subdivided into four phases using five events defined based on the vertical center of mass (COM) position and acceleration patterns (Figure 3). Start of Squat (SS) was the instant the participant started squatting down and Transition 1 (TR1) was when the vertical COM acceleration became zero during the downward motion. Bottom of Squat (BS) was when the COM was located at the lowest vertical position. Transition 2 (TR2) was when the vertical COM acceleration became zero during the upward motion. End of Squat (ES) was when the participant finished the squat and stood upright again. Based on these five events, four phases were defined: Acceleration of Downward Motion (AD; SS to TR1), Deceleration of Downward Motion (DD; TR1 to BS), Acceleration of Upward Motion (AU; BS to TR2), and Deceleration of Upward Motion (DU; TR2 to ES).

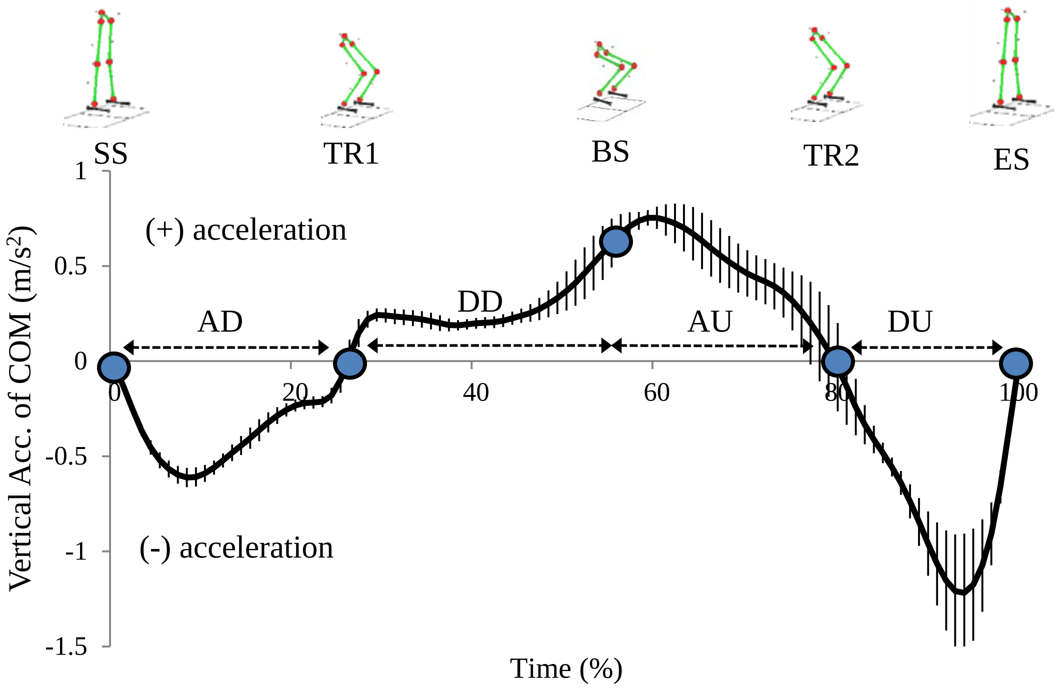


Figure 3: Squat motion events and phases defined: Start of Squat (SS), Transition 1 (TR1), Bottom of Squat (BS), Transition 2 (TR2), End of Squat (ES), Acceleration of Downward Motion (AD), Deceleration of Downward Motion (DD), Acceleration of Upward Motion (AU), and Deceleration of Upward Motion (DU).

Data Analysis

Local Coordinate System

For data analysis, local reference frames attached to each lower extremity segment were first defined (Table 5). The X-, Y-, and Z-axis of the segments were aligned with the mediolateral, anteroposterior, and longitudinal axes of the segments, respectively. To define the local reference frames, an anatomical plane was first defined using two axes (first axis and temporary second axis). The third axis was then defined by using the cross product of the first and temporary second axis unit vectors. The true second axis was lastly determined by using the cross product of the first and third axis unit vectors.

Table 5

Definitions of Local Reference Frames of Each Segment

Segment	Primary axis	Temporary second axis	Plane
Pelvis	L ASIS to R ASIS (+X axis)	Mid-PSIS to mid-ASIS (+Y axis)	Transverse
R Thigh	RKJ to RHJ (+Z axis)	RHJ to R upper thigh marker (+X axis)	Frontal
L Thigh	LKJ to LHJ (+Z axis)	L upper thigh marker to LHJ (+X axis)	Frontal
R Shank	RAJ to RKJ (+Z axis)	RKJ to R upper shank marker (+X axis)	Frontal
L Shank	LAJ to LKJ (+Z axis)	L upper shank marker to LKJ (+X axis)	Frontal
R Foot	RTO to RHL (+Z axis)	RHL to RAJ (+Y axis)	Sagittal
L Foot	LTO to LHL (+Z axis)	LHL to LAJ (+Y axis)	Sagittal

Abbreviations. R (right), L (left), RHJ (right hip joint), RKJ (right knee joint), RAJ (right ankle joint), LHJ (left hip joint), LKJ (left knee joint), LAJ (left ankle joint), RTO (right toe), LTO (left toe), RHL (right heel), LHL (left heel), Med (medial), and Lat (lateral).

Joint Coordinate System

The Joint Coordinate System (JCS) convention proposed by Grood and Suntay (1983) to eliminate the temporal sequence dependency of Euler angle techniques was used in this study. Three axes defined by the JCS in this study were X axis in the proximal segment, Z axis in the distal segment, and Y axis in a mutually orthogonal floating axis (Figure 4).

