Abstract

Muscles produce force, resulting in moments about a joint, causing movement of the body. Muscle forces are estimated with a Hill-type model incorporating four parameters; optimal fiber length (OFL), tendon slack length, physiological cross sectional area (PCSA), pennation angle, and maximal isometric force (F^{max}) scaled to individual subjects. **Purpose:** The purpose of this study was to determine if subject specific musculotendon parameters estimated in vivo using ultrasound would better estimate moments produced about a joint compared to previous scaling methods. **Methods:** 7 recreationally active and resistance trained males and females with no history of lower extremity injury participated. Subjects performed single-leg squats while kinematic, kinetic, and muscle activation data was recorded. Two models for each subject were used in SIMM to estimate knee moments, activations, and muscle forces: scaled (SC) and ultrasound-based (US). Ultrasound imaging of the primary knee muscles were used to derive subject-specific muscle parameters. Scaled muscle parameters were scaled from the model's generic muscle parameter values. **Results:** The scaled model produced approximately 50% more error compared to the ultrasound model (RMSE: US= 2.71Nm vs. SC= 6.08 Nm) when comparing inverse dynamics knee moments to each model. EMG analysis showed less error in the ultrasound vs. scaled models when compared to experimental muscle activation (RMSE: US= 0.16 mV ± .07, SC= 0.23 mV ± .09) ($p< .05$). Hamstring activation error was not statistically different between models (RMSE: US= 0.13 mV ± .07 vs. SC= 0.11 mV ± .04) ($p> .05$). Correlations between model and experimental EMG were weak to modest in both models for all muscles [Quadriceps: (US r= 0.50 mV ± .45, $p< 0.01$, SC r= 0.55 mV ± .27, $p<0.01$), Hamstrings: (US r= 0.44 mV ± .25, $p< 0.01$; SC r= 0.23 mV ± .30, $p< .01$)]

Conclusion: Advances in methodologies used in the field of biomechanical musculoskeletal modeling could be applied to a variety of pathological patients enabling researchers and physicians to better understand how pathology relates with muscle function. More research is warranted in the attempt of deriving a more physiological relevant muscle modeling technique.

Comparison of Scaled vs. Ultrasound Based Musculoskeletal Models on Knee Muscle Moments During Single-Leg Squatting

A Thesis Proposal Presented to
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Comparison of Scaled vs. Ultrasound Based Musculoskeletal Models on Knee Muscle Moments During Single-Leg Squatting

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Table of Contents:

Figures and Tables:

Chapter 1: Thesis Introduction

Advances in methodologies used in the field of biomechanical musculoskeletal modeling could be used in a broad range of applications, from sports medicine and athlete rehabilitation to geriatric therapy. New methods could be applied to a variety of pathological patients enabling researchers and physicians to better understand how pathology relates with muscle function.

Muscle force production results in moments about a joint, causing movement of the body. Muscle forces are estimated based on both muscle anatomy and several physiological parameters. These parameters are used as inputs in a Hill-muscle model. The Hill-muscle model consists of 3 elements, a contractile element, and two non-linear elastic elements, one in series and another in parallel (Hill, 1939). The contractile element represents muscle's contractile parts (actin and myosin cross-bridge), and the elastic element represents connective tissue's passive and intrinsic elastic energy storage. In order to calculate muscle forces the Hill model uses several musculotendon parameters, maximal isometric force (F^{max}), pennation angle, optimal fiber (fascicle) length, and tendon slack length.

Musculotendon parameters used in the Hill muscle model to estimate force have traditionally been calculated using scaling (Delp et al 1990, Winby et al 2008) or scaling and optimization methods (Lloyd and Buchanan 2005). Scaling is a process by which the modeling system fits a generic musculoskeletal model to a specific subject based on a percent difference in musculotendon length in the scaled verses generic model. Muscle force producing parameters (OFL, TSL, Pennation angle,) are scaled proportionally to the difference between a generic model and

measured parameters vs. the subject measurements. Scaling adjusts several variables in the musculoskeletal model, including: 1) length of bone and muscle, 2) center of mass position of each bone, 3) and mass of each segment. Optimization is the process of calibrating the scaled model based on the objective function of the model. The optimization process involves manipulation of the muscle parameters within the model for a best-fit solution when compared to the inverse dynamics model (Sartori et al, 2010).

Delp scaled musculotendon parameters derived from cadaver studies (Eijden et al 1985, Brand et al 1982, Eycleshymer et al 1970, Delp et al, 1990) to generate subject-specific musculotendon parameters with the assumption that the muscle forces produced would also be "subject-specific". The validity of these parameters was assessed by comparing joint moments predicted by the model to literature based isometric moments derived using the dynamometer (Delp et al, 1990). Lloyd and Buchanan expounded upon earlier studies and used in vivo measurements (EMG) along with the optimized musculotendon parameters (F^{max}, tendon slack length, etc.) as inputs to determine muscle forces and to calculate moments about the lower extremities (Lloyd and Buchanan 2005). This was an attempt to make the joint moment estimations about a joint more physiologically accurate. While these methods ultimately led to reasonable joint moment estimations, musculotendon parameters were never actually measured in individual subjects thus, the validity of these "subject-specific" musculotendon parameters is questionable.

Previous studies used scaled musculotendon parameters based on anatomical studies of cadavers or a combination of scaled and optimized musculotendon parameters to calculate muscle forces and joint moments based on linear relationships between height and weight (Sartori et al, 2010). The limitation of scaled modeling is that it may not adequately reflect differences in musculotendon parameters between subjects. Optimizing methods were improved with the addition of EMG-driven musculoskeletal modeling techniques (Lloyd and Buchanan 2005), which yield an accurate estimation of joint moments when compared to inverse dynamics joint moments. But it is feasible to believe that some of the parameters used in these scaled and then optimized models may still be inappropriate. A case of compensating errors may exist in these parameters on a muscle-to-muscle basis, as the net knee moments may still be accurate compared to inverse dynamics moments. If we could use subject specific musculotendon parameters, we may have confidence in the resultant muscle forces and joint moments produced by the subject-specific models.

Ultrasound imaging technologies have now evolved to the point that it is possible to incorporate measurements of muscle architecture to derive the musculotendon parameters necessary to model muscle forces through the Hill-muscle model. Ultrasound allows us to measure muscle volume, fascicle length, and pennation angle. Muscle size is believed to be the best determinate of strength (Powell et al, 1984). Muscle size is based on physiological cross sectional area (PCSA) which is linearly related to force production (Powell et al, 1984). The formula for calculating PCSA is muscle volume divided by optimal muscle fiber

(fascicle) length (Lieber et al, 1990). Tendon slack length the length of a tendon, where if it were stretched anymore it would start to produce force is dependent on the musculotendon length and fascicle length. Subject-specific parameters derived from the ultrasound can be used to determine tendon slack length. With these ultrasound derived musculotendon parameters, muscle forces calculated using the Hill-muscle model can be estimated in vivo. Therefore ultrasound imaging would allow for subject specific musculotendon parameters to be used to estimate muscle forces. Previous methods used a linear relationship for scaling musculotendon parameters to individual subjects, and therefore the muscle forces may not be indicative of the actual muscle forces produced. (Delp, 1990). A more reasonable approach to musculoskeletal modeling may be to incorporate parameters measured from individual subjects and not based on manipulating i.e. scaling or scaling and optimizing mean literature estimates.

Hypothesis:

Traditional methods of muscle force estimations in vivo have been based on musculotendon parameters that are either scaled or scaled and then optimized. Alternatively, incorporating parameters that are derived from ultrasound measurements on a subject-by-subject basis may lead to better representations of muscle forces and joint moments. This leads to an overall idea that an ultrasound-based musculoskeletal modeling approach would lead to more accurate subject-specific estimations of knee joint moments compared to traditional scaled musculoskeletal modeling techniques. For this thesis, our primary hypothesis states that ultrasound-based musculotendon parameters will more accurately predict

inverse dynamics knee moments compared to traditionally scaled musculotendon parameters. This hypothesis was applied to a single-leg squat. Secondarily, compared to optimized EMG signals generated from scaled models, we hypothesize that optimized EMG signals generated via the ultrasound based model will more accurately reflect experimental EMG. Although it is not feasible to directly measure muscle forces in vivo to which we could validate the model predicted muscle forces, we will also descriptively compare the ultrasound-based verses scaled muscle forces predicted during the single-leg squat in order to put into perspective the results of the primary and secondary hypotheses.

Scope of study:

The proposed method of estimating knee muscle forces and knee moments using ultrasound-based musculotendon parameters attempts to make a more physiologically relevant musculoskeletal model. More study in the area of applying subject specific data to musculoskeletal modeling is warranted to advance methodologies used in the field of biomechanical musculoskeletal modeling. New modeling techniques could be used in a broad range of applications from sports medicine and athlete rehabilitation to geriatric therapy. These methods could be applied to a variety of pathological patients to better understand how pathology relates with muscle function.

Limitations:

1) The ability to estimate musculotendon parameters via ultrasound imaging is limited to the ability to accurately measure muscle cross-sectional area, fascicle length, and pennation angles.
2) Validity of model estimates and muscle forces is determined indirectly by assessing joint moments to inverse dynamics and model predicted (optimized) activation to experimental activation.

Using the Ultrasound:

- The ability to track the full length of the muscle with the transducer and attain adequate fascicle images at both the proximal and distal ends of the muscle is a difficult task and requires much practice.
- The ability to track the same muscles from subject to subject can become difficult. Subject anatomy while the same can vary in shape and clarity, based on percent body fat, muscle definition.
- The amount of pressure the tester applies to the transducer while imaging can affect the quality of the image.

Direct Measurements Attained from Images:

- Following the same fascicle throughout the muscle can be difficult.
- Following the correct muscle borders when measuring the cross sectional area can be difficult. From subject to subject visibility of muscle border and clarity vary.

Assumptions:

- Assumption that fascicle length is representative of fiber length
- Model validity is assumed when joint moments are similar to inverse dynamics moments

Operational Definitions:

1) Cross sectional area (CSA) – A two-dimensional measure of the amount of space within the muscle.
2) Intraclass correlation coefficient (ICC) - Is a measure of reliability.
3) In vivo – In living tissue.
4) Musculotendon unit – Computed length from origin to insertion taking into account the path of the muscle and tendon
5) Optimal Fiber Length (OFL) – The length at which a muscle produces the maximal amount of force it is capable of assuming 100% activation.
6) Pennation Angle: The muscle's fascicle angle from the aponeurotic tendon.
7) Physiological cross sectional area (PCSA) – the average cross-sectional area of an individual muscle, measured by dividing its volume by it's optimal length
8) Raw fiber Length: Is the direct measure of the muscle fascicle length.
9) Scaled model: The process by which the modeling system fits a generic musculoskeletal model to a specific subject based on a linear relationship with subject height, weight, and center of mass.

10) Tendon Slack Length: The length of a tendon, where if it were stretched anymore it will start to produce force.
11) Electromyography (EMG): Technique for evaluating and recording the electrical activity produced by skeletal muscles

Chapter 2: Literature Review

The purpose of this study is to determine if subject specific musculotendon parameters estimated in vivo using new techniques in ultrasound will better estimate individual muscle moments compared to previous scaling methods. In this review of literature the following topics will be covered: 1) History and Advances in Muscle Force Estimation, 2) Limitations of Musculotendon Modeling, 3) Evidence for Using Subject Specific Musculotendon Parameters, 4) Using ultrasound Imaging to Generate Subject Specific Measurements, 5) Summary.

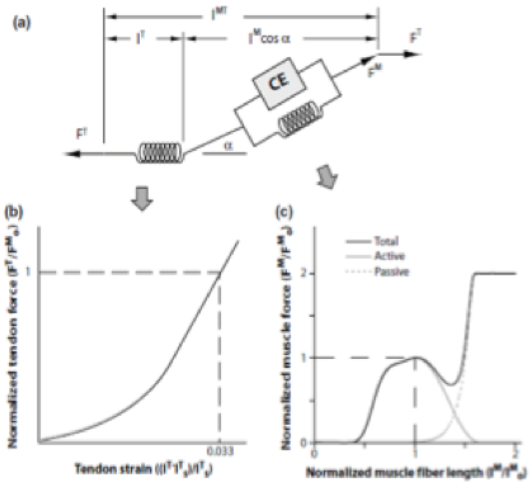
History and Advances in Muscle Force Estimation:

Muscles produce force, which results in the production of moments about a joint, causing movement of the body. These muscle forces are estimated based on both the muscles anatomy and several physiological parameters. Calculations of forces provided from each individual muscle around the joint have been traditionally estimated using a Hill-muscle model. A.V. Hill's classic paper outlines key principles of muscle physiology and diagramed how to develop quantitative models of the physiological processes. Hill outlined several musculotendon parameters which he termed as relations that determine the effect of load on speed of shortening, allow the form of the isometric contraction to be predicted, and are the basis of the visco-elasticity of skeletal muscle (Hill, 1939)

The Hill-muscle model consists of 3 elements, a contractile element, and two non-linear elastic elements, one in series and another in parallel (Figure 1a). The contractile element represents the muscles contractile parts (sarcomeres), and the 2

elastic elements represent the connective tissues' passive and intrinsic elastic energy storage. The parallel element represents the passive force of the connective tissues (epimysium, perimysium, endomysium) and is responsible for passive muscle behavior when it is stretched, even when the contractile element is not activated. The series element represents the tendon and the intrinsic elasticity of the myofilaments. This intrinsic elasticity accounts for what is often referred to as an inertial catch mechanism (Zajac, 1989, Hill, 1939); this mechanical property of the muscle enhances storage and release of elastic energy. In order to calculate muscle forces the Hill model uses several musculotendon parameters described in Zajac's 1989 article. The article described the estimation of muscle force dependent on several parameters, including maximum isometric force, optimal fiber length, tendon slack length, pennation angle, and activation time in a Hill type model. Zajac used a Hill type muscle model to simulate the muscle's contraction dynamics (Zajac, 1989, Hill, 1939).

Figure 1: Hill Muscle Model & Force Curves ***(Zajac, 1989)***



a.) Example of a Hill type muscle model containing a contractile element (CE), and two elastic elements (springs); one in series, one in parallel.

B & C.) Examples of Zajac's muscle force curves showing the passive non-linear force produced by the tendon (b) and the total force produced by the muscle tendon unit (c).

Since the development and application of the Hill-muscle model to estimate muscle forces additions have been made to the methodological approach; Delp et al (1990) developed a computer-based model of the human lower extremity to study how surgical changes in musculoskeletal geometry and musculotendon parameters affect muscle force and muscle moments about the joint. Delp et al (1990) defined the line of action of 43 musculotendon actuators by using three-dimensional bone surface representations based on anatomical relationships obtained by dissecting cadavers. The isometric-force length and force velocity relationships for each musculotendon unit were incorporated in the model (Zajac 1989). The three-dimensional models kinematics were defined by modeling the hip, knee, ankle, subtalar, and metatarsophalangeal joints, this helped to compute the force and the joint moment developed by each musculotendon unit. The data calculated with the model compared well with other literature-based data obtained from the experimentally measured isometric joint moments. By manipulating the model's parameters it theoretically would allow physicians to visualize how surgical techniques change muscle moment production. Thus, the model allows for quick assessment and aids in surgical procedure design. The knee model was simplified and only represented motion in the sagittal plane (Delp et al 1990). This computer-based model developed by Delp et al 1990 is commonly used in musculoskeletal research today to calculate joint moments and muscle forces. Delp's generic model is often scaled and applied to an individual to estimate subject-specific muscle forces. The scaling protocol includes the linear scaling of the subjects musculotendon

parameters based on the subjects' full model musculotendon length vs. the generic musculotendon length (Delp et al, 1990).

