

**ESTIMATED MUSCLE FORCES DURING RUNNING AND CUTTING  
FOR ACL DEFICIENT, ACL RECONSTRUCTED, AND HEALTHY  
SUBJECTS USING A TWO-JOINT EMG-DRIVEN MODEL**

by

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A thesis submitted to the Faculty of the University of Delaware in partial fulfillment of the requirements for the degree of Master of Science in Mechanical Engineering

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## **ABSTRACT**

The purpose of this study was to extend an EMG-Driven Model to two joints in order to compare predicted joint moments and estimated muscle forces during dynamic movements between healthy subjects and those who had injured their anterior cruciate ligaments (ACL) and were either ACL-deficient (ACL-D) or had their ACL reconstructed (ACL-R). Eight subjects volunteered for this study, with one subject as both ACL-D and ACL-R since data were collected both pre-surgery and post-surgery. The forces of 12 muscles of the thigh and shank were estimated for both straight-ahead running and running with a side cut at 45°. Comparisons were made for estimated muscle forces at their peak force during gait. In order to be compared between different subjects, the muscle forces were normalized to the theoretical maximum muscle force as computed by the model. Motion capture data were acquired for each subject for walking, running, and run-cut tasks. Additionally, maximum voluntary isometric and isokinetic contractions were collected using a dynamometer. The collected data were processed and combined with a musculoskeletal model in order to be input into the EMG-Driven Model. Normalized muscle force for the biceps femoris long-head were observed to be decreased for the ACL-D and ACL-R groups compared to the healthy group, while normalized muscle force for the biceps femoris short-head was increased for the ACL-D group over the healthy and ACL-R groups. Normalized muscle force for all vastii muscles were decreased for the ACL-D group only for running tasks compared to all other groups and to cutting trials. These findings suggest that the force of the biceps femoris short-

head increases for ACL-D patients in order to resist anterior tibial translation during dynamic tasks. The decrease in force of the biceps femoris long-head coincides with the increase in force of the biceps femoris short-head in order to maintain proper force and moment contribution from the lateral hamstrings as a whole. The decrease in vastii force for the ACL-D subjects agrees with previous studies that have reported quadriceps weakness for ACL-D patients. This result could be related to pain prevention, a lack of confidence in the injured leg, and decreased joint proprioception for the ACL-D subjects. The subject with pre and post surgery data underwent a quadruple semitendinosus-gracilis graft and exhibits two major differences from the previous results: the forces of the biceps femoris long-head and the semimembranosus increase post surgery. Five of the eight muscles crossing the knee show an increase in peak force post-surgery during running, indicating that joint contact forces also increase post-surgery. In summary, we have presented an application of an EMG-Driven Model expanded to two-joints in order to successfully predict joint moments and estimate muscle forces for dynamic movements between healthy subjects and those with ACL injury. It is hoped that this model will be applied to investigate altered muscle force patterns between healthy subjects and pathological subject groups.

## **Chapter 1**

### **INTRODUCTION**

Rupture of the anterior cruciate ligament (ACL) is one of the most common knee injuries in the United States, with between 80,000 to 250,000 reported cases each year [Gianotti 2009, Griffin 2006]. The injury occurs during large external rotation of the tibia and internal rotation of the femur during excessive weight bearing at the knee [Ireland 1999]. Injuries can be the result of contact, a strong blow to the knee joint, or the result of extreme and uncontrolled kinematic patterns.

#### **1.1 Effects of ACL Rupture**

ACL deficient (ACL-D) patients experience laxity in the knee joint, as the primary functions of the ACL are to prevent anterior tibial translation (ATT) and medial rotation of the tibia in relation to the femur [Daniel 1994]. The laxity is characterized by episodes of giving way in the knee during excessive weight bearing activities [Beard 1993, Houck 2003]. Other observed outcomes of ACL injury include quadriceps weakness [Lewek 2002, Williams 2005a], decreased proprioception [Beard 1993], and increased co-contraction of the quadriceps and hamstrings [Williams 2005b]. Long-term effects of ACL rupture include a high incidence of osteoarthritis (OA) developing years after ACL rupture, independent of whether or not the ACL was surgically reconstructed [Lohmander 2004, Louboutin 2009].