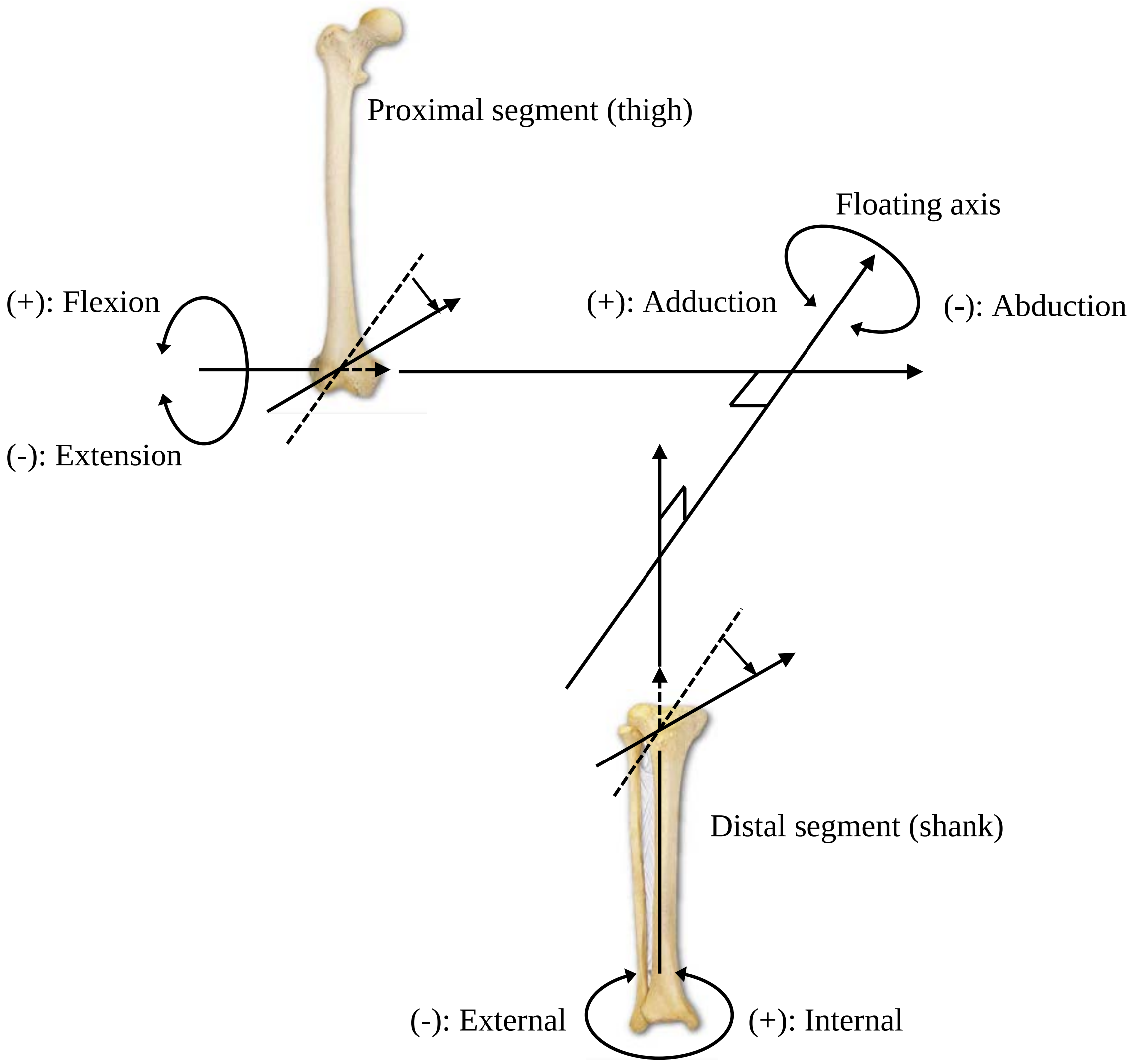


Figure 4: A diagram of Joint Coordinate System (JCS) using thigh and shank segments. Flexion/extension is about the X axis in the proximal segment (thigh) and external/internal rotation is about the Z axis in the distal segment (shank). Abduction/adduction is about the floating axis.

For the computation of the relative orientation angles of the lower extremity segments to their respective linked proximal segments, the XYZ (mediolateral-anteroposterior-longitudinal) rotation sequence was used. The first rotation was about the X-axis of the proximal segment while the second rotation was about a floating intermediate Y-axis. The third rotation was performed about the Z-axis of the distal segment. The JCS was composed of these three rotation axes.

The JCS was used to extract the components of the resultant joint moments acting on the hip, knee and ankle joints. The flexor/extensor (hip and knee) and plantar-flexor (ankle) moments were about the first rotation axis. The internal/external rotator moment (hip and knee) was about the third rotation while internal/external rotator moment (ankle) was about the second rotation axis. Adductor/abductor moment (hip and knee) was about the second rotation moment and inverter/everter moment (ankle) was about the third rotation axis.

Kinetic Analysis

Resultant joint moments acting on the hip, knee, and ankle joints in the sagittal and frontal planes were calculated through an inverse dynamics procedure (equation 1 and 2):

$$\mathbf{F} = \sum_{s} \frac{d\mathbf{P}_s}{dt} - \sum_{s} \mathbf{W}_s - \mathbf{F}_E \qquad [1]$$

$$\mathbf{N} = \sum_{s} \left(\frac{d\mathbf{L}_s}{dt} + \mathbf{r}_{js} \times \frac{d\mathbf{P}_s}{dt} \right) - \sum_{s} \left(\mathbf{r}_{js} \times \mathbf{W}_s \right) - \left(\mathbf{r}_{je} \times \mathbf{F}_E + \mathbf{N}_E \right) \qquad [2]$$

where $\mathbf{F}$ is the resultant joint force acting at the joint, $\frac{d\mathbf{P}_s}{dt}$ is the rate of change in linear momentum of the segment, $\mathbf{W}_s$ is the weight of each segment, $\mathbf{F}_E$ is the ground reaction force, $\mathbf{N}$ is the resultant joint moment acting at the joint, $\mathbf{r}$ is the position vector drawn from the joint center to the point at which the muscle force occurs, $\frac{d\mathbf{L}_s}{dt}$ is the rate of change in local angular momentum of the segment due to the rotation of the segment about its own COM, $\mathbf{r}_{js}$ is the position vector drawn from the joint center to segment's COM, $\mathbf{r}_{js} \times \frac{d\mathbf{P}_s}{dt}$ is the remote angular momentum of the relative motion of the segment COM to the joint, $\mathbf{r}_{je}$ is the position vector drawn from the joint center to the point at which the ground reaction force acts, and $\mathbf{N}_E$ is the ground reaction moment (free moment acting at the center of pressure). The resultant joint moments calculated were normalized to participant's body mass plus barbell load.

Statistical analysis

The peak moment values extracted from five trials in each squat condition were averaged and used in the statistical analysis. Two three-way (3 × 2 x 2) mixed designs MANOVAs were conducted to compare normalized resultant joint moments in the sagittal and frontal planes separately, with the stance width (within-subject: NS, MS, and WS), toe angle (within-subject: TF and TO), and gender (between-subject: male and female) being factors. For significant factor effect or interaction, post-hoc tests were performed with Bonferroni adjustment. The level of significance was set at .05 for all statistical tests.

Prior to running statistics, both univariate and multivariate outliers were eliminated using z-scores provided by descriptive statistics and Mahalanobis distance (Mahalanobis, 1936) provided by a dummy multiple regression using a dummy variable as the dependent variable and the dependent variables for the MANOVA as the independent variables, respectively.

CHAPTER IV

RESULTS

Resultant Joint Moments in the Sagittal Plane

Only the stance width * toe angle interaction (Wilk's λ = 0.657, F = 3.051, p = .017) was significant, whereas stance width * gender (Wilk's λ = 0.961, F = 0.237, p = .961), angle * gender (Wilk's λ = 0.890, F = 1.566, p = .214), and width * angle * gender (Wilk's λ = 0.842, F = 1.091, p = .387) interactions were not significant (Table 6).

Overall, the toe out condition revealed larger hip extensor moment (HEX), knee extensor moment (KEX), and ankle plantar-flexor moment (AP) than the toe forward condition across stance widths. The toe out condition showed significantly larger HEX (medium stance, p = .018; wide stance, p = .006), KEX (medium stance, p = .009; wide stance, p = .035), and AP (narrow stance; p = .008) than the toe forward condition. In addition, significant inter-stance differences in HEX (narrow stance < wide stance, p < .001; medium stance < wide stance, p < .001), KEX (narrow stance < medium stance, p = .005; narrow stance < wide stance, p = .044), and AP (narrow stance > wide stance, p = .03) were observed in the toe out condition.

Table 6

Comparison of the Normalized Sagittal-Plane Resultant Joint Moments

Variables		Toe forward			Toe out			Sig. effects
		Narrow	Medium	Wide	Narrow	Medium	Wide	
HEX (Nm/kg)	F	-0.88 ± 0.17	-0.88 ± 0.17	-0.89 ± 0.15	-0.87 ± 0.15	-0.89 ± 0.15	-0.93 ± 0.13	
	M	-0.96 ± 0.24	-0.94 ± 0.23	-0.96 ± 0.21	-0.99 ± 0.20	-0.99 ± 0.20	-1.05 ± 0.19	
	C	-0.92 ± 0.21	-0.91 ± 0.20	-0.92 ± 0.18	-0.93 ± 0.18	-0.94 ± 0.18[c]	-0.99 ± 0.17[a,b,c]	
KEX (Nm/kg)	F	0.66 ± 0.13	0.68 ± 0.15	0.68 ± 0.13	0.68 ± 0.12	0.70 ± 0.12	0.71 ± 0.11	Width ×Angle interaction
	M	0.73 ± 0.13	0.74 ± 0.15	0.73 ± 0.14	0.75 ± 0.14	0.78 ± 0.14	0.77 ± 0.13	
	C	0.70 ± 0.14	0.71 ± 0.15	0.70 ± 0.14	0.72 ± 0.13	0.74 ± 0.13[a,c]	0.74 ± 0.12[a,c]	
AP (Nm/kg)	F	-0.35 ± 0.09	-0.38 ± 0.13	-0.37 ± 0.11	-0.41 ± 0.12	-0.40 ± 0.11	-0.36 ± 0.09	
	M	-0.35 ± 0.10	-0.35 ± 0.08	-0.35 ± 0.09	-0.37 ± 0.12	-0.37 ± 0.13	-0.35 ± 0.14	
	C	-0.35 ± 0.09	-0.36 ± 0.10	-0.36 ± 0.10	-0.39 ± 0.12[c]	-0.38 ± 0.12	-0.36 ± 0.12[a]	

Abbreviations: HEX, hip extensor moment; KEX, knee extensor moment; AP, ankle plantar-flexor moment; F, female; M, male; C, combined. [a] significantly ($p < 0.05$) different from the matching narrow stance condition; [b] significantly different from the matching medium stance condition; [c] significantly different from the matching toe forward condition.