Lloyd and Buchanan expounded upon earlier studies and used measurements of EMG along with scaled musculotendon parameters to calculate moments about the lower extremities (Lloyd and Buchanan, 2005). They used EMG signals as inputs in a forward dynamics driven musculoskeletal model, to estimate and predict joint moments and muscle forces. The joint moments predicted were verified through comparison to inverse dynamics joint moments. Forward dynamics calculates motion using known internal variables (EMG, forces, and moments) to estimate external forces and moments during motion.

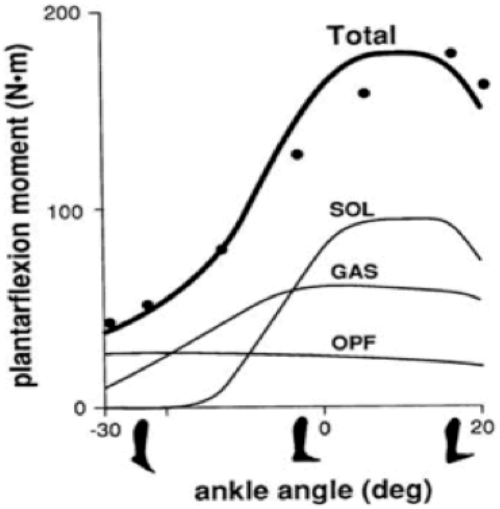
Lloyd and Buchanan's model was justified because it preserved the way healthy subjects activate their muscles and was used to estimate corresponding joint compressive forces, and/or ligament forces produced by this activation (Lloyd and Buchanan, 2005). The actual estimation of the joint moments required three steps involving complex non-linear relationships. Muscle activation dynamics facilitated the change of EMG signals to model the muscle activation. Muscle contraction dynamics characterized how muscle activations were transformed into muscle forces. It is important to note that Lloyd and Buchanan optimized musculotendon parameters after they were initially scaled; this allowed estimates of joint moments predicted by the model to closely resemble values determined experimentally (Lloyd and Buchanan 2005).

Limitations of Musculotendon Modeling:

Previous studies utilize musculotendon parameters that are based on cadaver measurements. These cadaveric-based parameters were then scaled (Delp et al, 1990) or scaled then optimized (Lloyd and Buchanan, 2005). The limitation of scaling or scaling then optimizing muscle parameters used in the Hill-model is that neither takes into account subject-specific musculotendon parameters. Thus, validity of the muscle forces generated by may be questionable.

Delp's computer based model of the lower extremity used model predicted muscle moments and compared generated moments to experimentally measured isometric moments (Figure 2). Delp attempted to use cadaver studies to produce the physiological parameters used in his study in hopes of making the data more physiologically relevant (Delp et al, 1990). Not all of the parameters could be determined from the study of cadavers. Tendon slack length (which was not reported in previous cadaver studies) had to be manipulated to attain a curve that best fit the line of the isometrically measured joint moments (Delp et al, 1990).

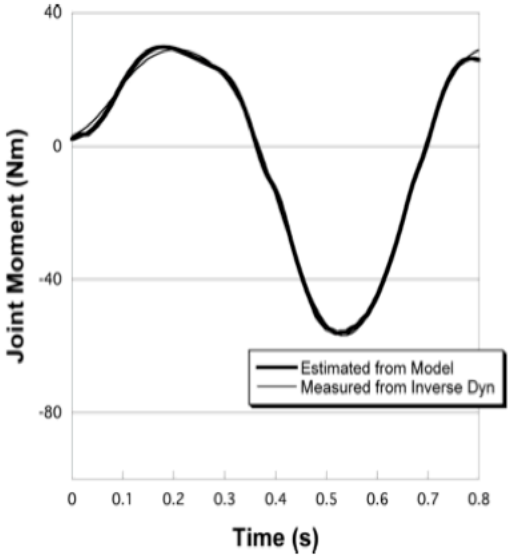
Figure 2: Estimated Plantar Flexion Moments ***(Delp et al, 1990)***



a) The dotted line represents the isometrically measured torques. The solid line is their data where tendon slack was manipulated to get the best-fit line.

Lloyd and Buchanan expanded upon Delp's earlier work and attempted to make the scaled model more physiologically accurate by including muscle activation measurements: attained using EMG, into the scaled model. Lloyd and Buchanan allowed their musculotendon parameters to be optimized to attain the best-fit line for their data (Figure 3). Optimizations were used to acquire what the tendon slack length, optimal fiber length, pennation angle at optimal fiber length, and maximal isometric force should be for a given muscle on each individual subject. The optimization of the parameters reduced the error of the model.

Figure 3: Ankle Joint Moment Comparisons ***(Lloyd & Buchanan, 2005)***



a) This graph shows the model estimation of ankle joint moments compared to the torques generated using inverse dynamics.

This begs the question: If the parameters are optimized and/or manually manipulated how is the data physiologically relevant or subject specific? If we could use subject specific parameters to model musculotendon forces the result may lead to more accurate estimations of muscle forces at a given muscle and joint. Thus, this

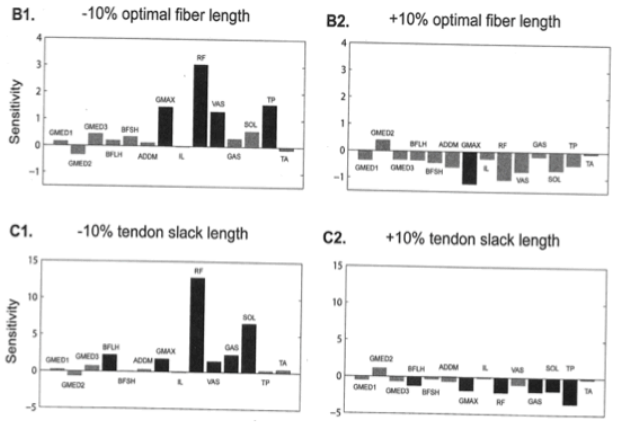
subject-specific incorporation into muscle models could advance current methods used to calculate muscle forces via musculoskeletal modeling techniques.

Evidence for Using Subject Specific Musculotendon Parameters:

The accuracy of the musculotendon parameters specific to an individual has a profound effect on the moments produced by the musculotendon model and the validity of the estimation. Ming Xiao's 2010 article examined the roles that the different parameters play in the estimation of moments by altering the parameters slightly in the model and examining the results – commonly known as a sensitivity analysis.

The sensitivity of individual muscle forces to slight changes in the parameters was observed during gait (Xiao et al, 2010). The number of muscles included in the model, F^{max}, optimal fiber length, and tendon slack length were the parameters that were altered in the model to observe the different moment outputs. F^{max}, tendon slack length, and optimal fiber length were all altered + or – 10% of their model estimated values (Figure 4). It was observed that the parameters that had the greatest effect on the muscle force estimation were tendon slack length and optimal fiber length about the ankle plantar flexors and the knee extensors. The changes in one muscle were usually compensated for by changes in a synergist muscle from the same group or by changes to an antagonist muscle group. The article pointed out that optimal fiber length and tendon slack length seem to be the most significant parameters affecting muscle forces during gait (Xiao et al, 2010).

Figure 4: Sensitivity Analysis of Optimal Fiber Length ***(Xiao & Higginson,2010)***



a) The parameters were altered by + or – 10% of the estimated OpenSim value. A muscle was considered to be sensitive to slight changes in parameters if the absolute value of the sensitivity was greater than one, this is represented by the dark bars.

In Xiao's sensitivity analysis the muscles that were most sensitive to the manipulations in the muscle parameters were the muscles that were most involved during gait (soleus, rectus femoris, gluteus maximus, and biceps femoris long head; all biarticular muscles). Xiao's article highlighted the muscle parameters that were most sensitive to changes in muscle force, optimal fiber length (OFL) and tendon slack length (TSL). Based on Xiao's article a separate sensitivity analysis was done on a different task - the single legged squat in our lab here at ECU. The manipulation of muscle parameters by as little as ±10% in a single muscle altered the estimated forces of multiple other agonists and antagonists, as well as the individual muscle manipulated. A study by Ward et al, in 2009 showed that from person to person muscle parameters can differ by as much as 20%. Given, that a difference of 10% in a musculotendon parameter can significantly alter the muscle force, the 20% coefficient of variation reported in humans highlights the importance of subject specific muscle parameters to more physiologically relevant force estimations.

Historically parameters had been scaled to predict moments (Delp, 1990) (Winby et al, 2008) and optimized to create more physiologically relevant moments

(Lloyd & Buchanan, 2005). Xiao pointed out the importance of subject specific parameters to moment estimations (Xiao et al, 2010). If more accurate parameters could be predicted for a specific subject then the moments estimated by the scaled model would be more physiologically relevant.

Using Ultrasound Imaging to Generate Subject Specific Measurements:

Ultrasound imaging technologies have now evolved to the point that it is possible to directly measure many of the musculotendon parameters necessary to model muscle forces through the Hill-muscle model.

Muscle volume and fiber length are easily obtained by using new ultrasound techniques (Ahtianen et al, 2009, De Oliveira et al, 2010). This is important because physiological cross-sectional area (PCSA) is calculated from these two variables and it is well accepted that maximum strength is related to muscle size (Powell et al, 1984). Ahtianen et al in 2009 used panoramic ultrasound techniques to detect training induced changes in cross sectional area of the vastus lateralis (VL). In this study they compared measures obtained via ultrasound to the same measurements obtained using magnetic resonance imaging (MRI). MRI is considered to be the gold standard in direct muscle cross-sectional area and volume measurements. Results concluded that panoramic ultrasound methods provided repeatable measures of muscle CSA and detected training induced changes in muscle CSA with a comparable degree of precision to MRI. The intraclass correlation coefficient (ICC) and the standard error of the mean (SEM) for comparing ultrasound verses MRI methods of obtaining the CSA of the VL were as follows: The ICC was .905 and the SEM was .87

cm squared respectively. It should also be noted that the MRI produced systematically larger absolute CSA values (Ahtianen et al, 2009).

The Ahtianen et al, article provides insight into the techniques used to measure musculotendon parameters and is expounded upon in De Oliveira et al's, journal article published in 2010. In the paper F^{max} was calculated by estimating PCSA obtained using ultrasound techniques. F^{max} is the maximum voluntary contraction force and is a subject specific parameter that if varied slightly can lead to different model forces (Xiao et al, 2010). In the experiment ultrasound was used to measure the Soleus, gastrocnemius medialis, and gastrocnemius lateralis of 8 physically active subjects. Using a dynamometer, measurements were taken at 20% and 60% of the subjects MVC (De Oliveira et al, 2010). Muscle fiber length, pennation angle and muscle thickness were measured and muscle volume and PCSA were obtained. The measure of muscle thickness allowed them to calculate muscle volume through a series previously validated regression equations from Miyatani et al (2004).

OpenSim was used to calculate the muscle forces and muscle moments. A Hill type EMG driven model of the ankle joint was used to compare the estimated moments from the ultrasound obtained parameter to the OpenSim model. OpenSim uses a model containing several different scaled musculotendon parameters; including optimum fiber length, optimal pennation angle, and tendon slack. The optimal pennation angle was obtained from Manal et al. (2006). None of the subject specific physiological parameters other then F^{max} were used in the muscle model. The root mean square error comparing the models to the dynamometer obtained

muscle moment data were obtained (De Oliveira et al, 2010). A statistically significant reduction in the root mean square error was observed when the F^{max} derived via ultrasound was used, compared to the F^{max} obtained from the scaled OpenSim values(De Oliveira et al, 2010).

Ultrasound allows us to measure muscle volume, fascicle length, and pennation angle and it is theoretically feasible to derive Fmax, optimal fiber length, and pennation angle at optimal fiber length from these ultrasound measurements. Because tendon slack length is unique to each individual muscle and is dependent on musculotendon length and fascicle length, the subject's specific parameters derived from the ultrasound can be used to determine tendon slack length. With these musculotendon parameters muscle forces using the Hill-muscle model can be estimated in vivo. Specific musculoskeletal parameters derived from ultrasound should estimate moments about the knee more accurately than traditional musculotendon scaling methods.

Summary:

Muscle forces have been traditionally estimated using a three element Hill driven muscle model. The Hill model uses specific musculotendon parameters to estimate these forces, optimal fiber length, tendon slack length etc. The musculotendon parameters have traditionally been calculated using scaling (Delp et al 1990, Winby et al 2008) or scaling and optimization methods (Lloyd and Buchanan 2005). The manipulation of muscle parameters by ±10% in a single muscle can significantly alter estimated forces of multiple other agonists,

antagonists, as well as the individual muscle manipulated by as much as 20% (Ward et al, 2009). These parameters can be calculated on an individual basis using ultrasound-based measurements of muscle architecture. This evidence collectively highlights that subject specific muscle parameters may lead to more physiologically relevant force estimations.

The central idea behind our research is that an ultrasound-based musculoskeletal modeling approach would lead to better subject-specific estimations of knee moments compared to traditional musculoskeletal modeling techniques. For this thesis, our primary hypothesis states that ultrasound-based musculotendon parameters would more accurately predict inverse dynamics knee moments compared to traditionally scaled musculotendon parameters. This hypothesis was applied to a single-leg squat. Secondarily, compared to optimized EMG signals generated from the scaled models, we hypothesized that optimized EMG signals generated via the ultrasound based model would better reflect experimental EMG. Although it is not feasible to directly measure muscle forces in vivo to which we could validate the model predicted muscle forces, we also descriptively compared the ultrasound-based verses scaled muscle forces predicted during the single-leg squat in order to put into perspective the results of the primary and secondary hypotheses.

Chapter 3: Methods

Subjects:

Seven healthy young college aged volunteers were chosen as subjects (N=7). They were recruited through classroom volunteering. All subjects read and signed a document of informed consent. Additional criteria includes, no previous history of lower extremity injury, and comfortable performing squatting tasks. There were three male subjects (male N=3), age 24 ± 1.2, height 1.8 m ± .08 m, mass 77.6 kg ± 3 kg. There were four female subjects (female N= 4), age 23 ± 2, height 1.7 m ± .07 m, mass 59 kg ± 8 kg. Group mean age was 23 ± 1.7, height 1.7m ± .09m, and mean mass was 67 kg ± 11.5 kg.

Study Design:

The study procedures were separated into 2 days of data collection. On the first day we collected ultrasound data. On the second we collected motion capture measurements, along with maximal EMG and moment output using a dynamometer. Dynamometer data was used to normalize EMG to a % of maximal effort.

Ultrasound Protocol:

Images attained using ultrasound were used to measure the cross sectional area, fascicle length, and the pennation angle of the quadriceps, hamstrings, and the gastrocnemius muscle groups. The transducer was moved horizontally along the muscle to capture a cross-sectional area (CSA) panoramic image of individual muscles (Figure 5). The transducer was moved parallel to the muscle

fascicle/longitudinally along the thigh to capture the fascicle length and the pennation angle of each muscle (Figure 6B & 6D).

