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The Effects of Squat Depth on Lower Extremity Joint Kinetics, ACL Loading, PCL Loading, Tibiofemoral Compressive Forces and Activations of Eight Lower Extremity Muscles

Zachary A. Sievert
Old Dominion University

THE EFFECTS OF SQUAT DEPTH ON LOWER EXTREMITY JOINT KINETICS, ACL LOADING, PCL LOADING, TIBIOFEMORAL COMPRESSIVE FORCES AND ACTIVATONS OF EIGHT LOWER EXTREMITY MUSCLES

by

Zachary A. Sievert
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Approved by:

Joshua T. Weinhandl (Chair)

Laura Hill (Member)

Stacie Ringleb (Member)

ABSTRACT

THE EFFECTS OF SQUAT DEPTH ON LOWER EXTREMITY JOINT KINETICS, ACL LOADING, PCL LOADING, TIBIOFEMORAL COMPRESSIVE FORCES AND ACTIVATONS OF EIGHT LOWER EXTREMITY MUSCLES

Zachary A. Sievert
Old Dominion University, 2014
Director: Dr. Joshua T. Weinhandl

Performing resistance training exercises are important in many facets of life. Resistance training can be used from elite athletes preparing for the Olympics to the average person exercising for overall health. The squat is a common training exercise that can be utilized to improve lower extremity strength and performance. There has been much debate on the effects of squatting depth on lower extremity joint kinetics and kinematics. Therefore the purpose of this thesis is to examine the effects of depth and load on the muscle activation of eight lower extremity muscles and three-dimensional (3D) joint moments about the right knee, hip and ankle. The study will also examine the effects of load (BW, 50% 1RM and 80% 1RM loads) on anterior cruciate ligament loading, posterior cruciate ligament loading, and tibiofemoral compressive forces during 90° knee flexion squats.

Twenty-one recreationally active volunteers were recruited but only seventeen were used for analysis due to failure to complete the protocol or data processing difficulties. Recreationally active was defined as participating in exercise three times a week for at least 30 minutes per session and performing at least one session of lower body resistance training per week. Subjects were asked to perform two days of testing. During the first day of testing subjects one repetition maximum (1RM) was determined. On day two subjects performed five squats to knee flexion angles of 90° and 120° at loads

of body weight, 50% 1RM and 80% 1RM while right leg muscle activity, kinematics and kinetics were recorded. 90^{0} knee flexion squat were further analyzed with a musculoskeletal model to determine anterior and poster cruciate ligament loading as well as tibiofemoral contact forces.

It was found that PCL loading and tibiofemoral compressive forces had significant increases as external load increased during the 90° squats. During the testing all the muscles tested, the gluteus maximus, medial hamstring, lateral hamstring, rectus femoris, vastus medialis, vastus lateralis, medial gastrocnemius and tibialis anterior tested had increases in activation as external load increased from body weight to 50% 1RM to 80% 1RM. All muscles tested had increases in activations as condition changed expect for the medial hamstring and medial gastrocnemius from the 90° squat to the deep condition. All the reported moments, hip extention, hip external rotation, knee extension, knee external rotation, knee adduction, ankle plantarflexion and ankle inversion had significant increases as weight increased from body weight to 50% 1RM to 80% 1RM and as condition changed from the 90° squat to the deep condition.

Overall it may be suggested that participating in squat exercises can be beneficial. First, as load and depth increase muscle activation in the lower extremities increases which can increase strength and hypertrophy in those muscles. Along with the potential of increased muscle activation there seems to be no increased risk in damaging the ligaments, but the risk of cartilage damage remains unclear.

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NOMENCLATURE

EMG	Electromyography
MVIC	Maximal Voluntary Isometric Contraction
PCL	Posterior Cruciate Ligament
ACL	Anterior Cruciate Ligament
MCL	Medial Collateral Ligament
LCL	Lateral Collateral Ligament
1RM	One Repetition Maximum
BW	Body Weight

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Chapter 1: Development of the Problem

Background and Rationale

Performing resistance training exercises are important in many facets of life. Resistance training is used from elite athletes preparing for the Olympics to the average person exercising for overall health. A commonly performed resistance exercise is the squat. The squat can be defined as the coordinated actions of the torso and lower extremities to lower the body into a seated position and return to an upright position while keeping the shins and torso as upright as possible (Escamilla, 2001). In everyday life, squatting is a natural motion that occurs frequently as individuals kneel, sit down, etc. and requires a large range of knee motion. In athletics, the knee experiences large degrees of flexion while also experiencing increased forces and torques compared to normal everyday activities (Gullet, 2009).

The squat exercise is good for maintenance and improvement of muscles that support and move the knee. There are many variations of the squat exercise. The back squat is either performed to 90° of knee flexion or to maximum knee flexion (~120°), which is considered a deep squat. Deep knee flexion can cause damage to the knee (Escamilla et al., 1998). However, relatively little is known concerning the relationship between knee flexion angle, tibiofemoral contact forces, and knee ligamentous forces during the squat exercise.

The squat exercise affects knee ligaments during flexion and extension. The anterior cruciate ligament (ACL) is primarily loaded through the first 30° of knee flexion and last 30° of knee extension during the squat exercise. The posterior cruciate ligament (PCL) is primarily loaded during knee flexion (Amis, Gupte, Bull, & Edwards, 2006).

During the squat, stance can alter the strain placed on the knee ligaments (Klein, 1961). It was found that stance width did not change muscle activation in any thigh muscle except the gluteus maximus (Paoli, Marcolin, & Petrone, 2009).

However, there is disagreement in the literature about muscle activations and varying stance width. Paoli et al. (2009) did not find significant differences in muscle activation with stance width. On the other hand, Schwarzenegger (1985) argued stance width does change the activation of the vastus medalis. Furthermore, increasing knee flexion between 90° and maximum knee flexion can cause increased muscle activation of the medial and lateral hamstrings, vastus medialis, vastus lateralus, the rectus femoris, and the medial and lateral heads of the gastrocnemius. As knee flexion increases and muscle length of the quadriceps deviates from optimal length, there may be a need for greater activation levels to recruit more motor units to develop the same level of tension as at optimal length.

Muscle activity will increase as load increases (Boyden, Kingman, & Dyson, 2000; Newton et al., 1997; Paoli et al., 2009). As external load increases the amount of fibers that need to be recruited will increase, with greater needs of recruitment the greater amount of activation will take place (Clark, Lambert, & Hunter, 2012).

It is possible that continuous stretch of a muscle fiber may increase peak contractile force (Oatis, 2009). Therefore, it may be speculated that muscle fiber contractile force could be increased as the squat depth is increased over time as muscle fibers are continually stretched. Although it is possible for muscle fiber contractile force to increase when continuously lengthened, it has also been shown that muscle tension is minimal or non-existent when sarcomere length is stretched too far in an isometric contraction (Lieber & Bodine-Fowler, 1993). Therefore, there is an optimal length when muscle will produce the maximal amount of force. For example, it has been shown in cat soleus muscles, that at a 45° ankle angle the muscle could achieve the most amount of force with the least amount of activation (Rack & Westbury, 1969). As the ankle angle increased a higher stimulus rate was needed to produce more force. When the ankle reached 120° the soleus needed almost twice as much stimulus to produce a smaller force than the ankle at a 45° angle (Rack & Westbury, 1969).

Statement of the Problem

Research has shown that throughout the performance of a squat the length of the supporting ligaments of the knee joint will change (Amis & Dawkins, 1991; Amis et al., 2006). Additionally, muscle activation level increases as squat depth increases (Schoenfeld, 2010). With the increased muscle tension, the tibiofemoral contact and ligamentous forces are likely increased as well. However, there is limited research examining the changes in these forces between traditional squat exercises (90°of knee flexion) compared to deep squats (>120° of knee flexion). Research also does not agree on how squat depth affects changes in muscle activation of various lower limb muscles.

Statement of Purpose

The purpose of this study was to examine the effects of deep knee flexion squats on the muscle activation of eight lower extremity muscles and three-dimensional (3D) joint moments about the right knee, hip and ankle. It was also the goal of this study to examine the differences and compare the effects of deep knee flexion squats and squats performed at 90° knee flexion squats. The study will also examine the effects of load (BW, 50% 1RM and 80% 1RM loads) on anterior cruciate ligament loading, posterior cruciate ligament loading, and tibiofemoral compressive forces during 90° knee flexion squats.

Null Hypotheses

Load

There will not be significant increase in ACL loading, PCL loading and tibiofemoral compressive forces as the external load increases among the 90°squats.

Condition

Muscle activation patterns of the medial and lateral hamstrings, vastus medialis, vastus lateralus, rectus femoris, medial gastrocnemius and the gluteus maximus will not increase from deep squats compared to 90° squats. Next, there will not be significant increases in joint moments about the right knee, hip and ankle between the deep and 90° squats.

Research Hypotheses

Load

There will be significant increase in ACL loading, PCL loading and tibiofemoral compressive forces as the external load increases among the 90° squats.

Condition

Muscle activation patterns of the medial and lateral hamstrings, vastus medialis, vastus lateralus, rectus femoris, medial gastrocnemius and the gluteus maximus will significantly increase during the deep squats compared to 90° squats. There will be significant increase in muscle activations patterns of the medial and lateral hamstrings, vastus medialis, vastus lateralus, rectus femoris, medial gastrocnemius and the gluteus maximus as both deep and 90° increase from BW to 50% 1RM to 80% 1RM. There will be significant increases in three dimensional peak joint moments about the right knee, hip and ankle during deep squats compared to the 90° squats. There will be significant increase in three dimensional joint moments about the right knee, hip and ankle as both deep and 90° increase from BW to 50% 1RM to 80% 1RM.

Independent Variables

- Squat depth (90° of knee flexion; >120° of knee flexion).
- Squat load (0%, 50%, and 80% of each subject's 1RM).

Dependent Variables

- Peak right knee, hip and ankle joint moments of force
- Peak tibiofemoral compressive force
- Peak anterior cruciate ligament loading
- Peak posterior cruciate ligament loading
- Peak muscle activation

Limitations of the Study

- The subjects' ability to perform a 90° and deep squat.
- Fatigue.
- Subjects' reliability to give maximal effort.
- The subjects' squatting technique: stance width, foot alignment.
- Bar placement on the subjects' back.

Delimitations of the Study

- Subject demographics will be controlled, gender: ten males, ten females, and Age: 18-30.
- Subjects must resistance train legs using a back squat at least once a week.

- Clothes and shoes will be monitored and controlled.
- No weight belts will be permitted.

Assumptions of the Study

- Subjects will perform 90° degree and deep squats to the best of their abilities without stopping early.
- Subjects will not experience peripheral fatigue from the study protocol.
- During inverse dynamics it is assumed that all body segments are a perfectly rigid body.
- During inverse dynamics it is assumed the center of mass location is a fixed point for each segment.
- During inverse dynamics it is assumed that each joint is a frictionless pin joint.
- During inverse dynamics it is assumed that the moment of inertia remain constant during the movement.

Significance of the Study

Many studies, have been done on the effects of knee flexion on the ligament loading (Amis & Dawkins, 1991; Amis et al., 2006; Klein, 1961; Paoli et al., 2009). Furthermore, while the effects of squatting exercises on ligament loading up to 90° knee flexion are understood (Amis & Dawkins, 1991; Amis et al., 2006; Escamilla et al., 1998; Escamilla, Fleisig, Zheng, et al., 2001; Escamilla, Lander, & Garhammer, 2000; Schoenfeld, 2010), there have been few studies to directly compare squat mechanics between traditional and deep squats (Amis & Dawkins, 1991; Schoenfeld, 2010). Research is in disagreement on how squat depth impacts muscle activation (Bryanton, Kennedy, Carey, & Chiu, 2012; Caterisano et al., 2002; Clark et al., 2012; Escamilla, Fleisig, Lowry, Barrentine, & Andrews, 2001; Wretenberg, Arborelius, Weidenhielm, & Lindberg, 1993a). The results of this study can help strength and conditions coaches training athletes, recreational and competitive lifters performing squats for overall health, and health care providers (i.e., physical therapists) developing rehabilitation protocols for individual with injuries (e.g., ACL reconstruction, knee osteoarthritis, etc.).

Operational Definition of Terms

- Squat: the coordinated actions of the torso and lower extremities to perform the task of keeping the shins as near vertical as possible while keeping the torso upright and bringing the body into a sitting motion and returning back to an upright position.
- Squat depth: the degree of maximum knee flexion that the subject will perform during the squat exercise.
- Joint moment of force: rotational effect of an eccentric force on an object.
- Electromyography (EMG): an electrical recording of muscle activity.
- Kinetics: the study of the forces acting on an object.
- Kinematics: description of motion.
- Maximal voluntary isometric contraction (MVIC): the maximum isometric contraction a subject can voluntarily produce with a given muscle.

Chapter 2: Review of the Literature

Introduction

During a squat exercise there are several anatomical components that make the exercise possible. The knee joint, comprised of the articulation of the femur and the tibia, performs a great deal of flexion and extension which are essential to proper squat performance. Within the knee there are structures that help guide normal joint motion and transmit forces up the kinetic chain. The internal structures of the knee that are most influenced during a squat are the ACL, PCL and articulating cartilage. The ACL and PCL experience strain throughout knee flexion and extension phases as muscle forces cause translation of the tibia relative to the femur. Contact forces between the tibia and the femur must be considered in addition to ligament loading. The aim of this review of literature is to provide an understanding on the effects of squatting depth on ACL and PCL strain, as well as tibiofemoral compressive forces.

Squatting can be described as the coordinated actions of the torso and lower extremities to perform the task of keeping the shins as near vertical as possible while keeping the torso upright and bringing the body into a sitting motion and returning back to an upright position. With this scenario the squat can be considered a closed kinematic chain (Escamilla, 2001) which means the femur is rotated about a nearly fixed tibia. The back squat exercise is usually performed in two variations: 1) the half squat performed from 0 to 90° knee flexion, and 2) the full squat performed from 0° of flexion until the individual cannot flex the knee anymore (Escamilla, 2001).

The knee joint is a vital anatomical component in human movement. It provides necessary flexion and extension of the lower limb to perform squat exercises and daily activities of daily living. The extensor muscles that cross the knee joint, primarily the quadriceps, must safely absorb the load through eccentric contraction during the negative flexion phase and then raise the load through concentric contraction during the positive extension phase. As such, they must develop high levels of tension that would in turn cause anterior translation of the tibiofemoral joint, placing strain on the ACL. During the squat, knee flexion takes place from the muscles on the posterior aspect of the thigh, primarily the hamstrings. During this negative phase of the squat the hamstrings contract causing knee flexion while during the positive or upward phase of the squat the hamstrings will contract to help faciltate hip extension with the gluteus maximus. The action of the hamstrings during the positive and negative phases of the squat will cause posterior translation of the tibia relative to the femur placing strain on the PCL. These high muscle forces from both the quadriceps and the hamstrings will also cause compression of the tibiofemoral joint and yield large tibiofemoral contact forces.

It is recommended that squats be performed to a parallel depth (Escamilla, 2001). This recommendation is thought to come from safety issues with squat depth and the potential negative effects it will have on the joints. Although many consider squat depth below 100° dangerous, there are some disagreements among researchers whether the benefits of deep squats outweigh the injurious possibilities. To train the knee extensors (i.e., quadriceps) to their maximum ability, a deeper squat depth or greater knee flexion is needed (Bryanton et al., 2012). The greater knee flexion is needed to generate greater knee extensor muscle effort, but this may in turn place greater stress on the knee joint (Bryanton et al., 2012). During the squat exercise, the extensor muscles of the hip and knee perform large amounts of work. However, as load increases the net hip extensor moment of force increases more than the net knee extensor moment of force (Flanagan & Salem, 2008). Conversely, the net knee joint moment of force increases more than the net hip joint moment of force when squat depth is increased (Wrentenberg, 1993).

Anatomy of the knee

The knee is a large and complex joint composed of many bones and ligaments that must work together to create normal joint motion and provide functional stability. The knee can be classified as a condylar joint by its ability to perform primarily flexion and extension, with internal and external rotation occurring as a result (Hamill, 2009). Frontal plane abduction and adduction occur as well, although these motions are not considered natural motions of the knee.

The knee is composed of three main bones; the femur, tibia, and patella (Tortora, 2012) The boney anatomy of the knee is designed for easy articulation between the tibia and femur to create movement. The tibial plateau is formed by the medial and lateral tibial condyles, which allow the epicondyles of the femur to rest and articulate. These tibial condyles are separated by an intercondylar notch. The groves of the medial and lateral condyles of the tibia and the femoral epicondyles allow motions at the knee, such as flexion and extension. This tibiofemoral space and differing femoral epicondyle radii of curvature permits small amounts of rotation.

There are four main ligaments that help stabilize the knee and guide normal joint motion. These ligaments can be further separated into collateral and cruciate ligaments. The lateral and medial collateral ligaments are extra-capsular. The medial collateral ligament attaches at the medial aspect of the medial condyle of the femur and inserts to the medial condyle of the tibia (Tortora, 2012). The medial meniscus attaches to the medial collateral ligament. The lateral collateral ligament attaches to the lateral aspect of the condyle of the femur and inserts on the lateral aspect of the fibular head (Tortora, 2012). The medial collateral ligament protects the knee against adduction movements, while the lateral collateral ligament helps protect the knee from abduction movements. The anterior and posterior cruciate ligaments are intra-capsular. The ACL attaches to a position anterior to the intercondylar notch of the tibia and extends posteriorly and laterally to the posterior portion of the medial surface of the lateral condyle of the femur, which helps against anterior translation of the tibia against the femur (Tortora, 2012). The PCL attaches to the posterior aspect of the intercondylar notch and moving anteriorly and medially to the anterior portion of the lateral surface of the medial condyle of the femur. The PCL helps against posterior translation of the tibia against the femur. (Tortora, 2012).

Finally, intra-capsular cartilaginous tissues including the articulating cartilage and the menisci help aid in movement. Articular cartilage on the femoral condyles and tibial plateaus, along with synovial fluid, help reduce friction between the bones permitting movement with minimal resistance. The menisci are positioned between the femur and tibia and act to increase the contact area between the tibia and femur, thereby, dispersing the force between the tibia and femur.

Anterior Cruciate Ligament

It is estimated that between 80,000-250,000 ACL injuries and 100,000 ACL reconstructions are performed annually in 15-25 year olds (Griffin et al., 2006). Research has shown that neuromuscular fatigue may be a contributing factor in non-contact ACL injury (Chappell et al., 2005). Muscular fatigue can cause changes in biomechanics and muscle responses. Furthermore, fatigue has been associated with excessive external rotation which can increase ACL loading (Schoenfeld, 2010). During a normal squat, the ACL is mainly stressed during the first 30°. However, when a subject is fatigued ACL loading is increased throughout the whole squat (Schoenfeld, 2010).

The ACL is responsible for countering anterior translation and medial rotation of the tibia relative to the femur. The ACL is comprised of two different components, an anteromedial and posterolateral bundle. Amis and Dawkins (1991) found there to be a third, intermediate bundle separating the posterolateral and the anteromedial bundles. However, the focus here will be on the two dominate ACL portions, the anteromedial and posterolateral bundles.

The anteromedial and posterolateral bundles of the ACL have different functions to help support knee during flexion and extension, as well as during rotation. For example, the posterolateral bundle is tight or tense during knee extension while the anteromedial is tense during knee flexion. This suggests that the anteromedial bundle stabilizes the knee against anterior movement and rotation when the knee is flexed while the posterolateral bundle stabilizes the knee at full extension (Kato et al., 2012). Thus, as the knee flexes and extends during a squat, loading of the two bundles will vary with length (Yoo et al., 2010). In the ACL, the anteromedial bundle is smaller than the posterolateral. In the first 30° of knee flexion, both bundles shorten and from there each bundle performs separate functions. The anteromedial bundle increases in length after 30° of knee flexion until 120° of flexion. The posterolateral bundle continues to decrease in length until 90-100° of flexion and then starts to increase in length (Amis & Dawkins, 1991). During the squat exercise, the ACL is primarily being strained during the first 60° of knee flexion (Escamilla, Fleisig, Zheng, et al., 2001).

Posterior Cruciate Ligament

The PCL is a thicker and stronger ligament compared to the ACL. The PCL is responsible for countering the posterior translation of the tibia relative to the femur. The PCL lengthens during flexion and loosens or slackens during extension. The PCL has an origin on the posterior tibia below the articular surface. The PCL attaches to the distal end of the articular cartilage of the medial femoral condyle. Some research has shown the attachment site of the PCL can vary between people (Amis et al., 2006). Similar to the ACL, the PCL can be further separated into two bundles, an anterolateral bundle and a posteromedial bundle. The anterolateral bundle is stronger and thicker than the posteromedial bundle. The anterolateral bundle attaches to the intercondylar notch and is loose during knee extension while tightening during flexion. However, during hyperflexion of the knee, the anterolateral bundle becomes less effective at protecting against posterior tibial translation. The posteromedial bundle attaches to the medial femoral condyle, and attaches proximal to the anterolateral bundle. The posteromedial bundle is tight when the knee is extended and loosens as the knee flexes. As the knee goes to hyperflexion the posteromedial bundle stretches again.

Tibiofemoral Forces

Force generated from the muscles surrounding the knee cause translation and compression in the joint, which generates tibiofemoral compressive force. Many factors, such as an individual's weight, sex, and activity being performed, influence the amount of compressive force (Taylor, Heller, Bergmann, & Duda, 2004). Experimentally measuring compressive force or tibiofemoral joint reaction force in the knee is difficult. There have been numerous computer simulations and mathematical equations that have estimated tibiofemoral contact force (Taylor et al., 2004). Wilk et al. (1996), Sahli, Rebai, Elleuch, Tabka, and Poumarat (2008) and Taylor et al. (2004) indicated that compressive forces increased as the knees flexed and decreased as the knee extended. On average, forces were slightly greater during ascent compared with descent of the squat exercise.

There have been numerous researchers who have examined the effects of squat depth on tibiofemoral shear and compressive forces (Escamilla, 2001; Wilk et al. (1996)), (Escamilla et al. (2000); Sahli et al. (2008)). Wilk et al. (1996) reported tibiofemoral compressive forces are larger in closed kinetic chain movements compared to open kinetic chain exercises. In this study, it was found that the compressive forces reached a maximum of 6139 ± 1708 N and the peak compressive forces accumulated between 89 to 95° of knee flexion during the eccentric portion of the squat. Maximum reported knee flexion was 115°.

Furthermore, Wilk et al. (1996), also found posterior shear force was generated during the entire squat exercise. They defined posterior shear force as the force displacing the tibia posteriorly relative to the femur. Maximal posterior shear force occurred at 88 to 102° of knee flexion and produced 1783 ± 634 N. However, in the study, subjects only used their 12-repetition maximum for their prescribed load. Although, Wilk et al. (1996) found peak compressive force at 85-95° knee flexion, a study by Escamilla (2001) found that their subjects had the most compressive forces at 130°.

Electromyography

There have been many studies conducted on muscle activity during squatting (Bryanton et al. (2012); Caterisano et al., 2002; Clark et al. (2012); Escamilla, Fleisig, Zheng, et al., 2001; Westing, Cresswell, and Thorstensson (1991)) but much of the research has opposing viewpoints on the different effects of squat depth on muscle activations. This portion of the literature review will focus on how past research was conducted using deep knee flexion squats with external loads and the effects on muscle activity.

Muscle can be broken down into many smaller components that comprise the whole system. Muscle is made up of many different fascicles, which contain multiple muscle fiber cells. Muscle fiber cells can be separated further into myofibrils, which then can be broken down into multiple sarcomere containing actin and myosin filaments. Muscle is activated or stimulated through nervous activity. Neural activity to the muscle comes from motor neurons that innervate the muscle fibers. The motor neuron and all the fibers a neuron innervates is considered the motor unit. The signal an EMG probe receives is the summation of all the electrical signals being activated throughout the muscle belly in range of the EMG sensor.

Muscle fiber length is important to the abilities of a muscle. Some fibers have the potential to run the entire muscle length, where as some other fibers may be short and only be located in the proximal, distal, or middle of the entire muscle. It has been suggested by Kamen and Gabriel (2010) that as muscle fiber length increases, the muscle fiber conduction velocity will decrease. As the muscle lengthens and the conduction velocity decreases, the amplitude of the EMG signal can decrease as well with passive stretching of the muscle (Kamen & Gabriel, 2010). As external load increases and additional recruitment of muscle fibers is needed, there will be more activation of motor units. There is reduced EMG amplitude during the eccentric phase of actions (Westing et al., 1991). As muscle fiber length changes from optimal length, more muscle activation is required to contract the muscle as external load stays constant. Although Kamen and Gabriel (2010) have shown that muscle activation will change as muscle length changes, other studies have shown the opposite. For example, Clark et al. (2012) concluded that muscle activity does not change from varying squat depths at moderate loads. Clark, Lambert, and Hunter (2012) suggest that as the muscles are stretched to different lengths, there was no difference in activation.

It is important to have some understanding of how muscle may respond when stretched to different lengths. A study conducted by Rack and Westbury (1969), in which a cat soleus was used, states that stimulus rates at different lengths will have different effects on the muscle tension. They found that when the muscle was deviated from its optimal length more stimuli were needed to produce the same tension than when at its optimal length. As muscle is stretched past its optimal length, it may require more stimuli to produce the same force. This can be applied to human muscle during the squat exercise. As the quadriceps and hamstrings are stretched during the squat exercise, muscle activation may change.

Others studies have focused on how muscle activation changes as external load changes. It has been shown that as external load increases muscle activity increases (Clark et al., 2012). It has also been shown that the quadriceps activity can increase by 20% as external load increases from 60 to 75% of a subject's 1 RM (Clark et al., 2012).

One study by Caterisano et al. (2002) looked at EMG of thigh muscles during squats with loads as percentages of body weight and found muscle contributions as a percentage of total EMG of all thigh muscles during a 90°, parallel and full squat. Although the study looked at submaximal levels of external load, they found no muscle had a significant difference between any squat depths. The exception was that the vastus medialis decreased in activity level from parallel to full squats. The only muscle found to significantly increase in activation was the gluteus maximus. The book *Strength Training for Coaches* (Pauletto, 1991) suggests a deeper squat will activate the hamstrings more than a partial squat. Although Pauletto (1991) states that deeper squats will active the hamstrings more, he also states it is unnecessary for knee flexion to exceed parallel thighs to the ground. However, Pauletto (1991) does not provide an explanation for this theory. Another study using 65% of the subjects' 1 RM found EMG was significantly different between 90° and parallel, but not parallel to deep for the vastus lateralis and rectus femoris. The biceps femoris, however, had significant increases between 90° and parallel, but not parallel and deep (Wrentenberg, 1993).