Resultant Joint Moments in the Frontal Plane

There were no significant interactions among the factors on the normalized frontal-plane resultant joint moments (stance width * toe angle: Wilk's λ = 0.925, F = 0.457, p = .835; stance width * gender: Wilk's λ = 0.938, F = 0.376, p = .889; angle * gender: Wilk's λ = 0.904, F = 1.307, p = .287; width * angle * gender: Wilk's λ = 0.929, F = 0.434, p = .851). However, Width (Wilk's λ = 0.214, F = 20.798, p < .001) and angle (Wilk's λ = 0.179, F = 56.753, p < .001) revealed significant factor effects (Table 7).

There were significant differences in HAB, KAB, and AE among the stance widths. In all parameters, the narrow stance exhibited the largest resultant joint moment values while the wide stance demonstrated the smallest. The toe out condition showed significantly larger KAB (p < .001) and AE (p = .002) but smaller HAB (p = .004) than the toe forward condition.

Table 7

Comparison of the Normalized Frontal-Plane Resultant Joint Moments

Variables		Toe forward Narrow	Toe forward Medium	Toe forward Wide	Toe out Narrow	Toe out Medium	Toe out Wide	Sig. effects
HAB (Nm/kg)	F	-0.30 ± 0.09	-0.26 ± 0.07	-0.25 ± 0.09	-0.28 ± 0.07	-0.24 ± 0.06	-0.21 ± 0.06	Width (N > M > W), Angle (F > O)
	M	-0.27 ± 0.07	-0.23 ± 0.07	-0.21 ± 0.09	-0.26 ± 0.07	-0.22 ± 0.07	-0.19 ± 0.08	
	C	-0.28 ± 0.08	-0.25 ± 0.07	-0.23 ± 0.09	-0.27 ± 0.07	-0.23 ± 0.07	-0.20 ± 0.07	
KAB (Nm/kg)	F	-0.12 ± 0.05	-0.10 ± 0.05	-0.08 ± 0.06	-0.16 ± 0.07	-0.14 ± 0.07	-0.12 ± 0.07	Width (N > M > W), Angle (O > F)
	M	-0.19 ± 0.08	-0.17 ± 0.07	-0.15 ± 0.09	-0.27 ± 0.10	-0.24 ± 0.10	-0.22 ± 0.10	
	C	-0.16 ± 0.07	-0.14 ± 0.07	-0.11 ± 0.08	-0.22 ± 0.10	-0.19 ± 0.10	-0.17 ± 0.10	
AE (Nm/kg)	F	-0.10 ± 0.04	-0.10 ± 0.04	-0.08 ± 0.04	-0.12 ± 0.05	-0.11 ± 0.05	-0.09 ± 0.04	Width (N > M > W), Angle (O > F)
	M	-0.09 ± 0.03	-0.08 ± 0.03	-0.06 ± 0.03	-0.11 ± 0.05	-0.09 ± 0.04	-0.08 ± 0.04	
	C	-0.09 ± 0.04	-0.09 ± 0.04	-0.07 ± 0.04	-0.11 ± 0.05	-0.10 ± 0.04	-0.09 ± 0.04	

Abbreviations: HAB, hip abductor moment; KAB, knee abductor moment; AE, ankle everter moment; N, narrow; M, medium; W, wide; F, forward; O, out

Significant factor effects at $p < .05$.

Patterns of Resultant Joint Moments throughout the Squat Motion

The maximum values of HEX, KEX, and AP were observed at the beginning of AU (Figure 5). The maximum values of HAB and KAB occurred approximately at TR1 during the downward phase and TR2 during the upward phase while the smallest HAB and KAB occurred approximately at BS. The peak AE was observed around BS.

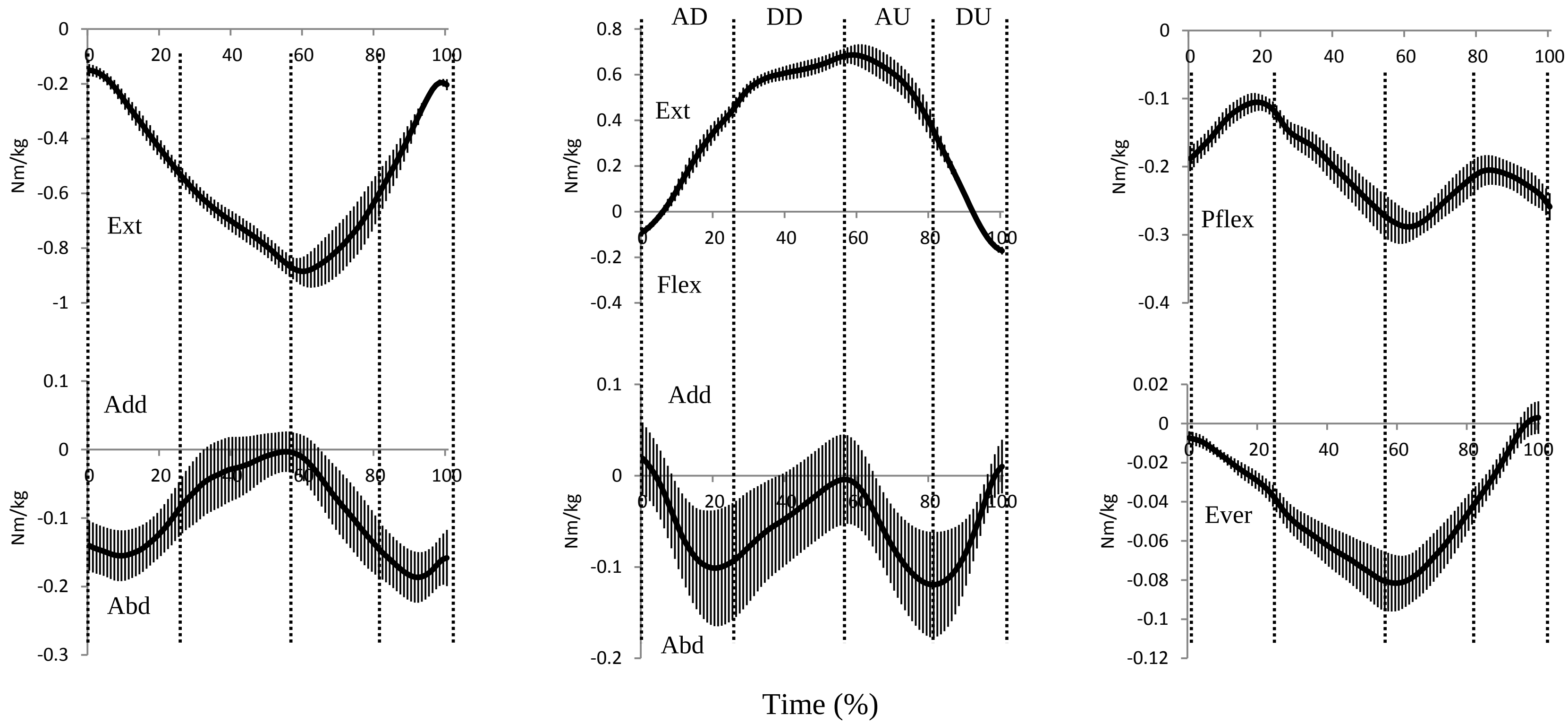


Figure 5: Ensemble-average normalized resultant joint moment patterns of the hip (1st column), knee (2nd column), and ankle (3rd column). The 1st and 2nd rows represent the normalized resultant joint moments in the sagittal and frontal planes, respectively.