## **1.2 ACL Reconstructive Surgery**

The complications arising from ACL rupture motivate the need for ACL reconstructive surgery. The two most common ACL graft types are bone patella tendon bone (BPTB) and quadruple semitendinosus-gracilis autografts (QSTG) [Jansson 2003]. A less common practice is using a cadaveric allograft. The large majority of patients that undergo ACL reconstructive surgery display positive results, most being able to return to their pre-injury levels of activity [Daniel 1994] and kinematics return to healthy values [Bulgheroni 1997]. Since the two autografts are taken from the ipsilateral knee, unfavorable outcomes do arise, such as changed muscle morphology [Williams 2005a], altered muscle activation [Williams 2005a, Williams 2004], and quadriceps weakness [Konishi 2002]. Not all people who rupture their ACLs require surgery as they are able to compensate for the laxity in the knee joint [Daniel 1994, Eastlack 1999, Fitzgerald 2000, Rudolph 2001]. These people are called copers for their ability to adjust to the absence of an ACL.

## **1.3 Onset of Osteoarthritis Following Injury to the ACL**

One of the major long-term outcomes of ACL injury is the development of OA in the ipsilateral knee. This is seen both in patients who have had their ACL reconstructed and in copers. Previous research has shown that untreated ACL ruptures can lead to OA that needs to be treated with total knee arthroplasty [Louboutin 2009]. Studies speculate that altered muscle force patterns contribute to the development of OA [Slemenda 1998] and it is well documented that ACL injury leads to altered muscle activation patterns [Williams 2005b]. This motivates exploring the muscle

force patterns used by ACL-D and ACL reconstructed (ACL-R) subjects in order to know the extent of change as a result of ACL rupture.

#### **1.4 Options for Estimating Muscle Forces**

Direct measurement of muscle forces is not an option, a model must be used in order to estimate muscle forces. Models have been developed relying on muscle properties, such as physiological cross-sectional area [Dhafer and Kahn 2002, Elias 2004], muscle moment arms [Brechtner and Powers 2002], or static optimization routines distributing forces based on joint moments [Crowningshield and Brand 1981, Seireg and Arvikar 1973]. Unfortunately, these models rely on assumptions of muscle activation patterns. An electromyography (EMG) driven model [Buchanan 2004, 2005; Lloyd and Besier 2003] that predicts muscle forces combining muscle activation dynamics, muscle contraction dynamics, and a musculoskeletal model presents a worthwhile option since it incorporates both neuromuscular control and physical muscle properties. The model has been used successfully in other studies that explored subject groups with abnormal muscle activation patterns such as stroke patients during walking [Shao 2009] and subjects with patella-femoral pain during running [Besier 2009]. In order to account for the bi-articulating nature of the gastrocnemius muscles, which are both plantar flexors and knee flexors, the EMG-Driven Model was expanded to two joints. The gastrocnemius muscles generate a large plantar flexion force to help push off during dynamic tasks. This study will be the first to apply the EMG-Driven Model for two joints, ankle and knee, whereas the previous two studies focused on either the ankle [Shao 2009] or the knee [Besier 2009] only.

## **1.5 Focus of Study**

This study aims to expand the EMG-Driven Model to a two-joint model in order to predict joint moments and estimate leg muscle forces in healthy, ACL-D, and ACL-R subjects during running and cutting tasks. The results of this study will provide insight to the changes in both muscle group function and individual muscle function as a result of ACL rupture and subsequent surgery.

### **1.5.1 Aim 1: Joint Moment Predictions**

The EMG-Driven Model will be expanded to two joints to accommodate the two gastrocnemius muscles which cross both the knee and ankle. During dynamic tasks, the gastrocnemius muscles mostly act as plantar flexors, motivating the expansion of the model. Comparing joint moment predictions between healthy, ACL-D, and ACL-R subjects will show if the expanded model can be applied towards different subject populations and will provide insight into its use for future studies. We predict there will be no difference in Pearson product-moment correlation,  $R^2$ , and normalized root-mean square error,  $nRMSE$ , of joint moment predictions between groups.