Bryanton et al. (2012) performed a study in which relative muscular effort was determined from the ratio of net joint moment to maximum voluntary isometric contraction. They found the maximum isometric strength in relation to joint angles for all muscle groups used. Bryanton et al. (2012) reported that barbell load had minimal effects on knee extensor relative muscular effort compared to the squat depth which had greater effects on relative muscular effort. It was reported that greatest relative muscular effort was between 105° and 119° of knee flexion.

Clark et al. (2012) discussed the need to account for variation in EMG signaling and suggested the best way to normalize the EMG signal is to have a reference point. In this review, Clark et al. (2012) stated that it may be beneficial to compare EMG signals of the activity to the EMG signals of a maximal isometric contraction. It was also noted that other researchers have been comparing there EMG signals to submaximal loads. In one study, EMG normalization was found to be most appropriately normalized to more dynamic methods (Albertus-Kajee, Tucker, Derman, & Lambert, 2010). Although the study used a different approach to normalizing EMG during cycling, the study may show evidence that more dynamic movements need to be normalized against dynamic movements.

A standard inverse dynamics calculation can be used to determine moment of force about a joint to estimate forces generated by muscles causing the movements during activates. This joint moment represents the sum of all the muscles acting on that joint (Erdemir, McLean, Herzog, & van den Bogert, 2007). Inverse dynamics is used in biomechanics to estimate the mechanical effort used to perform a movement (Bryanton et al., 2012). However, from inverse dynamics alone one cannot directly calculate the force generated by each individual muscle. Instead, individual muscle forces during movements can be calculated from static optimization. To do so, first inverse dynamics from experimental data calculates joint moments. Then a musculoskeletal model is developed to demonstrate the muscle properties. Next, a linear or non-linear algorithm in conjunction with the musculoskeletal model is used to estimate the individual muscle forces. There are other forms of optimization available within biomechanics that can be used to determine muscle force generated during movements. For example, dynamic optimization, or other alternative optimizations such as EMG-based models or computed muscle control (CMC) can be used (Erdemir et al., 2007). Although static, dynamic, and alternative optimization can give values to muscle force during human movements, there are limitations to the information that optimization algorithms can provide. A major limitation with static optimization is that during a movement the optimization algorithms tend to only calculate the agonist muscle forces at a joint. Whereas in real life there are agonist and antagonist muscles forces acting on a joint. This limitation can be overcome by using a dynamic optimization routine, which will predict coactivation of agonist and antagonist muscles. Dynamic optimization routines are timely and can be computationally inefficient. Alternative optimizations, such as CMC, may offer a better option. CMC can be used with OpenSim (v3.1, https://simtk.org), an open source musculoskeletal modeling software system which allows for the user to make three dimensional (3D) musculoskeletal models of human subjects from previously recorded motion capture data (Delp et al., 2007). OpenSim also allows users to implement tendons and ligaments on the bones of their musculoskeletal models. This will allow for an estimation of ligamentous loading and joint contact forces during human movement. Although dynamic optimization can establish agonist and antagonist muscles activation, static optimization has been shown to be just as valid when computational time starts to exceed appropriate time allowances for optimizations (Anderson & Pandy, 2001). The estimation of muscle forces for this study can be accomplished through static optimization (Anderson & Pandy, 2001). To estimate joint compressive forces a joint reaction analysis was conducted as established by Steele, Demers, Schwartz, and Delp (2012).

Conclusion

There are many aspects of the squat to consider when examining the kinetics and kinematics of the squat exercise. Escamilla et al. (2010) and Schoenfeld (2010) report that there is no additional benefit for knee flexion past parallel thigh level because there is possible risk of knee ligament injury. However, neither researcher showed evidence to prove that going past parallel thigh level will in fact cause harm to the knee.

There are still many questions that need to be addressed regarding squatting mechanics. How does squat depth affect muscle activation at varying knee flexion angles? How does a full squat, with knee flexion at 130°, affect the ligaments and cartilage compared to 90° of flexion if the same subject is under external loads? There has been research conducted on the tibiofemoral contact forces as knee flexion reaches 130-150° of flexion, but the research is lacking on the how injurious these actions may be. The primary reason for the difficulties in measuring tibiofemoral contact forces at these high degrees of knee flexion is because researchers have to use invasive procedures to obtain exact tibiofemoral contact force measurements (Escamilla, 2001). Although it is difficult to determine ACL and PCL loading and the compressive and shear tibiofemoral joint forces, it is possible to estimate these forces through computer simulations and muscle modeling.

Chapter 3: Methods

Subjects

There were 21 participants in this study, but 4 were excluded. The excluded participants were either unable to perform the required squatting conditions or the static optimization of their model could not be performed. Ten men and seven women (average age: 22.8±1.88 yrs, height: 1.69±0.06 m, mass: 73.5±12.72 kg) who were recreationally active, free of injury within the last six months and had no history of lower extremity surgery were recruited for the study. Recreationally active was defined as participating in exercise three times a week for at least 30 minutes per session and performing at least one session of lower body resistance training per week. The 21 subjects were recruited based on an *a priori* power analysis that detected a two N/kg change in ACL loading with a α level of 0.05 and 80% power (Weinhandl et al., 2013). Thirteen total subjects were sufficient to meet the requirements, but additional subjects were collected for possible dropouts. Each subject had at least six months of previous experience with proper squatting technique. Prior to data collection, research approval from the Institutional Review Board of Old Dominion University and written informed consent from all subjects was obtained.

Experimental Protocol

Written informed consent as well as health history and activity level questionnaires were obtained from each subject. The questionnaires ensured all subjects met each of the inclusionary requirements to participate. The questionnaire was used to evaluate each subject for activity level, injury history, surgical history, and any medical condition. The subject had their height and weight recorded on testing days. Each subject was required to wear tight form fitting spandex shorts, and a sports bra or form fitting shirt. The testing protocol was split into two days separated by at least 48 hours and no more than seven days.

All testing was completed in the Neuromechanics Lab (ODU, Norfolk, VA, USA). On the first day of testing subjects reported to the laboratory for 1 repetition maximum (1 RM) testing. Once the subject dressed in proper clothing, they were given a pair of lab shoes (Air Max Glide, Nike, Beaverton, OR, USA). The subject performed a dynamic warm up which consisted of riding on a braked cycle ergometer and performing warm up squats at a load determined by the subject. Subjects chose to stretch at their discretion during the warm up. The subject announced when they felt adequately warm to begin 1 RM testing. 1 RM testing was administered from the protocol advised by Baechle and Earle (2008). The protocol called for the subject to warm up and then rest for at least one minute. After resting subjects performed three to five reps at a weight they were comfortable squatting. The subjects rested for two minutes, where 30-40 pounds were added to the previous lad. After two to four minutes of rest subjects performed two to three repetitions of the new load. Again the subject rested two to four minutes while another 30-40 pounds or an estimated load were added to the previous load so the subjects were near their 1 RM. Subjects were instructed to perform one repetition at each new load until the subjects 1 RM was determined. After resting two to four minutes the subject performed a squat at the new load for 1 repetition. When the subject was successful at the repetition, five to twenty pounds was added to the previous load. The subject was instructed to rest again for two to four minutes. The subject continued to increase the load until the subject can no longer successfully complete a 1 RM. If the subject was unsuccessful, the load was decreased by five to ten pounds and another 1 RM attempt was completed. The load will continually be decreased until the subject has successfully completed a 1 RM. After the subject completed the 1 RM, it was required for the subject to stretch the lower extremities and lower back. Stretching was key after performing a 1 RM because stretching after exercise can help reduce muscle soreness and help recovery (Baechle & Earle, 2008). Once all requirements were completed for day one testing subjects were scheduled for a new time to complete the second day of testing. The second session was scheduled with at least 48 hours of rest and within seven days of completion of the first session. The subject was required to restrain from performing lower extremity tasks outside normal daily activities, in particular long distance running and resistance exercises of the lower extremities.

On the second day of testing, the subjects wore spandex shorts, a sports bra or form fitting shirt if applicable, and lab shoes. Once the subject was dressed appropriately the researcher placed eight EMG surface electrodes on the subject's muscle bellies of the right rectus femoris, vastus medialis, vastus lateralis, medial hamstring, lateral hamstring, gluteus maximus, medial gastrocnemius, and tibialis anterior. Before the electrodes were placed on the muscle bellies, hair was removed from the skin from the electrode placement site, along with abrasions to remove epithelial cells and then the area was cleaned with alcohol. The electrodes were placed on each muscle belly by the guidelines set by Cram, Kasman, and Holtz (1998). For the gluteus maximus, the electrode was placed midway between the greater trochanter and the sacrum. The rectus femoris electrode was placed halfway between the knee and the anterior inferior ridge of the iliac spine. The vastus lateralis was placed on the muscle belly at an oblique angle laterally from the midline of the thigh. The vastus medialis had an electrode placed on the muscle belly above the knee cap and placed at an oblique angle medially from the midline. The medial and lateral hamstrings had an electrode place on each muscle belly half way between the gluteal fold and the back of the knee. The medial gastrocnemius had an electrode placed on the muscle belly at an oblique angle medial to the midline so that the electrode is perpendicular to the fibers. The tibialis anterior had an electrode placed on the muscle belly laterally from the tibia and one third of the distance down from the down from the knee. Once all electrode were placed on the desired muscles, they were secured with pre wrap (Mueller Sports Medicine, Prairie du sac, WI) and then secured again with athletic tape (Collins Sports Medicine, Raynham, MA). The subject had 18 retro reflective markers placed bilaterally on anatomical landmarks which included: the acrimoclavicular joint, the most superior aspect of the iliac crest, the greater trochanter, and the lateral and medial epicondyles of the femur, the lateral and medial malleoli, and the first and fifth metatarsals. Once the anatomical markers were placed on the skin, two elastic Velcro straps (McDavid, Woodridge, IL, USA) were placed on each leg, one on the middle of the thigh, and the second on the middle of the shank. These Velcro straps held tracking plates, with four tracking markers. The tracking plates were made from semi-rigid, molded Orthoplast (Johnson & Johnson, Raynham, MA, USA). Next tracking plates were placed on the heel of each shoe, on the subjects back which were placed in the middle of the scapulae, and a pelvis tracking plate placed on an elastic Velcro strap around the waist.

Once all markers, plates, and electrodes were secured properly and in working condition the subject completed maximum voluntary isometric contractions (MVIC) for each muscle. They were then asked to stand on the laboratory force plates with one foot on each plate while a static calibration trial was recorded. Once the calibration trial was completed the anatomical landmark markers were removed. The subject was instructed to warm up the same way as the first session. When the subject was adequately warmed up, they performed a series of squats to $90^{\circ} \pm 10$ of knee flexion and a series of deep squats ($120^{\circ} \pm 10$). Five body weight squats were performed for each 90° and deep conditions. The subject then performed five squats at 50% of their established 1 RM for the 90° and deep conditions. The subject then completed five repetitions of 80% of their established 1 RM for the 90° and deep conditions. The subjects were required to rest for two to four minutes between each of the 90° and deep squats at each load.

Instrumentation

Three-dimensional marker coordinate data were collected at 200 Hz using an eight-camera motion analysis system (Vicon, Centennial, CO). All force data were collected at 2000 Hz using two Bertec force plates (Bertec Co., Columbus, OH). All EMG data were collected at 2000 Hz using a Trigno 16 channel wireless surface electrode electromagnetic system (Delsys Inc., Boston, MA).

Data Analysis

A kinetic model made of eight segments of a trunk, pelvis, right and left thigh, shank and feet was created from the standing calibration trial (Weinhandl, Joshi, & O'Connor, 2010). Data were reduced with Visual 3D (v5.00, C-Motion Inc., Rockville, MD). Marker data and ground reaction force data were filtered through a low-pass fourth order Butterworth zero lag filter with 12 Hz and 50 Hz cut off frequencies. Three-dimensional ankle, knee, and hip angles were calculated using a joint coordinate system approach (Grood & Suntay, 1983). The hip joint centers were defined as 25% of the distance from the ipsilateral to the contralateral greater trochanter markers (Weinhandl & O'Connor, 2010). The knee joint centers were defined as the midpoint between the lateral and medial markers on the condyles of the femur (Grood & Suntay, 1983). The ankle joint centers were defined as the midpoint between the medial and lateral malleoli. The x axis of the pelvis (flexion and extension) was defined as a vector between the iliac crest markers. The frontal plane of the pelvis was defined as a plane of best fit between the iliac crest and greater trochanter markers. The y axis of the pelvis (adduction and abduction) was defined as the vector that is orthogonal to the frontal plane. The z axis (internal and external rotation) was defined as the vector orthogonal to the x and y axes. The x axis of the thigh (flexion and extension) was defined as a vector from the medial to lateral epicondyle markers. The frontal plane of the thigh was defined as a plane containing the x axis and the hip joint center marker. The y axis of the thigh (adduction and abduction) was defined as the vector that is orthogonal to the frontal plane. The z axis (internal and external rotation) was defined as the vector orthogonal to the x and y axes. The x axis of the shank (flexion and extension) was defined as a vector from the medial to lateral malleoli markers. The frontal plane of the shank was defined as a plane of best fit between the epicondyles and malleoli markers. The y axis of the shank (adduction and abduction) was defined as the vector that is orthogonal to the frontal plane. The z axis (internal and external rotation) was defined as the vector orthogonal to the x and y axes. The x axis of the foot (flexion and extension) was defined as a vector from the medial to lateral malleoli markers. The frontal plane of the foot was defined as a plane of best fit between the malleoli markers and the metatarsal markers. The y axis of the foot (adduction and abduction) was defined as the vector that is orthogonal to the frontal plane. The z axis (internal and external rotation) was defined as the vector orthogonal to the x and y axes. Three-dimensional joint kinetics were calculated using a Newton-Euler approach (Bresler & Frankel, 1950) with body segment parameters estimated from De Leva (1996) and reported in the distal segment reference frame. EMG data were pre-amplified and high-pass filtered using fourth-order, zero lag, recursive Butterworth filter with a cutoff frequency of 20Hz to remove movement artifact. The signal was full-wave rectified and normalized to maximize recorded signal of each muscle over the trial. The full-wave rectified signal was then be low pass filtered with a cutoff frequency of 5 Hz to create a linear envelope. EMG were then normalized against MVICs of the individual muscles. Matlab (MathWorks, Natick, MA) was used to post process EMG data and extract discrete kinetic variables.

Opensim (v3.1, SimTK, Stanford, CA) was used to compute estimated joint reaction forces and estimated muscle forces throughout each squat. To estimate muscle forces static optimization was employed with a cost function to minimize the sum of squared muscle activations (Anderson & Pandy, 2001). Joint reaction forces were estimated using the JointReaction analysis tool in OpenSim (Steele et al., 2012). Finally, ACL and PCL loading were estimated using a previously established knee model (Laughlin et al., 2011; Weinhandl et al., 2013).

Statistical Analysis

A two by three (condition (normal and deep) by load (BW, 50%, and 80%)) repeated measures ANOVA was used to test the statistical significance of the dependent variables. If two-way interactions were found, independent and paired t-tests were implemented. All statistical analyses was conducted using SPSS (v21.0, SPSS Inc., Chicago, IL).

Chapter 4: Results

Moments

Subjects produced a peak hip extension moment of -0.60±0.16 $Nm\cdot kg^{-1}\cdot m^{-1}$, -1.29±0.26 $Nm\cdot kg^{-1}\cdot m^{-1}$, -1.67±0.28 $Nm\cdot kg^{-1}\cdot m^{-1}$ for the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced a peak moment of -0.75±0.21 $Nm\cdot kg^{-1}\cdot m^{-1}$, -1.53±0.27 $Nm\cdot kg^{-1}\cdot m^{-1}$, and -1.97±0.36 $Nm\cdot kg^{-1}\cdot m^{-1}$ (Figure 1). The statistical analysis indicated a significant interaction of weight*condition (p=0.001). Post hoc analyses revealed that all comparisons were significantly different (p<0.001).

During the study it was found that there were peak hip adduction moments of 0.08±0.07$Nm\cdot kg^{-1}\cdot m^{-1}$, 0.16±0.10$Nm\cdot kg^{1}\cdot m^{-1}$, 0.20±0.13$Nm\cdot kg^{-1}\cdot m^{-1}$ for the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced a peak moment of 0.14±0.08$Nm\cdot kg^{-1}\cdot m^{-1}$, 0.29±0.14$Nm\cdot kg^{-1}\cdot m^{-1}$, and 0.41±0.22 $Nm\cdot kg^{-1}\cdot m^{-1}$ (Figure 2). There was a significant main effect for weight by condition (p<0.001) and a significant interaction between weight*condition (p=0.001). There were significant differences between 90° 80 and deep 80 (p<0.001), 90° 50% and 90° BW (p<0.001), 90° 80% and 90° BW (p<0.001), 90° 80% and 90° 50% (p<0.001), 90° 50% and 90° BW (p<0.001), 90° 80% and 90° BW (p<0.001), deep 50% and deep BW (p<0.001), deep 80% and deep BW (p<0.001), deep 80% and deep 50% (p=0.002).

There were peak hip external rotation moment of -0.07±0.04 $Nm\cdot kg^{-1}\cdot m^{-1}$, -0.08±0.05$Nm\cdot kg^{-1}\cdot m^{-1}$, -0.11±0.07 $Nm\cdot kg^{-1}\cdot m^{-1}$ for the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced a peak moment of -0.16±0.07 $Nm\cdot kg^{-1}\cdot m^{-1}$, -0.17±0.09 $Nm\cdot kg^{-1}\cdot m^{-1}$, and -0.21 ± 0.11 $Nm \cdot kg^{-1} \cdot m^{-1}$(Figure 3). There was a significant main effect for both weight and condition ($p<0.001$) but there was no significant interaction within weight*condition ($p=0.834$)

It was found that there were peak knee extension moments of 0.62 ± 0.13 $Nm \cdot kg^{-1} \cdot m^{-1}$, 0.91 ± 0.19 $Nm \cdot kg^{-1} \cdot m^{-1}$, 1.07 ± 0.22 $Nm \cdot kg^{-1} \cdot m^{-1}$ for the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced a peak moment of 0.59 ± 0.13 $Nm \cdot kg^{-1} \cdot m^{-1}$, 1.00 ± 0.20 $Nm \cdot kg^{-1} \cdot m^{-1}$, and 1.17 ± 0.26 $Nm \cdot kg^{-1} \cdot m^{-1}$ (Figure 4). There was a significant interaction between weight*condition ($p=0.043$). There was significant difference between and 90° 80% and deep 80% squats ($p=0.003$). The post hoc analysis showed that 90° 50% and 90° BW, 90° 80% and 90° BW, 90° 80% and 90° 50%, deep 50% and deep BW, deep 80% and deep BW, deep 80% and deep 50% all had significant differences in hip extension moments ($p<0.001$).

During the 90° BW, 50% and 80% squats, there was a peak knee adduction moment of 0.07 ± 0.06 $Nm \cdot kg^{-1} \cdot m^{-1}$, 0.06 ± 0.06 $Nm \cdot kg^{-1} \cdot m^{-1}$, 0.07 ± 0.08 $Nm \cdot kg^{-1} \cdot m^{-1}$ and during the deep BW, 50% and 80% squats subjects produced a peak moment of 0.18 ± 0.11 $Nm \cdot kg^{-1} \cdot m^{-1}$, 0.20 ± 0.14 $Nm \cdot kg^{-1} \cdot m^{-1}$, and 0.25 ± 0.19 $Nm \cdot kg^{-1} \cdot m^{-1}$ (Figure 5). There was a significant main effect for condition ($p<0.001$) and no significant main effect or interaction for weight ($p=0.080$) and weight*condition ($p=0.104$).

Peak knee external rotation moments of -0.05 ± 0.02 $Nm \cdot kg^{-1} \cdot m^{-1}$, -0.08 ± 0.02 $Nm \cdot kg^{-1} \cdot m^{-1}$, -0.011 ± 0.04 $Nm \cdot kg^{-1} \cdot m^{-1}$ occurred during the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced a peak moment of -0.06 ± 0.02 $Nm \cdot kg^{-1} \cdot m^{-1}$, -0.10 ± 0.02 $Nm \cdot kg^{-1} \cdot m^{-1}$, and -0.12 ± 0.04 $Nm \cdot kg^{-1} \cdot m^{-1}$

(Figure 6). There was significant main effects for weight ($p<0.001$) and condition ($p=0.001$), but there was no significant interaction between weight*condition ($p=0.393$).

Subjects experienced peak ankle plantar flexion moments of -0.20 ± 0.05 $Nm\cdot kg^{-1}\cdot m^{-1}$, -0.42 ± 0.10 $Nm\cdot kg^{-1}\cdot m^{-1}$, -0.67 ± 0.14 $Nm\cdot kg^{-1}\cdot m^{-1}$ for the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced an average peak moment of -0.23 ± 0.08 $Nm\cdot kg^{-1}\cdot m^{-1}$, -0.51 ± 0.13 $Nm\cdot kg^{-1}\cdot m^{-1}$, and -0.74 ± 0.17 $Nm\cdot kg^{-1}\cdot m^{-1}$ (Figure 7). There was a significant main effect for both weight and condition ($p\leq0.001$) but no significant interaction between weight*condition ($p=0.061$).

There was a peak ankle inversion moment of 0.01 ± 0.01 $Nm\cdot kg^{-1}\cdot m^{-1}$, 0.01 ± 0.02 $Nm\cdot kg^{-1}\cdot m^{-1}$, $0.01\pm0.02 Nm\cdot kg^{-1}\cdot m^{-1}$ for the 90° BW, 50% and 80% squats and during the deep BW, 50% and 80% squats subjects produced a peak moment of 0.03 ± 0.02 $Nm\cdot kg^{-1}\cdot m^{-1}$, 0.04 ± 0.05 $Nm\cdot kg^{-1}\cdot m^{-1}$, and 0.05 ± 0.05 $Nm\cdot kg^{-1}\cdot m^{-1}$ (Figure 8). There statistical analysis indicated there was a significant interaction between weight*condition ($p=0.007$). Post hoc analysis revealed significant differences between the 90° BW and deep BW squats ($p<0.001$), 90° 50% and deep 50% squats ($p=0.002$), and 90° 80% and deep 80% squats ($p=0.001$).

Ligamentous/Compressive

The subjects had an average ACL loading of 0.14 ± 0.04 BW, 0.18 ± 0.1 BW, 0.22 ± 0.11 BW for the 90° BW, 50%, 80% squats (Figure 9). There was a significant main effect for weight ($p=0.032$), which post hoc then revealed no significant differences for any comparisons measured. There was an average of 1.29 ± 0.63 BW, 2.8 ± 1.07 BW, 3.62 ± 1.50 BW of PCL loading for the 90° BW, 50%, 80% squats. There was a significant main effect for weight ($p<0.001$) and post hoc produced significant differences between 90° BW and 90° 50% ($p<0.001$) and 90° BW and 90° 80% ($p<0.001$). During the study there was an average tibiofemoral compressive of 3.74±0.96 BW, 6.98±1.59 BW, 8.07±2.03 BW for the 90° BW, 50%, 80% squats (Figure 10). There was a significant main effect for weight ($p<0.001$) and post hoc analysis revealed that there was a significant difference between 90° BW and 90° 50% ($p<0.001$) and 90° BW and 90° 80% ($p<0.001$).

Electromyography

The gluteus maximus had activations of 0.16±0.07 %MVIC, 0.64±0.34 %MVIC, and 1.04±0.62 %MVIC for the 90 BW, 50%, and 80% squats (Figure 11). During the deep BW, 50%, and 80% squats there were activations of 0.21±0.11 %MVIC, 0.95±0.68 %MVIC, and 1.34±0.97 %MVIC. There was a significant interaction between condition*weight ($p=.023$). There was significant differences between the 90 BW and 50% ($p<0.001$), 90 BW and 80% ($p<0.001$), 90 50% and 80% ($p<0.001$), deep BW and 50% ($p<0.001$), deep BW and 80% ($p<0.001$), deep 50% and 80% ($p=0.003$).

Medial hamstrings activation during the study was 0.09±0.05 %MVIC, 0.21±0.12 %MVIC, and 0.32±0.17 %MVIC for the 90 BW, 50%, and 80% squats (Figure 12). During the deep BW, 50%, and 80% squats there were activations of 0.10±0.06 %MVIC, 0.39±0.56 %MVIC, and 0.35±0.17 %MVIC. There was no significant main effect for condition ($p=0.152$) but there was a main effect for weight ($p=0.005$). There was no significant interaction for condition*weight ($p=0.275$).

During the 90 BW, 50%, an 80% squats the biceps femoris experienced an average of 0.16±0.11 %MVIC, 0.40±0.34 %MVIC, and 0.54±0.36 %MVIC activation

(Figure 13). During the deep BW, 50%, and 80% squats there was an average of 0.19±0.10 %MVIC, 0.45±0.26 %MVIC, and 0.65±0.26 %MVIC activations. There were significant main effects for condition ($p=0.049$) and weight ($p<0.001$). There was not a significant interaction between condition*weight ($p=0.163$).

During the 90 BW, 50%, and 80% squats the rectus femoris had average activations of 0.55±0.39 %MVIC, 0.98±0.57 %MVIC, and 1.37±0.70 %MVIC (Figure 14). During the deep BW, 50% and 80% squat there was average activations of 0.91±0.97 %MVIC, 1.31±0.67 %MVIC, and 1.73±0.76 %MVIC. There was significant main effects for condition ($p=0.001$) and weight ($p<0.001$). There was no significant interaction for condition*weight ($p=0.962$).

The vastus medialis had average activations of 0.74±0.26 %MVIC, 1.13±0.39 %MVIC, and 1.49±0.64 %MVIC for the 90 BW, 50% and 80% squats (Figure 15). The deep BW, 50%, and 80% squats had average activations of 0.80±0.40 %MVIC, 1.40±0.50 %MVIC, and 1.69±0.76 %MVIC. There was a significant interaction between condition*weight ($p=0.022$). There were significant differences between 90 and deep 50% ($p=0.002$), 90 BW and 50% ($p<0.001$), 90 BW and 80% ($p<0.001$), 90 50% and 80% ($p=0.006$), deep BW and 50% ($p<0.001$), deep BW and 80% ($p<0.001$) and deep 50% and 80% ($p=0.003$).