CHAPTER V

DISCUSSION

This study investigated the effects of different foot placements on kinetics of the lower extremity joints during the squat. Normalized resultant joint moments of the hip, knee, and ankle in the sagittal and frontal planes were calculated and compared across the three stance widths and two toe angles. Ensemble-average normalized resultant joint moment patterns were analyzed to identify how the normalized resultant joint moment patterns changed throughout the squat performance.

Resultant Joint Moments in the Sagittal Plane

Toe Angle Comparison across Matching Stance Widths

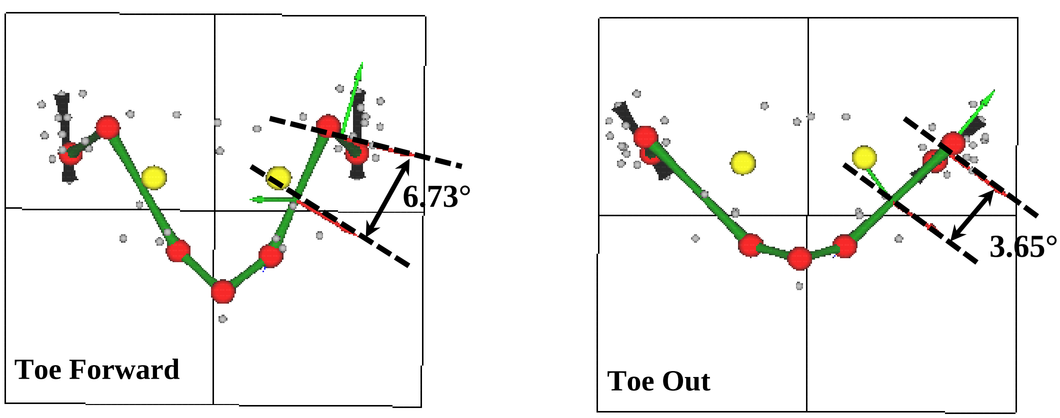


Figure 6: A top view of the squat showing alignment of the femur and tibia relative to a local mediolateral axis fixed to thigh and shank segments between the toe forward and toe out conditions in the wide stance.

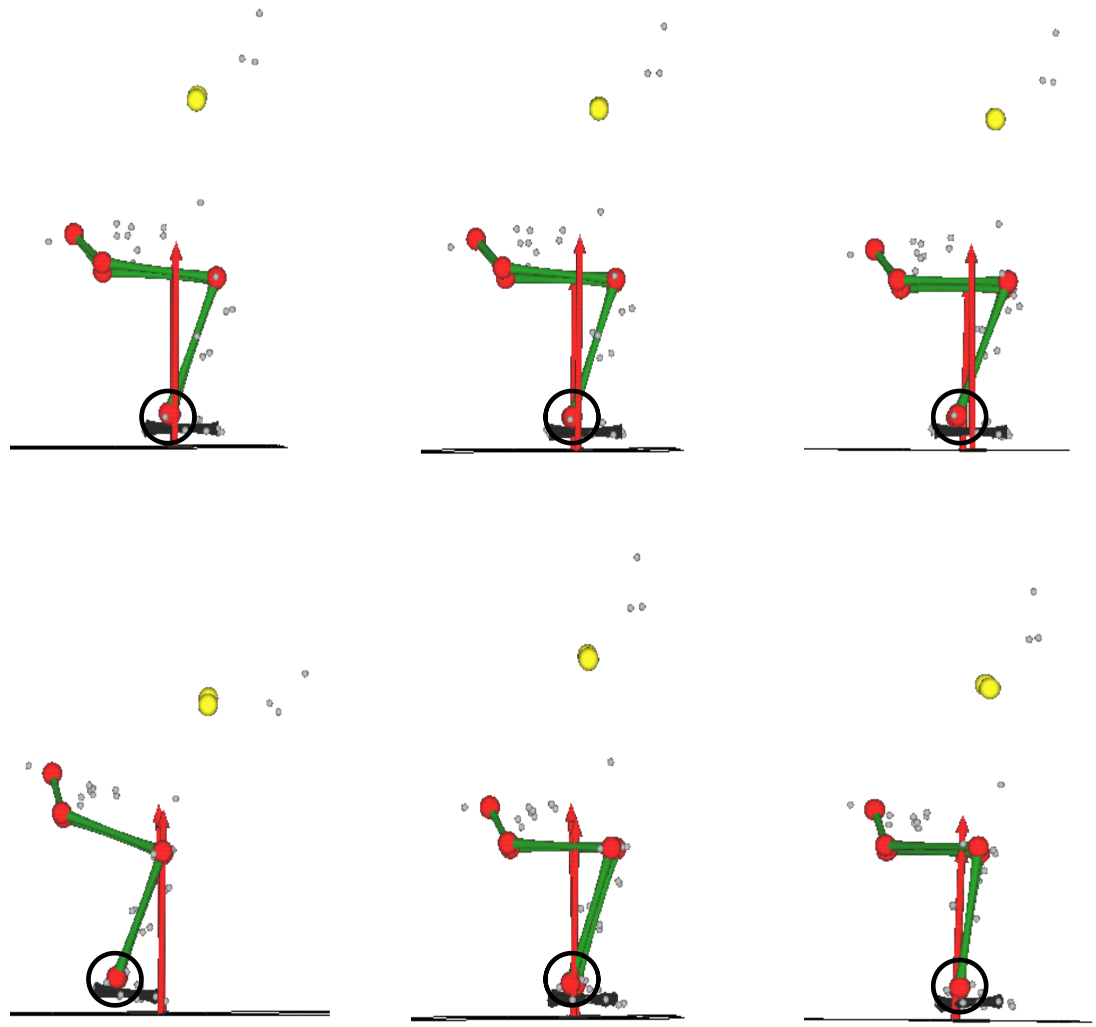


Figure 7: Points of action of the ground reaction force vector in various foot position conditions: narrow stance (left), medium stance (middle), wide stance (right), foot forward (top), and foot outward (bottom).

The significantly larger HEX and KEX observed in the medium and wide stances for the toe out condition could be explained by a discrepancy in alignment of the femur and tibia relative to a local mediolateral axis fixed to thigh and shank segments (Figure 6) and center of pressure (COP) shift in the anteroposterior direction of the feet between the toe forward and toe out conditions during the squat (Figure 7).

During the toe out condition, as stance width increased, hip horizontal abduction increased because of external rotations of the femur and tibia and it caused a better alignment of the femur and tibia between the mediolateral axes of the segments than that during the toe forward condition (Figure 6). A proper alignment of the femur and tibia is required for proper knee flexion/extension movements during the squat as the knee (tibiofemoral joint) is considered a hinge joint. A proper alignment of the femur and tibia would also allow the hip to travel more posteriorly, which causes an increase in HEX, in order to properly perform the squat during the toe out condition (Fry et al., 2003). The COP would then be shifted closer to the ankle during the toe out condition (Figure 7).