All measurements were taken from the subject's right leg. While the subject was standing; starting with the lower leg we captured an ultrasound image of the lateral and medial heads of the gastrocnemius by first palpating the lower leg and then placing the transducer vertically along medial and lateral gastrocnemius muscles of the lower leg starting distally and moving proximally.

Next the subject was asked to relax while lying face down on a padded table while images for the Hamstrings were taken. First to estimate volume we measured the length of the leg from the gluteal fold to the center of the knee. The hamstrings were palpated to find the long head of the biceps femoris; starting at the distal musculotendinous junction as visually verified through the ultrasound images; ten equidistant measurements were then placed along the biceps femoris to the proximal musculotendinous junction. Starting distally and moving proximally; cross sectional area images (Figure 5) of the long head of the biceps femoris were collected by running the transducer horizontally from the lateral side of the hamstrings to the medial side along the previously placed tick marks. After the cross sectional area images of the biceps femoris were collected the fascicle length of the short head of the biceps femoris, semimembranosus, and semitendinosus were collected. After palpating the muscles the transducer was ran longitudinally along the individual muscles starting distally and moving proximally to collect the fascicle length of each individual muscle.

Figure 5: CSA Image of Biceps Femoris Long Head

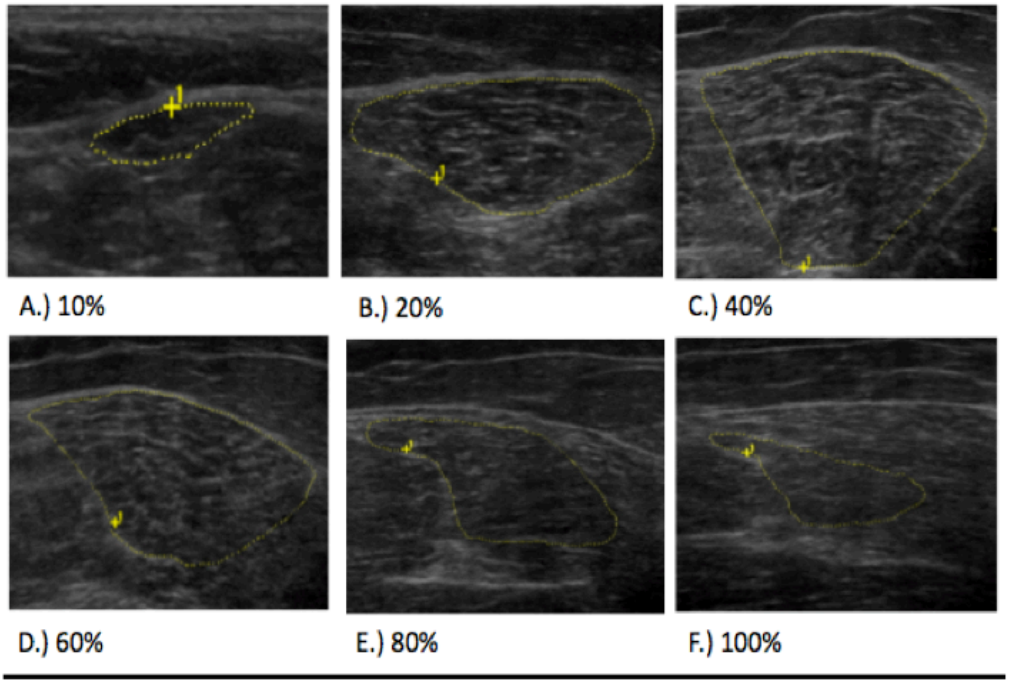
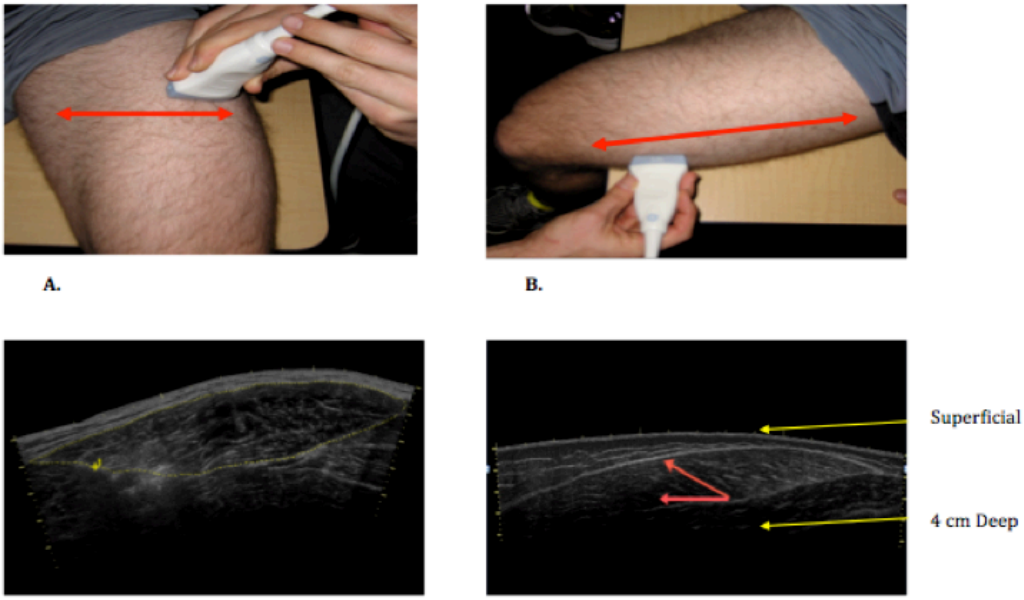


Figure 5 – CSA image of Biceps femoris long head at 10, 20, 40, 60, 80, 100% of the muscle belly.

Finally the subject was asked to lay supine on the padded table. Measurements of the leg were taken from the greater trochanter at the hip to the lateral epicondyle at the knee. Next the leg was palpated to locate the rectus femoris; at the distal end of the rectus femoris the ultrasound transducer was used to find the point where the tendon ends and the muscle begins. From this point 60% of the muscle length was calculated and marked off at the rectus femoris. This point is considered to be the maximal cross-sectional area of the rectus femoris (Morse et al, 2007). Cross-sectional area measurements for the rectus femoris were taken by moving the transducer horizontally from the lateral side of the leg to the medial side at the point marked for 60% of the muscle length (Figures 6A & 6C). After rectus femoris cross sectional area was collected panoramic images of the vastus lateralis, rectus femoris, vastus intermedius, and vastus medialis were collected by palpating

the muscles and then running the transducer vertically along the individual muscles starting distally and moving proximally (Figure 6b & 6d).

Figure 6: Using Ultrasound to Collect Transverse and Sagittal Plane Images of Muscles



C. Rectus Femoris Transverse Plane

The yellow outline represents the cross sectional area of the Rectus Femoris in the middle of the muscle belly. Figure 1 shows how the image was taken with the Ultra Sound transducer running horizontally across the thigh.

D. Vastus Lateralis Sagittal plane

The Red arrow represents the fascicle length and pennation angle relative to the aponeurotic tendon. Figure 2 shows how the image was taken with the transducer running vertically up the thigh.

Motion Capture and EMG Protocol:

EMG electrodes were applied to the vastus lateralis, vastus medialis, rectus femoris, long head of the biceps femoris, semimembranosus/semitendinosus, tibialis anterior, medial and lateral gastrocnemius to collect muscle activation data during all of the trials. Electrode placement sites were prepped before application of

electrodes. The areas were shaved with a disposable razor, then were cleaned using an abrasive pad to rid the area of any dead skin and wiped clean using alcohol wipes.

After application of the EMG electrodes, reflective makers used for the static calibration test were applied to the subjects. A vest was worn with markers already placed on it. Markers were placed on the anterior superior iliac spine (ASIS), the greater trochanter, on the vertebral column at L4-L5 vertebrae, marker plates were placed on the thigh and shank, each with three reflective markers respectively. Markers were then placed medial and laterally on the knee, on the foot at the toe, heel, metatarsal heads 1 & 5 and the lateral and medial ankle. Finally markers were placed at the elbow, wrist and on the forehead. A static calibration trial was taken with both feet on the force plate, shoulder width apart, and arms stretched out to the side in line with the shoulders. Calibration markers were removed. The subject was then asked to perform 5 single legged squats at a self-selected pace with one foot on the force plate.

HUMAC Protocol:

The chair was pre-positioned based on the subjects estimated height, but once the subject sat in it the dynamometer's chair it was fitted specifically for the subject. Moment and EMG measurements were recorded for the quadriceps and hamstrings at 15, 30, 45, 60, 75, and 90 degrees of knee flexion. The starting position was randomly selected from the 6 different degree criterion; following measurements were taken in that set order. Gastrocnemius and tibialis anterior

EMG was recorded during this process with a maximal plantar and dorsiflexion recording taken at the end.

EMG Processing:

All surface EMG signals were high pass filtered at 10Hz, full-wave rectified, and then low-pass filtered at 6Hz (Besier et al, 2009). The high and low pass filters are "recursive" or bi-directional meaning that time delays in the EMG signal induced by smoothing are corrected for. The filtered EMG signals were normalized to the peak EMG signal for each muscle identified at any time during HUMAC testing protocol.

Calculations of Ultrasound Based Musculotendon Parameters:

To calculate cross sectional area, the outer border of the muscle was traced using the measurement functions available within the ultrasound software (Figure 5). Using the same function in the ultrasound software fascicle length and pennation angle relative to the aponeurotic tendon were also labeled and measured (Figure 6d).

Optimal Fiber Length:

Raw fiber length is the direct measure of the muscle fiber length. We derived fiber length from fascicle length. We assume that the fascicle length is representative of fiber length. Optimal fiber length is the length at which the muscle will produce its maximal amount of force. Optimal fiber length is calculated by

multiplying the raw fiber length measured from ultrasound by the ratio of optimal sarcomere length ($S_{opt} = 2.7$um) to resting sarcomere length. Ward et al (2008) provided the data for resting sarcomere lengths for the quadriceps, hamstrings, and gastrocnemius muscles. Of special note is that this derivation of optimal fiber length assumes that the measured fascicle lengths are linearly correlated to the sarcomere lengths with the limb in the same position.

$$Ofl = L_m \left(\frac{S_{opt}}{S_m}\right)$$

Pennation Angle at Optimal Fiber Length:

Since pennation angle changes with fiber length, assuming the muscle belly has a constant thickness, we used the following equation (Lloyd and Besier 2003) to solve for pennation angle at optimal fiber length:

$$\Theta_{raw} = \sin^{-1}\left(\frac{FL_{opt}\, sin\Theta_{opt}}{FL_{raw}}\right)$$

Maximal Isometric Force (F-max):

The muscle volume is accounted for by examining the area under the curve of CSA and femur length and will be measured in units of cm^3. Since the point of origin for the quadriceps is so far up the hip, it becomes difficult to take complete ultrasound measurements without invading the subject's personal space. For this reason muscle volume for the quadriceps will be estimated based on a series of regression equations attained from (Morse et al, 2007). Volumes of vastus lateralis, vastus intermedius, and vastus medialis were estimated based on relative percentage of total quadriceps volumes and the rectus femoris volume (Appendix A). For the hamstrings, we estimated the biceps femoris long head volume by

integrating the area under the CSA vs. muscle length curve. Biceps femoris short head, semimembranosus, and semitendinosus volumes were based on relative volume estimates pooled from the literature (Appendix B).

After the muscle volumes were calculated physiological cross sectional area (PCSA) was determined. PCSA is assumed to be linearly related to force (Powell et al, 1984) and is measured in units of cm^2. PCSA is calculated by dividing the muscle volume by the optimal muscle fiber length. Next F-max which is the maximal isometric force exerted by the muscle was calculated by multiplying PCSA by specific tension, (PCSA X 35 N/cm^2) (Erskine et al, 2009). The number that we used for the specific tension of each muscle (35 N/cm^2) was a normalized number attained from Erskine et al.

Tendon Slack Length:

Tendon slack length is the length of a tendon (passive elements of the Hill-model), where if it were stretched anymore it will start to store energy. The Hill-muscle model shows that this tendon length occurs at the optimal fiber length: if a muscle is stretched beyond its optimal fiber length, the passive tendon forces will start to contribute to the overall musculotendon force (active and passive contributions combined, Figure 1c). Tendon slack length cannot be measured directly using ultrasound, but it can be derived based on knowledge of optimal fiber length and the available lengthening of the entire musculotendon unit (Garner and Pandy 2003, Winby 2008). Because the musculoskeletal model assumes the summed total length of tendon in series with muscle (corrected for pennation angle)

is to equal the entire musculotendon length, tendon slack length can be computed if the optimal fiber length (corrected for pennation angle at optimal fiber length), and musculotendon length are known. To calculate TSL, we assumed that the muscle is in a passive state i.e. no active contraction. First, the ultrasound based optimal fiber length was adjusted by a multiplication factor of 1.27 based on the premise that optimal fiber length increases at low activation levels (Huijing 1996, Lloyd and Besier 2003) and reaching a low of 1.27 when not active. Pennation angle was also adjusted with this same correction factor. With these two adjusted parameters incorporated into the subject-specific model, the total musculotendon and muscle fiber lengths during a neutral stance were estimated using SIMM and the subject-specific TSL was calculated using the following equation derived from Garner and Pandy (2003):

$$Tendon\ Slack\ Length = MT_{length} - \left(FL_{opt\ adj} * \sqrt{FL_{norm}^2 - sin\Theta_{opt\ adj}^2}\right)$$

Where $FL_{opt\ adj}$ and $\Theta_{opt\ adj}$ are the optimal fiber length and optimal pennation angle adjusted for no activation, MT_{length} is the musculotendon length estimated (via SIMM) with the hip and knee at zero degrees of flexion i.e. neutral, and FL_{norm} is the fiber length (estimated via SIMM) normalized to optimal fiber length and with the hip and knee at zero degrees of flexion i.e. neutral. Since fascicle lengths were measured via ultrasound in the same neutral stance and also in the passive condition, if the tendon slack length calculation was correct, then the SIMM estimated fiber length would equal the ultrasound measured fascicle length. Others have similarly calculated TSL such that the model estimated fiber length is equal to

cadaver measured fiber lengths (Arnold et al 2010). TSL is thought to be a static parameter i.e. unchanging regardless of muscle length or activation level (Manal and Buchanan 2004, Garner and Pandy 2003, Winby 2008). Thus, once tendon slack length was computed in this manner, all musculotendon parameters: Fmax, TSL, optimal fiber length and pennation angle at optimal fiber length were incorporated into the subject-specific musculoskeletal model.