The vastus lateralis had average activations of 0.84±0.22 %MVIC, 1.33±0.37 %MVIC, and 1.77±0.49 %MVIC for the 90 BW, 50% and 80% squats (Figure 16). During the deep BW, 50% and 80% squats there were average activations of 0.98±0.24 %MVIC, 1.98±0.56 %MVIC, and 2.30±0.79 %MVIC. There was a significant interaction between condition*weight ($p<0.001$). There were significant differences between 90 and deep 50% ($p<0.001$), 90 and Deep 80% ($p=0.004$), 90 BW and 50% ($p<0.001$), 90 BW and 80% ($p<0.001$), 90 50% and 80% ($p<0.001$), deep BW and 50% ($p<0.001$), and deep BW and 80% ($p<0.001$).

The medial gastrocnemius had activations of 0.23±0.15 %MVIC, 0.43±0.31 %MVIC, and 0.61±.41 %MVIC for the 90 BW, 50%, and 80% squats (Figure 17). During the deep BW, 50%, and 80% squats there were activations of 0.23±0.14 %MVIC, 0.55±0.40 %MVIC, and 0.797±0.524 %MVIC. There was a significant interaction between condition*weight ($p=.002$). There was significant differences between the 90 80% and deep 80% squats ($p=0.003$), 90 BW and 50% ($p=0.003$), 90 BW and 80% ($p<0.001$), 90 50% and 80% ($p<0.001$), deep BW and 50% ($p<0.001$), deep BW and 80% ($p<0.001$), and deep 50% and 80% ($p<0.001$).

The tibialis anterior had an average activation of 0.94±0.53 %MVIC, 0.585±0.400 %MVIC, and 0.69±0.48 %MVIC for the 90 BW, 50% and 80% squats (Figure 18). There were also average activations of 1.15±0.54 %MVIC, 0.84±0.55 %MVIC, and 0.87±0.54 %MVIC for the deep BW, 50% and 80% squats. There was significant main effects for both condition ($p<0.001$) and weight ($p<0.001$). There was not a significant interaction between condition*weight ($p=0.600$).

Chapter 5: Discussion

The objective of the study was to find the effects of different knee flexion angles during squats on ACL, PCL, and tibiofemoral compressive forces. To do this, subjects were asked to perform squats with loads of BW, 50%, and 80% of their 1 RM while performing these loads at a 90° and 120° knee flexion angle. It was hypothesized that: 1) as load increased ACL, PCL, and tiobiofemoral compressive forces would significantly increase but would not be in ranges of dangerous loading; 2) muscle activation patterns of the medial and lateral hamstrings, vastus medialis, vastus lateralus, rectus femoris, medial gastrocnemius and the gluteus maximus will increase in deep squats compared to 90° squats; and 3) there will be significant increases in three dimensional in peak joint moments of the knee, hip and ankle from 90° and deep squats.

During the testing there was significantly higher hip adduction moments between each deep and 90° condition as well as a significant difference during each 90° squat at each increasing weight and during each deep squat at each increasing weight. With increases load there is a demand for increased control of the lower extremities during the squatting movement. During the study there was not a significant interaction between weight and condition for either the biceps femoris or medial hamstring, but the biceps femoris had significant increases in activation between the two conditions and both the biceps femoris and medial hamstring had significant increases in activation as weight increased. Although the biceps femoris had higher activations than the medial hamstrings there was still an increased adduction moment about the hip. The increased hip adduction is possibly being generated from other muscle in the lower extremities that were being contracted to cause the adduction, such as gracilis, adductor magnus, adductor longus, adductor brevis, and pectineus. 3D joint moments increased frontal plane moments between a squat around 110° and a squat performed to a depth around 121° (Swinton, Lloyd, Keogh, Agouris, & Stewart, 2012). Although there was increased gluteus maximus muscle activation as the weight increased in each condition, this increased gluteal activation may seem to only affect the increase in the extension moment. Furthermore the gluteus maximus was thought to only be a hip abductor and hip extensor, however, the lower fibers of the gluteus maximus have the ability to perform some adduction at the hip (Biel, 2010). Therefore, the increased gluteal activation could have contributed to the increased hip adduction moment.

From the previous literature and this study it was shown that as squat depth and load increase the activations and relative muscular effort of the hip extensors are increased. Hip extension moments were significantly different among each condition between 90° and deep and among each increase in load. The biceps femoris increased significantly between each condition and significant differences as weight increased but no significant interaction between weight and condition. This finding does not agree with (Escamilla, 2001; Schoenfeld, 2010; Wretenberg, Arborelius, Weidenhielm, & Lindberg, 1993b), it did agree with (Pauletto, 1991).The medial hamstring had no significant interaction between weight and condition but had a significant main effect for weight and no significant main effect for condition which coincides with previous research (Escamilla, 2001; Schoenfeld, 2010; Wretenberg et al., 1993b).The increased hip extension moments could be attributed to the increased gluteus maximus activation, the increased activation of the bicep femoris, and the medial hamstring during increases in load but not during increased knee flexion. The increased hip extension moments between 90° and deep squats was also represented in (Caterisano et al., 2002; Swinton et al., 2012; Wretenberg et al., 1993b). Relative muscular effort, defined as the muscle force required to perform a task relative to the maximum muscle force is a measure of relative muscular intensity (Bryanton et al., 2012). For example, if an 85% 1RM load is used, the muscle group should be generating 85% of its maximum force. As squat depth increased there were significant differences in the hip extensor moments and as load increased that hip extensor relative muscular effort increased (Bryanton et al., 2012). When comparing 90-104° squats there was significant differences between 60% compared to 50%, 70% compared to 50%, 80% compared to 50%, 60%, 70%, and 90% compared to 50, 60, 70 and 80% 1 RM (Bryanton et al., 2012). During squats ranging from 105-119° Bryanton et al. (2012) found that relative muscular effort was significantly different between 70% compared to 50 and 60%, 80% compared to 50, and 60%, and 90% 1 RM to 50, 60, and 70% 1 RM.

The increase in the hip extensor moments were likely caused more by the uniarticulate gluteus maximus rather than the biarticulate hamstrings. The hamstring may need to facilitate the coordinated actions of the hip and knee during the squat, where one end of the hamstrings is being stretched while the other is contracting. This may reduce the force generating capabilities of the hamstrings and the increased demands with increased loads and depths may have to be facilitated by the gluteus muscles at the hip. Almost at the same instance during peak hip adduction moments, the peak hip external hip rotation takes place. Although this external rotation takes place at the same instance of the peak hip adduction, there was no significant changes among any condition change throughout the study. Swinton et al. (2012) also saw no change among transverse joint moments as squat depth changed.

The observed increases in knee extension moments correspond with the observed increased activations of the vastus lateralis and vastus medialis muscles. The absence of an equivalent increase in rectus femoris activity is likely due to the biarticular nature of the muscle compared to the uniarticulate vasti muscles. Along with this idea, it was suggested that the rectus femoris is more responsible for hip flexion rather than knee extension (Biel, 2010). Both this study and Wretenberg et al. (1993b) found that the peak hip and knee extension moments occur at the same point during the deepest aspect of the squat. Wretenberg et al. (1993b) also found that the vastus lateralis and rectus femoris increased in activation from the 90° squat to the deep squat but the was no significant changes in the biceps femoris during any of the different squat depths. It is also stated that the authors believe that the joint moments around the knee and hip are the highest at the deepest part of the squat is due to the largest positive acceleration during the deepest part of the knee flexion. Caterisano et al. (2002) reported significant findings on the percent contribution of each muscle based off peak integrated EMG analysis of the muscle throughout the squat movement across three different squat depths. Caterisano et al. (2002) found that as squat depth went from a parallel (90°) to a full (136°) squat the gluteus maximus increased in contribution and that the vastus medialis had a significant decrease in contribution. The other muscles within the study, biceps femoris and vastus lateralis, had no significant change among squat depth. Escamilla (2001) states that musculature responsible for knee extension is not activated any differently with depths beyond 90°.

Similarly, there were significant increases observed for plantarflexion moments and medial gastrocnemius activity as squat depth and external load increased. Since the gastrocnemius is a biarticulate muscle and one of the primary contributors to plantarflexion, the increase in muscle activation may come from the increased responsibilities at the knee helping the hamstrings with knee flexion while causing plantarflexion at the ankle. Swinton et al. (2012) found that there was significant differences in ankle extension moments between the two conditions in his study between depths around 110 compared to depths at 121°. The differences were found at 50, 60 and 70% 1RM between the two conditions. In the current study both the deep and 90° BW squats produced the greatest activation in the tibialis anterior. The tibialis anterior had significant decreases as external load increased and between squat depths. As medial gastrocnemius activation increased and tibialis anterior activation decreased the plantarflexion moments increased about the ankle. The increased tibailis anterior activation during bodyweight squats could also be caused by an increase in dorsiflexion due to greater forward lean compared to the 50% and 80% 1RM squats.

ACL loading had significant interaction as weight increased during the 90° squats. Although there was a significant main effect for weight, post hoc tests revealed no significant differences for any comparisons measured, which could have been a result of the study being underpowered. Although not significant ACL loading did increase as load increased. Previous literature stated that during knee flexion ACL loading is highest from full extension to 30° and decreases until 60° of knee flexion (Amis & Dawkins, 1991; Escamilla et al., 2000; Schoenfeld, 2010; Yoo et al., 2010). It has been quantified that maximal ACL loading has been established as a load between 1700 and 2200 N at failure

(Noyes, Butler, Grood, Zernicke, & Hefzy, 1984; Woo, Hollis, Adams, Lyon, & Takai, 1991). This study found that the highest average ACL loading occurred during the 80% 1RM squats and was around 150 N. Therefore, ACL loading did not reach maximal levels, even at 80% 1RM squats.

PCL loading had a significant main effect for weight, and post hoc analyses revealed significant differences in PCL loading between 90° squats at BW and 50% but no significant difference between 50% and 80%. Previous literature found PCL loading to start at 30° and continue until 120° where there is a decrease in PCL loading (Amis et al., 2006; Escamilla et al., 2000; Schoenfeld, 2010). Donnelly, Berg, and Fiske (2006) determined that during moderate weighted squats PCL loading will exceed 2700 N while maximal strength capabilities of a healthy young PCL is 4000 N (Race & Amis, 1994). It was found during testing that the average peak PCL loading was highest during the 80% 1RM squats and was found to be around 2500 N. During the BW squats PCL loading was around 900 N and at 60% 1RM squats and was approximately 1950 N.

During this study tibiofemoral compressive force was found to have a main effect for weight, so as weight increased tibiofemoral compressive forces increased. There were significant increases in tibiofemoral compressive forces between each load of the 90° squats. Previous literature has demonstrated that tibiofemoral compressive forces increase as load increases and as knee flexion increases (Escamilla et al., 2000; Schoenfeld, 2010). Tibiofemoral compressive forces ranged from approximatley 500 to 8000 N, but subjects that were found experiencing 8000 N of compressive force were lifting 2.5 times the subjects body weight and experienced the force at 130° of knee flexion (Escamilla et al., 2000). It has also been reported that tibiofemoral compressive forces can reach an even higher amount, around 10,000 N (Ariel, 1974). Tibiofemoral compressive forces were approximately 5600 N at the 80% 1RM squats, which was the highest among the three loads. The BW squats resulted in 2800 N of tibiofemoral compressive forces. At 60% 1RM tibiofemoral compressive forces were around 4800 N. As the load increased the tibiofemoral compressive forces increased. There was no load that produced the amount of compressive forces that was found in previous literature (Ariel, 1974; Escamilla et al., 2000). Tibiofemoral compressive forces are expected to be higher during loaded barbell squats, as increased weight on the subject will create larger compressive forces in the knee joint as the flexion angle changes over time of the squat. With this increased weight from the load, the musculature around the knee will be responsible for the controlled flexion and extension of the knee to prevent injuries moments and complete the movement as desired by the participant. Squat kinematics and kinetics can affect ACL, PCL, and tibiofemoral compressive forces. As muscle contracts to perform the desired movement they create tibiofemoral compressive and shear forces. As external loads increase so will the muscular demand and therefore increases the shear and compressive forces caused by the muscles. As knee flexion changes, different forces will change from compressive to shear and vice versa, what can affect these changes is the displacement of the knee over the toes. As the displacement increases the anterior shear force can increase creating a greater loading in the ACL (Escamilla et al., 2000; Schoenfeld, 2010).

Limitations

A main limitation of the current study was different variations of the starting position of the subjects. Foot position, stance width and bar placements were not all the same among the subjects. It is possible these different factors can attribute to differences among the subjects testing outcomes. Previous studies demonstrated no difference in muscle activation of the lower extremity during different foot angles and stance widths (Escamilla, Fleisig, Zheng, et al., 2001; McCaw & Melrose, 1999) but the same studies showed that there was increases in tibiofemoral compressive and shear force within the knee as stance width changed. Not only can stance with affect compressive forces but gender and weight can have influences on compressive forces as well (Taylor et al., 2004). Along with compressive forces it was shown that women are more susceptible to ACL injury risk during dynamic tasks (Hewett et al., 2005). However, during a less dynamic task such as the squat it is believed that factors influencing ligamentous loading and compressive forces will not differ between men and women. Thus, gender disparities in injury rates, particularly ACL injury, may not be present during the squat exercise. Bar placement on the subjects back has been demonstrated to increase moments about subject's joints with the change in position of the subject's center of mass (Gullett, Tillman, Gutierrez, & Chow, 2009). Subject bar placement was determined by what was most comfortable and was not recorded or measured.

Fatigue is another factor that may increase the risk of decreased muscular control leading to increased compressive and ligamentous loading (Chappell et al., 2005; Schoenfeld, 2010). To help prevent fatigue within the study subjects were given adequate rest between each change in condition and between each load increase. Although subjects were screened before the study, subjects have the potential to be less than truthful on their questionnaire. If a subject had not met the required amount of training experience and performed the squatting protocol, even with adequate rest fatigue could still affect those subjects. Furthermore, untruthful responses on the prescreening questionnaire about previous injuries and training experience could have affected squat kinematics and kinetics, and subsequent ligamentous loading and tibiofemoral contact forces.

Conclusion

As squat depth changes there were changes in the gluteus maximus, vastus lateralis, vastus medialis, medial gastrocnemius, tibialis anterior and the biceps femoris. It can be concluded that as athletes or everyday exercise enthusiasts use the squat exercise the greater the knee flexion employed the greater activation of muscles can be generated. As a training tool this could potentially increase muscle hypertrophy and strength overtime. As squat depth increases and muscles are continually activated it can potentially increase the strength of those muscles. As squat load increased all muscles tested increased in activations, therefore when used as a training mechanism, increasing squat load can ultimately increase the activations of muscle used in the squat and potentially increase the strength of these muscles.

The increases in activations help explain the increase in joint moments. Specifically, the increased activation suggest an increase in force production from the muscles. That increased force production in the muscles will create greater shear and compressive forces within the lower extremity joints. However, as the loads increased in the 90° squat up to 80% of the subject's 1 RM, ligamentous loading never reached maximal levels in either the ACL or PCL. In fact, ACL loading was approximately 7-9% of the estimated maximum, while PCL loading was approximately 63% of the estimated maximum. Given the relatively low amounts of ACL loading, it is evident that increases in squat load will not increase risk of ACL injury. On the other hand, PCL loading was much greater and increased as squat load increased. Nevertheless, these loads were still substantially below the ligament's estimated maximum and unlikely to cause ligament failure. Finally, tibiofemoral compressive forces reached 8 times BW (approximately 4500 N) but it is unclear how this loading might influence injury risk as the amount of force needed to cause articular cartilage damage is unknown.

Overall it may be suggested that participating in squat exercises can be beneficial. First, as load and depth increase muscle activation in the lower extremities can increase and can increase strength and hypertrophy in those muscles. Along with the potential of increased muscle activation there seems to be no increased risk in damaging the ligaments, but the risk of cartilage damage remains unclear.

Table 1. Mean ± standard deviation of peak right leg moments (Nm·kg^{-1}·m^{-1}) during the squat

	90 BW	90 50	90 80	Deep BW	Deep 50	Deep 80	Condition	Weight	Interaction
HE [ABCDEFGHI]	-0.60±0.16	-1.29±0.26	-1.67±0.28	-0.75±0.21	-1.53±0.27	-1.97±0.36	p<0.001	p<0.001	p=0.001
HER	-0.07±0.04	-0.08±0.05	-0.11±0.07	-0.16±0.07	-0.17±0.09	-0.21±0.11	p<0.001	p<0.001	p=0.834
HAD [ABCDEGHI]	0.08±0.07	0.16±0.10	0.20±0.13	0.14±0.08	0.29±0.14	0.41±0.22	p<0.001	p<0.001	p=0.001
KE [CDEFGHI]	0.62±0.13	0.91±0.19	1.07±0.22	0.59±0.13	1.00±0.20	1.17±0.26	p=0.001	p<0.001	p=0.043
KER	-0.05±0.02	-0.08±0.02	-0.11±0.04	-0.06±0.02	-0.10±0.02	-0.12±0.04	p=0.001	p<0.001	p=0.393
KAD	0.07±0.06	0.06±0.06	0.07±0.08	0.18±0.11	0.20±0.14	0.25±0.19	p<0.001	p=0.080	p=0.104
PF	-0.20±0.05	-0.42±0.10	-0.67±0.14	-0.23±0.08	-0.51±0.13	-0.74±0.17	p=0.001	p<0.001	p=0.061
INV [ABC]	0.01±0.01	0.01±0.02	0.01±0.02	0.03±0.02	0.04±0.05	0.05±0.05	p=0.001	p=0.009	p=0.007

[A]Deep BW is significantly different than 90° BW (p<0.006)
[B]Deep 50% is significantly different than 90° 50%
[C]Deep 80% is significantly different than 90° 80%
[D]90° 50% is significantly different than 90° BW
[E]90° 80% is significantly different than 90° BW
[F]90° 80% is significantly different than 90° 50%
[G]Deep 50% is significantly different than deep BW
[H]Deep 80% is significantly different than deep BW
[I] Deep 80% is significantly different than deep 50%

Table 2. Mean ± standard deviation of peak ligamentous and compressive forces (BW) during the squat

Structure	90° BW	90° 50%	90° 80%	Weight
Anterior Cruciate Ligament	0.14±0.04	0.18±0.10	0.22±0.11	p=0.032
Posterior Cruciate Ligament[1,2]	1.29±0.63	2.80±1.07	3.62±1.50	p<0.001
Tibiofemoral Compressive[1,2,3]	3.74±0.96	6.98±1.59	8.07±2.03	p<0.001

[1] 90° BW is significantly different than 90° 50%
[2] 90° BW is significantly different than 90° 80%
[3] 90° 50% is significantly different than 90° 80%

Table 3. Mean ± STDV activations of peak muscle activation (%MVIC) during the squat

Muscle	90° BW	90° 50%	90° 80%	Deep BW	Deep 50%	Deep 80%	Condition	Weight	Interaction
GMax[DEFGHI]	0.16±0.07	0.64±0.34	1.04±0.62	0.21±0.11	0.95±0.68	1.34±0.97	p=0.005	p<0.001	p=0.023
BF	0.16±0.11	0.40±0.34	0.54±0.36	0.19±0.10	0.45±0.26	0.65±0.26	p=0.049	p<0.001	p=0.163
MH	0.09±0.05	0.21±0.12	0.32±0.17	0.10±0.06	0.39±0.56	0.35±0.17	p=0.152	p=0.005	p=0.275
RF	0.55±0.39	0.98±0.57	1.37±0.70	0.91±0.97	1.31±0.67	1.73±0.76	p=0.001	p<0.001	p=0.962
VM[BCDEFGHI]	0.74±0.26	1.13±0.39	1.49±0.64	0.80±0.40	1.40±0.50	1.69±0.76	p=0.006	p<0.001	p=0.022
VL[BCDEFGH]	0.84±0.22	1.33±0.37	1.77±0.49	0.98±0.24	1.98±0.56	2.30±0.79	p=0.001	p<0.001	p<0.001
MG[CDEFGHI]	0.23±0.15	0.43±0.31	0.61±0.41	0.23±0.14	0.55±0.40	0.80±0.52	p=0.200	p<0.001	p=0.002
TA	0.94±0.53	0.59±0.40	0.69±0.48	1.15±0.54	0.84±0.55	0.87±0.54	p<0.001	p<0.001	p=0.600

[A]Deep BW is significantly different than 90° BW (p<0.006)
[B]Deep 50% is significantly different than 90° 50%
[C]Deep 80% is significantly different than 90° 80%
[D]90° 50% is significantly different than 90° BW
[E]90° 80% is significantly different than 90° BW
[F]90° 80% is significantly different than 90° 50%
[G]Deep 50% is significantly different than deep BW
[H]Deep 80% is significantly different than deep BW
[I] Deep 80% is significantly different than deep 50%

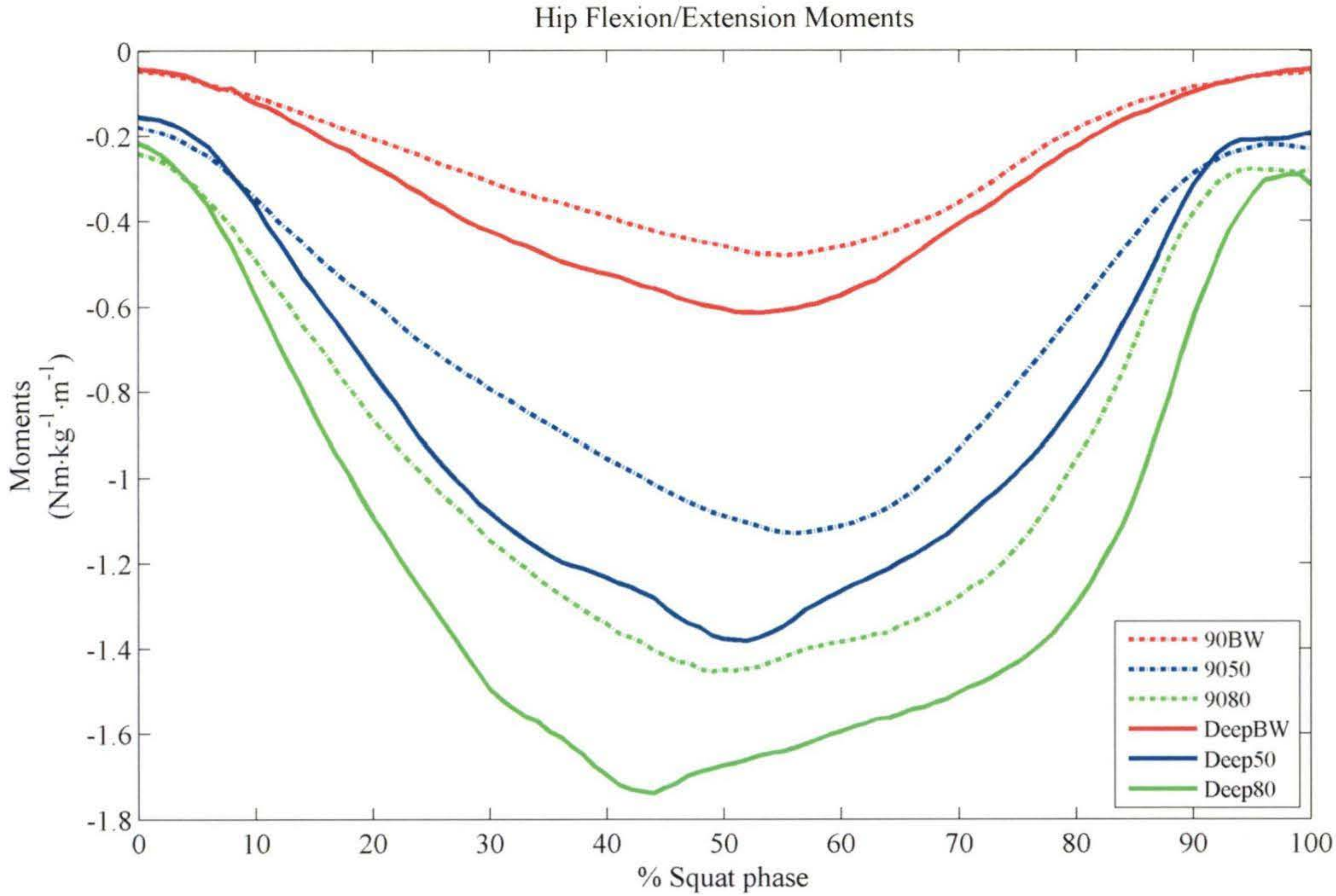

Figure 1. Mean hip flexion and extension moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represent extension moments while positive represent flexion moments.