During the toe forward condition, hip horizontal abduction also increased as the stance width increased; however, the feet remained pointing forward causing more knee adduction and internal rotation compared to the toe out condition due to an improper alignment of the femur and tibia (Figure 6). It was speculated that the squat could be better performed with the toes facing out during relatively wider stance widths because the toe out condition allows the tibia and femur to be externally rotated together with a proper alignment of the femur and tibia, whereas the toe forward condition inhibits the femur and tibia from external rotation causing an improper alignment of the femur and tibia.

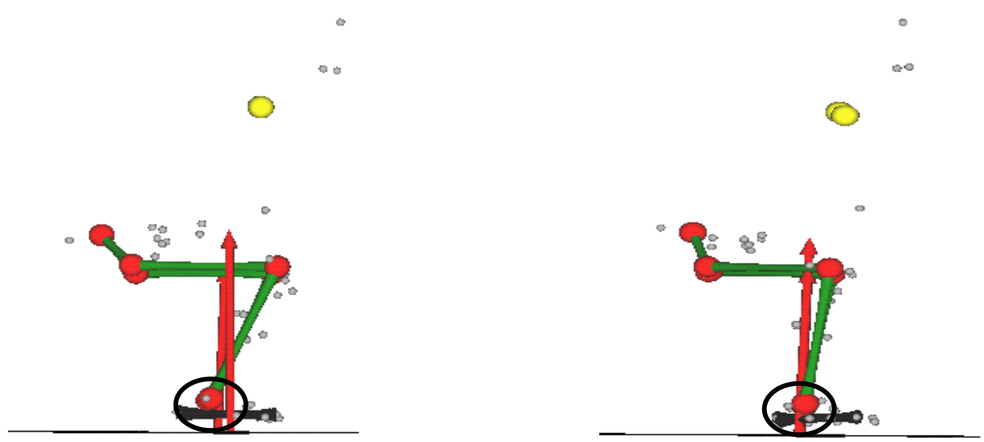
The significantly larger AP observed in the narrow stance during the toe out condition could also be explained by the COP shift. Unlike what was observed in the medium and wide stances, the squat was performed without hip horizontal abduction in both the toe out and toe forward conditions of the narrow stance. In the toe forward condition, as the squat was performed without external rotations of the femur and tibia, participants were able to naturally move the hip posteriorly during the narrow stance. In the narrow stance and toe out condition, however, the hip was inhibited from traveling posteriorly, which resulted in a knee out situation. Thus, the narrow stance during the toe out condition could cause the COP to be placed over a more anterior part of the feet than that during the toe forward condition, which would result in the significantly larger AP (Figure 7).

Stance Width Comparison between Toe Forward and Toe Out Conditions

The significantly larger HEX and KEX and smaller AP with increasing stance width during the toe out condition could explain that participants relied more on hip and knee extensor muscles and less on ankle plantar-flexor muscles with increasing stance width during the toe out condition. The results of HEX, KEX, and AP could be attributed to the COP shift and a squat posture change in response to the COP shift. The COP was shifted closer to the ankle joint as stance width increased during the toe out condition (Figure 7), which might cause AP to decrease and HEX to increase due to a larger hip backward movement (Fry et al., 2003). The trunk would also be more erect with increasing stance width during the toe out condition due to increased hip horizontal abduction and hip external rotation causing a decrease in trunk extensor moment and an increase in KEX (Swinton et al., 2012). Therefore, it could be said that the decreased AP and trunk extensor moment caused by the COP shift toward the ankle and the increased hip horizontal abduction and hip external rotation could be transferred to HEX and KEX as all the lower extremity joints and lower trunk function as a system during the squat to remain balanced (Fry et al., 2003).

The observed HEX and KEX results were consistent with the outcomes of a previous study (Escamilla et al., 2001a) which reported that KEX and HEX increased as stance width increased during the toe out condition. However, these results were not in line with the outcomes of another study (Swinton et al., 2012) which indicated that HEX increased but KEX decreased as stance width increased during the squat. This discrepancy could be attributed to the fact that the study emphasized posterior movement of the hip during the wide stance by intentionally restricting anterior movement of the knee and maintaining as vertical a shin position as possible, while at the same time maximizing posterior displacement of the hip. This controlled squat technique used in the previous study could result in the discrepancy in KEX between the two studies.

Unlike HEX and KEX, AP seen in the present study was inconsistent with the results of the previous study (Escamilla et al., 2001a). The AP was observed overall across all stances during the toe out condition with occurrence of dorsiflexor moment in a few participants between TR1 and BS in this study, whereas AP was only observed in the narrow stance and dorsiflexor moment was only observed in the medium and wide stances in the previous study. This result could be due to a discrepancy in the group of participants recruited between the two studies. Recreationally trained individuals were recruited for this study, whereas professional powerlifters were recruited for the previous study. This heterogeneity of the group of participants could be the major reason for AP discrepancy observed between two studies. The style used by powerlifters to perform a ‘powerlifting squat’ is completely different from the manner in which recreationally trained individuals perform the squat, (known as the traditional or Olympic squat) because of the difference in purpose between the two performance styles (Swinton et al., 2012).



Traditional **Powerlifting**

Figure 8: Comparison of points of action of the ground reaction force vector between traditional squat and powerlifting squat.

The goal of the traditional squat is to strengthen, to a relatively equal extent, the knee extensor (quadriceps), hip extensor (gluteus maximus and hamstrings), ankle plantar-flexor (gastrocnemius and soleus), and lumbar stabilizer muscles (erector spinae). Recreationally trained individuals tend to emphasize the use of the knee extensor muscles by placing a weighted bar on the upper trapezius and using a relatively narrow stance width than seen in the powerlifting squat (Swinton et al., 2012). This technique allows the COP to be located in the middle of the foot and causes less trunk flexion compared to the powerlifting squat, which is an ideal squat posture for achieving the goals of the traditional squat (Figure 8).

The goal of the powerlifting squat, however, is to lift as much weight as possible with each successful completion of the squat. Powerlifters tend to emphasize the use of the hip extensor muscles, which is a larger and stronger group of muscles than the quadriceps and gastrocnemius, by placing the weighted bar on the lower trapezius and using a relatively wider stance width than used for the traditional squat (Swinton et al., 2012). This technique allows the COP to shift closer to the ankle joint and causes more trunk flexion compared to the traditional squat, which is an ideal squat posture for achieving the goal of the powerlifting squat (Figure 8). Thus, the narrow, medium, and wide stances were defined as 75%, 100%, and 140% relative to the shoulder width respectively, in the current study, whereas 107%, 142%, and 169% were the defining values in the previous study, with similar degrees of toe angles (less than 7° differential) between the two studies. Because of the relatively wider stances and lower position of the weighted bar used in the previous study, the COP could be located posterior to the ankle joint during the squat in the medium and wide stances which resulted in dorsiflexor moment.

The result of AP observed in this study was supported by a study indicating that wider stance showed a significantly lower AP than a narrow stance during the squat (Swinton et al., 2012). The comparison of the AP between the narrow and wide stances during the squat observed in both the current study and the Swinton et al. study could both be explained by the fact that ankle plantar-flexor muscle (gastrocnemius) activity increased with decreasing stance width during the squat (Escamilla et al., 2001b).