Calculations of Muscle Forces and Muscle Moments:

The static calibration trial was used to scale a generic musculoskeletal model (Delp et al, 1990) to each subject. In this scaled model, F^{max}, tendon slack length, optimal fiber length, and pennation angle were all scaled according to the generic musculotendon length. Once we scaled the model, muscle forces were calculated via the Hill muscle model at each of the recorded knee angles using muscle activation amplitudes. Vastus intermedius activation was modeled as the average activation from the vastus medialis and lateralis. Biceps femoris (short head) muscle activation and semimembranosus activation were equal to the recorded biceps femoris (long head) and semitendinosus muscles respectively. These modeling assumptions have been used previously (Lloyd and Buchanan 2003). Muscle moments were then calculated as the product of the calculated muscle force and the muscle moment arm. Muscle force and static optimization programs use an optimization function to make sure that the muscle torque values are equal between the inverse dynamics model and the muscle joint moments themselves. Optimization uses EMG (muscle activation) to minimize the sum of muscle

activations squared to the least amount required for muscle torque output that will match activation patterns (minimizing cost function). All quadriceps, hamstring, and gastrocnemius muscle moments were summed to yield a net knee torque. This process of calculating muscle forces and moments was repeated for the squatting trials using the individual musculotendon parameters derived from ultrasound. Essentially, for each subject there will be two models: a model with scaled musculotendon parameters and a model with ultrasound based musculotendon parameters. Muscle forces and moments estimated from the squatting trials were computed for each model and compared to inverse-dynamics based knee moments.

Data Analyses:

Two types of data analyses were used. Correlational and paired sample t-test analyses compared scaled and ultrasound model predicted knee joint moments to inverse dynamics based knee moments. Correlational and paired sample t-tests were also used to compare scaled and ultrasound model predicted muscle activations to experimental muscle activations, to assess whether the differences in knee moments and muscle activations produced any meaningful differences in quadriceps and hamstring muscle forces, paired sample t-tests compared scaled vs. ultrasound models.

Chapter 4: Results

Descriptive Results:

All of the raw parameters used in the individual models are presented in appendix C. Qualitatively, there were inconsistent differences between models for all muscles across subjects. To better understand absolute differences between the ultrasound and scaled parameters they are summarized in table 1.

Table 1: Ultrasound and Scaled Model Differences

	Fmax			OFL			TSL			Penn		
	Mean		**SD**	**Mean**		**SD**	**Mean**		**SD**	**Mean**		**SD**
VL	17.84	±	12.57	13.89	±	6.25	10.14	±	12.01	24.56	±	8.71
VMO	27.13	±	20.02	5.61	±	7.26	16.67	±	8.66	26.49	±	16.98
VI	112.70	±	32.96	15.82	±	8.56	14.33	±	12.98	133.90	±	48.58
RF	53.26	±	35.93	12.75	±	8.93	8.49	±	9.13	10.63	±	8.87
Quadriceps Avg.	**52.73**	**±**	**25.37**	**12.02**	**±**	**7.75**	**12.41**	**±**	**10.70**	**48.90**	**±**	**20.78**
SM	40.81	±	14.25	14.65	±	13.02	9.81	±	7.01	17.91	±	10.26
ST	58.73	±	40.87	27.95	±	10.39	28.79	±	23.72	19.06	±	4.73
BFLH	30.08	±	18.79	15.80	±	9.92	9.39	±	13.71	10.20	±	6.78
BFSH	60.35	±	56.44	28.07	±	14.96	49.67	±	30.55	57.43	±	41.85
Hamstrings Avg.	**47.49**	**±**	**32.58**	**21.62**	**±**	**12.07**	**24.42**	**±**	**18.75**	**26.15**	**±**	**15.91**
MGAS	45.03	±	18.36	17.14	±	22.48	9.77	±	12.48	30.25	±	29.30
LGAS	22.99	±	21.83	26.86	±	9.39	11.50	±	15.05	88.75	±	55.13
Gastrocnemius Avg.	**34.01**	**±**	**20.09**	**21.87**	**±**	**15.93**	**15.23**	**±**	**13.77**	**48.38**	**±**	**42.22**

All data represent the absolute difference (%)

Between the ultrasound and Scaled models in the quadriceps Fmax varied 53%, OFL 12%, TSL 12%, and pennation angle 49%. In the hamstrings Fmax varied 47%, OFL 22%, TSL 24%, and pennation angle 26%. In the gastrocnemius Fmax varied 34%, OFL 22%, TSL 15%, and pennation angle 48%.

Results for Primary Hypothesis: *"Ultrasound-based musculotendon parameters will more accurately predict inverse dynamics knee moments compared to traditionally scaled musculotendon parameters."*

Figure 7: Ultrasound and Scaled Model Joint Moment Curves vs. Inverse dynamics Model

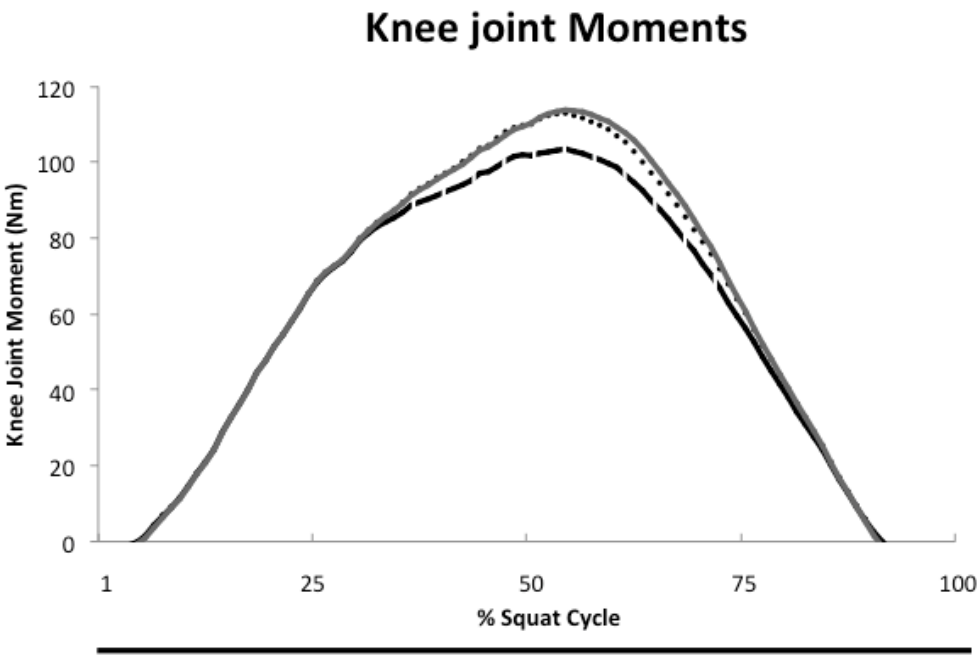


Figure 7 – This figure shows the ensemble average curve of the knee joint moment difference Ultrasound, Scaled, and Inverse Dynamic models. Solid line represents Inverse dynamics model, dashed line represents scaled model, & dotted line represents ultrasound model [0% refers to beginning (0° knee flexion), 50% = peak knee flexion, and 100% = ending (0° knee flexion)].

Joint Moments:

A qualitative analysis of the ensemble average joint moment curves of both ultrasound and scaled models vs. the invers dynamics model reveals that the peak joint moments are where the only differences are scene between the two models moment estimations (Figure 7). Using a correlation analysis it was evident that both the scaled and the ultrasound based models estimated knee joint moments strongly

correlated with the inverse dynamics knee joint torque (US r= .997 Nm: p< 0.01, SC r= .993 Nm: p< 0.01 for both models). The average root mean square error (RMSE) across all trails for the ultrasound-based model was 2.71 Nm ± 0.73 Nm (Figure 8: p < .05). This represents an absolute percent difference of 1.61 % of the peak joint moment (Figure 7).The scaled model had a RMSE of 6.08 Nm ± 1.03 Nm. This represents an absolute percent difference of 3.19 % of the peak joint moment (Figure 7).

Figure 8: Ultrasound vs. Scaled Joint Moment Root Mean Square Error

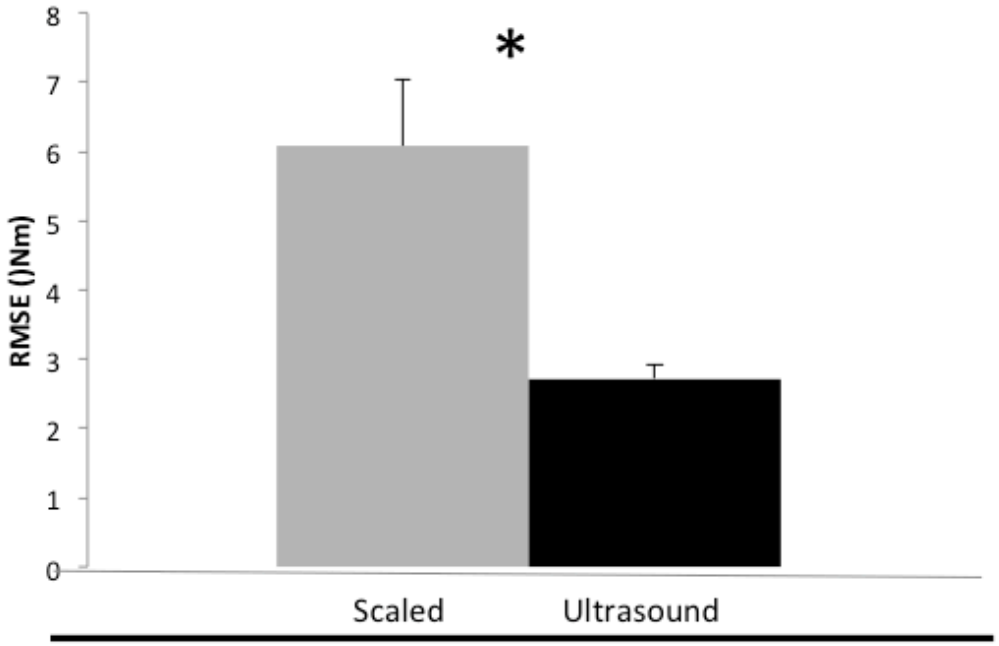


Figure 8 – Grey represents scaled mode's RMSE for joint moments, black represents ultrasound model's RMSE for joint moments (* = p<.05).

Results for Secondary Hypothesis: *"That EMG signals generated via the ultrasound based model will better reflect experimental EMG."*

Figure 9: Average Quadriceps EMG

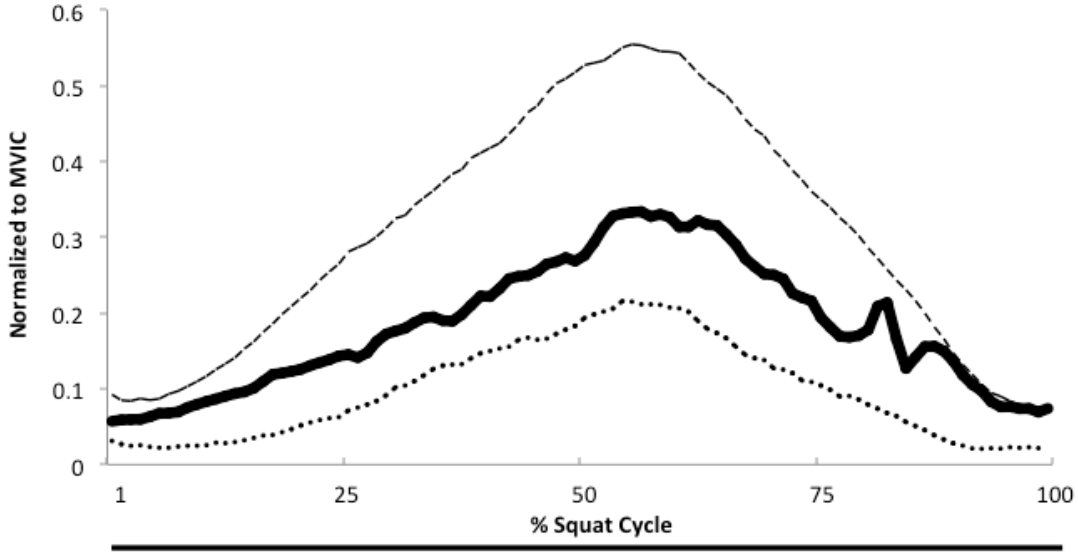


Figure 9 – Figure depicts ensemble average curves showing the difference between average quadriceps EMG in both the Scaled and Ultrasound models vs. Raw EMG. Solid line represents raw EMG, dashed line represents scaled EMG, & dotted line represents ultrasound EMG (Vertical axis: Normalized to MVIC) [0% refers to beginning (0° knee flexion), 50% = peak knee flexion, and 100% = ending (0° knee flexion)].

Quadriceps EMG:

Figure 9 depicts an ensemble average curve showing the difference between average quadriceps EMG in both the ultrasound and scaled models vs. Raw EMG. The ultrasound model derived EMG were moderately correlated to the experimentally obtained muscle activations (US r= 0.50 mV ± .45, p< 0.01) (Table 2). The scaled model muscle activations were also moderately correlated with experimental muscle activations (SC r= 0.55 mV ± .27, p<0.01) (Table 2). The RMSE for the quadriceps EMG signals were smaller in the ultrasound model (RMSE: US=

0.16 mV ± .07) compared to the scaled model (RMSE: SC= 0.23 mV ± .09, $p < .05$) (Figure 10).

Table 2: Quadriceps EMG Correlations

	RF	VM	VI	VL	Quad
Scaled	0.44**	0.40**	0.54**	0.54**	0.55**
Us	0.18**	0.42**	0.53**	0.51**	0.50**

Table 2 - Scaled vs. Ultrasound individual muscle and total quadriceps EMG correlations (** = p<.01).

Figure 10: Quadriceps Raw EMG vs. Ultrasound & Scaled EMG Root Mean Square Error

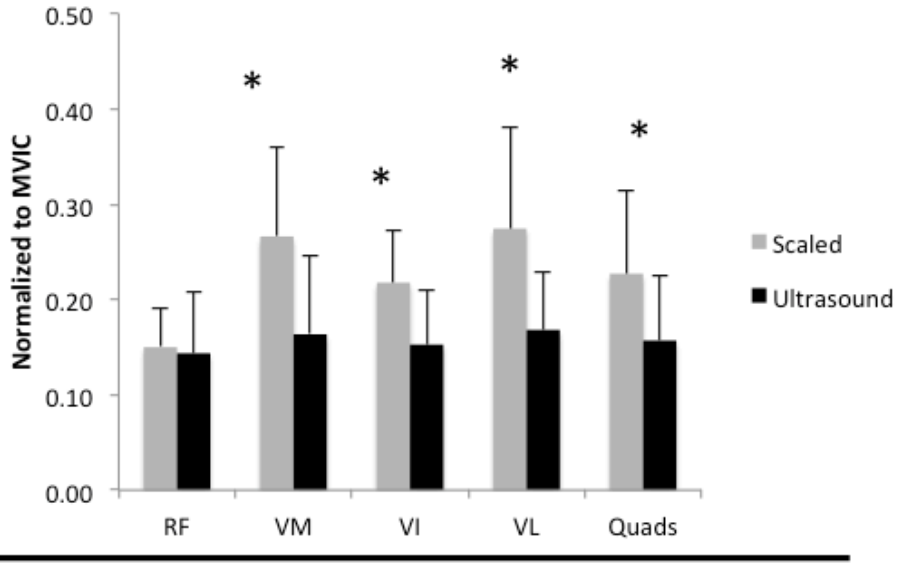


Figure 10 – Grey represents scaled mode's EMG, black represents ultrasound model's EMG RMSE (* = p<.05).