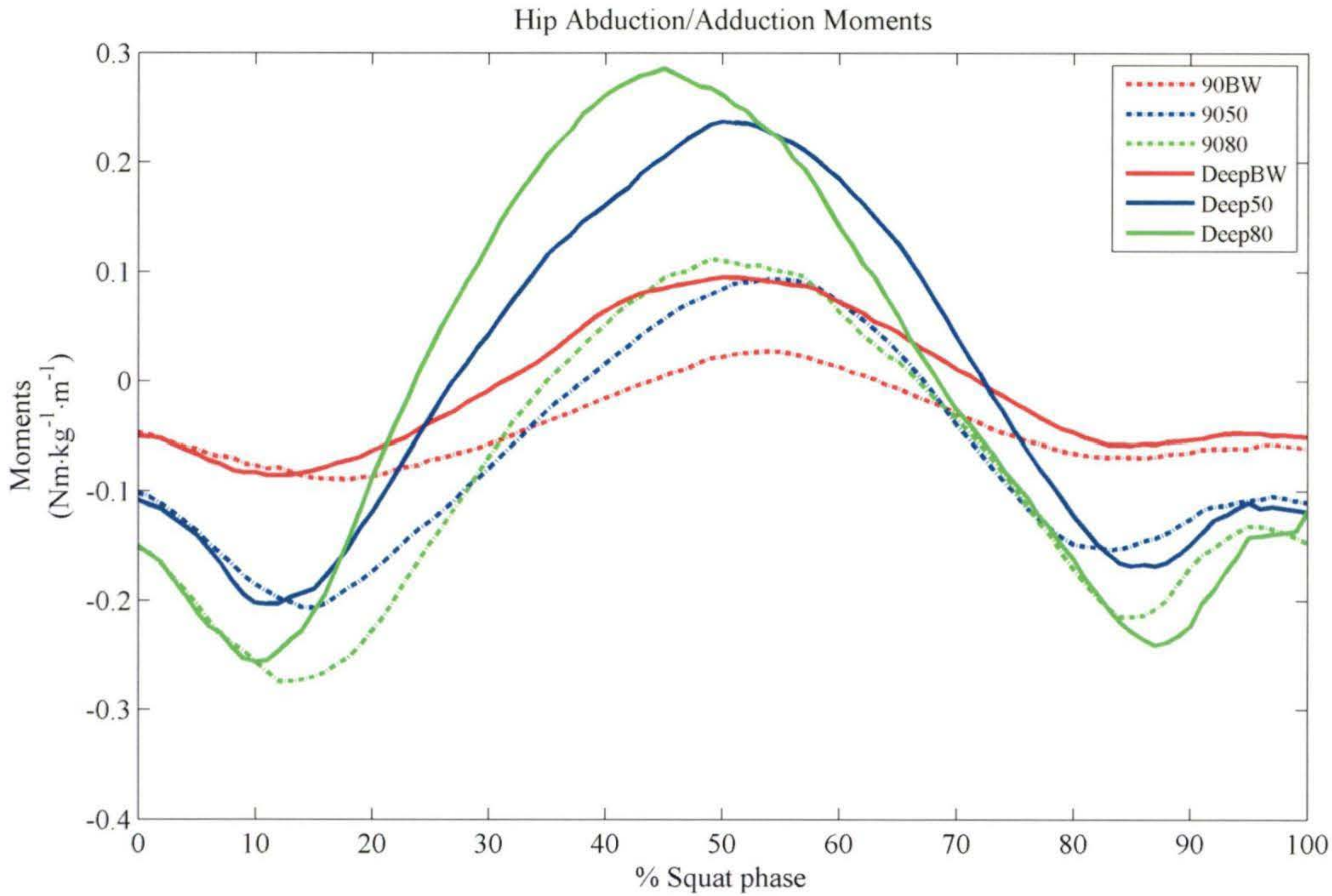

Figure 2. Mean hip adduction and abduction moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents abduction moments while positive represents adduction moments.

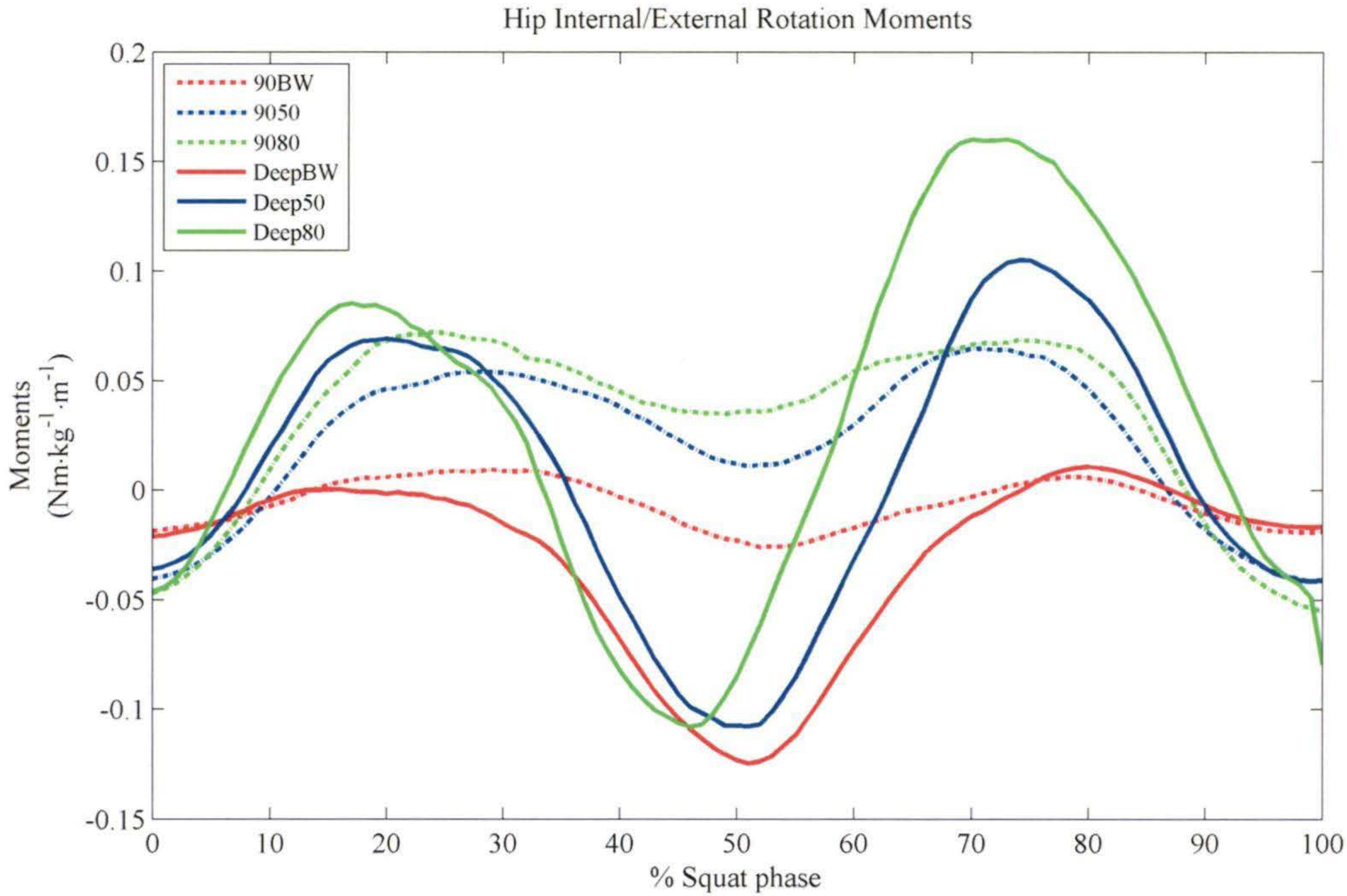

Figure 3. Mean hip internal and external rotation moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents external rotation moments while positive represents internal rotation moments.

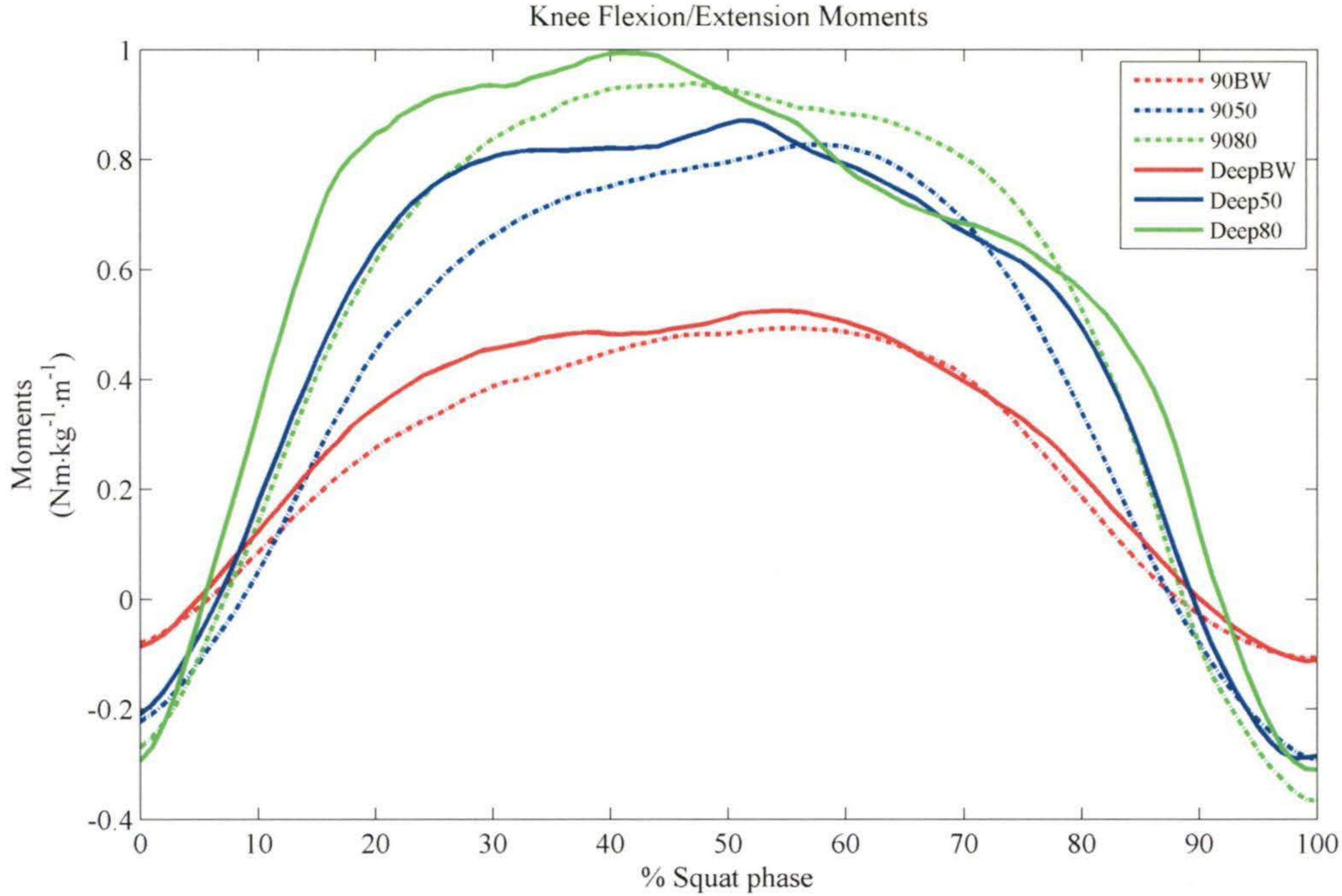

Figure 4. Mean knee flexion and extension moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents flexion moments while positive represents extension moments.

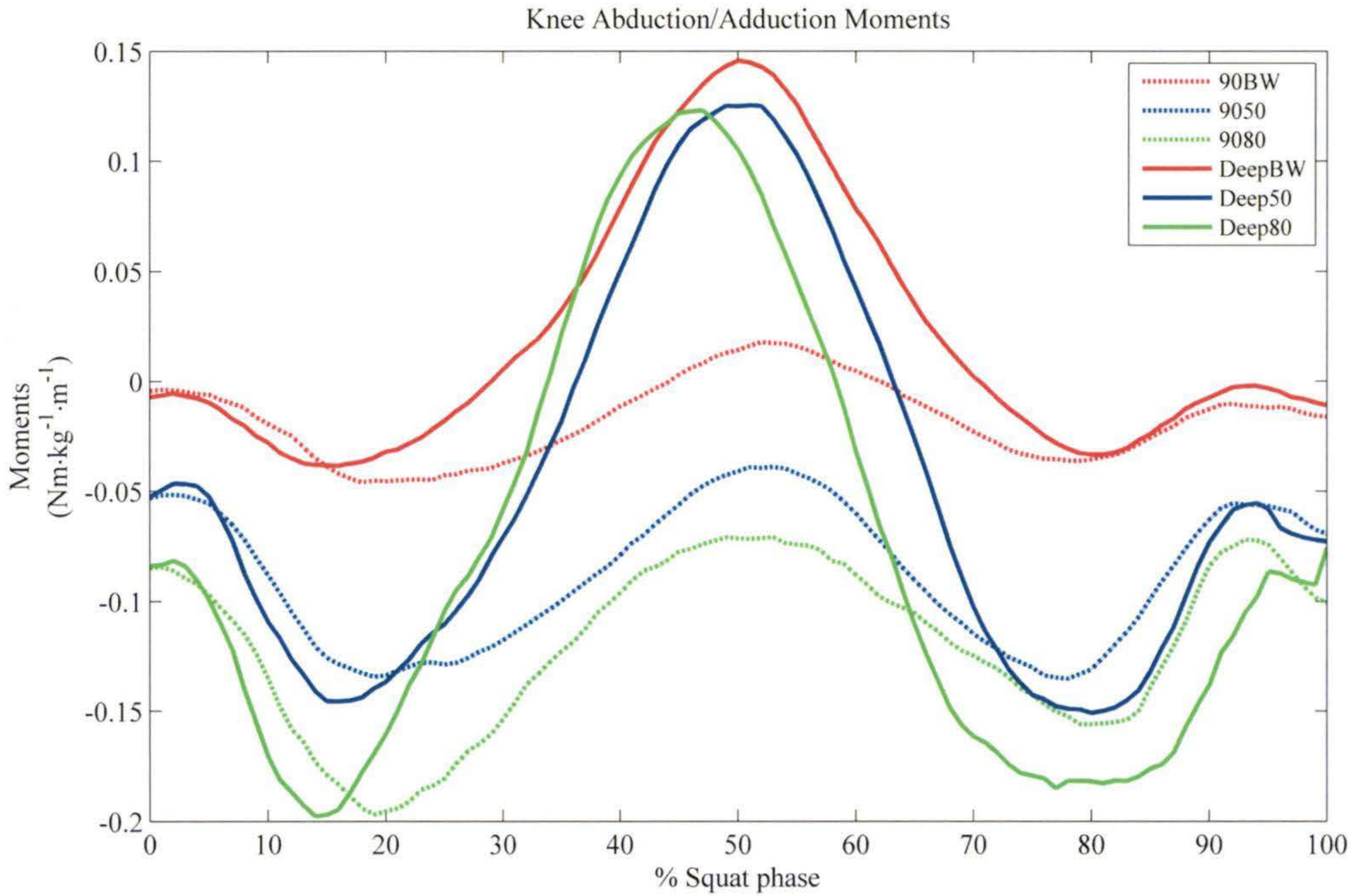

Figure 5. Mean knee adduction and abduction moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents abduction moments while positive represents adduction moments.

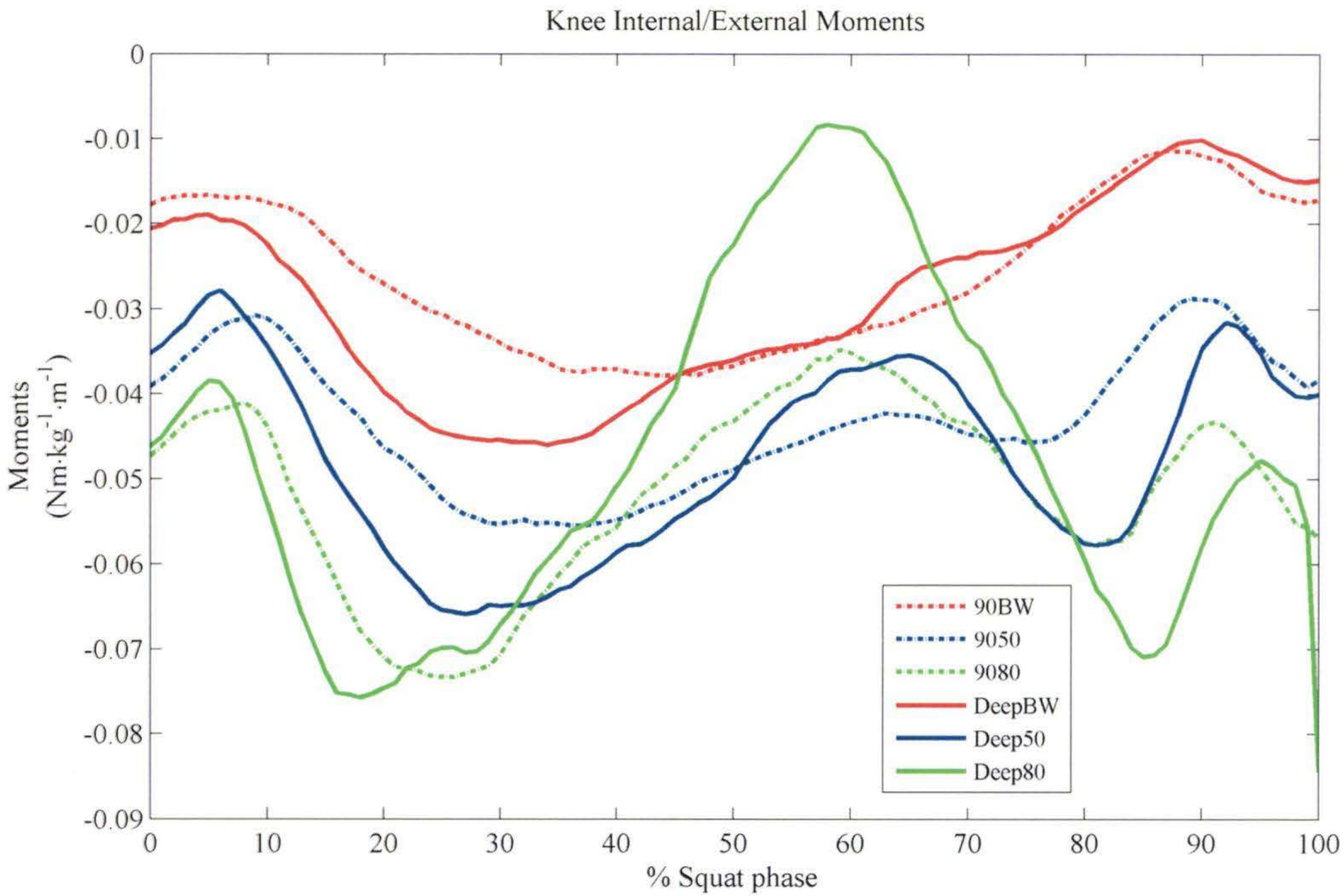

Figure 6. Mean knee internal and external rotation moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents external rotation moments while positive represents internal rotation moments.

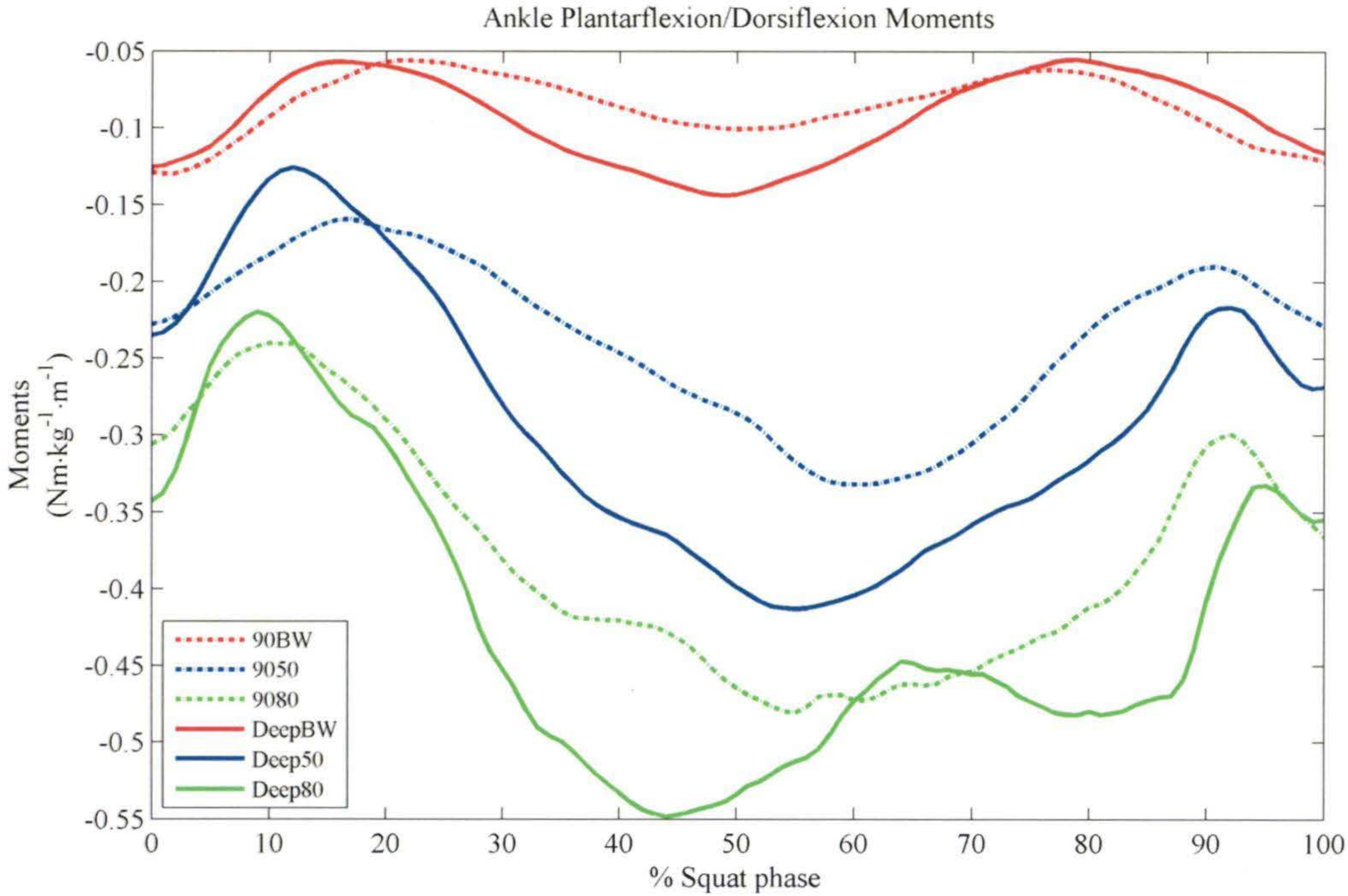

Figure 7. Mean ankle plantarflexion and dorsiflexion moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents plantarflexion moments while positive represents dorsiflexion moments.

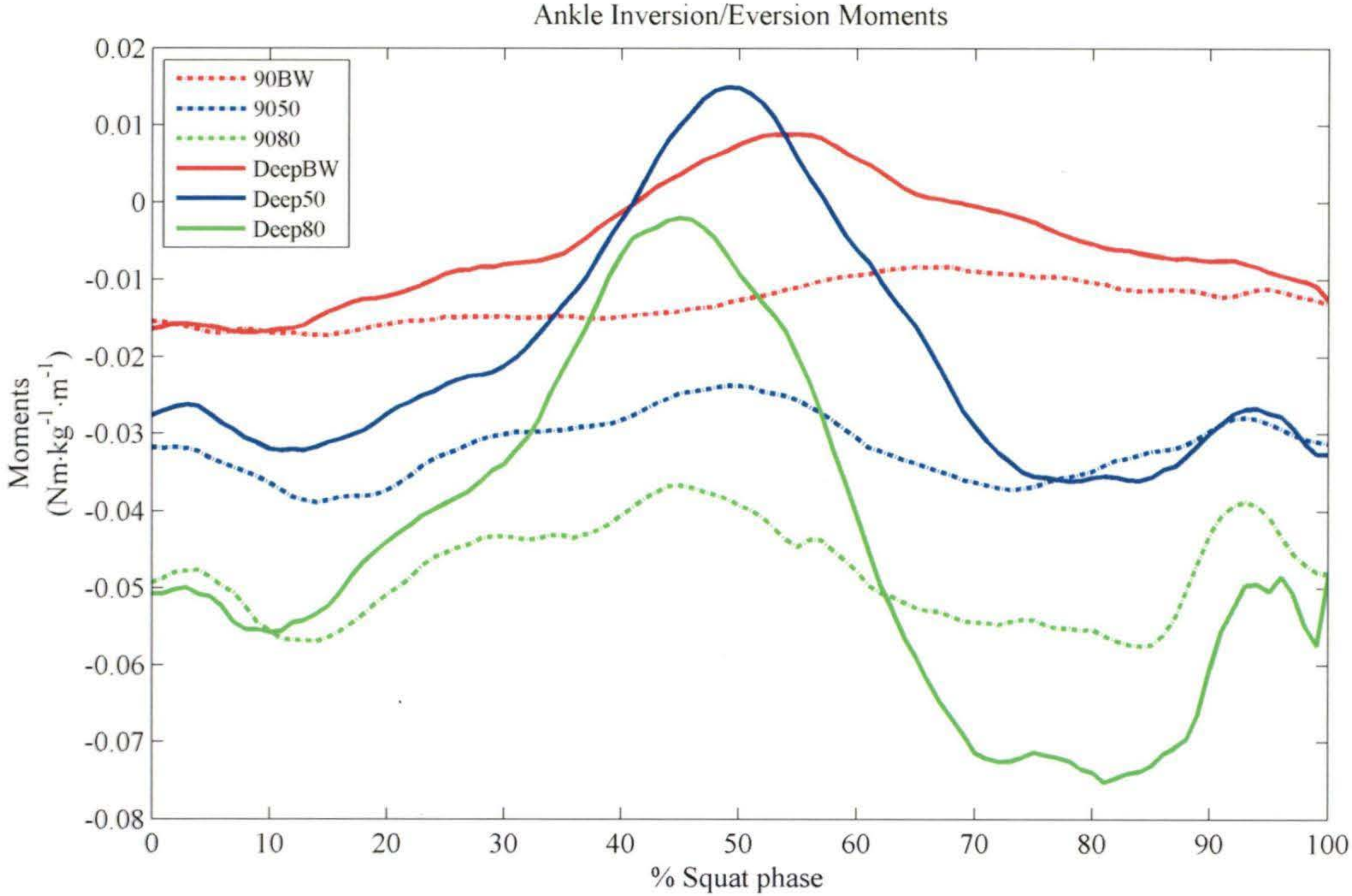

Figure 8. Mean ankle inversion and eversion moments for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents eversion moments while positive represents inversion moments.