Resultant Joint Moments in the Frontal Plane

Stance Width Comparison

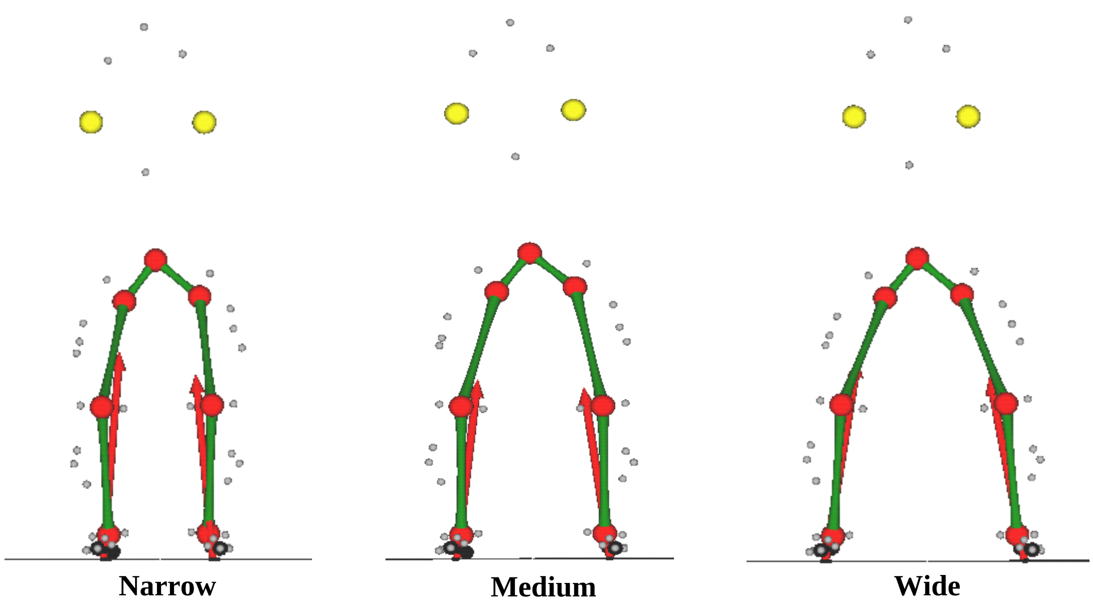


Figure 9: A change in the GRF vector in the frontal plane across stance widths.

The significantly largest HAB, KAB, and AE observed in the narrow stance and smallest HAB, KAB, and AE in the wide stance could be explained by a change in the direction of the GRF vector. The direction of the GRF vector changed in response to the change in stance width during the squat (Figure 9). The COP was placed more on the lateral side of the feet with increasing stance width. As a result, the moment arms from the joints to the line of action of the GRF vector in the frontal plane decreased with increasing stance width. This could cause smaller moments in the frontal plane with increasing stance width. The result of the HAB was not in line with the outcome of the Swinton et al., (2012) study which indicated that HAB increased as stance width increased during the squat. This discrepancy could be explained by the fact that as previously discussed the previous study emphasized posterior movement of the hip during the wide stance by intentionally maximizing posterior displacement of the hip.

Toe Angle Comparison

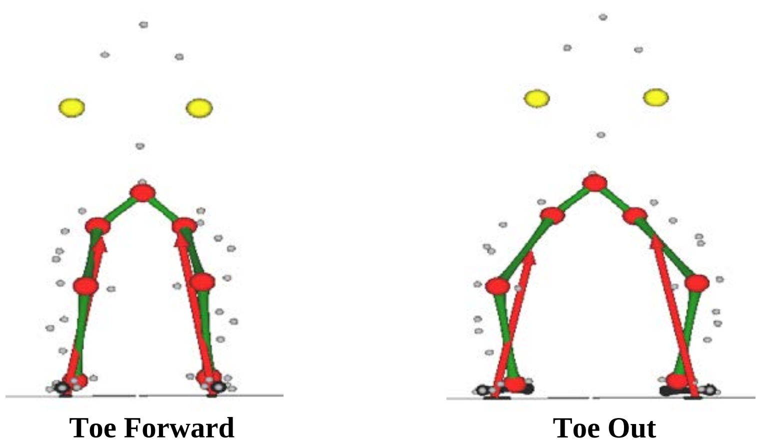


Figure 10: Comparison of the direction of the GRF vector between toe forward and toe out conditions.

The significantly larger KAB and AE and smaller HAB observed in the toe out condition compared to the toe forward condition could be attributed to the fact that the knee was more adducted and the ankle was more inverted during the toe out condition, whereas the hip was more adducted during the toe forward condition (Figure 10). This is because hip horizontal abduction increased when the femur and tibia became externally rotated during the toe out condition. In addition, the discrepancy in the direction of the GRF vector between the toe forward and toe out conditions could be used to support these results. The COP was placed more on the lateral side of the feet during the toe out condition than during the toe forward condition, which could result in the change in the direction of the GRF vector more medially directed during the toe out condition (Figure 10). Both of these reasons could cause more knee adduction and more ankle inversion during the toe out condition and more hip adduction during the toe forward condition. Therefore, significantly higher KAB and AE were observed in the toe out condition and the significantly higher HAB was observed in the toe forward condition.

Patterns of Resultant Joint Moments throughout the Squat Motion

Patterns of Resultant Joint Moments in the Sagittal Plane

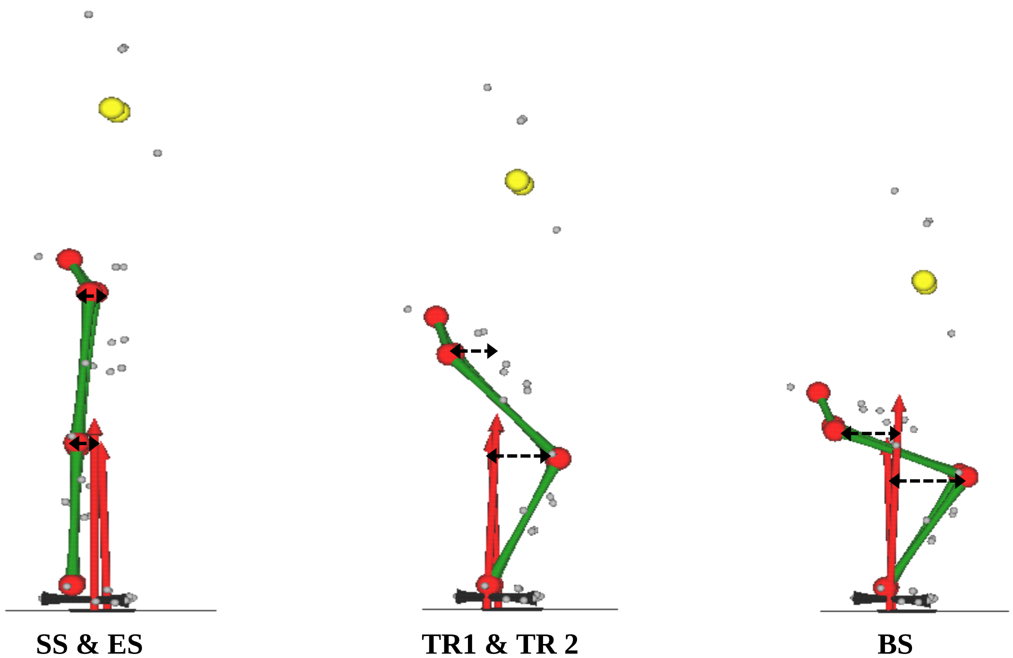


Figure 11: A change in moment arm from the line of action of the GRF vector to each joint and the magnitude of GRF at each event.

Similar patterns of resultant joint moments of the hip, knee and ankle in the sagittal plane were observed in both the toe forward and toe out conditions. HEX and KEX showed similar patterns of both HEX and KEX increased as participants squatted down while those decreased as participants stood upright (Figure 5). This can be explained by a change in moment arm from the line of action of the GRF vector to each joint and the magnitude of GRF during the squat (Figure 11). The longest moment arm and largest GRF were observed in BS, followed by TR1 and TR2, and then SS and ES. Therefore, it could be stated that the maximum HEX and KEX would be observed around BS.