### **1.5.2 Aim 2: Peak Flexion-Extension Knee Joint Moment**

As a combination of the expected quadriceps weakness and the expected increase in hamstrings force, it is expected that the peak flexion-extension knee moment will be decreased for ACL-D and ACL-R subjects compared to healthy subjects. The flexion-extension knee joint moment is a cumulative measure reflecting

the overall effect of changes of individual muscle forces, giving insight to the system wide effects of ACL rupture.

### **1.5.3 Aim 3: Estimated Hamstrings Forces**

Reports of increased quadriceps and hamstrings co-contraction in ACL-D subjects motivates the investigation of hamstrings forces for ACL-D and ACL-R subjects. The attachment of the hamstring muscles at the knee joint occur at the tibia and fibula, wrapping around the posterior of the knee joint. Thus they are positioned to resist ATT in the absence of the ACL. Previous studies have shown that the hamstrings produce the most force during weight acceptance [Besier 2009]. We predict that forces of the hamstrings muscles will be highest for the ACL-D subject group, with no difference between the healthy and ACL-R subject groups.

### **1.5.4 Aim 4: Estimated Quadriceps Forces**

As stated in the introduction, quadriceps weakness has been reported for both ACL-D and ACL-R subjects. During dynamic tasks, the extension moment of the knee is provided mostly by the vastii muscles. We predict that the previously reported quadriceps weakness will manifest in decreased peak vastii forces for ACL-D and ACL-R subjects compared to healthy subjects.

## **1.6 Thesis Outline**

The next chapter will describe the methods of this study with a thorough description of the EMG-Driven Model. Chapter 3 will present the predicted joint moments and estimated muscle forces output from the EMG-Driven Model. Chapter 4

will provide discussion on the results presented in chapter 3. Chapter 5 lists the limitations of this study. Chapter 6 summarizes the results of this study and recommends future uses for the two-joint EMG-Driven Model.

## **Chapter 2**

### **METHODS**

#### **2.1 Subject Selection**

Eight subjects, five male and three female, were selected for this study, with one subject undergoing data collections as both ACL-D and ACL-R (QSTG graft). Table 1 provides a summary of all subjects. All ACL-D and ACL-R subjects were regular (>50 hours per year) Level I or II sports participants at the time of injury. Exclusion criteria included concomitant ligament injury to the involved knee, concomitant meniscal repair, concomitant fracture of either lower extremity, use of an assistive walking device, knee joint effusion, hip or ankle pathology, or injuries to the uninvolved knee. All subjects gave full consent before submitting to the testing protocol, which was approved by the Human Subjects Review Board of the University of Delaware (Appendix A).

Gender	Group	Age at test (years)	Height (m)	Weight (kg)	Time after injury/surgery (weeks)	Affected Limb	Graft type
Female	Healthy	24	1.50	50.45	-	Left	-
Male	Healthy	24	1.78	70.99	-	Left	-
Male	Healthy	26	1.66	66.02	-	Right	-
Female	ACL-D	37	1.63	61.36	14	Left	-
Male*	ACL-D	18	1.75	68.18	10	Left	-
Male	ACL-D	17	1.80	75.74	8	Right	-
Male	ACL-R	17	1.78	66.40	27	Right	Allograft
Male*	ACL-R	19	1.73	74.77	30	Left	QSTG
Female	ACL-R	16	1.71	58.97	30	Left	BPTB