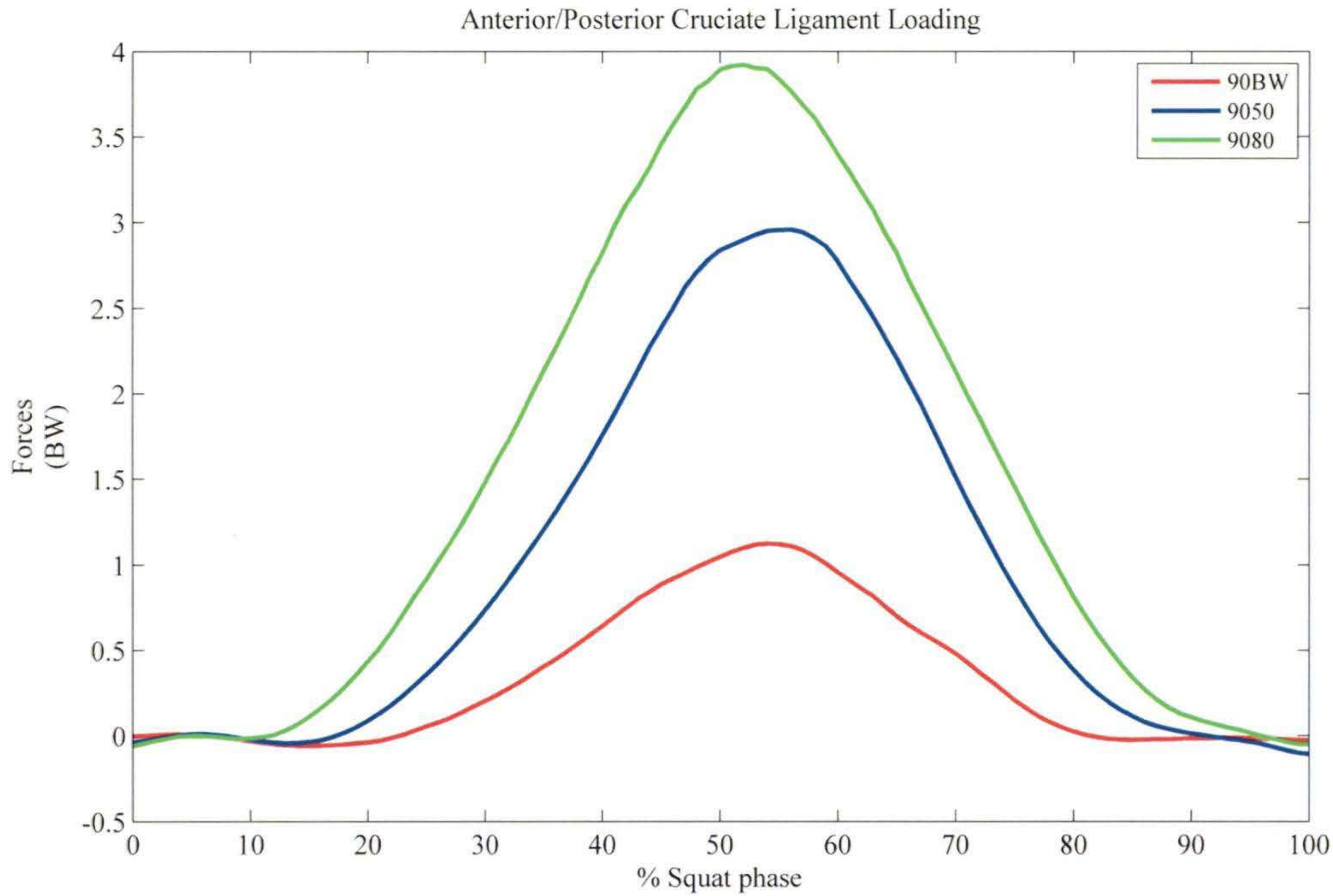

Figure 9. Mean cruciate ligament loading for 90BW (Red line), 9050 (Blue line), and 9080 (Green line). Positive forces represent posterior cruciate ligament loading while negative represents anterior cruciate ligament loading.

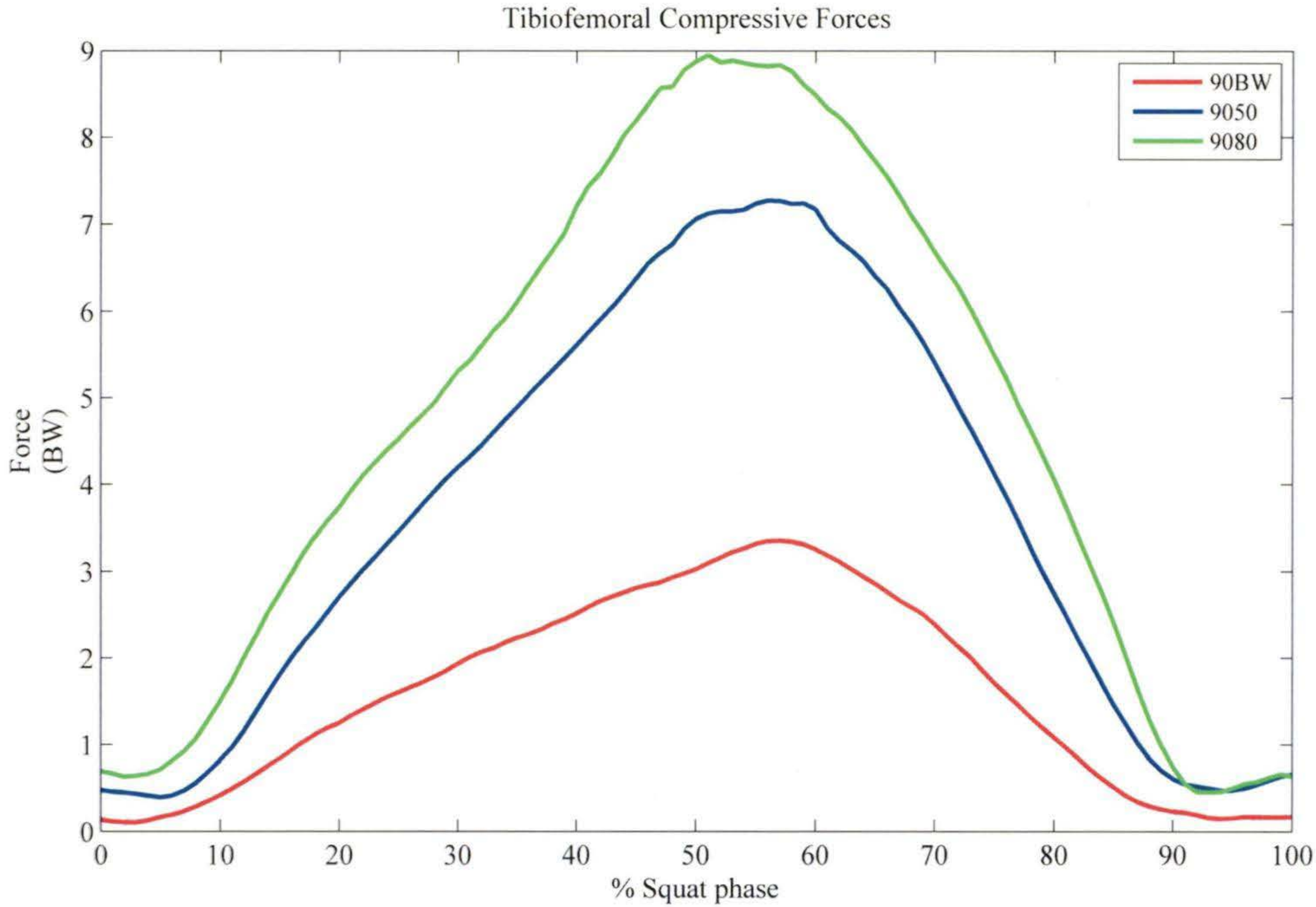

Figure 10. Mean tibiofemoral compressive forces for 90BW (Red line), 9050 (Blue line), and 9080 (Green line). Positive represents compression of the femur on to the tibia.

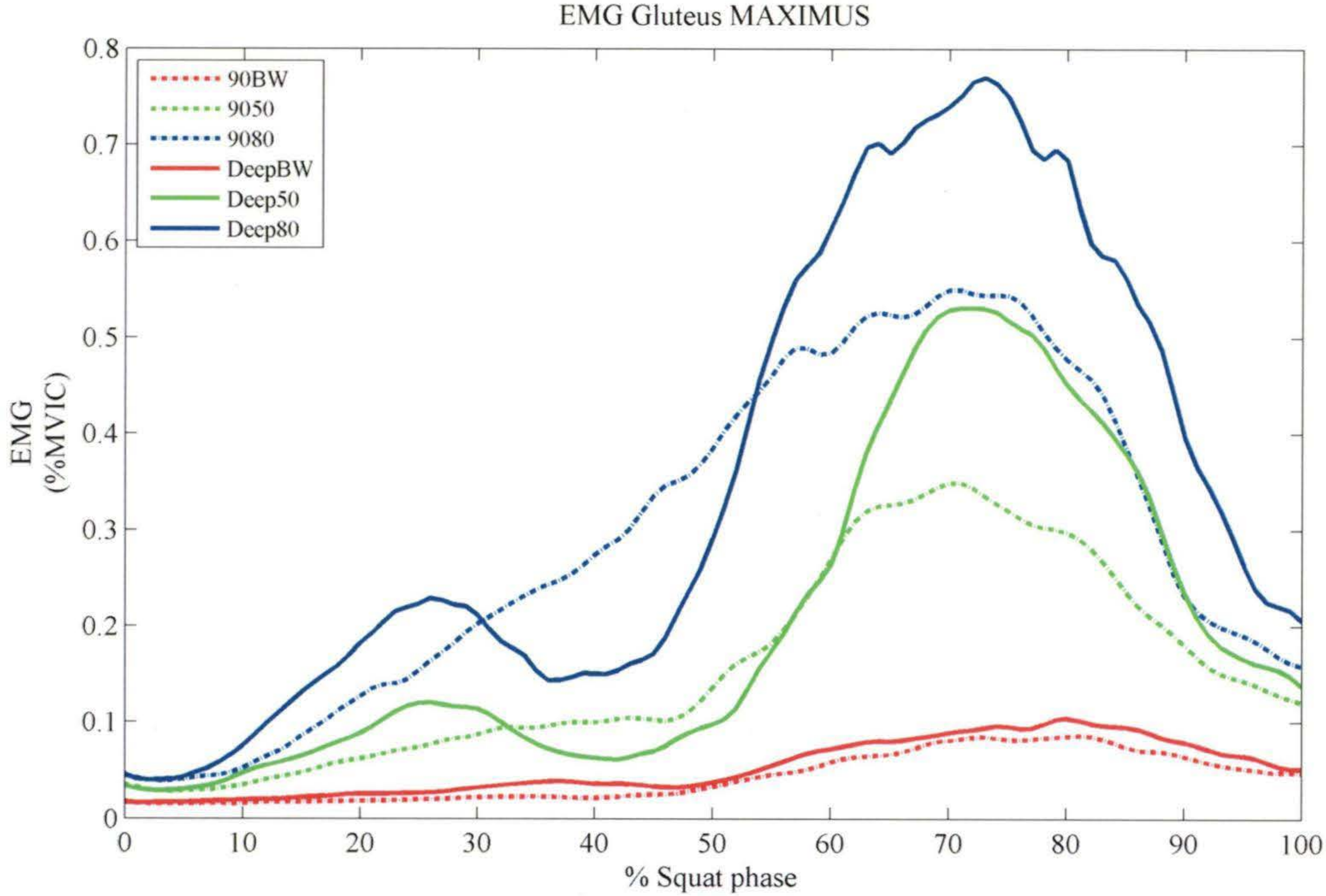

Figure 11. Mean EMG for gluteus maximus 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

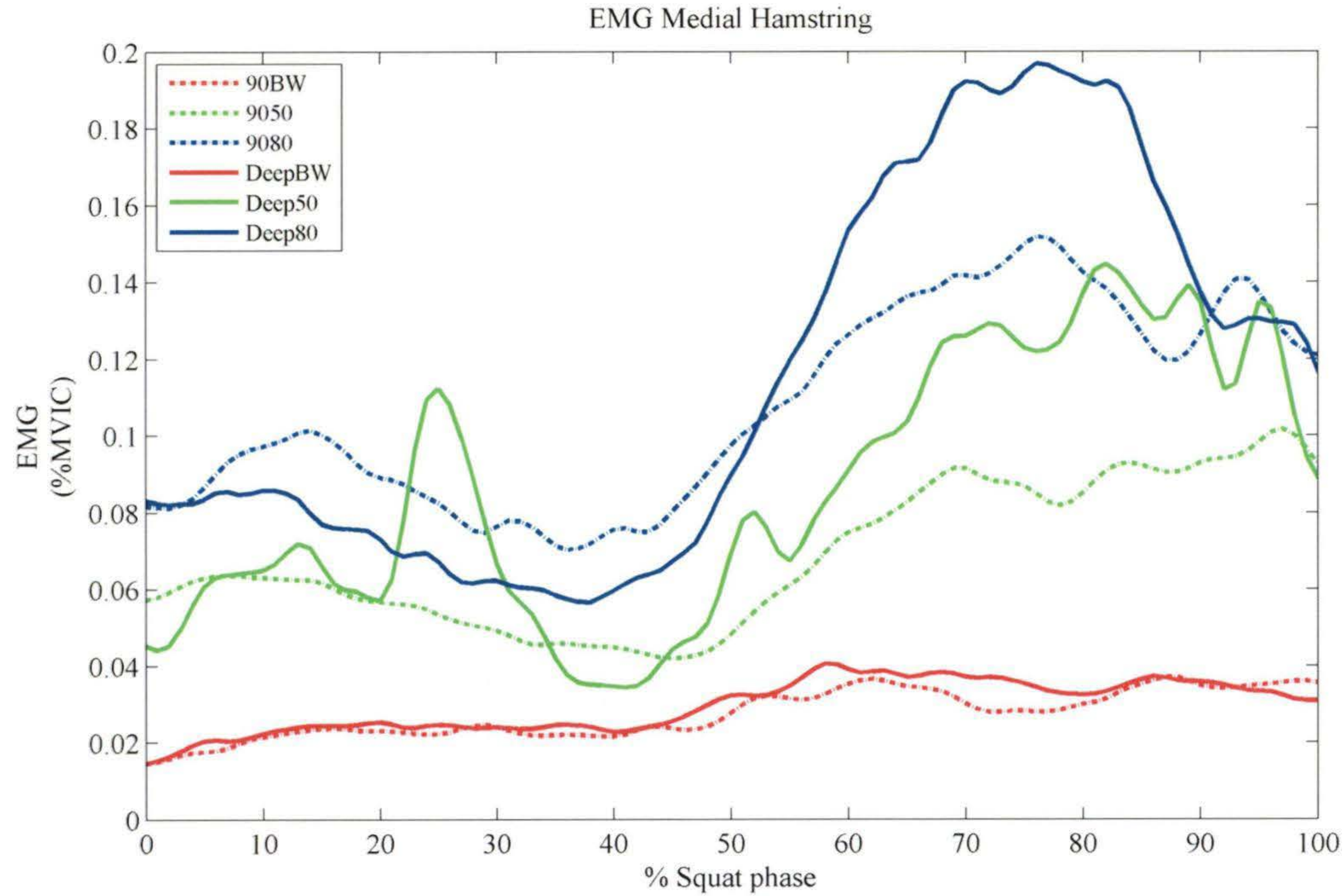

Figure 12. Mean EMG for the medial hamstring 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

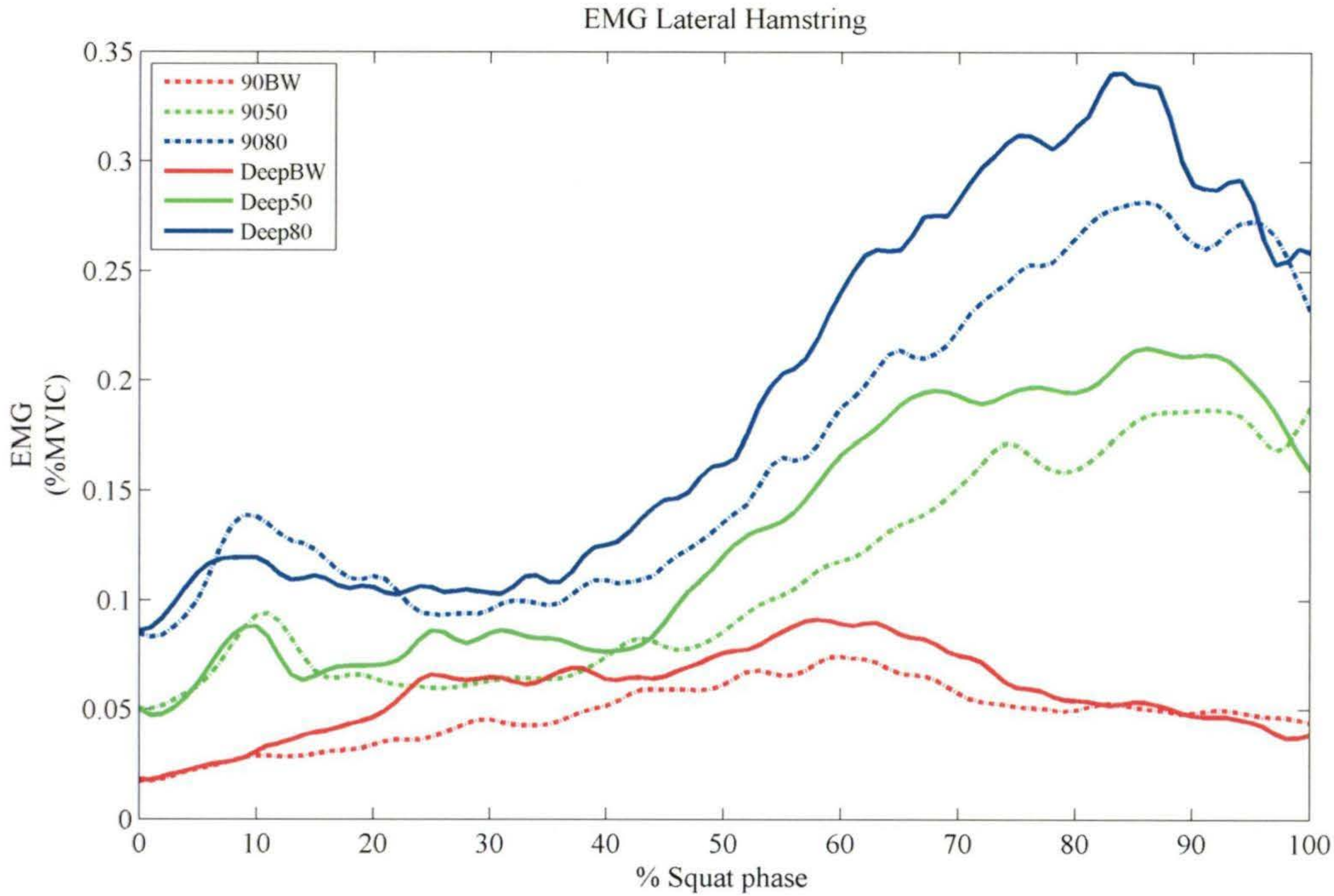

Figure 13. Mean EMG for the lateral hamstring 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

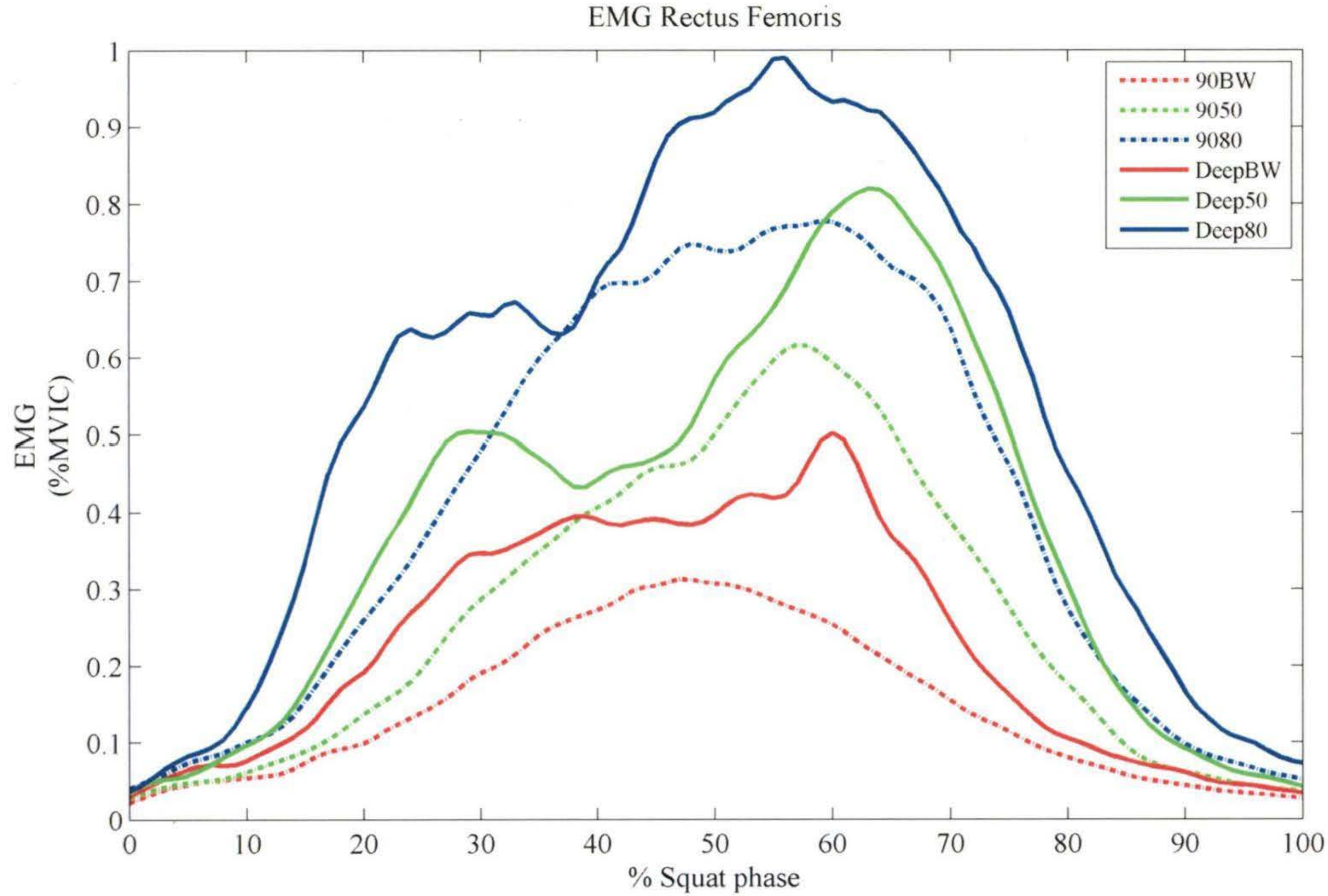

Figure 14. Mean EMG for the rectus femoris 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

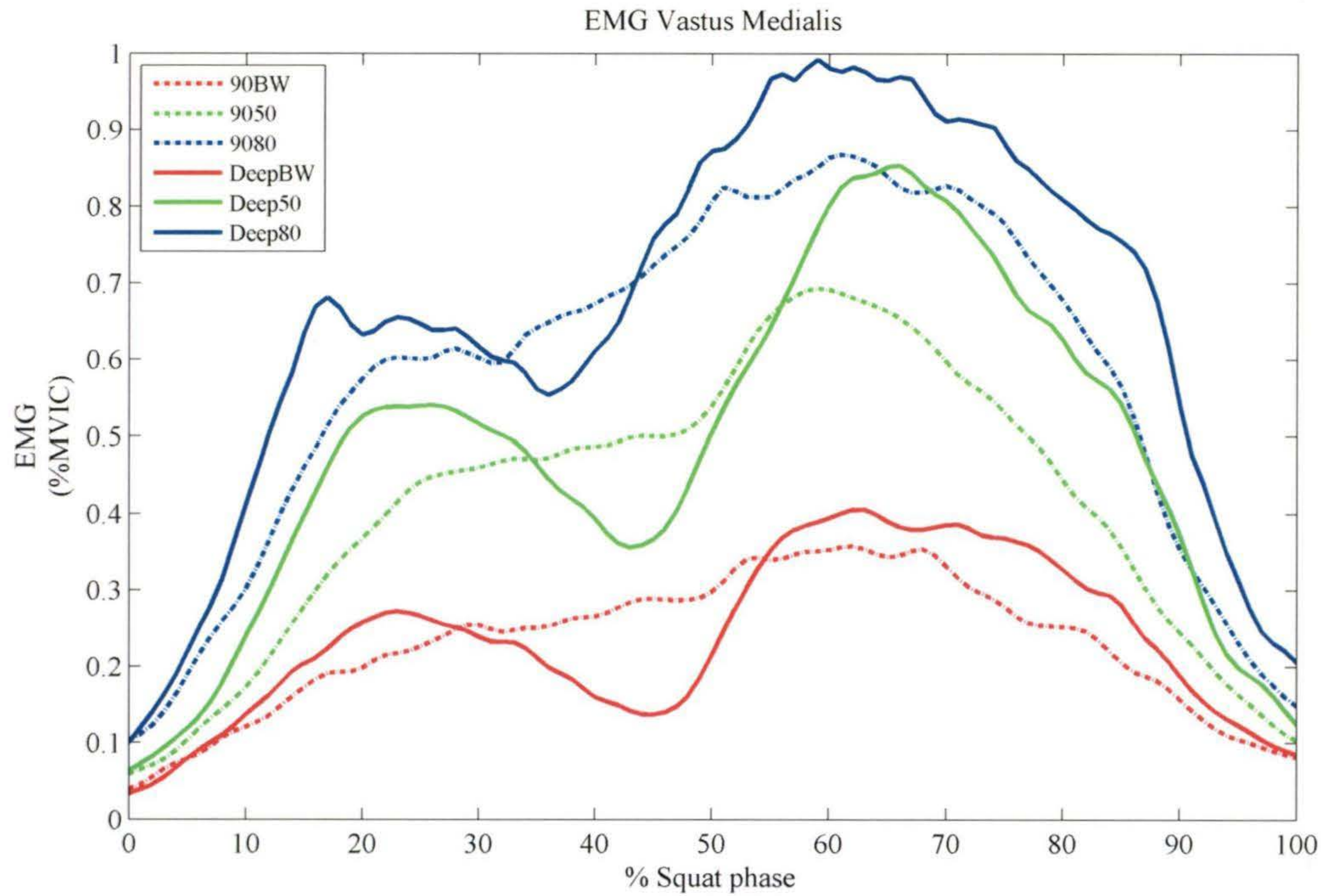

Figure 15. Mean EMG for vastus medialis 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