More specifically, the maximum values of HEX and KEX were observed in the beginning of AU, which was between BS and TR2 (Figure 5). These results can be due to the stretch-shortening cycle (SSC) occurring during an immediate transition from an eccentric contraction (downward motion) to a concentric contraction (upward motion) during the squat (Manabe, Shimada, & Ogata, 2007). The SSC occurring during the immediate transition phase (between BS and TR2) might help the lower extremity joints generate the highest moment values by generating the highest vertical acceleration of the whole lower extremity (Figure 3). These results were consistent with several squat studies (Almosnino et al, 2013; Manabe et al., 2007; McCaw & Melrose, 1999) which reported that the largest muscle activities and resultant joint moments of the lower extremity occurred in the beginning of AU.

For AP, the maximum value of AP occurred in the beginning of AU, however, the minimum value of AP was shown around TR1 or TR2 unlike HEX and KEX. This could be due to the point of action of the GRF vector around TR1 and TR2 which was located directly over the ankle at TR1 and TR2 (Figure 11). This would then minimize the moment arm from the line of the action of the GRF to the ankle.

Patterns of Resultant Joint Moments in the Frontal Plane

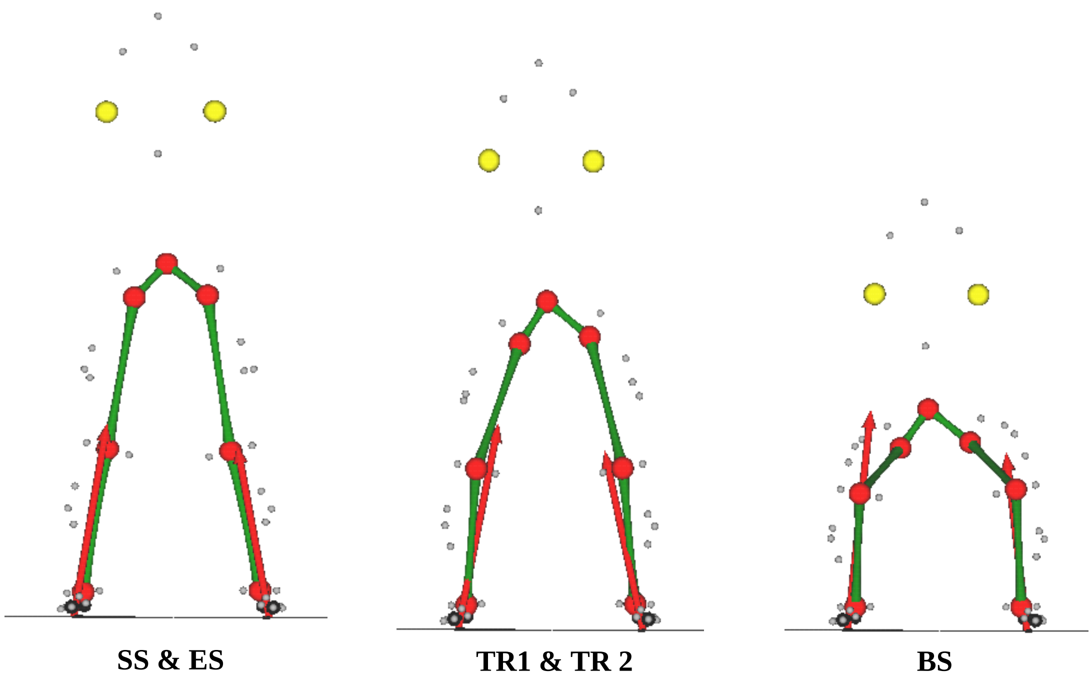


Figure 12: A change in the direction of the GRF vector in the frontal plane.

The direction of the GRF vector showed continual change throughout the squat performance, especially in the frontal plane (Figure 12). In SS and ES, the GRF vector pointed toward the center of mass (COM) of the body along the leg because as the feet were pushed outward, the COP moved toward the lateral side of the feet. As participants squatted down, the knee was adducted while moving out. At the same time the COP moved toward the lateral side of the feet, changing the direction of the GRF vector toward the inside of the body, and generating an increase in the moment arm from the joints to the line of action of the GRF vector in the frontal plane. Both the increased moment arm and GRF vector could generate the increase in adduction of the knee and hip and inversion of the ankle.

The maximum values of HAB and KAB observed approximately at TR1 and TR2 could be attributed to the fact that the joints were positioned furthest from the line of action of the GRF vector at TR1 and TR2 in the frontal plane. The minimum values of HAB and KAB were observed when participants were positioned approximately at BS because the direction of the GRF vector changed from toward the body to vertical as the participants reached BS. It minimized the moment arms resulting in minimum HAB and KAB at BS. The decreased HAB and KAB could then be transferred to the AE resulting in the maximum value of AE around BS because all the lower extremity joints function as a system to remain balanced.

Conclusion

The purpose of this study was to investigate the effects of foot placement (stance width and foot angle) on kinetic factors (normalized sagittal-and frontal-planes resultant joint moments) in the lower extremity joints during the squat using 3D motion analysis. From the findings of this study, the following conclusions were derived:

- For normalized sagittal-plane resultant joint moments, the toe out condition showed significantly larger HEX and KEX in the medium and wide stances and larger AP in the narrow stance than the toe forward condition. In addition, significant inter-stance differences in HEX (narrow stance < wide stance; medium stance < wide stance), KEX (narrow stance < medium stance; narrow stance < wide stance), and AP (narrow stance > wide stance) were observed in the toe out condition.

- For normalized frontal-plane resultant joint moments, in terms of stance width, there were significant differences in HAB, KAB, and AE among the stance widths. In all parameters, the narrow stance exhibited the largest resultant joint moment values while the wide stance demonstrated the smallest. In terms of toe angle, the toe out condition showed significantly larger KAB and AE but smaller HAB than the toe forward condition.
- The maximum values of HEX, KEX, and AP were observed at the beginning of AU. The maximum values of HAB and KAB occurred approximately at TR1 during the downward phase and TR2 during the upward phase while the smallest HAB and KAB occurred approximately at BS. The peak AE was observed approximately at BS.
- Based on the results of this study, an optimal foot placement during the squat would be toes out with wide stance condition. The purpose of squat performance needs to be considered to determine foot placement. For a rehabilitation purpose, foot placement should be decided by the type of injury. For athletic performance enhancement or muscle strengthening purposes, foot placement needs to be matched by the type of sport and purpose of exercise.

Recommendation

Based on results of this study, there are two recommendations for future studies:

- To account for sex differences in squat width, using the pelvic width would result in a more accurate measurement for stance width because females have significantly narrower shoulders but a wider pelvic width than males.
- This study dealt with only healthy participants. For a wider application of squat movement, it would be recommended to understand differences in squat performance between a healthy population and a population diagnosed with injuries such as ACL injury or patellofemoral pain.

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APPEMDIX A

INFORMED CONSENT FORM

TEXAS WOMAN'S UNIVERSITY
CONSENT TO PARTICIPATE IN SQUAT RESEARCH

Title: **Effects of foot placement on muscular imbalance and musculoskeletal factors in the knee joint during the squat**

Principal Investigator: Sangwoo Lee……..……………………………….940-xxx-xxxx
Research Advisor: Young-Hoo Kwon, Ph.D.……………………………. 940-898-2598

Explanation and Purpose of the Research

You are being asked to participate in a research study for Mr. Sangwoo Lee's dissertation at Texas Woman's University. The purpose of this study is to investigate the effects of the foot placement on muscular imbalance and musculoskeletal factors (muscle and joint contact forces) in the knee joint during the squat. You have been asked to participate in this study because you identified yourself as a potential participant that meets the inclusion criteria for this study.

Description of Procedures

You will be asked to read, fill out and sign this informed consent prior to the initiation of the study in the PH 123, Biomechanics Laboratory in Texas Woman's University. The principal investigator will record your age, height and weight. You will then enter the Motion Analysis Laboratory (PH 124) and be requested to change into specific clothing required for biomechanical research –full body close fitting dark spandex material. A separate preparation room will be provided to change the clothes.