Hamstring EMG:

Figure 11: Average Hamstring EMG

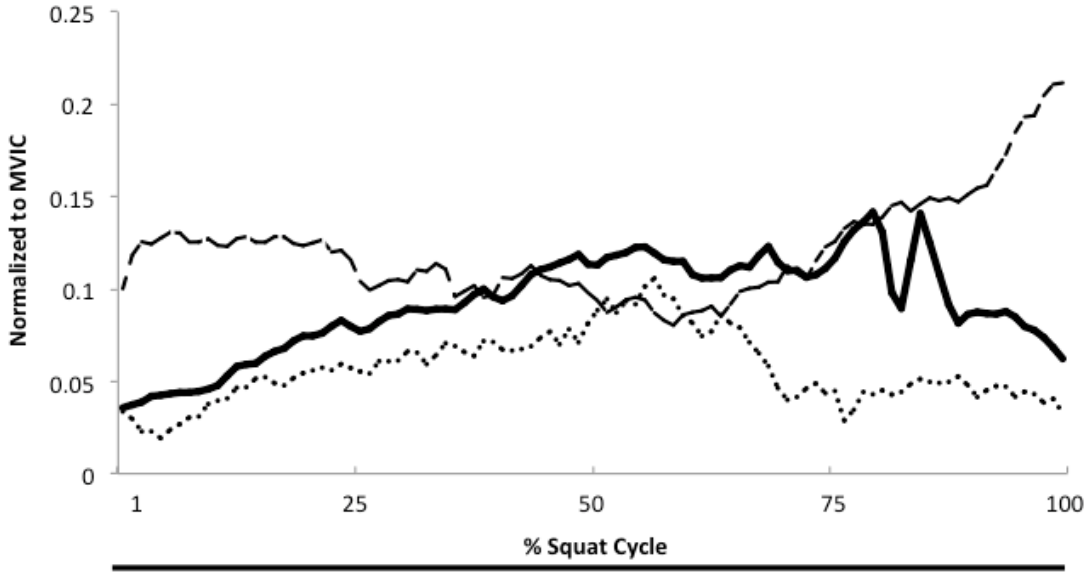


Figure 11 – Figure depicts ensemble average curves showing the difference between average hamstring EMG in both the Scaled and Ultrasound models vs. Raw EMG. Solid line represents raw EMG, dashed line represents the scaled EMG, & dotted line represents ultrasound EMG (Vertical axis: Normalized to MVIC) [0% refers to beginning (0° knee flexion), 50% = peak knee flexion, and 100% = ending (0° knee flexion)].

Figure 11 depicts an ensemble average curve showing the difference between average hamstrings EMG in both the ultrasound and scaled models vs. Raw EMG. Hamstring EMG correlation in the ultrasound was moderate while in the scaled model correlations were weak (US r= 0.44 mV ± .25, p< 0.01; SC r= 0.23 mV ± .30, p< .01, Table 3). There were no significant differences between the RMSE for the scaled vs ultrasound models (RMSE: US= 0.13 mV ± .07 vs. SC= 0.11 mV ± .04, p>.05, Figure 12).

Table 3: Hamstring EMG Correlation

	SM	ST	BFLH	BFSH	Ham
Scaled	-0.05**	0.26**	0.23**	0.18**	0.23**
Us	0.26**	0.25**	0.39**	0.29**	0.44**

Table 3 - Scaled vs. Ultrasound individual muscle and total Hamstring EMG correlations (** = p<.01).

Figure 12: Hamstrings Raw EMG vs. Ultrasound & Scaled EMG Root Mean Square Error

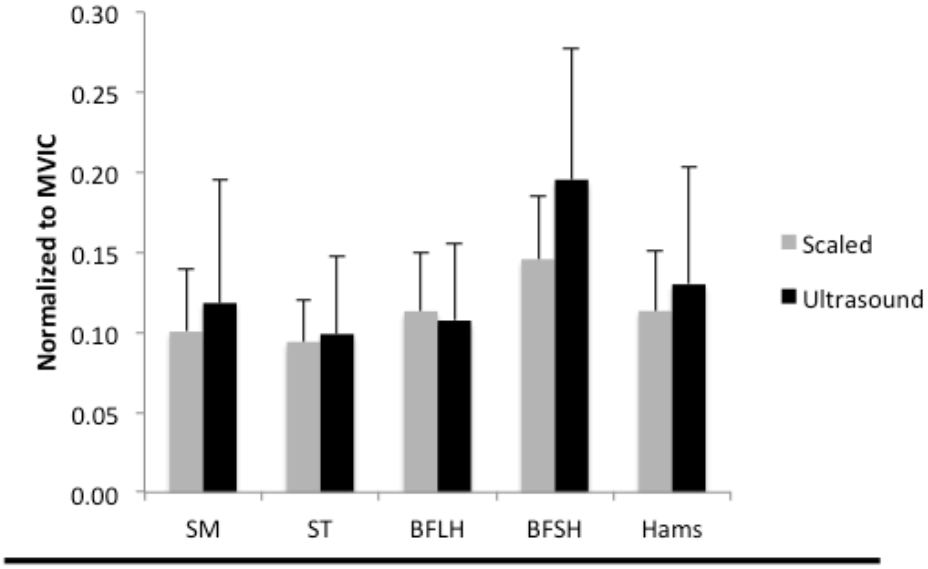


Figure 12 - Grey represents scaled mode's EMG, black represents ultrasound model's EMG RMSE.

Supplemental Analysis A: **"***On/Off Analysis."*

Figure 13: Individual & Total Muscle On/Off Analysis

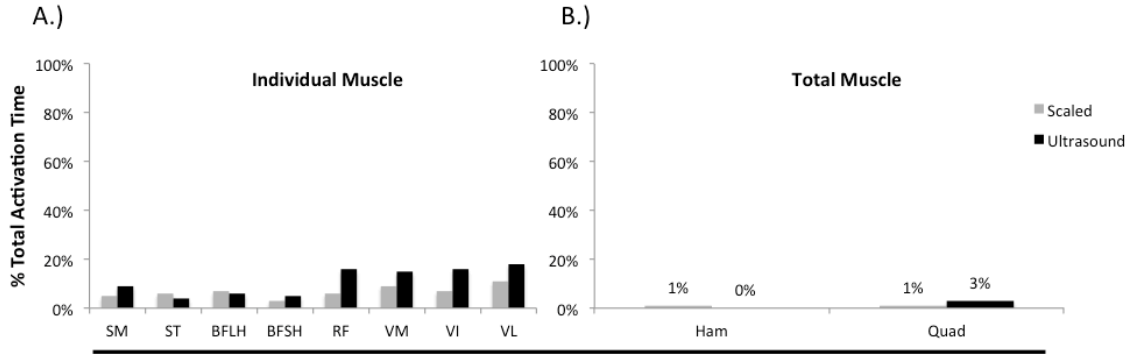


Figure 13 – Figures represents the amount of time that the models shut individual muscles off while raw EMG shows that they are active. A.) Depicts individual muscle on/off times. Grey represents the scaled model, black represents the ultrasound model. B.) Depicts Total muscle on/off times. Grey represents the scaled model, black represents the ultrasound model.

Qualitatively looking at individual trials of muscle EMG estimation, optimized activation signals would sporadically drop to zero ("off") when the experimental activations were clearly "on". This was seen in trials from both scaled and ultrasound models. Figure 13A qualitatively examines individual muscles on off times. The scaled model shut off the hamstrings 1% of the time throughout all trials and shut off the quadriceps 1% of the time (Figure 13B). The ultrasound model shut off the hamstrings 0% of the time throughout all trials and shut off the quadriceps 3% of the time (Figure 13B). In muscle modeling activation signals dropping to zero would essentially shut the muscle off for a period of time.

Supplemental Analysis B: *Are the muscle forces different with different musculotendon parameters?*

Figure 14: Ultrasound vs. Scaled Average and Peak Muscle Forces

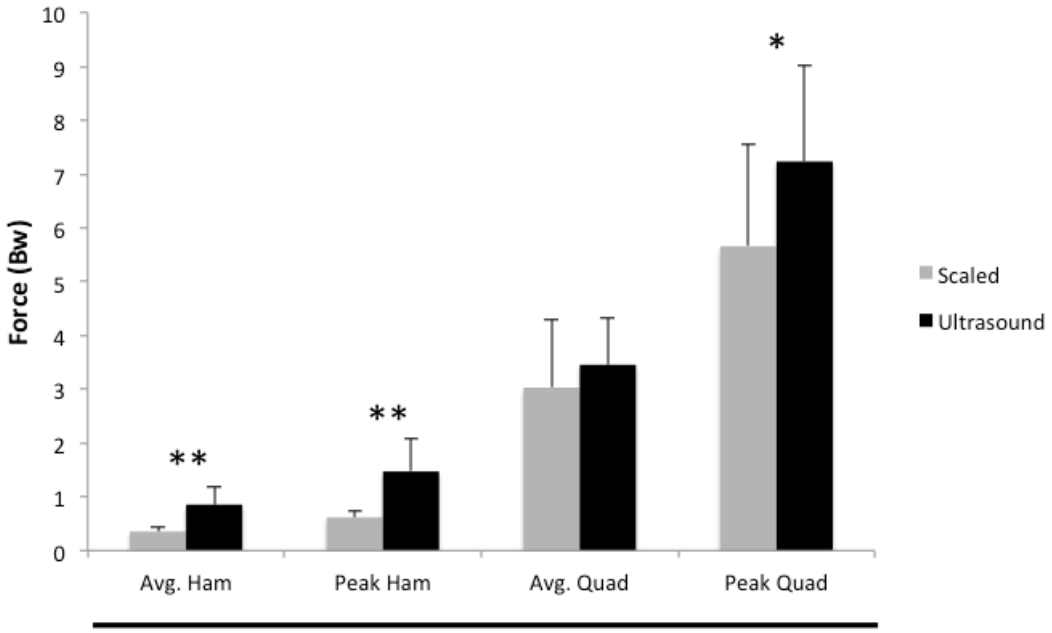


Figure 14 – Figure depicts difference between estimated average and peak muscle forces in both scaled and ultrasound models. Grey represents the scaled, black represents Ultrasound model (* = p<.05, ** = p<.01), (Bw = Force normalized to body weight).

It is not feasible to directly measure muscle forces in vivo to validate model predicted muscle forces. We descriptively compared muscle forces predicted during the single-leg squat in ultrasound-based vs. scaled model in order to put into perspective the results of the primary and secondary hypotheses. Figure 14 represents the difference between average and peak muscle forces in both the scaled and ultrasound models. The percent difference in average Quadriceps forces between the Ultrasound and scaled models is 14% (p>.05). The percent difference in average hamstring forces between the ultrasound and scaled models was 140.3% (p< .01). The percent difference in peak Quadriceps forces between the Ultrasound

and scaled model was 27.9% ($p < .05$). The percent difference in peak hamstring forces between the ultrasound and scaled models was 138% ($p < .01$).

The ultrasound models force estimation was greater for both the hamstrings and quadriceps. The absolute percent difference of quadriceps muscle forces between both the scaled and ultrasound model is 5% across all points. For the hamstrings the absolute percent difference of muscle forces between both the scaled and ultrasound models is 19% across all points (Figure 15). When examining the average difference between the peak muscle forces there was 76% difference between the scaled and ultrasound quadriceps muscle force curves and 43% difference between the peak muscle forces in the hamstrings scaled and ultrasound muscle force curves.

Figure 15: Quadriceps and Hamstring Muscle Forces – Difference Between Ultrasound and Scaled Models

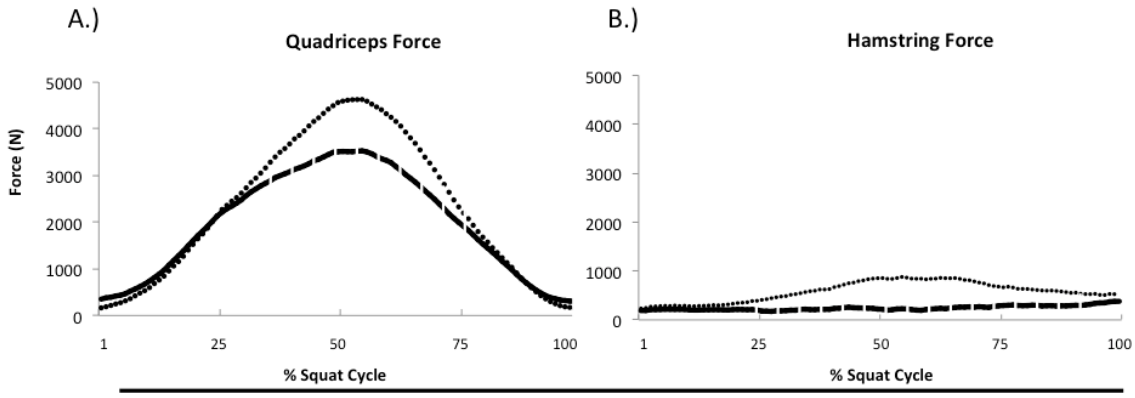


Figure 15 – A.) Depicts ensemble average difference between scaled and ultrasound quadriceps muscle forces. B.) Depicts ensemble average difference between scaled and ultrasound hamstring muscle forces. Dotted line represents the ultrasound models, dashed line represents the scaled model [0% refers to beginning (0° knee flexion), 50% = peak knee flexion, and 100% = ending (0° knee flexion)].

Results Summary:

Both the scaled and ultrasound models did a good job of estimating the knee joint moments but the scaled model had approximately 50% more error compared to the ultrasound model (RMSE: US= 2.71Nm vs. SC= 6.08 Nm) ($p < .05$). EMG analysis of the quadriceps revealed moderate correlation to experimental activation for both models, but less RMSE in the ultrasound model (RMSE: US= 0.16 mV ± .07, SC= 0.23 mV ± .09) ($p < .05$). EMG analysis of the hamstrings revealed moderate correlation for the ultrasound model and a weak correlation for the scaled model (US r= 0.44 mV ± .25, $p< 0.01$ vs. SC r= 0.23 mV ± .30, $p>0.01$). The RMSE was not different between models (RMSE: US= 0.13 mV ± .07 vs. SC= 0.11 mV ± .04) ($p > .05$).

The Ultrasound model more accurately predicted knee joint moments (RMSE: US= 2.71Nm vs. SC= 6.08 Nm) ($p < .05$), better resembled the experimentally collected EMG, and produced higher quadriceps and hamstring muscle forces compared to the scaled models. It seems that using ultrasound to derive parameters used in muscle modeling helps to preserve the subject specificity of the model.

Chapter 5: Discussion

The purpose of this study was to determine if subject specific musculotendon parameters estimated in vivo using ultrasound would better estimate moments produced about a joint compared to previous scaling methods. To do this we used muscle parameters that were derived from direct measurements of individual subjects attained via ultrasound (Ahtianen et al, 2009, Xiao et al, 2010) The overall results support using subject specific parameters. The joint moments predicted using the ultrasound parameters showed 55% less error than the moments predicted using scaled parameters when compared to traditional inverse dynamics joint moments (Scaled RMSE= 6.08 ± 1.03, Ultrasound RMSE= 2.71 ± 0.73). Both ultrasound and scaled model's knee joint moments had a strong correlation to the inverse dynamics model (SC r = .993, US r = .997). While both models did a good job overall of predicting the inverse dynamics knee joint moments, the ultrasound-based model more accurately predicted inverse dynamics knee moments compared to the scaled model and had 55% less error. The scaled model had an absolute percent difference of 3.19% of the peak joint moment compared to the inverse dynamics model. The ultrasound model has an absolute percent difference of only 1.61 % of the peak joint moment compared to the inverse dynamics model.