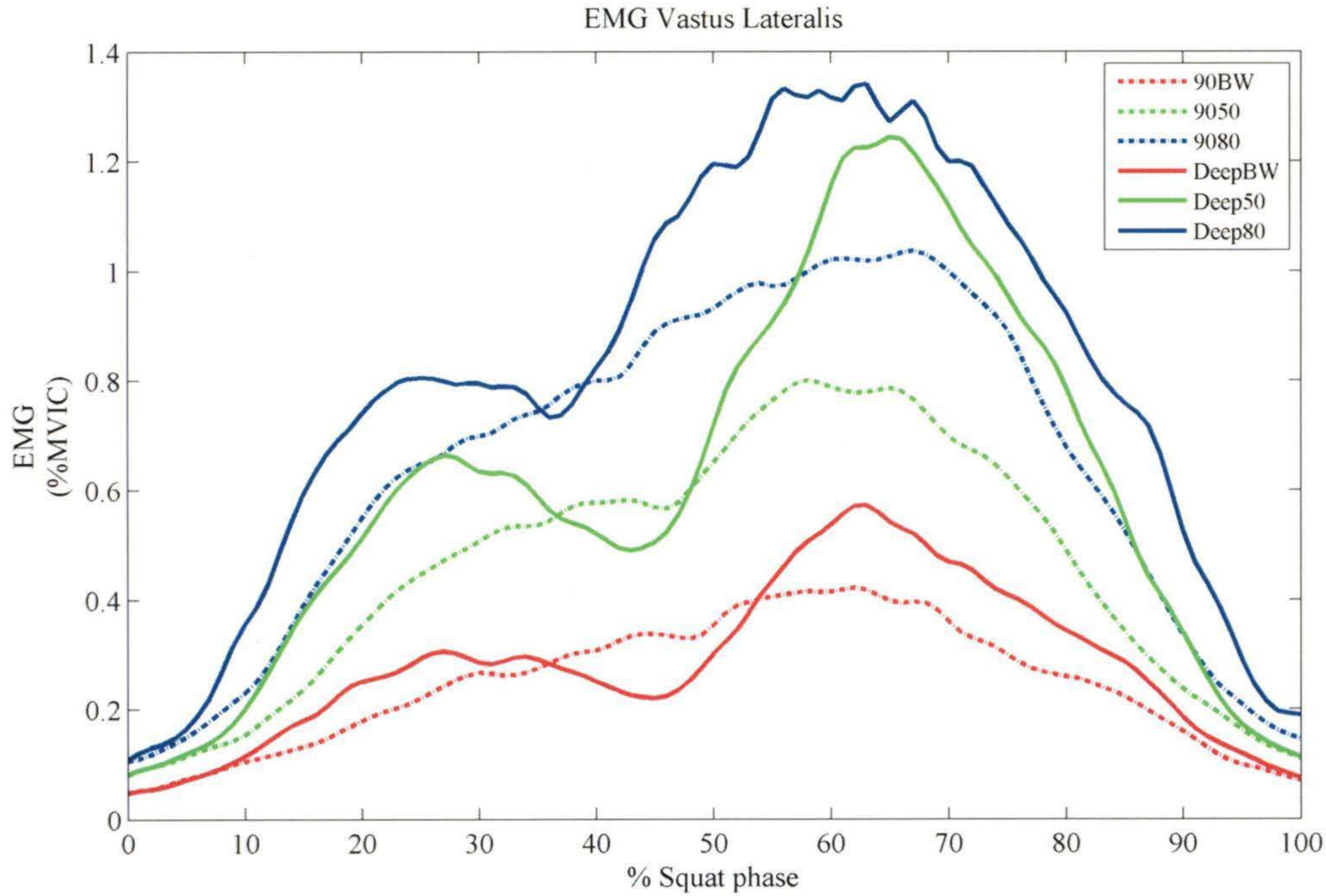

Figure 16. Mean EMG for vastus lateralis 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

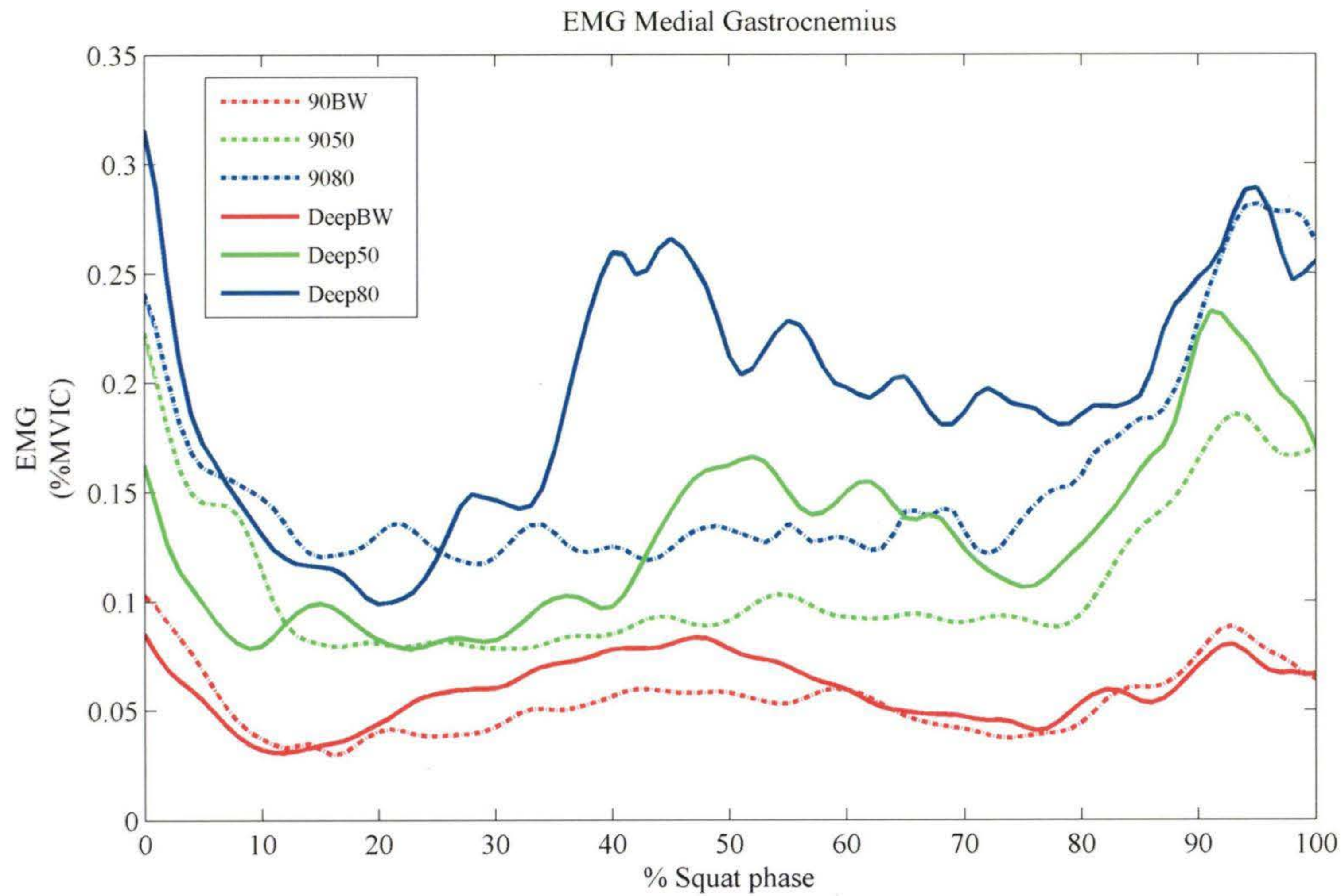

Figure 17. Mean EMG for the medial gastrocnemius 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

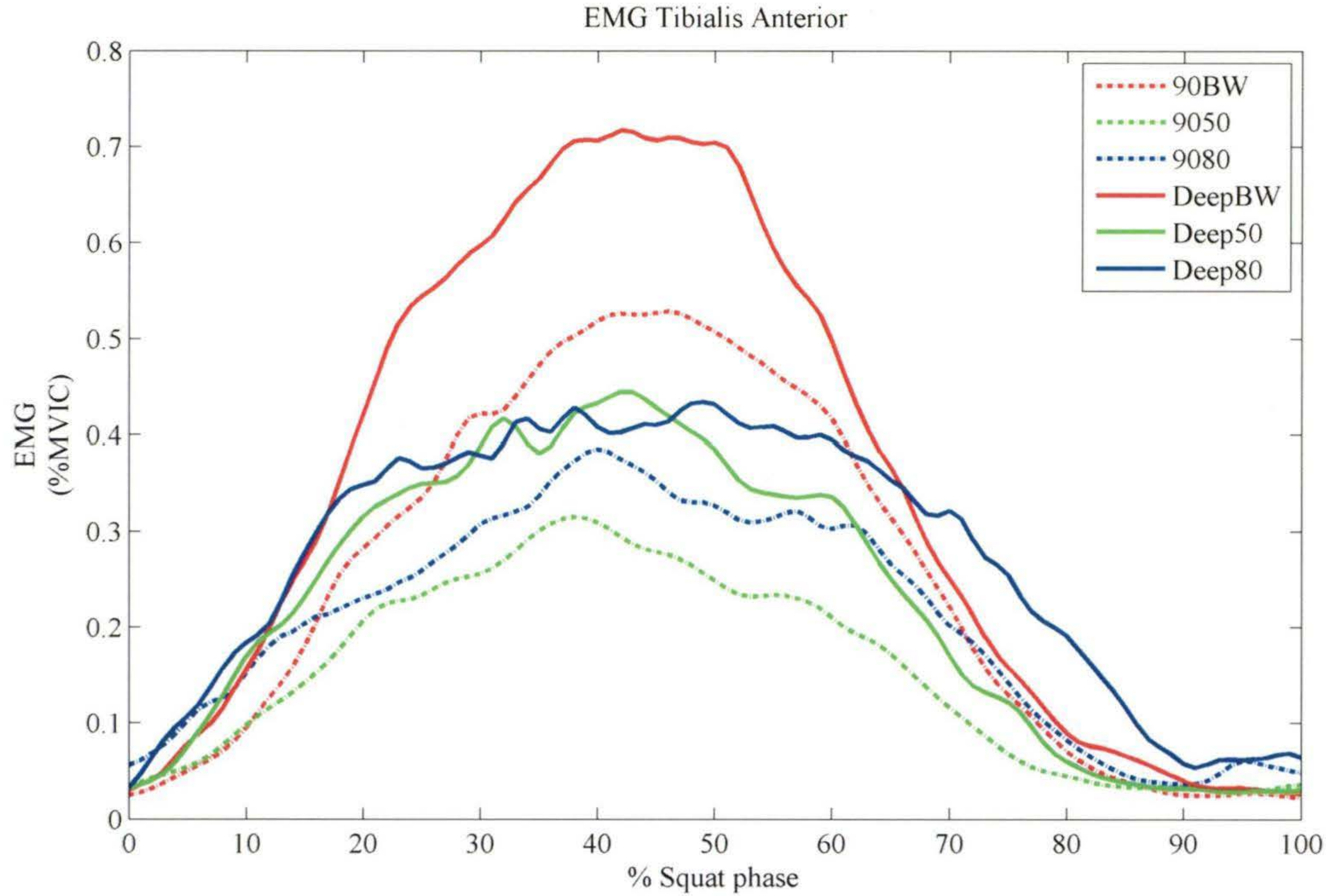

Figure 18. Mean EMG for the tibialis anterior 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line).

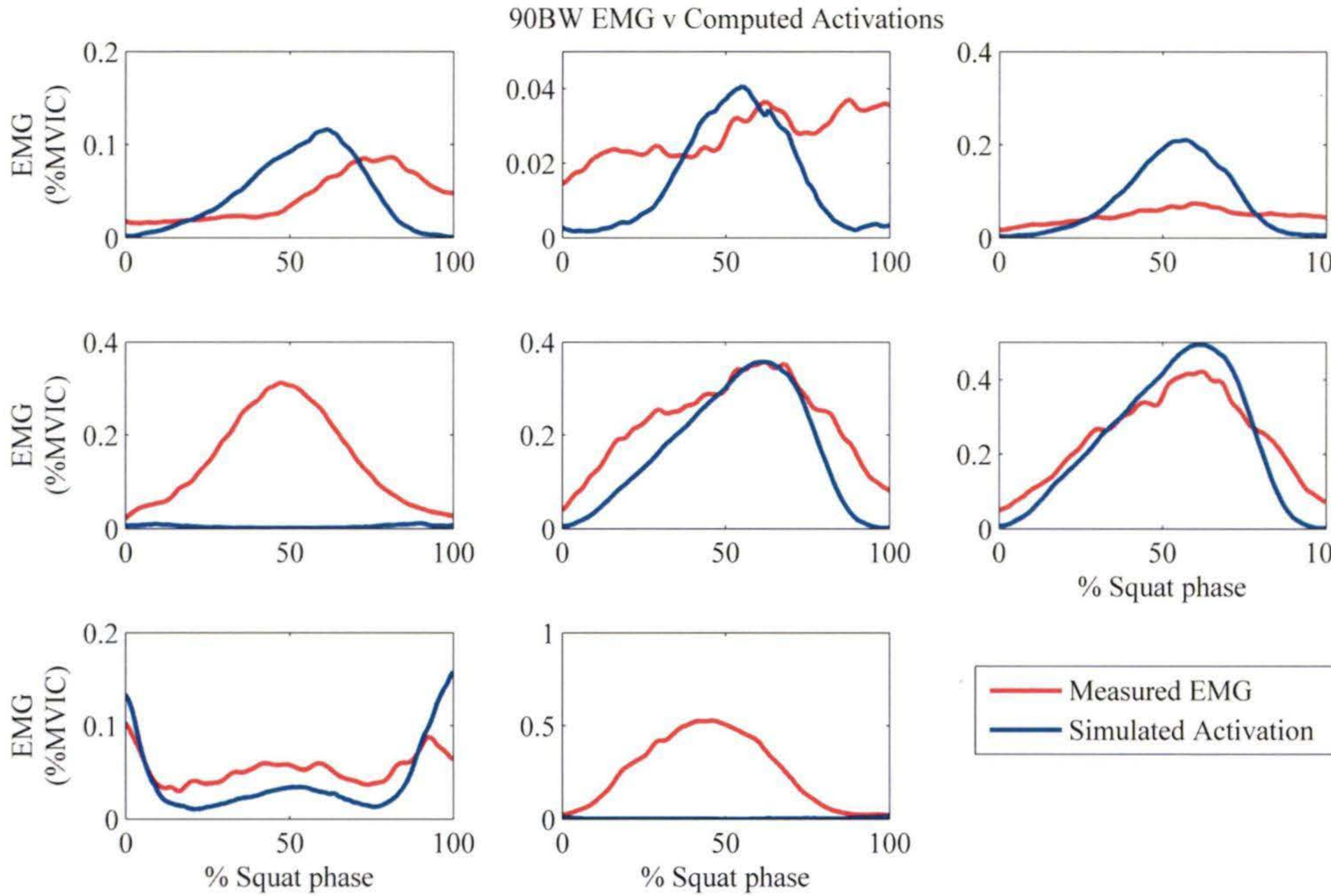

Figure 19. Mean EMG and simulated activations for 90BW (top row) Gluteus Maximus, Medial Hamstring, Lateral Hamstring, (middle row) Rectus Femoris, Vastus Medialis, Vastus Lateralis, (bottom row) Medial Gastrocnemius, and Tibialis Anterior.

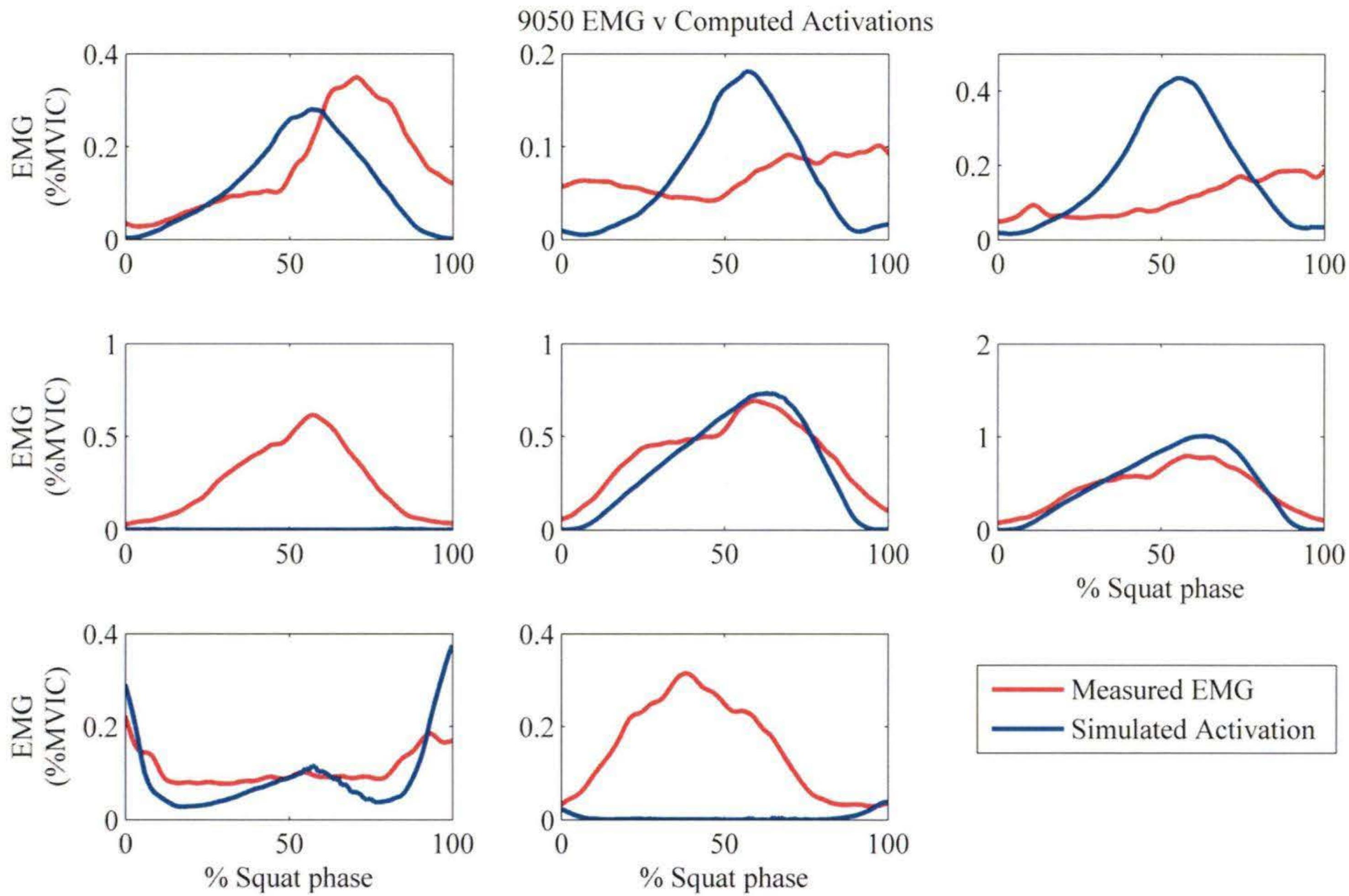

Figure 20. Mean EMG and simulated activations for 9050 (top row) Gluteus Maximus, Medial Hamstring, Lateral Hamstring, (middle row) Rectus Femoris, Vastus Medialis, Vastus Lateralis, (bottom row) Medial Gastrocnemius, and Tibialis Anterior.

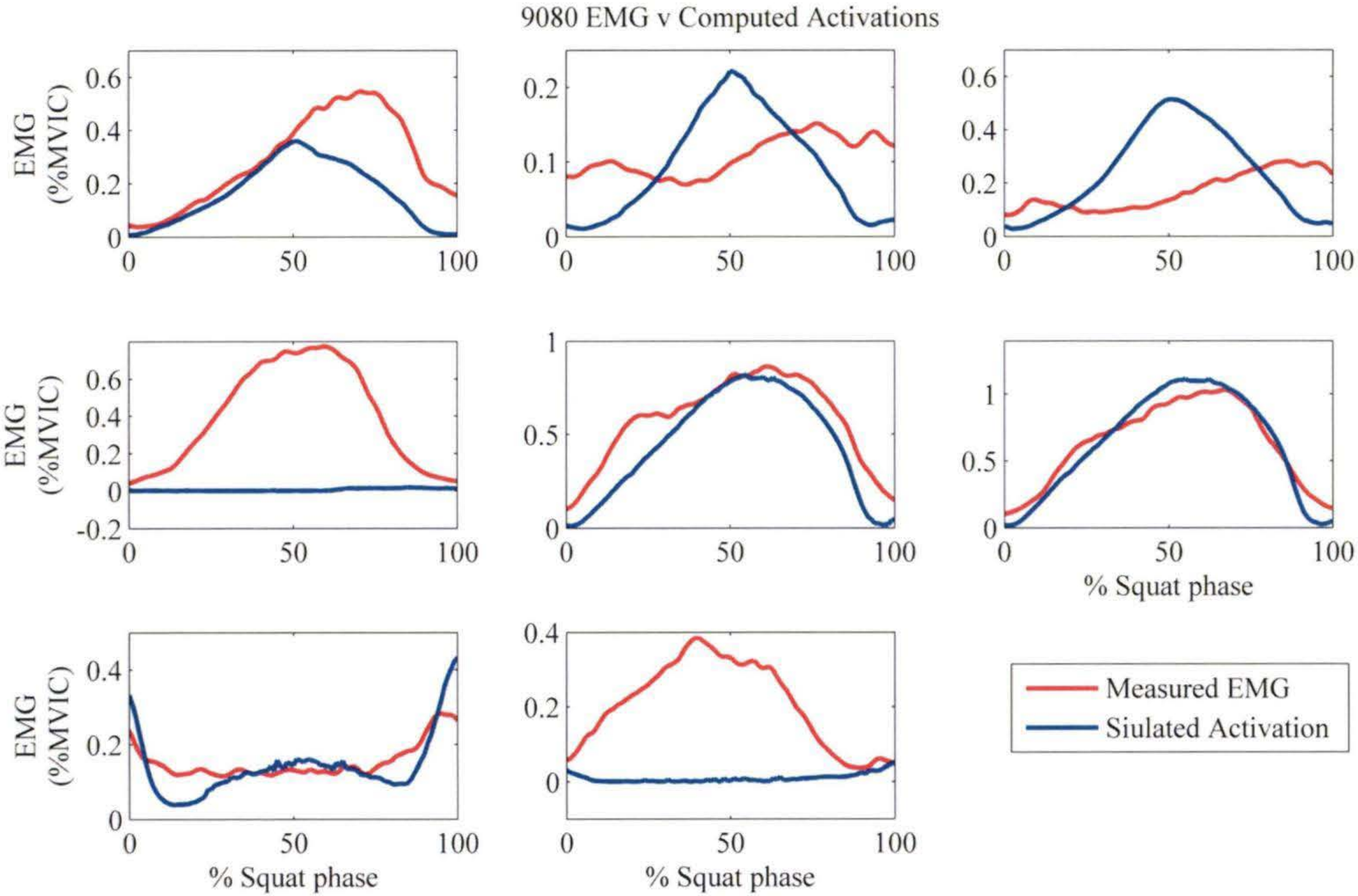

Figure 21. Mean EMG and simulated activations for 9080 (top row) Gluteus Maximus, Medial Hamstring, Lateral Hamstring, (middle row) Rectus Femoris, Vastus Medialis, Vastus Lateralis, (bottom row) Medial Gastrocnemius, and Tibialis Anterior.

REFERENCES

Albertus-Kajee, Y., Tucker, R., Derman, W., & Lambert, M. (2010). Alternative methods of normalising EMG during cycling. *Journal of Electromyography and Kinesiology, 20*(6), 1036-1043. doi: 10.1016/j.jelekin.2010.07.011

Amis, A. A., & Dawkins, G. P. (1991). Functional anatomy of the anterior cruciate ligament. Fibre bundle actions related to ligament replacements and injuries. *Journal of Bone and Joint Surgery. British Volume, 73*(2), 260-267.

Amis, A. A., Gupte, C. M., Bull, A. M., & Edwards, A. (2006). Anatomy of the posterior cruciate ligament and the meniscofemoral ligaments. *Knee Surgery, Sports Traumatology, Arthroscopy, 14*(3), 257-263. doi: 10.1007/s00167-005-0686-x

Anderson, F. C., & Pandy, M. G. (2001). Static and dynamic optimization solutions for gait are practically equivalent. *Journal of Biomechanics, 34*(2), 153-161.

Ariel, G. (1974). Method for biomechanical analysis of human performance. *Research Quarterly, 45*(1), 72-79.

Baechle, T. R., & Earle, R. W. (2008). *Essentials of strength training and conditioning*: Human kinetics.

Biel, A. R. (2010). *Trail Guide to The Body: A hands-on guide to locating muscles, bones and more*: Books of Discovery.

Boyden, G., Kingman, J., & Dyson, R. (2000). A comparison of quadriceps electromyographic activity with the position of the foot during the parallel squat. *The Journal of Strength & Conditioning Research, 14*(4), 379-382.

Bresler, B., & Frankel, J. (1950). The forces and moments in the leg during level walking. *Trans. Asme, 72*(27), 25-35.

Bryanton, M. A., Kennedy, M. D., Carey, J. P., & Chiu, L. Z. (2012). Effect of squat depth and barbell load on relative muscular effort in squatting. *Journal of Strength and Conditioning Research, 26*(10), 2820-2828. doi: 10.1519/JSC.0b013e31826791a7

Caterisano, A., Moss, R. F., Pellinger, T. K., Woodruff, K., Lewis, V. C., Booth, W., & Khadra, T. (2002). The effect of back squat depth on the EMG activity of 4 superficial hip and thigh muscles. *Journal of Strength and Conditioning Research, 16*(3), 428-432.

Chappell, J. D., Herman, D. C., Knight, B. S., Kirkendall, D. T., Garrett, W. E., & Yu, B. (2005). Effect of fatigue on knee kinetics and kinematics in stop-jump tasks. *American Journal of Sports Medicine, 33*(7), 1022-1029.

Clark, D. R., Lambert, M. I., & Hunter, A. M. (2012). Muscle activation in the loaded free barbell squat: a brief review. *Journal of Strength and Conditioning Research, 26*(4), 1169-1178. doi: 10.1519/JSC.0b013e31822d533d

Cram, J., Kasman, G., & Holtz, J. (1998). Introduction to surface electromyography. 1998. *Maryland, USA: Aspen Publication.*

De Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics, 29*(9), 1223-1230.

Delp, S. L., Anderson, F. C., Arnold, A. S., Loan, P., Habib, A., John, C. T., . . . Thelen, D. G. (2007). OpenSim: open-source software to create and analyze dynamic simulations of movement. *IEEE Transactions on Bio-medical Engineering, 54*(11), 1940-1950. doi: 10.1109/TBME.2007.901024

Donnelly, D. V., Berg, W. P., & Fiske, D. M. (2006). The effect of the direction of gaze on the kinematics of the squat exercise. *J Strength Cond Res, 20*(1), 145-150. doi: 10.1519/R-16434.1

Erdemir, A., McLean, S., Herzog, W., & van den Bogert, A. J. (2007). Model-based estimation of muscle forces exerted during movements. *Clinical Biomechanics, 22*(2), 131-154. doi: 10.1016/j.clinbiomech.2006.09.005

Escamilla, R. F. (2001). Knee biomechanics of the dynamic squat exercise. *Medicine and Science in Sports and Exercise, 33*(1), 127-141.