As a safety precaution, you will perform a warm-up (such as stretching and prior squat) on your own comfort level before the data collection to assure that the core body temperature is raised and the muscles are prepared. Warming up your connective tissues reduces the possibility of injury and the risk of premature fatigue.

All experimental procedures will be conducted in the Motion Analysis Laboratory (PH 124). A real-time motion capture system will be used to capture the motions of squat (reflective markers placed on your body). Data will be collected through the electromyography system (an instrument used to measure electrical activity produced by skeletal muscles) and force-plates (an instrument used to measure the forces generated by a body on standing or moving) during the squat.

Reflective markers (10 mm diameter) will be placed on specific parts of your body so that each part is properly defined on the camera system to track the motion being recorded. A total of 32 reflective markers will be placed: pelvis (4), greater trochanter (2), thigh (6), lateral epicondyles (2), medial epicondyles (2), shank (6), lateral malleoli (2), medial malleoli (2), heels (2), toes (2), and metatarsal (2). Electrodes (38 mm diameter) will also be placed on thigh muscles. A total of 9 surface electrodes will be placed: rectus femoris (2), vastus lateralis (2), vastus medialis (2), bicep femoris (2), and knee (1). There is a diagram attached to this consent form that shows exactly where the markers will be placed on your body.

When the motion capture system is ready for data collection, you will be asked to perform a static trial (standing in T-pose with arms spread out laterally) to locate your joint centers. After the capture of the static trial is over, you will be asked to do the squat trials with six different foot placement conditions based on the stance width and toe angle. Three stance width conditions will be used: narrow stance (NS, 75% of shoulder width), medium stance (MS, 100% of shoulder width), and wide stance (WS, 140% of shoulder width). In addition, two toe angle conditions will be used: toes forward (TF, pointing toes directly forward) and toes out (TO, pointing 30° out from the forward direction).

You will perform all the conditions in a randomized sequence to prevent order effects. A total of 30 squat repetitions (five repetitions in each trial condition) will be performed with a barbell placed on trapezius. The barbell will weigh 75 % of your one repetition maximum (maximum amount of weight one can lift in a single repetition for a given exercise). Each trial will be initiated by the word “Go” in order to control the pace of each participant. You will take a rest for approximately 3 minutes in between the squat repetitions to minimize the random error resulting from fatigue. If the rest period between each trial is not long enough, you may request a break at any time. The total time commitment for this single-session data collection is two hours, which includes time for reading and signing the consent form prior to data collection and the data collection itself.

Potential Risks

Loss of Confidentiality: You will be identified as a participant by a unique ID code. All computer files associated with you will be identified solely by this code and will not contain any identifying information. The Participant Master Identification Code Sheet will be kept separate from all other data collected and will be accessible only to the principal investigator and research assistants. No other identifying information will be collected.

Upon completion of data collection sessions, Participant Master Identification Code Sheet will be destroyed by its erasure. As computer files will not contain any identifying information, erasure of computer files is not considered necessary to protect confidentiality after the destruction of the master-cross reference list. All files are recorded and stored directly to the computer; therefore, will be erased from the computer and/or DVDs will be destroyed three years from the conclusion of the study. There is a potential loss of confidentiality through all email transactions. Confidentiality will be protected to the extent that is allowed by the law.

Muscle soreness: Potential muscle and joint soreness will be minimized by asking you to stretch and warm-up at the beginning of the data collection session. If muscle or joint soreness should occur, then you will be instructed through a post-event cool-down and stretching routine. If soreness persists, then you will be advised to seek medical attention. However, TWU does not provide medical services or financial assistance for injuries that might happen because of participant participation in the current research.

Coercion: Participant in this study is voluntary, and you may withdrawal at any time at your discretion.

Injury to a muscle or joint or risk of falling: Potential of injury will be minimized by allowing each participant adequate time for between the testing condition(s). If you feel uncomfortable with any of trials in testing condition(s) you are free to discontinue your involvement in this research project. In addition, research assistants will be strategically positioned around the squat rack to assist the participants in maneuvering down the squat descent. If an injury should occur all proper and necessary medical and/or first aid procedures will be followed as dictated by the type or extent of the injury.

Embarrassment: All data collection sessions will be attended by mixed genders; however, your preparation and placement of reflective markers will be handled by practitioners of the same gender. No other participants will be present during data collection sessions. The researchers will try to prevent any embarrassment issues that may occur prior to incident. You will be advised to let the researchers know at once if there is a problem or if you are uncomfortable. Each practitioner will be instructed to assist you in meeting your needs.

Fatigue: You will be allowed to rest in between trials if you feel tired. If you feel fatigued you have the right to stop the study completely at any time.

Skin irritation due to skin preparation: Prior to the data collection, you will be asked whether you have any skin allergies. This step is necessary to ensure good skin-reflective marker contact for adherence during motion trials. Erroneous results can occur if this step is not taken. Care will be taken in skin preparation to minimize this risk. You will be asked if you have any skin sensitivity and if you are sensitive to such treatments you cannot be recruited.

The researchers will try to prevent any problem that could happen because of this research. You should let the researchers know at once if there is a problem and they will help you. However, TWU does not provide medical services or financial assistance for injuries that might happen because you are taking part in this research.

Participation and Benefits

Participation in this study is voluntary and you are free to withdraw at any time without penalty. There is no monetary compensation for this study. If you would like to know the results of this study you can provide your mailing address, or email address, and they will be sent directly to you following completion.

Questions Regarding the Study

You will be given a copy of this signed and dated consent form to keep. If you have any questions about the research study you should ask the researchers; their phone numbers are at the top of this form. If you have questions about your rights as a participant in this research or the way this study has been conducted, you may contact the Texas Woman's University Office of Research and Sponsored Programs at 940-898-3378 or via e-mail at IRB@twu.edu.

_______________________________ _____________________

Signature of Participant Date

*If you would like to know the results of this study tell us where you want them to be sent:

Email: ___________________________

or

Address:

____

APPEMDIX B

1 REPETITION MAXIMUM TESTING PROTOCOL

Here is the protocol for 1 RM developed by the National Strength & Conditioning Association (Baechle & Earle, 2000).

1 RM TESTING PROTOCOL

1. Instruct the athlete to warm up with a light resistance that easily allows 5-10 repetitions.

2. Provide a 1-minute rest period.

3. Estimate a warm-up load that will allow the athlete to complete 3-5 repetitions by adding

- 10-20 lb (4-9 kg) or 5-10 % for upper body exercise
- 30-40 lb (14-18 kg) or 10-20 % for lower body exercise

4. Provide a 2-minute rest period.

5. Estimate a conservative, near-maximum load that will allow the athlete to complete 2-3 repetitions by adding

- 10-20 lb (4-9 kg) or 5-10 % for upper body exercise
- 30-40 lb (14-18 kg) or 10-20 % for lower body exercise

6. Provide a 2-4 minute rest period.

7. Make a load increase

- 10-20 lb (4-9 kg) or 5-10 % for upper body exercise
- 30-40 lb (14-18 kg) or 10-20 % for lower body exercise

8. Instruct the athlete to attempt a 1 RM.

9. If the athlete was successful, provide a 2-4 minute rest period and go back to step 7.

If the athlete failed, provide a 2-4 minute rest period, decrease the load by subtracting

- 5-10 lb (2-4 kg) or 2.5-5 % for upper body exercise
- 15-20 lb (7-9 kg) or 5-10 % for lower body exercise

And then go back to step 8.

Continue increasing or decreasing the load until the athlete can complete one repetition with proper exercise technique. Ideally the athlete's 1 RM will be measured within 5 testing sets.