While the EMG analysis showed mixed results in terms of which model performed better compared to the experimental EMG, the average quadriceps EMG signals generated via the ultrasound-based model better reflected experimental EMG (RMSE= 0.16 ± .07) compared to scaled model (RMSE= 0.23 ± .09). The correlation between the two model's quadriceps' EMG and the experimental muscle

activation was moderate at best (Scaled r= .55 ± .27, p>0.01, Ultrasound r= .50 ± .45, p<0.01). Figure 9 illustrates the closer fit of the ultrasound model vs the scaled model to the ensemble average quadriceps EMG.

In contrast to the quadriceps activation RMSE, the scaled and ultrasound models were no different when comparing the RMSE of the hamstring muscles - Ultrasound RMSE = 0.13 ± .07, Scaled RMSE = 0.11 ± .04 (p>.05). However, the ultrasound model's ensemble average curve seemed to qualitatively better represent the experimental EMG compared to the scaled model shown in Figure 11.

Overall, neither scaled or ultrasound models produced optimized muscle activation signals directly matching to the experimental muscle activations (Figure 9 & 11). However, it did appear that the ultrasound-based models' optimized muscle activations were more closely representative of experimentally collected data compared to the scaled model in terms of the shape and amplitude of the curves. Qualitatively the ultrasound model EMG activation curves seem to be consistent with the level of activation required to meet the same force and joint moments. In the quadriceps the ultrasound model's activation curve seems to match the same bell shape as the muscle force and joint moment curves. The ultrasound model's hamstring activation curve follows the same representative curve as the raw EMG but with less activation. The lower levels of activation suggest the model strictly activates the hamstring as a synergist muscle group to the squatting movement. Hamstring activation simply controls the eccentric portion of the movement and aids in hip extension. The model estimations were done during a single legged squatting task, which is a quadriceps dominant movement. This explains the greater

levels of activation in the quadriceps vs. the hamstrings; and may also explain why the hamstrings activation was dramatically lower than the quadriceps in the ultrasound model. Both models used optimized EMG; the only difference between the two models is the parameters used. Using ultrasound-derived parameters seems to preserve the EMG activation signals better than traditional scaling methods.

Despite having reasonable error in producing knee joint moments, the ultrasound models produced higher quadriceps and hamstring muscle forces combined with lower muscle activations (Quadriceps, Figure 9) compared to the scaled models. The different muscle parameters are the only factors that were different between the two models: kinematics, ground reaction forces, center of pressure, experimental muscle activations, musculotendon kinematics, muscle moment arms were all identical in both models. A qualitative comparison of the parameters in both the scaled and ultrasound models suggests that the maximal isometric forces and optimal fiber lengths for the quadriceps were the driving factors leading to the muscle force and muscle activation discrepancies. The parameter maximal isometric force (F-max) for the quadriceps were 40% greater in the ultrasound vs scaled models and the hamstring F-maxes were about 19% greater in the ultrasound models as well (Appendix C) (Ward et al, 2009). Fmax was calculated based on a relative percentage of total quadriceps volume (Appendix A) based on the rectus femoris volume computed using a regression equation (Morse et al, 2007). Optimal fiber length (OFL) is the denominator in the formula used to calculate F-max. The ultrasound model contained quadriceps OFL estimates that were regularly smaller verses the scaled model (Appendix C). Based on the F-max's

dependence on muscle volume and optimal fiber length: the lower quadriceps EMG in the ultrasound model verses the scaled model may be attributed to the scaled model having less physiological cross-sectional area available to produce the same force; therefore the scaled model required greater activation to produce quadriceps forces necessary to match the inverse dynamics knee moments. However, it should be noted that there were 40 parameters total that were changed between the scaled and ultrasound models (10 muscles, 4 parameters each). Thus it is difficult to attribute these muscle force discrepancies to just one or two factors alone.

The Ultrasound model more accurately predicted knee joint moments (RMSE: US= 2.71Nm vs. SC= 6.08 Nm) ($p < .05$), better resembled the experimentally collected EMG, and produced higher quadriceps and hamstring muscle forces compared to the scaled model. It seems that using ultrasound to derive parameters used in muscle modeling helps to preserve the subject specificity of the model. Ultrasound methodologies could be especially relevant in interventions that cause a change in muscle architecture; as well ass longitudinal changes over time and differences across populations. Ultrasound techniques make it possible to directly observe changes to the muscle architecture.

Scaling techniques may reduce the natural variability in muscle parameters. A coefficient of variation analysis of the parameter OFL in all 10 muscles crossing the knee shows that in the ultrasound model parameters varied on average 15.91% from subject to subject. The variability of parameters in the scaled model was much lower with parameters only varying 7.03%. In 2009 Ward et al, examined the accuracy of current muscle architecture measuring techniques. A coefficient of

variation analysis done on the parameter OFL across 20 cadavers revealed a natural variation of ~20% from person to person (Ward et al, 2009). The use of scaling techniques fails to preserve accurate muscle parameters. Using ultrasound based modeling techniques appears to better reflect natural variability in subjects.

Limitations and Directions for Future Research:

Use of optimization may have "clouded" actual differences between scaled and ultrasound-based models. Hamstring and quadriceps ultrasound-based models both underestimated the experimentally collected data. Quadriceps EMG signals generated via the ultrasound-based model gave a closer representation of experimental EMG signal and had 30% less error. Hamstring EMG signals generated via the ultrasound-based model preserved the muscle activation pattern of the experimental collected EMG and had a 47% stronger correlation.

The objective function of the models optimization algorithm may be the largest contributing factor for why the quadriceps muscle activations appeared to more closely represent the experimental EMG compared to the hamstring muscles. The optimization algorithm's objective function was to minimize the sum of muscle activations squared in order for the knee muscle moments to match the inverse dynamics calculated joint knee moments (minimizing cost function). Errors were seen with this optimization approach when evaluating the on/off analysis. Hamstring and quadriceps' muscle activation signals were periodically dropped to zero in some muscles when the experimental EMG was clearly NOT zero and this appeared in both the scaled and ultrasound-based models. The process of trying to

find the best possible fit for an individual muscle affects both agonist and antagonist muscle groups (Xiao et al, 2010). Models dropping the muscle's EMG activation to zero essentially shuts the muscle off for a period of time. As a reciprocal effect the antagonistic muscle groups muscle activations are then altered to compensate for the zero agonist muscle activation. Thus, for future research investigating whether ultrasound-based muscle parameters are better than scaled parameters, it would be advantageous to evaluate the joint moments produced via an EMG-driven approach where static optimization is not utilized.

The validity of ultrasound measurements could be considered a limitation. The ability to estimate musculotendon parameters via ultrasound imaging is limited to the collector's ability to accurately measure muscle cross-sectional area, fascicle length, and pennation angles. For example, applying too much pressure to the transducer while collecting the image can alter all of these measurements. In addition, increasing amounts of adipose tissue appeared to affect the crispness of the image. Thus, future research utilizing ultrasound measurements to derive muscle parameters should involve testers skilled in the use of equipment and a body composition criterion for subject selection.

A third limitation of the current study is that it was only applied to a simple squatting task. To test the robustness and usefulness of ultrasound-based muscle parameters, future research should examine the application of these subject specific techniques on the estimation of muscle forces and joint moments during more dynamic tasks (running, jumping, etc.).

There is no way to know which one parameter and muscle had the greatest effect on the model. The parameter that varied most between the ultrasound and scaled model is maximal isometric force (Fmax). In the quadriceps Fmax varied 53%, in the hamstrings Fmax varied 47%, and in the Gastrocnemius Fmax varied 34%. The parameter that varied the least was tendon slack length (TSL). In the quadriceps TSL varied 12%, in the hamstrings TSL varied 24%, and in the gastrocnemius TSL varied 15%. Due to the fact that we manipulated 40 parameters (4 parameters per muscle, across 10 muscles) simultaneously we cannot attribute the differences in moments, EMG, or forces, to simply one parameter.

Discussion Summary:

A model's ability to accurately predict joint moments and forces is reliant on specific parameters. Previous methods of muscle modeling estimated these parameters (Zajac, 1989, Delp et al, 1990, Lloyd & Buchanan, 2005). With the present investigation, the result of using a scaled modeling method is an increased error in net knee joint moments, muscle activations (quadriceps only), as well as lower overall forces estimations (quadriceps and hamstrings). Using ultrasound to derive parameters for muscle modeling techniques seems to preserve subject

specificity and give a better representation of muscle force distribution across all muscles (Ahtianen et al, 2009, Xiao et al, 2010). While the error in joint moment estimations in both models was low, overall muscle activations and muscle forces varied dramatically between models. While these outcomes may be due to the effect of inaccurate muscle force-producing parameters, future research should aim to further validate the ultrasound-based images and utilize an EMG-driven modeling approach which excludes use of static optimization (Xiao et al, 2010). Using parameters derived directly from individual subjects for muscle modeling appears to give a better representation of an individual's joint moments and muscle forces produced during a specific task, and further research in this area is supported (Xiao et al, 2010).

Conclusion:

Muscle modeling has proven to be an important tool in the assessment and understanding of muscle function since Delp introduced a computer-based method of estimating muscle forces (Zajac, 1989, Delp et al, 1990). Research in the application of musculoskeletal modeling has continued to make advances over the past ten years (Lloyd and Buchanan, 2005, Arnold et al, 2010, Xiao et al, 2010). Previous studies have shown estimation of muscle forces is dependent on specific parameter values (Xiao et al, 2010) and that musculoskeletal parameters may vary from subject to subject (Ward et al, 2008). Using ultrasound to measure muscle architecture and then using parameters derived from these measurements may better preserve subject specificity (Ahtiainen et al, 2009). Muscle models using

muscle parameters directly measured from a subject appears to make more accurate predictions of knee joint moments. Using ultrasound to derive parameters for muscle modeling techniques seems to preserve subject specificity and give a better representation of muscle force distribution across all muscles. More research is warranted in the attempt of deriving a more physiological relevant muscle modeling technique. In addition to manipulations of musculoskeletal modeling techniques, the next step would be estimating muscle forces and joint moments required for more dynamic tasks (running, jumping, etc.).

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Appendix: A

Quadriceps Volume Estimates

			Absolute Muscle Volumes (cm^3)					Relative Muscle Volume				
Authors	Subject characteristics	Age	Vas Lat	Vas Int	Vas Med	Rect Fem	Total	Vas Lat	Vas Int	Vas Med	VastiiTot	Rect Fem
Narici et al 1992	6 males	34.0±4.7	586.00	574.00	425.00	239.00	1824.00	32.13	31.47	23.30	86.90	13.10
Erskine et al 2009	27 untrained males	21.3±3.4	677.00	586.00	466.00	345.00	2074.00	32.64	28.25	22.47	83.37	16.63
	27 trained males	21.3±3.4	704.00	609.00	500.00	375.00	2188.00	32.18	27.83	22.85	82.86	17.14
Obrien et al 2009	10 untrained males	28.2±3.6	691.00	558.00	523.00	280.00	2052.00	33.67	27.19	25.49	86.35	13.65
	10 untrained females	27.4±4.2	456.00	374.00	350.00	179.00	1359.00	33.55	27.52	25.75	86.83	13.17
Morse et al 2007	18 men recreationally active	23.9±3.4	702.00	604.00	468.00	266.00	2040.00	34.41	29.61	22.94	86.96	13.04
Wickiewicz et al 1983	Cadaver specimen #1	NR	220.18	293.57	235.49	109.30	858.53	25.65	34.19	27.43	87.27	12.73
	Cadaver specimen #2	NR	339.50	98.10	221.55	96.94	756.10	44.90	12.97	29.30	87.18	12.82
	Cadaver specimen #3	NR	135.59	113.84	98.42	60.72	408.57	33.19	27.86	24.09	85.14	14.86
Ward et al 2009	21 subjects (M:F = 9:12)	83.9	396.95	181.53	252.81	116.79	948.08	41.87	19.15	26.67	87.68	12.32
Tate et al 2006	6 males	19.4	692.00	538.50	447.50	280.50	1958.50	35.33	27.50	22.85	85.68	14.32
	4 females	18	372.00	330.00	248.50	153.00	1103.50	33.71	29.90	22.52	86.14	13.86
Akima et al 2007	6 males (training group)	23.3±4.9	458.70	381.80	344.60	198.60	1383.80	33.15	27.59	24.90	85.64	14.35
	6 males (control group)	22.7±3.9	552.40	464.80	423.00	274.90	1715.20	32.21	27.10	24.66	83.97	16.03
Akima et al 2000	5 males (training group)	24.0±4.7	706.70	594.60	477.20	294.50	2073.00	34.09	28.68	23.02	85.79	14.21
	4 males (control group)	19.5±1.7	621.80	546.00	453.00	259.80	1899.30	32.74	28.75	23.85	85.34	13.68
Friederick and Brand 1990	Cadaver Specimen #1	37	514.00	606.00	555.00	238.00	1913.00	26.87	31.68	29.01	87.56	12.44
	Cadaver Specimen #2	63	133.00	135.00	123.00	60.00	451.00	29.49	29.93	27.27	86.70	13.30
		Grand Mean	497.71	421.60	367.34	212.61	1500.31	33.43	27.62	24.91	85.96	13.98
		Grand SD	196.60	187.86	138.88	95.34	602.48	4.43	4.72	2.23	1.41	1.40
		Grand CV	0.40	0.45	0.38	0.45	0.40	0.13	0.17	0.09	0.02	0.10
			Average Proportions Used to Derive Muscle Volume					**33**	**28**	**25**	**86**	**14**

Appendix: B

Quadriceps Volume Estimates

			Absolute Muscle Volumes (cm^3)					Relative Muscle Volume				
Authors	Subject characteristics	Age	Vas Lat	Vas Int	Vas Med	Rect Fem	Total	Vas Lat	Vas Int	Vas Med	VastiiTot	Rect Fem
Narici et al 1992	6 males	34.0±4.7	586.00	574.00	425.00	239.00	1824.00	32.13	31.47	23.30	86.90	13.10
Erskine et al 2009	27 untrained males	21.3±3.4	677.00	586.00	466.00	345.00	2074.00	32.64	28.25	22.47	83.37	16.63
	27 trained males	21.3±3.4	704.00	609.00	500.00	375.00	2188.00	32.18	27.83	22.85	82.86	17.14
Obrien et al 2009	10 untrained males	28.2±3.6	691.00	558.00	523.00	280.00	2052.00	33.67	27.19	25.49	86.35	13.65
	10 untrained females	27.4±4.2	456.00	374.00	350.00	179.00	1359.00	33.55	27.52	25.75	86.83	13.17
Morse et al 2007	18 men recreationally active	23.9±3.4	702.00	604.00	468.00	266.00	2040.00	34.41	29.61	22.94	86.96	13.04
Wickiewicz et al 1983	Cadaver specimen #1	NR	220.18	293.57	235.49	109.30	858.53	25.65	34.19	27.43	87.27	12.73
	Cadaver specimen #2	NR	339.50	98.10	221.55	96.94	756.10	44.90	12.97	29.30	87.18	12.82
	Cadaver specimen #3	NR	135.59	113.84	98.42	60.72	408.57	33.19	27.86	24.09	85.14	14.86
Ward et al 2009	21 subjects (M:F = 9:12)	83.9	396.95	181.53	252.81	116.79	948.08	41.87	19.15	26.67	87.68	12.32
Tate et al 2006	6 males	19.4	692.00	538.50	447.50	280.50	1958.50	35.33	27.50	22.85	85.68	14.32
	4 females	18	372.00	330.00	248.50	153.00	1103.50	33.71	29.90	22.52	86.14	13.86
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	6 males (control group)	22.7±3.9	552.40	464.80	423.00	274.90	1715.20	32.21	27.10	24.66	83.97	16.03
Akima et al 2000	5 males (training group)	24.0±4.7	706.70	594.60	477.20	294.50	2073.00	34.09	28.68	23.02	85.79	14.21
	4 males (control group)	19.5±1.7	621.80	546.00	453.00	259.80	1899.30	32.74	28.75	23.85	85.34	13.68
Friederick and Brand 1990	Cadaver Specimen #1	37	514.00	606.00	555.00	238.00	1913.00	26.87	31.68	29.01	87.56	12.44
	Cadaver Specimen #2	63	133.00	135.00	123.00	60.00	451.00	29.49	29.93	27.27	86.70	13.30
		Grand Mean	497.71	421.60	367.34	212.61	1500.31	33.43	27.62	24.91	85.96	13.98
		Grand SD	196.60	187.86	138.88	95.34	602.48	4.43	4.72	2.23	1.41	1.40
		Grand CV	0.40	0.45	0.38	0.45	0.40	0.13	0.17	0.09	0.02	0.10
			Average Proportions Used to Derive Muscle Volume					**33**	**28**	**25**	**86**	**14**

Appendix B – Shows quadriceps volume estimations compiled from the literature. Appendix C – Shows hamstring volume estimations compiled from the literature. The yellow shaded area is the average proportions used to derive muscle volumes.