Escamilla, R. F., Fleisig, G. S., Lowry, T. M., Barrentine, S. W., & Andrews, J. R. (2001). A three-dimensional biomechanical analysis of the squat during varying stance widths. *Medicine and Science in Sports and Exercise, 33*(6), 984-998.

Escamilla, R. F., Fleisig, G. S., Zheng, N., Barrentine, S. W., Wilk, K. E., & Andrews, J. R. (1998). Biomechanics of the knee during closed kinetic chain and open kinetic chain exercises. *Medicine and Science in Sports and Exercise, 30*(4), 556-569.

Escamilla, R. F., Fleisig, G. S., Zheng, N., Lander, J. E., Barrentine, S. W., Andrews, J. R., . . . Moorman, C. T., 3rd. (2001). Effects of technique variations on knee biomechanics during the squat and leg press. *Medicine and Science in Sports and Exercise, 33*(9), 1552-1566.

Escamilla, R. F., Lander, J. E., & Garhammer, J. (2000). Biomechanics of powerlifting and weightlifting exercises. *Exercise and sport science*, 585-615.

Escamilla, R. F., Zheng, N., MacLeod, T. D., Imamura, R., Edwards, W. B., Hreljac, A., . . . Andrews, J. R. (2010). Cruciate ligament tensile forces during the forward and side lunge. *Clinical Biomechanics, 25*(3), 213-221. doi: 10.1016/j.clinbiomech.2009.11.003

Flanagan, S. P., & Salem, G. J. (2008). Lower extremity joint kinetic responses to external resistance variations. *Journal of Applied Biomechanics, 24*(1), 58-68.

Griffin, L. Y., Albohm, M. J., Arendt, E. A., Bahr, R., Beynnon, B. D., Demaio, M., . . . Yu, B. (2006). Understanding and preventing noncontact anterior cruciate ligament injuries: a review of the Hunt Valley II meeting, January 2005. *American Journal of Sports Medicine, 34*(9), 1512-1532. doi: 10.1177/0363546506286866

Grood, E. S., & Suntay, W. J. (1983). A joint coordinate system for the clinical description of three-dimensional motions: application to the knee. *Journal of Biomechanical Engineering, 105*(2), 136-144.

Gullet, J. C. (2009). A Biomechanical Comparison Of Back and Front Squats IN healthy Trained Individuals. *Journal of Strength and Conditioning Research, 23*(1), 284-292.

Gullett, J. C., Tillman, M. D., Gutierrez, G. M., & Chow, J. W. (2009). A biomechanical comparison of back and front squats in healthy trained individuals. *Journal of Strength and Conditioning Research, 23*(1), 284-292. doi: 10.1519/JSC.0b013e31818546bb

Hamill, J. K., K.M. (2009). *Biomechanical Basis of HUman Movement* (3 ed.). Baltimore, MD: Lippincott Williams & Wilkins.

Hewett, T. E., Myer, G. D., Ford, K. R., Heidt, R. S., Jr., Colosimo, A. J., McLean, S. G., . . . Succop, P. (2005). Biomechanical measures of neuromuscular control and valgus loading of the knee predict anterior cruciate ligament injury risk in female athletes: a prospective study. *American Journal of Sports Medicine, 33*(4), 492-501. doi: 10.1177/0363546504269591

Kamen, G., & Gabriel, D. A. (2010). *Essentials of electromyography*: Human Kinetics Publishers.

Kato, Y., Ingham, S. J., Maeyama, A., Lertwanich, P., Wang, J. H., Mifune, Y., . . . Fu, F. H. (2012). Biomechanics of the human triple-bundle anterior cruciate ligament. *Arthroscopy, 28*(2), 247-254. doi: 10.1016/j.arthro.2011.07.019

Klein, K. K. (1961). The deep squat exercise as utilized in weight training for athletes and its effect on the ligaments of the knee. *Journal of the Association for Physical and Mental Rehabilitation, 15*(1), 6-11.

Laughlin, W. A., Weinhandl, J. T., Kernozek, T. W., Cobb, S. C., Keenan, K. G., & O'Connor, K. M. (2011). The effects of single-leg landing technique on ACL loading. *Journal of Biomechanics, 44*(10), 1845-1851. doi: 10.1016/j.jbiomech.2011.04.010

Lieber, R. L., & Bodine-Fowler, S. C. (1993). Skeletal muscle mechanics: implications for rehabilitation. *Physical Therapy, 73*(12), 844-856.

McCaw, S. T., & Melrose, D. R. (1999). Stance width and bar load effects on leg muscle activity during the parallel squat. *Medicine and Science in Sports and Exercise, 31*(3), 428-436.

Newton, R. U., Murphy, A. J., Humphries, B. J., Wilson, G. J., Kraemer, W. J., & Häkkinen, K. (1997). Influence of load and stretch shortening cycle on the kinematics, kinetics and muscle activation that occurs during explosive upper-body movements. *Eur J Appl Physiol Occup Physiol, 75*(4), 333-342.

Noyes, F. R., Butler, D. L., Grood, E. S., Zernicke, R. F., & Hefzy, M. S. (1984). Biomechanical analysis of human ligament grafts used in knee-ligament repairs and reconstructions. *Journal of Bone and Joint Surgery. American Volume, 66*(3), 344-352.

Oatis, C. A. (2009). *Kinesiology : the mechanics and pathomechanics of human movement* (2nd ed.): Lippincott Williams & Wilkins

Paoli, A., Marcolin, G., & Petrone, N. (2009). The effect of stance width on the electromyographical activity of eight superficial thigh muscles during back squat with different bar loads. *J Strength Cond Res, 23*(1), 246-250.

Pauletto, B. (1991). *Strength training for coaches*: Leisure Press.

Race, A., & Amis, A. A. (1994). The mechanical properties of the two bundles of the human posterior cruciate ligament. *Journal of Biomechanics, 27*(1), 13-24.

Rack, P. M., & Westbury, D. R. (1969). The effects of length and stimulus rate on tension in the isometric cat soleus muscle. *Journal of Physiology, 204*(2), 443-460.

Sahli, S., Rebai, H., Elleuch, M. H., Tabka, Z., & Poumarat, G. (2008). Tibiofemoral joint kinetics during squatting with increasing external load. *J Sport Rehabil, 17*(3), 300-315.

Schoenfeld, B. J. (2010). Squatting kinematics and kinetics and their application to exercise performance. *Journal of Strength and Conditioning Research, 24*(12), 3497-3506. doi: 10.1519/JSC.0b013e3181bac2d7

Schwarzenegger, A. (1985). *The Arnold Schwarzenegger Encyclopedia of Modern Bodybuilding*. New York Simon and Schuster Inc.

Steele, K. M., Demers, M. S., Schwartz, M. H., & Delp, S. L. (2012). Compressive tibiofemoral force during crouch gait. *Gait Posture, 35*(4), 556-560. doi: 10.1016/j.gaitpost.2011.11.023

Swinton, P. A., Lloyd, R., Keogh, J. W., Agouris, I., & Stewart, A. D. (2012). A biomechanical comparison of the traditional squat, powerlifting squat, and box squat. *Journal of Strength and Conditioning Research, 26*(7), 1805-1816. doi: 10.1519/JSC.0b013e3182577067

Taylor, W. R., Heller, M. O., Bergmann, G., & Duda, G. N. (2004). Tibio-femoral loading during human gait and stair climbing. *Journal of Orthopaedic Research, 22*(3), 625-632. doi: 10.1016/j.orthres.2003.09.003

Tortora, G. J. D., B. (2012). *Principles of Anatomy and Physiology* (13th ed.): John Wiley & Sons

Weinhandl, J. T., Earl-Boehm, J. E., Ebersole, K. T., Huddleston, W. E., Armstrong, B. S., & O'Connor, K. M. (2013). Anticipatory effects on anterior cruciate ligament loading during sidestep cutting. *Clinical Biomechanics, 28*(6), 655-663.

Weinhandl, J. T., Joshi, M., & O'Connor, K. M. (2010). Gender comparisons between unilateral and bilateral landings. *Journal of Applied Biomechanics, 26*(4), 444-453.

Weinhandl, J. T., & O'Connor, K. M. (2010). Assessment of a greater trochanter-based method of locating the hip joint center. *Journal of Biomechanics, 43*(13), 2633-2636. doi: 10.1016/j.jbiomech.2010.05.023

Westing, S. H., Cresswell, A. G., & Thorstensson, A. (1991). Muscle activation during maximal voluntary eccentric and concentric knee extension. *Eur J Appl Physiol Occup Physiol, 62*(2), 104-108.

Wilk, K. E., Escamilla, R. F., Fleisig, G. S., Barrentine, S. W., Andrews, J. R., & Boyd, M. L. (1996). A comparison of tibiofemoral joint forces and electromyographic activity during open and closed kinetic chain exercises. *American Journal of Sports Medicine, 24*(4), 518-527.

Woo, S. L., Hollis, J. M., Adams, D. J., Lyon, R. M., & Takai, S. (1991). Tensile properties of the human femur-anterior cruciate ligament-tibia complex. The effects of specimen age and orientation. *American Journal of Sports Medicine, 19*(3), 217-225.

Wrentenberg, P. (1993). Joint moments of force and quadriceps muscle activity during squatting exercise. *Scandinavian Journal of Medicine and Science in Sports, 3*, 244-250.

Wretenberg, P., Arborelius, U. P., Weidenhielm, L., & Lindberg, F. (1993a). Joint moments of force and quadriceps muscle activity during squatting exercise. *Scandinavian Journal of Rehabilitation Medicine, 25*(4), 153-159.

Wretenberg, P., Arborelius, U. P., Weidenhielm, L., & Lindberg, F. (1993b). Joint moments of force and quadriceps muscle activity during squatting exercise. *Scandinavian Journal of Rehabilitation Medicine, 25*(4), 153-159.

Yoo, Y.-S., Jeong, W.-S., Shetty, N. S., Ingham, S. J., Smolinski, P., & Fu, F. (2010). Changes in ACL length at different knee flexion angles: an in vivo biomechanical study. *Knee Surgery, Sports Traumatology, Arthroscopy, 18*(3), 292-297.

Appendix A: Informed Consent Document

INFORMED CONSENT DOCUMENT
OLD DOMINION UNIVERSITY

PROJECT TITLE: The effects of squat technique on lower extremity kinematics, kinetics and muscle activation patterns.

INTRODUCTION

The purposes of this form are to give you information that may affect your decision whether to say YES or NO to participation in this research, and to record the consent of those who say YES. This project is being carried out to determine the effects of knee flexion and the changes in force and EMG activity as knee flexion increase. The research will be conducted at Old Dominion University in the Neuromechanics Laboratory.

RESEARCHERS

Joshua Weinhandl, PhD, Responsible Project Investigator, Assistant Professor, Darden College of Education, Department of Human Movement Sciences, Old Dominion University
Zach Sievert, Graduate Student, Department of Human Movement Sciences, Old Dominion University

DESCRIPTION OF RESEARCH STUDY

The purpose of this study is to compare traditional and deep squats. A total of 20 subjects will participate in this study. You will report to the Neuromechanics Laboratory, SRC 1007, for two sessions. You will be asked to participate in one, one hour-long testing session and one, two hour-long session.

The first session will involve performing a warm up of a traditional back squat and a one repetition maximum in the traditional back squat. On the second testing day, you will be asked to perform five body weight squats. Then will be asked to warm up with a traditional back squat. After you feel adequately warm, you will be asked to perform five repetitions of the traditional back squat with 50% and five repetition at 80% of your 1 repetition maximum to 90° of knee flexion and five repetitions at 50% and five at 80% to 125° of knee flexion.

Data collection will consist of measurements of height and weight. Proper attire for physical activity is suggested, however clean, tight-fitting shorts will be provided. Single reflective markers will be placed on specific anatomical landmarks. Furthermore, eight EMG markers will be placed on eight right leg muscles. After reflective markers and EMG markers have been applied, three-dimensional kinematics and kinetics will be collected during the squat exercises. During the visit to the lab, you will be asked fill out a Medical History and Physical Activity Questionnaire. This includes questions pertaining to age, physical activity, lower body injury(ies), recent head injury(ies), or any medications that may cause dizziness, etc.

EXCLUSIONARY CRITERIA

To be eligible to participate you must be physically active three days a week for at least 30 minutes and perform at least one session of lower body resistance training per week.

You will not be able to participate in the study if you:

- have suffered any injuries to the lower extremities within the last six months.
- have ever had surgery to your ankles, knees, hips, and or lower back.
- are pregnant or think you may be pregnant.

RISKS AND BENEFITS

RISKS:

If you decide to participate in this study, then you may experience general muscle soreness and/or minor skin irritation due to the adhesive on the external markers. It is also possible, although unlikely, that you may experience musculoskeletal injury such as a muscle strain or ankle sprain.

To reduce the above risks, care will be taken when applying and removing the external markers. If you feel any soreness or irritation while participating in this study, please tell the investigators as soon as possible. The investigators will provide you with the appropriate information for treating these problems, based on our level of expertise. If you are injured while participating in this research study, you will initially be provided care by the investigator(s) and will then be referred to the Student Health Services (students) or your personal physician (non-students) for follow-up care.

There is a small risk of loss of confidentiality. To minimize this risk all information gathered from you will be confidential in nature and stored in the laboratory of the principle investigator (SRC 1007). Only the principal investigator and research personal will have access to these files.

BENEFITS:

There are no direct benefits from participating in this study. The information obtained in this study will expand our knowledge base leading to beneficial changes in the future

COSTS AND PAYMENTS

The researchers are unable to give you any payment for participating in this study. Students in Exercise Science courses may be offered extra credit for participation per individual course policy. Total point value for participation will be determined by course instructor. Students in courses offering extra credit for research participation may receive extra credit of equal point value by completing an alternative assignment

NEW INFORMATION

If the researchers find new information during this study that would reasonably change your decision about participating, then they will give it to you.

CONFIDENTIALITY

The researchers will provide confidentiality to all documents regarding patient information, including questionnaires and test results, by storing information in a safe, locked location. The results of this study may be used within reports, presentations or publication; but all personal identifiers will be disregarded. However, your records may be subpoenaed by court order or inspected by government bodies with oversight authority. Following the completion of the study, all subject information will be destroyed.

WITHDRAWAL PRIVILEGE

In the event that you no longer wish to participate, you have the right to discontinue participation for this study. Even if you initially wish to participate you can withdraw at any time. Your decision will not affect your relationship with Old Dominion University or cause a loss of benefits to which you might otherwise be entitled. In addition, the researcher has the right to withdraw your participation if they find potential problems with your continued participation.

COMPENSATION FOR ILLNESS AND INJURY

If you participate, then your consent in this document does not waive any of your legal rights. In the event that you are harmed or injured from participating in this study, neither Old Dominion University nor the researchers are able to give you any compensation, including money, insurance coverage, or free medical care. In the event that you suffer any injury from participation in this study, you may contact Dr. Joshua Weinhandl at (757) 683-4754, Dr. George Maihafer, the current IRB chair, at (757) 683-4520, or Office of Research (757) 683-3460 at Old Dominion University.

VOLUNTARY CONSENT

By signing this form, you are saying several things. You are saying that you have read this form or have had it read to you, that you are satisfied that you understand this form, the research study, and its risks and benefits. The researchers should have answered any questions you may have had about the research. If you have any questions later on, then the researchers should be able to answer them:

Joshua Weinhandl 757-683-4754

If at any time you feel pressured to participate, or if you have any questions about your rights or this form, then you should call Dr. George Maihafer, the current IRB chair, at 757-683-4520, or the Old Dominion University Office of Research, at 757-683-3460.

And importantly, by signing below, you are telling the researcher YES, that you agree to participate in this study. The researcher should give you a copy of this form for your records.

Subject's Printed Name & Signature	**Date**

INVESTIGATOR'S STATEMENT

I certify that I have explained to this subject the nature and purpose of this research, including benefits, risks, costs, and any experimental procedures. I have described the rights and protections afforded to human subjects and have done nothing to pressure, coerce, or falsely entice this subject into participating. I am aware of my obligations under state and federal laws, and promise compliance. I have answered the subject's questions and have encouraged him/her to ask additional questions at any time during the course of this study. I have witnessed the above signature(s) on this consent form.

Investigator's Printed Name & Signature	**Date**

Appendix B: Medical History & Physical Activity Questionnaire

Medical History & Physical Activity Questionnaire

Please answer the following questions to the best of your ability:

Gender:

☐Male ☐Female

Race/ethnicity *(please check all that apply)*:

☐American Indian/Alaska Native
☐Asian
☐Native Hawaiian or other Pacific Islander
☐Hispanic or Latino
☐Black or African American
☐White

Which leg would you use to kick a ball?

☐Right ☐Left

To be completed by investigator:

Age:_______ yr Height: _______ m Mass: _______ kg

Medical History Questionnaire

For your safety, a list of conditions that would make you unable to participate in this study has been prepared. Please read this list carefully and consider whether any of the conditions apply to you. If any of these conditions are true for you, you will not be able to participate in this study. For each condition, please indicate "yes" or "no" if this is true or not for you.

☐Yes ☐No Are you currently physically active at a moderate level for at least 30 minutes/day, at least 3 days of the week and perform at least one session of lower body resistance training per week?

☐Yes ☐No Do you have a medical condition that may impair your balance performance (i.e. concussion, neurological impairments, etc)?

☐Yes ☐No Are you taking medications/drugs that may make you dizzy or make you tired (i.e. cold medications, sleeping medications, muscle relaxants)?

☐Yes ☐No Have you ever had a lower extremity injury that caused you to decrease the amount of physical activity you undertake? If yes, please complete the following:

☐Yes ☐No Hip injury(ies)
If yes, approximately how many injuries?__________________

☐Yes ☐No Knee injury(ies)
If yes, approximately how many injuries?__________________

☐Yes ☐No Ankle/foot injury(ies)
If yes, approximately how many injuries?__________________

☐Yes ☐No Have you had, <u>in the last 6 months</u>, a lower extremity injury that caused you to decrease the amount of physical activity you undertake?

☐Yes ☐No Do you <u>currently</u> have any lower extremity pain or injury(ies)?

☐Yes ☐No Have you <u>ever</u> had major orthopedic surgery on your lower extremities?

☐Yes ☐No Are you pregnant or do you have reason to believe that you may be

pregnant

Appendix C: Joint Angles

Appendix C represents the joint angles among the right hip, knee, and ankle during the 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line) squatting conditions.

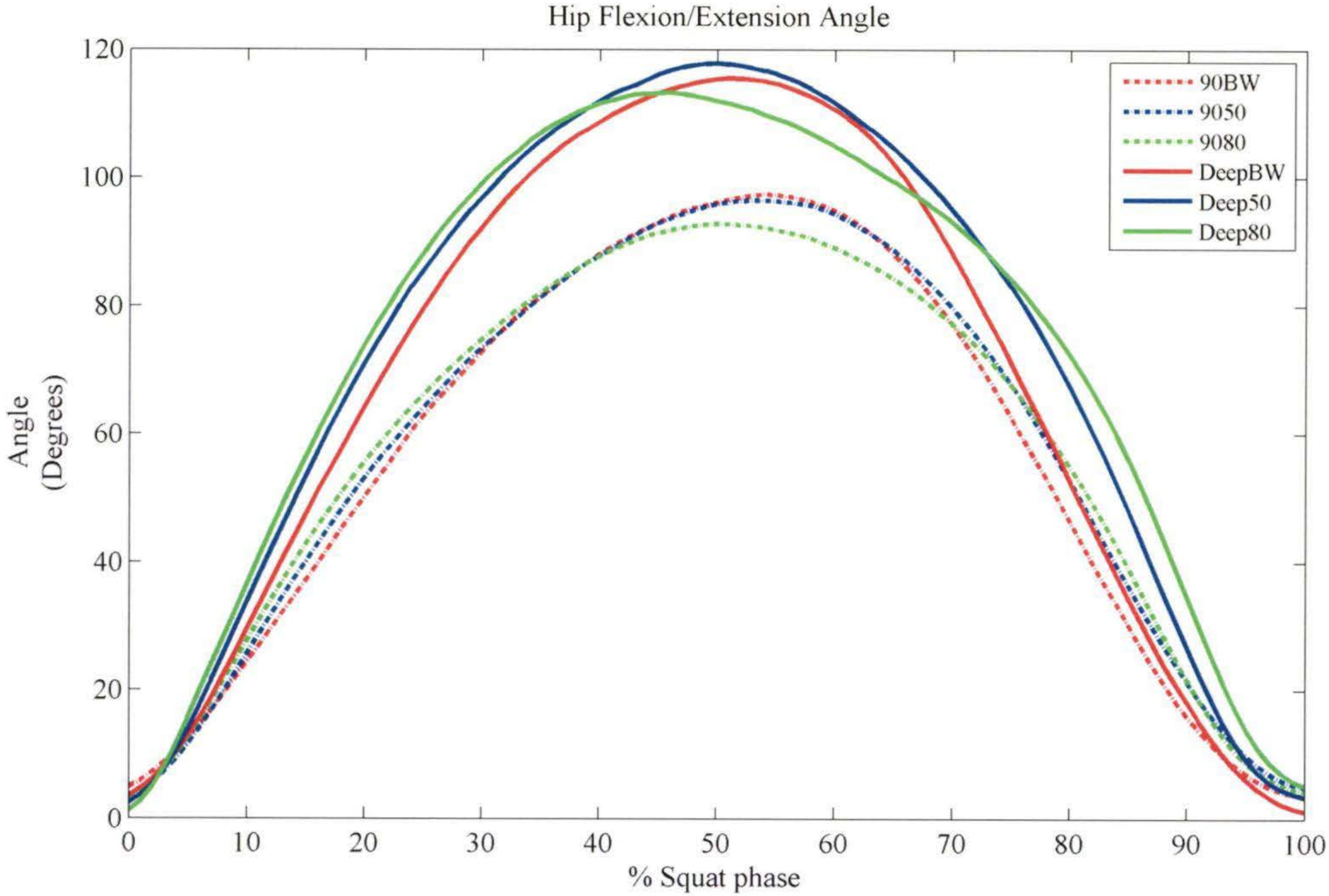

Figure C1. Mean hip flexion and extension angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents extension angles while positive represents flexion angles.

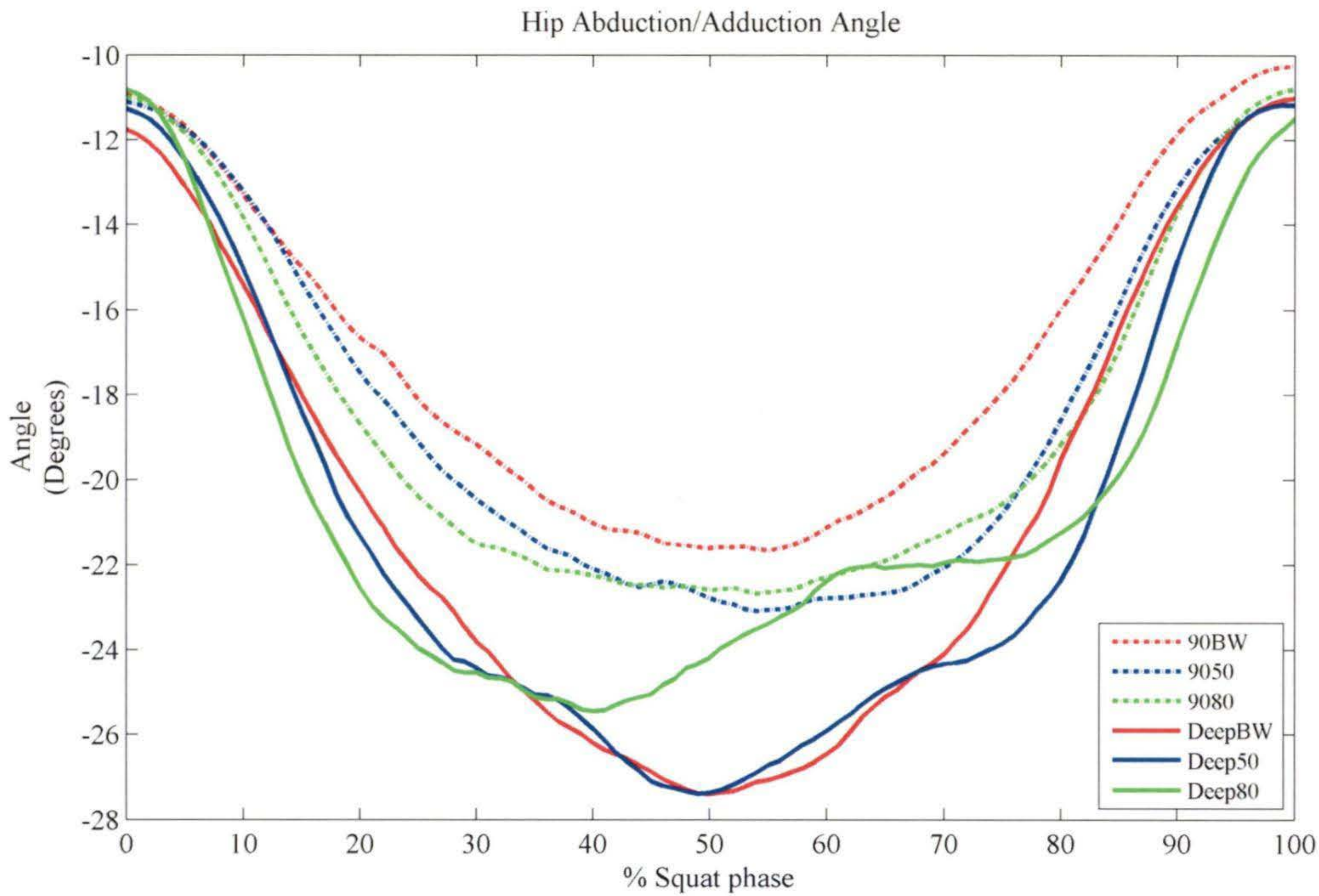

Figure C2. Mean hip adduction and abduction angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents abduction angles while positive represents adduction angles.