Appendix: C Scaled and Ultrasound Muscle Parameters

		Scaled Model Parameters				Ultrasound-Based Model Parameters					Model Differences (%) (Ultrasound-Scaled)/Scaled*100					Average Percent Differences			
SUBJECT 1		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2388.75	0.1049	0.1392	18.40	**VL**	3215.57	0.0820	0.1629	15.38	**VL**	34.61	-21.79	17.05	-16.41	**Quadriceps**	60.09	-13.54	6.85	11.66
	VMO	1533.71	0.1031	0.1201	29.60	**VMO**	1916.08	0.1054	0.1070	17.50	**VMO**	24.93	2.28	-10.87	-40.88	**Hamstrings**	64.74	-20.50	15.84	17.08
	VI	1093.70	0.1057	0.1147	4.50	**VI**	2504.72	0.0872	0.1315	9.06	**VI**	129.01	-17.50	14.66	101.33	**Gastrocnemius**	49.37	50.48	11.65	0.00
	RF	852.72	0.0764	0.3511	13.90	**RF**	1759.37	0.0633	0.3742	14.26	**RF**	106.33	-17.14	6.58	2.59					
	Total	5868.88				**Total**	9395.74				**Total**	60.09								
Hamstrings	**SM**	1269.31	0.0753	0.4184	15.10	**SM**	1967.64	0.0641	0.3956	17.64	**SM**	55.02	-14.91	-5.44	16.82					
	ST	323.98	0.2071	0.2704	12.69	**ST**	619.57	0.1529	0.3455	15.58	**ST**	91.23	-26.18	27.76	22.77					
	BFLH	748.77	0.1041	0.3471	11.60	**BFLH**	1221.75	0.0805	0.3792	13.27	**BFLH**	63.17	-22.63	9.25	14.40					
	BFSH	315.20	0.1098	0.1044	12.30	**BFSH**	568.64	0.0897	0.1376	14.06	**BFSH**	80.41	-18.30	31.80	14.31					
	Total	2657.26				**Total**	4377.60				**Total**	64.74								
Gastrocnemius	**MGAS**	902.04	0.0365	0.3307	17.00	**MGAS**	1444.11	0.0599	0.3701	17.00	**MGAS**	60.09	64.24	11.92	0.00					
	LGAS	365.30	0.0522	0.3139	8.00	**LGAS**	448.95	0.0714	0.3497	8.00	**LGAS**	22.90	36.71	11.38	0.00					
	Total	1267.34				**Total**	1893.06				**Total**	49.37								
SUBJECT 2		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2511.21	0.1102	0.1463	18.40	**VL**	2840.19	0.0947	0.1541	14.09	**VL**	13.10	-14.08	5.36	-23.45	**Quadriceps**	34.47	1.94	-0.82	39.60
	VMO	1610.37	0.1082	0.1261	29.60	**VMO**	1643.67	0.1284	0.0987	20.59	**VMO**	2.07	18.70	-21.74	-30.44	**Hamstrings**	27.96	-21.32	22.28	44.99
	VI	1146.42	0.1108	0.1202	4.50	**VI**	2792.81	0.0816	0.1455	13.85	**VI**	143.61	-26.33	21.06	207.78	**Gastrocnemius**	24.50	-11.24	0.83	110.96
	RF	931.17	0.0834	0.3834	13.90	**RF**	1059.04	0.1079	0.3528	14.53	**RF**	13.73	29.48	-7.98	4.50					
	Total	6199.17				**Total**	8335.71				**Total**	34.47								
Hamstrings	**SM**	1291.79	0.0767	0.4258	15.10	**SM**	1582.18	0.0633	0.4114	17.64	**SM**	22.48	-17.52	-3.39	16.82					
	ST	331.77	0.2121	0.2769	12.90	**ST**	660.38	0.1181	0.4074	15.58	**ST**	99.05	-44.30	47.13	20.78					
	BFLH	776.23	0.1079	0.3598	11.60	**BFLH**	784.11	0.1023	0.3838	13.27	**BFLH**	1.02	-5.23	6.68	14.40					
	BFSH	337.83	0.1177	0.1119	12.30	**BFSH**	476.30	0.0963	0.1552	28.04	**BFSH**	40.99	-18.22	38.69	127.97					
	Total	2737.62				**Total**	3502.96				**Total**	27.96								
Gastrocnemius	**MGAS**	1120.13	0.0453	0.4106	17.00	**MGAS**	1514.95	0.0441	0.4112	29.06	**MGAS**	35.25	-2.54	0.16	70.91					
	LGAS	491.39	0.0644	0.3877	8.00	**LGAS**	491.35	0.0516	0.3935	20.08	**LGAS**	-0.01	-19.94	1.51	151.00					
	Total	1611.52				**Total**	2006.30				**Total**	24.50								
SUBJECT 3		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2532.08	0.1032	0.1370	18.40	**VL**	3440.50	0.0851	0.1466	13.07	**VL**	35.88	-17.54	7.02	-28.97	**Quadriceps**	67.25	-6.01	1.44	21.64
	VMO	1505.43	0.1012	0.1178	29.60	**VMO**	2421.99	0.1011	0.1134	28.07	**VMO**	60.88	-0.10	-3.77	-5.17	**Hamstrings**	66.08	-7.98	12.54	23.13
	VI	1067.72	0.1032	0.1120	4.50	**VI**	2660.66	0.0924	0.1203	10.32	**VI**	149.19	-10.42	7.42	129.33	**Gastrocnemius**	-26.69	-7.49	0.07	35.96
	RF	887.38	0.0795	0.3654	13.90	**RF**	1499.73	0.0827	0.3474	12.70	**RF**	69.01	4.03	-4.93	-8.63					
	Total	5992.61				**Total**	10022.88				**Total**	67.25								
Hamstrings	**SM**	1208.93	0.0717	0.3985	15.10	**SM**	1966.03	0.0698	0.4009	20.92	**SM**	62.63	-2.65	0.60	38.54					
	ST	314.39	0.2010	0.2624	12.90	**ST**	634.94	0.1733	0.2684	11.12	**ST**	101.96	-13.78	2.29	-13.80					
	BFLH	728.10	0.1012	0.3375	11.60	**BFLH**	927.59	0.0806	0.3396	10.84	**BFLH**	27.40	-20.36	0.62	-6.55					
	BFSH	325.32	0.1133	0.1078	12.30	**BFSH**	751.00	0.1188	0.1581	21.44	**BFSH**	130.85	4.85	46.66	74.31					
	Total	2576.74				**Total**	4279.56				**Total**	66.08								
Gastrocnemius	**MGAS**	1170.62	0.0473	0.4291	17.00	**MGAS**	778.79	0.0575	0.4297	17.22	**MGAS**	-33.47	21.56	0.14	1.29					
	LGAS	513.21	0.0673	0.4049	8.00	**LGAS**	455.64	0.0427	0.4049	13.65	**LGAS**	-11.22	-36.55	0.00	70.63					
	Total	1683.83				**Total**	1234.43				**Total**	-26.69								
SUBJECT 4		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2737.62	0.1111	0.1510	18.40	**VL**	2680.20	0.1013	0.1521	11.49	**VL**	-2.10	-8.85	0.70	-37.55	**Quadriceps**	25.57	-8.89	3.25	27.17
	VMO	1765.84	0.1138	0.1082	29.60	**VMO**	2141.54	0.0991	0.1195	18.27	**VMO**	21.28	-12.93	10.42	-38.28	**Hamstrings**	-27.49	-19.96	12.30	15.33
	VI	1269.50	0.1041	0.1335	4.50	**VI**	2406.10	0.0962	0.1414	12.80	**VI**	89.53	-7.61	5.94	184.44	**Gastrocnemius**	-26.99	-8.07	-3.70	53.84
	RF	997.92	0.0972	0.4027	13.90	**RF**	1274.47	0.0912	0.3864	13.91	**RF**	27.71	-6.16	-4.05	0.07					
	Total	6770.87				**Total**	8502.31				**Total**	25.57								
Hamstrings	**SM**	1422.89	0.0844	0.4690	15.10	**SM**	875.28	0.0801	0.3989	17.46	**SM**	-38.49	-5.12	-14.94	15.63					
	ST	357.39	0.2285	0.2983	12.90	**ST**	338.98	0.1579	0.3555	10.01	**ST**	-5.15	-30.89	19.16	-22.40					
	BFLH	830.58	0.1154	0.3850	11.60	**BFLH**	513.74	0.1088	0.3828	11.71	**BFLH**	-38.15	-5.73	-0.58	0.95					
	BFSH	340.92	0.1188	0.1129	12.30	**BFSH**	412.42	0.0735	0.1644	20.56	**BFSH**	20.97	-38.11	45.55	67.15					
	Total	2951.77				**Total**	2140.43				**Total**	-27.49								
Gastrocnemius	**MGAS**	1107.60	0.0571	0.4060	17.00	**MGAS**	755.01	0.0600	0.3717	19.99	**MGAS**	-31.83	5.08	-8.45	17.56					
	LGAS	444.15	0.0635	0.3817	8.00	**LGAS**	377.92	0.0500	0.3857	15.21	**LGAS**	-14.91	-21.21	1.05	90.13					
	Total	1551.76				**Total**	1132.93				**Total**	-26.99								
SUBJECT 5		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2388.75	0.1049	0.1392	18.40	**VL**	2712.29	0.1018	0.1872	14.53	**VL**	13.54	-2.90	34.50	-21.03	**Quadriceps**	43.78	2.09	32.76	27.82
	VMO	1533.71	0.1030	0.1200	29.60	**VMO**	2231.81	0.0993	0.1530	23.99	**VMO**	45.52	-3.62	27.43	-18.95	**Hamstrings**	50.75	-30.04	61.92	11.43
	VI	1093.70	0.1057	0.1147	4.50	**VI**	2087.71	0.1104	0.1610	10.30	**VI**	90.89	4.45	40.34	128.89	**Gastrocnemius**	20.65	-17.20	39.76	90.34
	RF	852.72	0.0764	0.3511	13.90	**RF**	1406.55	0.0843	0.4521	17.01	**RF**	64.95	10.44	28.77	22.37					
	Total	5868.88				**Total**	8438.36				**Total**	43.78								
Hamstrings	**SM**	1269.31	0.0753	0.4184	15.10	**SM**	1829.80	0.0544	0.5019	16.35	**SM**	44.16	-27.73	19.96	8.28					
	ST	323.98	0.2071	0.2704	12.90	**ST**	531.05	0.1466	0.4642	14.66	**ST**	63.91	-29.23	71.65	13.64					
	BFLH	748.77	0.1041	0.3471	11.60	**BFLH**	897.39	0.0894	0.4843	12.85	**BFLH**	19.85	-14.10	39.54	10.78					
	BFSH	315.20	0.1098	0.1044	12.30	**BFSH**	747.63	0.0559	0.2261	13.90	**BFSH**	137.19	-49.11	116.53	13.01					
	Total	2657.26				**Total**	4005.88				**Total**	50.75								
Gastrocnemius	**MGAS**	1009.59	0.0408	0.3701	17.00	**MGAS**	1275.02	0.0404	0.5039	28.57	**MGAS**	26.29	-1.04	36.15	68.06					
	LGAS	406.87	0.0581	0.3497	8.00	**LGAS**	433.93	0.0387	0.5013	17.01	**LGAS**	6.65	-33.36	43.38	112.63					
	Total	1416.46				**Total**	1708.95				**Total**	20.65								
SUBJECT 6		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2291.95	0.1006	0.1335	18.40	**VL**	2609.67	0.0831	0.1271	12.58	**VL**	13.86	-17.35	-4.81	-31.63	**Quadriceps**	47.04	-14.54	-4.13	17.02
	VMO	1474.62	0.0991	0.1154	29.60	**VMO**	1827.82	0.0995	0.0953	28.07	**VMO**	23.95	0.40	-17.45	-5.17	**Hamstrings**	-22.68	-19.61	8.12	7.59
	VI	1056.14	0.1021	0.1108	4.50	**VI**	2391.98	0.0763	0.1196	9.97	**VI**	126.48	-25.29	7.99	121.56	**Gastrocnemius**	-49.88	-3.86	10.48	84.60
	RF	825.44	0.0739	0.3399	13.90	**RF**	1475.74	0.0621	0.3322	11.58	**RF**	78.78	-15.92	-2.25	-16.69					
	Total	5648.14				**Total**	8305.21				**Total**	47.04								
Hamstrings	**SM**	1196.64	0.0710	0.3944	15.10	**SM**	790.27	0.0709	0.3380	13.87	**SM**	-33.96	-0.15	-14.29	-8.15					
	ST	306.01	0.1956	0.2554	12.90	**ST**	345.73	0.1283	0.3180	14.85	**ST**	12.98	-34.43	24.51	15.12					
	BFLH	726.12	0.1009	0.3366	11.60	**BFLH**	500.48	0.0907	0.3288	11.14	**BFLH**	-31.07	-10.16	-2.31	-4.01					
	BFSH	314.51	0.1096	0.1042	12.30	**BFSH**	330.03	0.0726	0.1298	15.67	**BFSH**	4.93	-33.69	24.57	27.40					
	Total	2543.27				**Total**	1966.51				**Total**	-22.68								
Gastrocnemius	**MGAS**	899.16	0.0364	0.3296	17.00	**MGAS**	434.34	0.0380	0.3580	20.20	**MGAS**	-51.70	4.51	8.61	18.82					
	LGAS	361.85	0.0517	0.3110	8.00	**LGAS**	197.64	0.0454	0.3493	20.03	**LGAS**	-45.38	-12.22	12.34	150.38					
	Total	1261.01				**Total**	631.98				**Total**	-49.88								

LEGEND: VL = vastus lateralis, VMO = vastus medialis, VI = vastus intermedius, RF = rectus femoris, SM = semimembranosus, ST = semitenindosus, BFLH = biceps femoris long head, BFSH = biceps femoris short head, MGAS = medial gastrocnemius, LGAS = lateral gastrocnemius, Fmax = maximal isometric force (newtons), OFL = optimal fiber length (meters), TSL = tendon slack length (meters), Penn = pennation angle

Appendix C continued: Scaled and Ultrasound Muscle Parameters

		Scaled Model Parameters				Ultrasound-Based Model Parameters				Model Differences (%) (Ultrasound-Scaled)/Scaled*100					Average Percent Differences			
SUBJECT 7		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2574.96	0.1130	0.1500	18.40	**VL**	2270.91	0.0964	0.1478	20.77	**VL**	-11.81	-14.71	-1.51	12.88	**Quadriceps**	5.25	-7.26	-7.11	2.70
	VMO	1654.87	0.1112	0.1295	29.60	**VMO**	1467.65	0.1098	0.0972	15.82	**VMO**	-11.31	-1.24	-24.99	-46.55	**Hamstrings**	-27.65	-29.46	12.40	2.88
	VI	1182.72	0.1143	0.1240	4.50	**VI**	1894.21	0.0924	0.1276	7.38	**VI**	60.16	-19.14	2.92	64.00	**Gastrocnemius**	-71.79	-3.51	6.92	5.69
	RF	911.00	0.0816	0.3751	13.90	**RF**	1023.00	0.0865	0.3569	11.19	**RF**	12.29	6.07	-4.86	-19.53					
	Total	6323.56				**Total**	6655.76				**Total**	5.25								
Hamstrings	**SM**	1373.76	0.0815	0.4528	15.10	**SM**	975.88	0.0534	0.4073	11.91	**SM**	-28.96	-34.46	-10.06	-21.13					
	ST	348.43	0.2228	0.2908	12.90	**ST**	220.06	0.1853	0.3172	9.69	**ST**	-36.84	-16.81	9.06	-24.88					
	BFLH	801.43	0.1114	0.3715	11.60	**BFLH**	561.77	0.0753	0.3966	9.24	**BFLH**	-29.90	-32.40	6.74	-20.34					
	BFSH	330.89	0.1153	0.1096	12.30	**BFSH**	307.47	0.0759	0.1577	21.88	**BFSH**	-7.08	-34.18	43.86	77.89					
	Total	2854.51				**Total**	2065.19				**Total**	-27.65								
Gastrocnemius	**MGAS**	1093.10	0.0442	0.4007	17.00	**MGAS**	255.93	0.0535	0.4126	11.03	**MGAS**	-76.59	20.99	2.97	-35.12					
	LGAS	439.36	0.0628	0.3776	8.00	**LGAS**	176.41	0.0452	0.4186	11.72	**LGAS**	-59.85	-28.02	10.88	46.50					
	Total	1532.46				**Total**	432.34				**Total**	-71.79								
Averages		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax**	**OFL**	**TSL**	**Penn**		**Fmax %**	**OFL %**	**TSL %**	**Penn %**
Quadriceps	**VL**	2489.33	0.1068	0.1423	18.40	**VL**	2824.19	0.0921	0.1540	14.56	**VL**	13.87	-13.89	8.33	-20.88	**Quadriceps**	***40.49***	***-6.60***	4.61	21.09
	VMO	1582.65	0.1057	0.1196	29.60	**VMO**	1950.08	0.1061	0.1120	21.76	**VMO**	23.90	0.50	-5.85	-26.49	**Hamstrings**	***18.82***	***-21.27***	20.77	17.49
	VI	1129.99	0.1066	0.1186	4.50	**VI**	2391.17	0.0909	0.1353	10.53	**VI**	112.70	-14.55	14.33	133.90	**Gastrocnemius**	-11.55	-0.13	9.43	54.48
	RF	894.05	0.0812	0.3670	13.90	**RF**	1356.84	0.0826	0.3717	13.60	**RF**	53.26	1.54	1.61	-2.19					
	Total	6096.02				**Total**	8522.28				**Total**	40.49								
Hamstrings	**SM**	1290.38	0.0766	0.4253	15.10	**SM**	1426.73	0.0652	0.4077	16.54	**SM**	11.84	-14.65	-3.94	9.55					
	ST	329.42	0.2106	0.2750	12.87	**ST**	478.67	0.1518	0.3537	13.07	**ST**	46.73	-27.95	28.79	1.60					
	BFLH	765.71	0.1064	0.3549	11.60	**BFLH**	772.41	0.0896	0.3850	11.76	**BFLH**	1.76	-15.80	8.56	1.37					
	BFSH	325.69	0.1134	0.1079	12.30	**BFSH**	513.35	0.0832	0.1612	19.36	**BFSH**	58.32	-26.68	49.67	57.43					
	Total	2711.20				**Total**	3191.16				**Total**	18.82								
Gastrocnemius	**MGAS**	1043.18	0.0439	0.3824	17.00	**MGAS**	922.59	0.0505	0.4082	20.44	**MGAS**	-10.28	16.12	7.36	20.22					
	LGAS	431.73	0.0600	0.3609	8.00	**LGAS**	368.83	0.0493	0.4004	15.10	**LGAS**	-14.55	-16.37	11.50	88.75					
	Total	1474.91				**Total**	1291.43				**Total**	-11.55								

LEGEND: VL = vastus lateralis, VMO = vastus medialis, VI = vastus intermedius, RF = rectus femoris, SM = semimembranosus, ST = semitenindosus, BFLH = biceps femoris long head, BFSH = biceps femoris short head, MGAS = medial gastrocnemius, LGAS = lateral gastrocnemius, Fmax = maximal isometric force (newtons), OFL = optimal fiber length (meters), TSL = tendon slack length (meters), Penn = pennation angle

Appendix D: IRB Approval Form

EAST CAROLINA UNIVERSITY
University & Medical Center Institutional Review Board Office
1L-09 Brody Medical Sciences Building• 600 Moye Boulevard • Greenville, NC 27834
Office **252-744-2914** • Fax **252-744-2284** • **www.ecu.edu/irb**

TO: Anthony Kulas, PhD, LAT, ATC, ECU, Health Education and Promotion, Mailstop 529

FROM: UMCIRB KWB

DATE: May 10, 2011

RE: Expedited Category Research Study

TITLE: "Validation of Muscle Forces during Squatting and Landing Tasks"

UMCIRB #11-0298

This research study has undergone review and approval using expedited review on **05/03/2011**. This research study is eligible for review under an expedited category number four (4) which includes collection of data through non-invasive procedures (not involving general anesthesia or sedation) routinely employed in clinical practice, excluding procedures involving x-rays or microwaves. Where medical devices are employed, they must be learned/approved for marketing. (Studies intended to evaluate the safety and effectiveness of the medical device are not generally eligible for expedited review, including studies of cleared medical devices for new indication.). Examples: (a) physical sensors that are applied, whether to the surface of the body or at a distance, and do not involve input of significant amounts of energy into the subject or significant amounts of energy into the subject or an invasion of the subject's privacy; (b) weighing or testing sensory acuity; (c) magnetic resonance imaging; (d) electrocardiography, electroencephalography, thermography, detection of naturally occurring radioactivity, electroretinography, ultrasound, diagnostic infrared imaging, Doppler blood flow, and echocardiography; (e) moderate exercise, muscular strength testing, body composition assessment, and flexibility testing where appropriate given the age, weight, and health of the individual. The Chairperson (or designee) deemed this **unfunded** study **no more than minimal risk** requiring a continuing review in **12 months**. Changes to this approved research may not be initiated without UMCIRB review except when necessary to eliminate an apparent immediate hazard to the participant. All unanticipated problems involving risks to participants and others must be promptly reported to the UMCIRB. The investigator must submit a continuing review/closure application to the UMCIRB prior to the date of study expiration. The investigator must adhere to all reporting requirements for this study.

The above referenced research study has been given approval for the period of **05/03/2011** to **05/02/2012**. The approval includes the following items:

- Internal Processing Form (dated 04/18/2011)
- Informed consent (version date 04/28/2011)
- Recruitment Announcement
- Interview Questions
- UMCIRB COI

The Chairperson (or designee) does not have a potential for conflict of interest on this study.

The UMCIRB applies 45 CFR 46, Subparts A-D, to all research reviewed by the UMCIRB regardless of the funding source. 21 CFR 50 and 21 CFR 56 are applied to all research studies under the Food and Drug Administration regulation. The UMCIRB follows applicable International Conference on Harmonisation Good Clinical Practice guidelines.

Appendix E: Informed Consent Form

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CENTENNIAL
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UNIVERSITY HEALTH SYSTEMS
of Eastern Carolina

Informed Consent to Participate in Research

Information to consider before taking part in research that has no more than minimal risk.

Title of Research Study: Validation of Muscle Forces during Squatting and Landing Tasks

Principal Investigator: Anthony Kulas
Institution/Department or Division: East Carolina University Department of Health and Human Performance
Address: 249 Ward Sports Medicine Building
Telephone #: (252) 737-2884

Researchers at East Carolina University (ECU) study problems in society, health problems, environmental problems, behavior problems and the human condition. Our goal is to try to find ways to improve the lives of you and others. To do this, we need the help of volunteers who are willing to take part in research.

Why is this research being done?
The purpose of this research is to attempt to validate a novel method of computing your knee muscle forces during dynamic squatting and landing tasks. These muscle forces will be based on your actual muscle architecture as imaged through real-time ultrasound imaging technologies. Estimation and validation of muscle forces using ultrasound-based data would represent a significant advancement in science and allow us to have a better understanding of how humans generate muscle forces during dynamic movements. The decision to take part in this research is yours to make.

Why am I being invited to take part in this research?
You are being invited to take part in this research because you have experience in both squatting and jumping and landing activities and you are self-reportedly healthy. If you volunteer to take part in this research, you will be one of twelve (12) people to do so.

Are there reasons I should not take part in this research?
I understand I should not volunteer to be in this study if I am under 18 years of age, if I have a history within the past year of lower extremity injury or any history of lower extremity surgeries performed at any time, or if I have a known allergy to ultrasound gel.

What other choices do I have if I do not take part in this research?
Because enrollment in this research study is voluntary, you may simply choose not to participate.

Where is the research going to take place and how long will it last?
The research study will be conducted in the Ward Sports Medicine Building at East Carolina University. You will need to come to room 373 in the Ward Sports Medicine Building twice during the study. The total amount of time you will be asked to volunteer for this study is approximately 4 hours – 2 hours each session.

Title of Study: *Validation of Muscle Forces during Squatting and Landing Tasks*

What will I be asked to do?

- You are being asked to do the following:
  - During the first session, you will lie down on a padded treatment table while we collected ultrasound images of your thigh and lower leg muscles (Quadriceps, Hamstrings, and Gastrocnemius). This will take about 2 hours maximum.
  - During the second session, we will place surface electrodes over these same thigh and lower leg muscles, and place reflective markers on your legs, pelvis, and torso. You will then perform several single-leg squats, single-leg landings form a 45cm height, and single-leg forward hops onto a force plate mounted firmly to the ground and flush with the surrounding floor. We will simultaneously monitor your muscles with the electrodes and capture your motion and forces as you perform these three tasks. Finally, you will be instructed to extend your knee and flex your knee while seated in a dynamometer that measures your strength. This will conclude the second session which will take 2 hours. Following the second session, your direct involvement in this study will be concluded.

What possible harms or discomforts might I experience if I take part in the research?
It has been determined that the risks associated with this research are no more than what you would experience in everyday activities such as landing from a jump during sporting activities or during common squatting activities. The ultrasound is used for imaging purposes only, and you should not experience any symptoms throughout the testing period.

What are the possible benefits I may experience from taking part in this research?
This research will help us learn more about how to best estimate muscle forces specific to an individual. Previous research has estimated muscle forces during human movement based on generic models and are therefore not subject-specific. There may be no direct personal benefit from your participation but the information gained by doing this research will enhance our understanding of muscle force estimation for the scientific community. At your request, we will gladly provide you with a summary of the results once the study is concluded.

Will I be paid for taking part in this research?
No, we will not pay you for the time you volunteer while being in this study

What will it cost me to take part in this research?
It will not cost you any money to be part of the research.

Who will know that I took part in this research and learn personal information about me?
To do this research, ECU and the people and organizations listed below may know that you took part in this research and may see information about you that is normally kept private. With your permission, these people may use your private information to do this research.

- The University & Medical Center Institutional Review Board (UMCIRB) and its staff, who have responsibility for overseeing your welfare during this research, and other ECU staff who oversee this research.
- Additionally, the following people and/or organizations may be given access to your personal health information and they are:
  - Anthony S. Kulas, PhD, LAT, ATC
  - John Pope, BS

How will you keep the information you collect about me secure? How long will you keep it?
If you elect to enroll in this study by signing this informed consent document, you be assigned an alphanumeric code. Only this alphanumeric code, not your name, will appear on the saved ultrasound images, motion capture trials, or electronically saved strength measurements. The only person to have access to the master list of names will be the

two researchers identified above, John Pope and/or Dr. Anthony S Kulas. All paperwork and forms linking you to the study will be kept in Ward Sports Medicine Building, Room 249, Dr. Kulas's office, which remains locked except when in use. Because this is a validation study, your data may be used to guide future studies. In addition, your ultrasound images, strength data, or motion capture data collected in this study may be used for manuscript/presentation purposes. If used for these reasons, no information identifying you (your name or alphanumeric code) will be on any images/figures used for research purposes.

What if I decide I do not want to continue in this research?
If you decide you no longer want to be in this research after it has already started, you may stop at any time. You will not be penalized or criticized for stopping. You will not lose any benefits that you should normally receive.

Who should I contact if I have questions?
The people conducting this study will be available to answer any questions concerning this research, now or in the future. You may contact the Principal Investigator at (252) 737-2884 (days, between 8am-5pm).

If you have questions about your rights as someone taking part in research, you may call the UMCIRB Office at phone number 252-744-2914 (days, 8:00 am-5:00 pm). If you would like to report a complaint or concern about this research study, you may call the Director of UMCIRB Office, at 252-744-1971.

I have decided I want to take part in this research. What should I do now?
The person obtaining informed consent will ask you to read the following and if you agree, you should sign this form:

- I have read (or had read to me) all of the above information.
- I have had an opportunity to ask questions about things in this research I did not understand and have received satisfactory answers.
- I know that I can stop taking part in this study at any time.
- By signing this informed consent form, I am not giving up any of my rights.
- I have been given a copy of this consent document, and it is mine to keep.

_____ _____ _____
Participant's Name (PRINT) **Signature** **Date**

Person Obtaining Informed Consent: I have conducted the initial informed consent process. I have orally reviewed the contents of the consent document with the person who has signed above, and answered all of the person's questions about the research.

_____ _____ _____
Person Obtaining Consent (PRINT) **Signature** **Date**