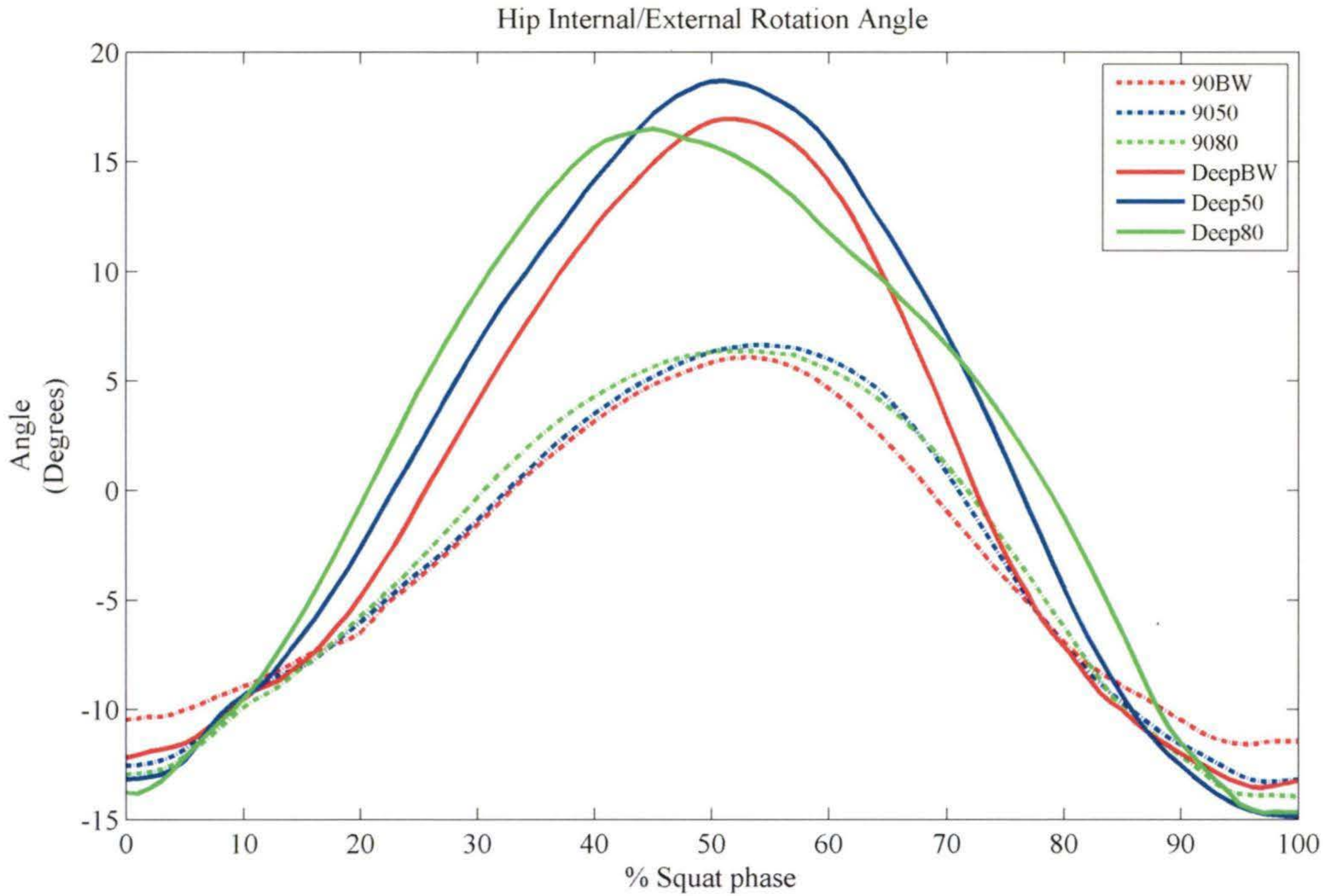

Figure C3. Mean hip internal and external angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents external rotation angles while positive represent internal rotation angles.

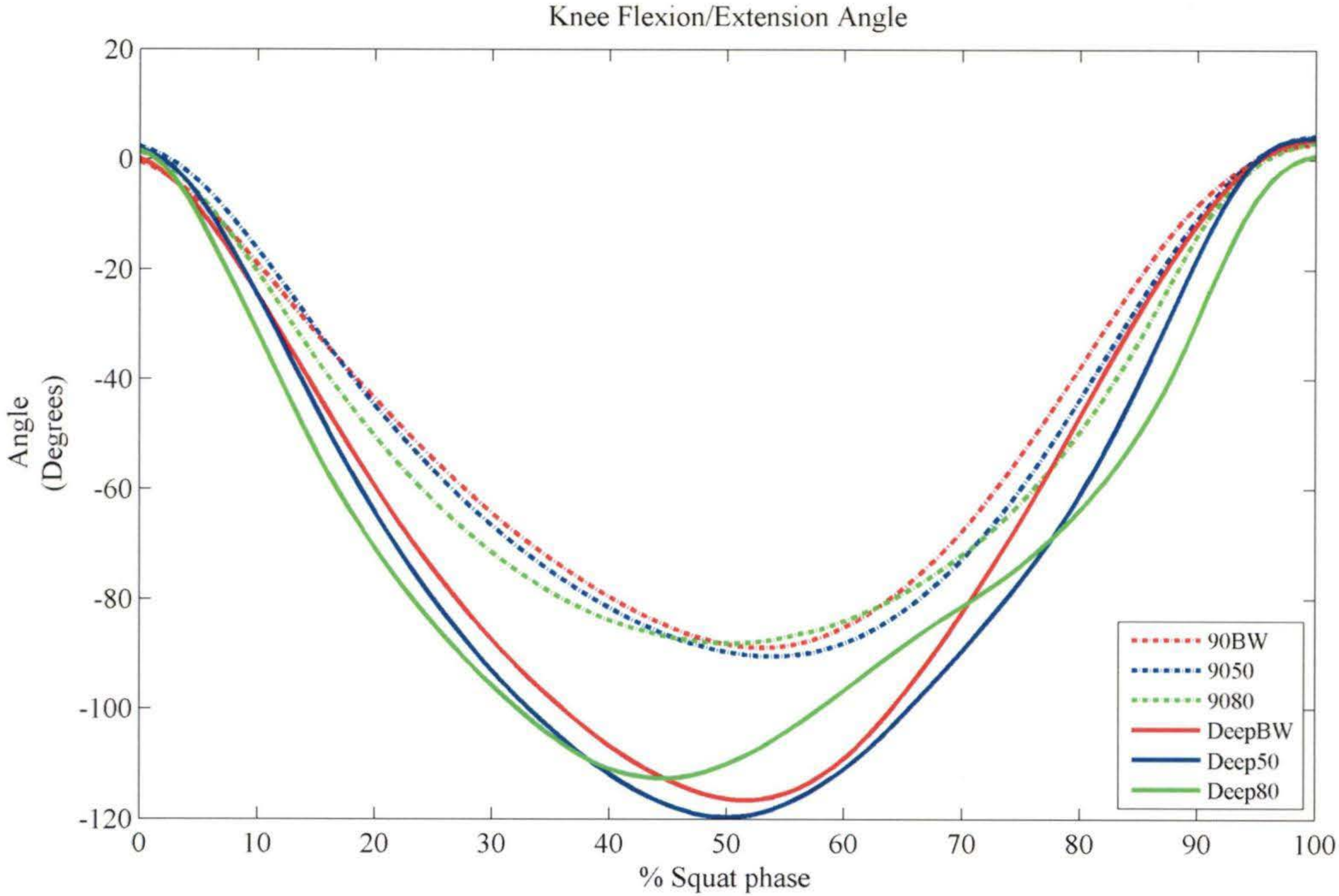

Figure C4. Mean knee flexion and extension angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents flexion angles while positive represents extension angles.

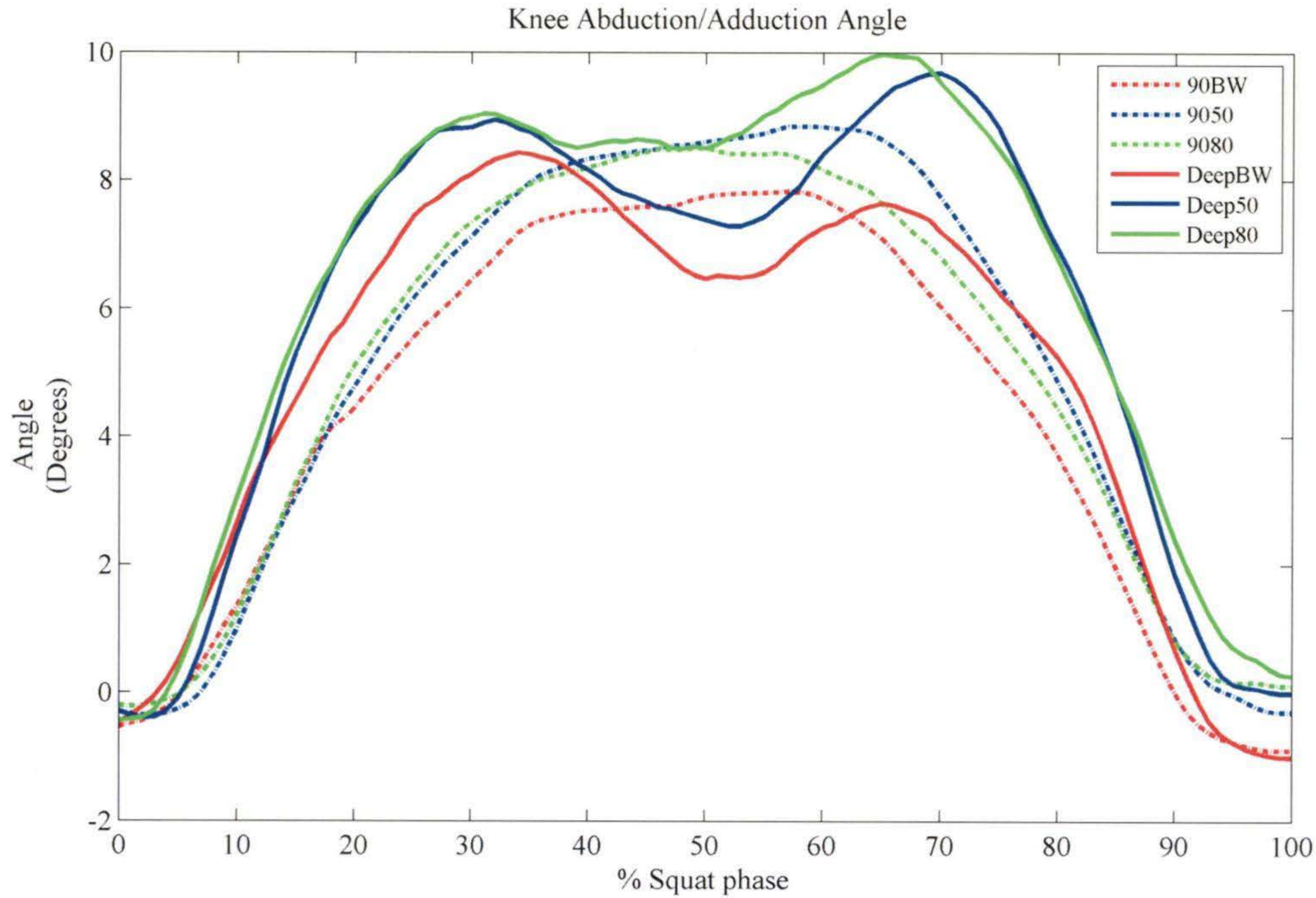

Figure C5. Mean knee abduction and adduction angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents abduction angles while positive represents adduction angles.

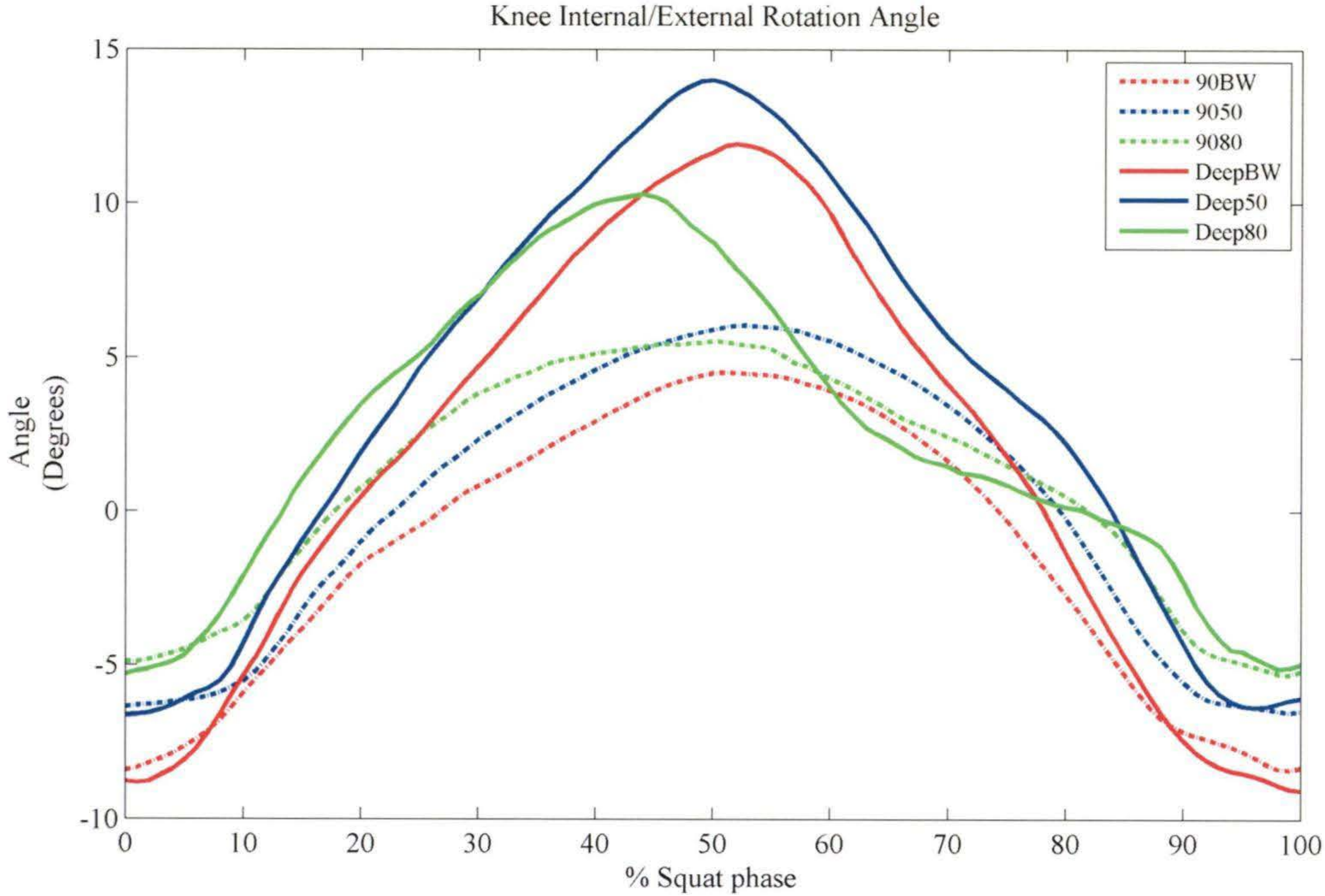

Figure C6. Mean knee internal and external rotation angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents external rotation while positive represents internal rotation

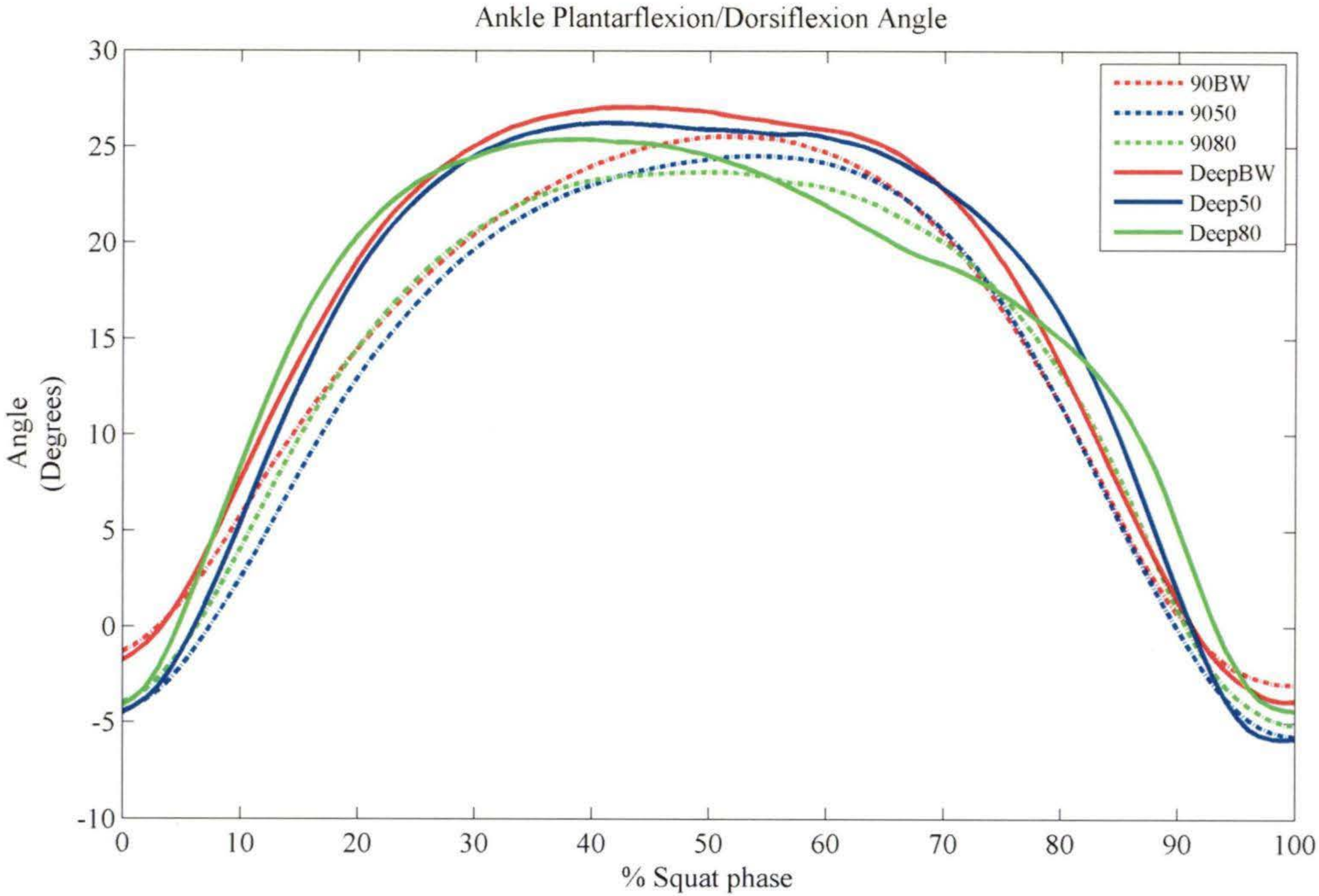

Figure C7. Mean ankle plantarflexion and dorsiflexion angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents plantarflexion while positive represents dorsiflexion.

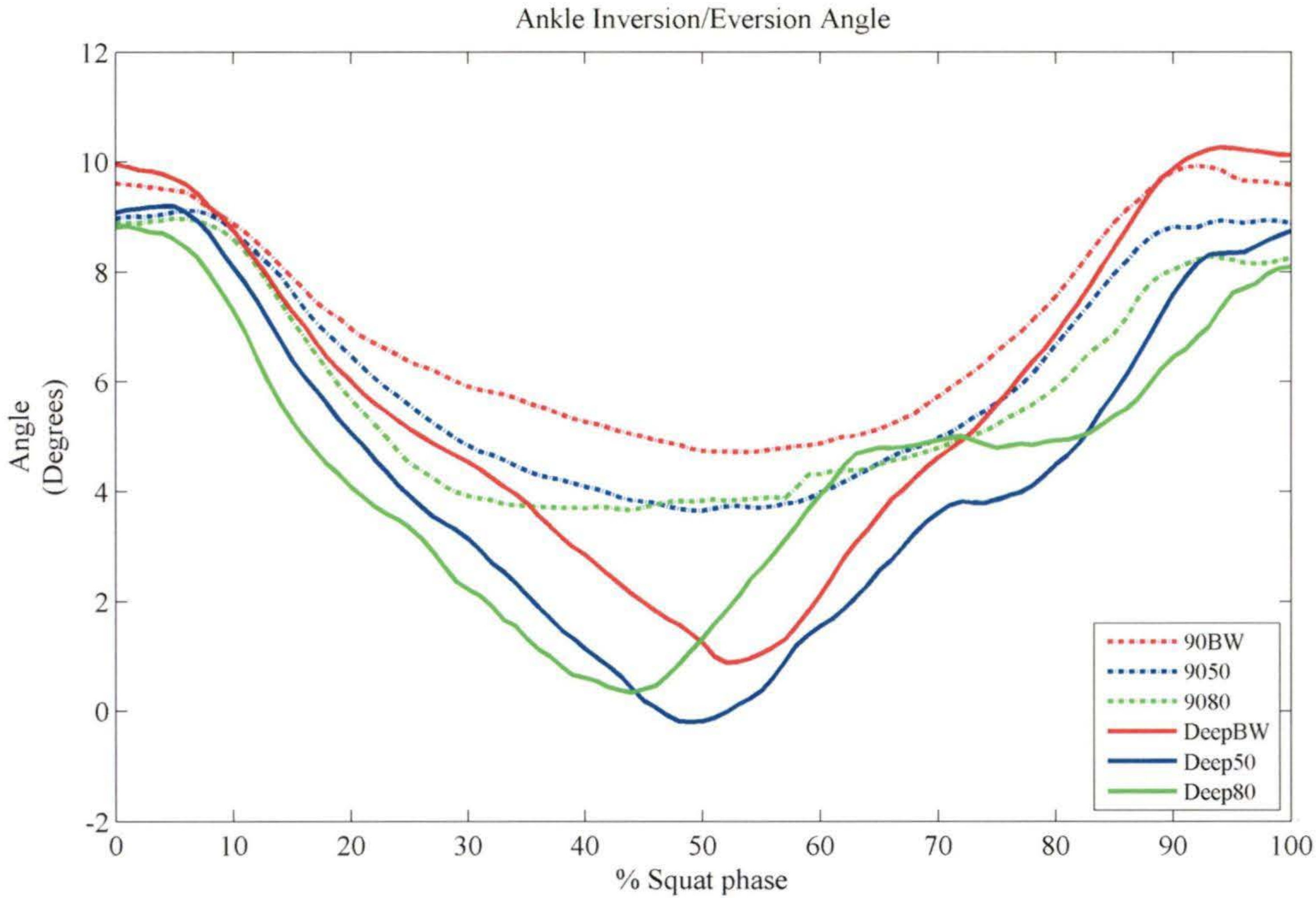

Figure C8. Mean ankle angles for 90BW (Red dash), 9050 (Blue dash), 9080 (Green dash), DeepBW (Red line), Deep50 (Blue line), Deep80 (Green line). Negative represents eversion angles while positive represents inversion angles.

Curriculum Vitae

Education:

Old Dominion University
M.S. (Exercise Science) – expected August 2014
Thesis: *"The Effects of Squat Depth on ACL, PCL, and Tibiofemoral Contact Forces."*

College of Mount St. Joseph
B.S. (General Studies) – May 2012

Relevant Coursework:

Exercise Physiology	Fall 2012
Advanced Cardiovascular Exercise Physiology	Spring 2013
Clinical Exercise Testing and Prescription	Spring 2013
Exercise Prescription for Chronic Disease	Spring 2014
Biomechanics	Fall 2012
Advanced Biomechanics	Fall 2013

Professional Experience:

Teaching:

Graduate Teaching Assistant, Old Dominion University — 2013-2014
Biomechanics Lab (EXSC 417)

Adjunct Instructor, Old Dominion University — 2013-Present
Anatomical Kinesiology Lecture (EXSC 322)
Anatomical Kinesiology Lab (EXSC 322)

Research:

Neuromechanics Laboratory, Old Dominion University — 2012-Present
Research Assistant
Administrate the day-to-day technical and scientific operations of the lab. Oversee funded research, undergraduate and graduate student research projects.

Publications:

Works in Review:

Weinhandl JT, Irmischer BS & **Sievert ZA**. Gender differences in unilateral landing mechanics from absolute and relative drop heights. *Journal of Biomechanics.*

Weinhandl JT, Irmischer BS, **Sievert ZA** & Fontenot KC. Influence of gender and limb dominance on lower extremity joint mechanics during unilateral land-and-cut maneuvers. *Journal of Biomechanics.*

Irmischer BS, Weinhandl JT & **Sievert ZA**. Effects of gait speed on hip joint forces. *Journal of Biomechanics.*

Presentations:

Sievert ZA, Irmischer BS, & Weinhandl JT. Gender differences in unilateral landing mechanics from absolute and relative drop heights. American Society of Biomechanics 37th Annual Meeting. Omaha, NE. September 4-7, 2013.

Current Research Projects

"The influence of squat depth on lower extremity kinematics, kinetics, and muscle activations."

"Effects of femoracetabular impingement on lower extremity gait mechanics."

"Influence of ankle joint immobilization of lower extremity joint coupling coordination."

Research Presentations:

Fontenot K, Pitman E, **Sievert ZA** & Weinhandl JT. The influence of gender and limb dominance on lower extremity joint mechanics during land-and-cut maneuvers. 7th World Congress of Biomechanics.July 6-11, 2014.

Irmischer BS, **Sievert ZA** & Weinhandl JT. Effects of walking speed on hip joint forces. 7th World Congress of Biomechanics. July 6-11, 2014.

Pitman E, Fontenot K, **Sievert ZA** & Weinhandl JT. Comparison of unilateral drop landings and land-and-cut maneuvers. 7th World Congress of Biomechanics. July 6-11, 2014.

Sievert ZA, Irmischer, BS & Weinhandl JT. Gender differences in unilateral landing mechanics from absolute and relative drop heights. American Society of Biomechanics 37th Annual Meeting. Omaha, NE. September 4-7

Professional Meetings Attended:

2013 – 37th Annual Meeting of the American Society of Biomechanics. Omaha, NE.

Professional Memberships:

American Society of Biomechanics 2013-Present

Computer Skills:

Microsoft Word, Excel, and PowerPoint
Vicon
Visual 3D
Opensim
Matlab