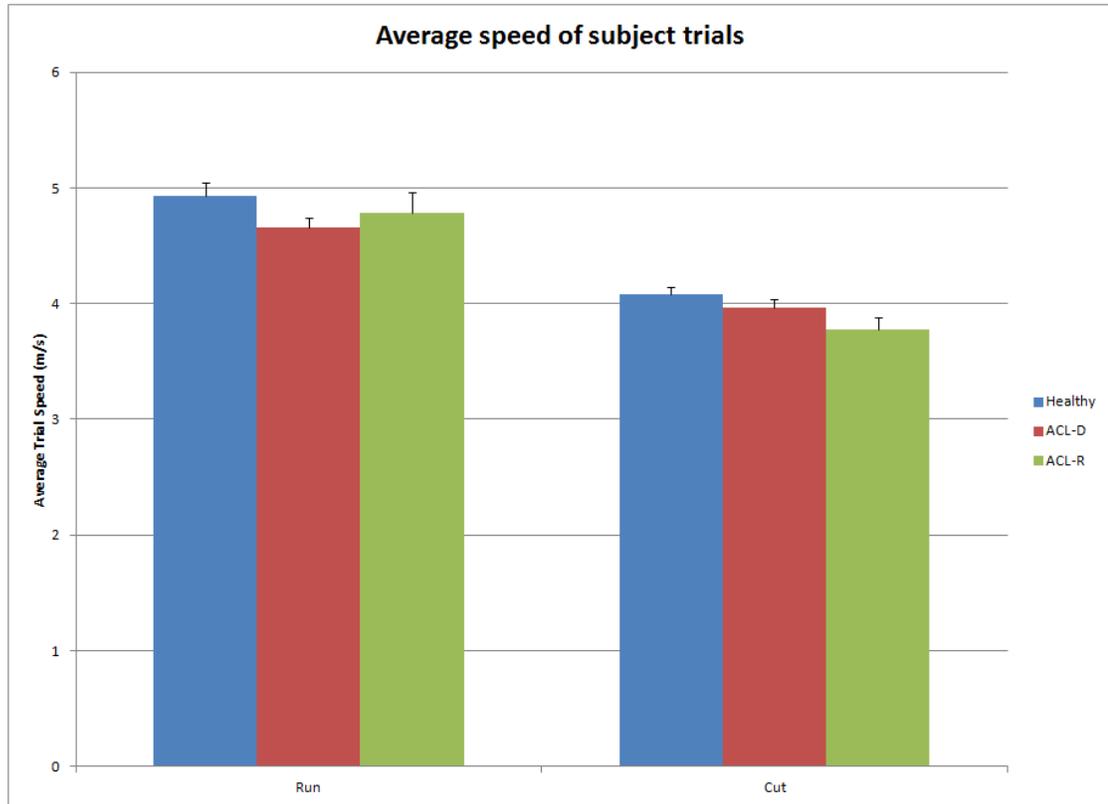
**Table 1: Summary of relevant clinical information of subjects, asterisk indicates individuals that provided multiple data collections**

## 2.2 Data Collection

Before performing any trials, subjects were fitted with reflective markers on anatomical landmarks and Ag/AgCL surface EMG electrodes (Neurotrode 120, Myotronics-Neuromed, Inc., Tukwila, WA) on muscles in standardized locations [Hermens 2000]. Subject body hair was shaved off the electrode sites to promote optimal electrode function. EMG was collected for the following muscles: rectus femoris (RF), vastus lateralis (VL), vastus medialis (VM), biceps femoris long head (BFL), semimembranosus (SM), medial gastrocnemius (MG), lateral gastrocnemius (LG), tibialis anterior (TA), and soleus (SL). EMG electrodes were connected to a differential amplifier with a gain of 20 and a two-pole (20-2000 Hz) bandpass filter. To ensure consistent contact points throughout trials, electrodes and preamplifiers were secured to a subject's body with tape (Hypafix™, Smith and Nephew, London, UK) and wrapped with 10 cm elastic wrap (Superwrap™, Fabrifoam, Inc., Exton, PA, USA). The raw EMG signal was anti-aliased at 500 Hz through a backpack unit (MA-

300-28, Motion Laboratory Systems, Baton Rouge, Louisiana, USA) and sampled at either 1000 Hz or 1080 Hz. Prior to testing, resting trials were recorded in order to remove baseline noise from the EMG signal.

Data collections consisted of anticipated straight-ahead running, anticipated running and side cutting at a 45° angle (away from the support leg, further referred to as "cutting"), anticipated straight-ahead walking, maximum voluntary isometric contraction (MVIC), and isokinetic contraction trials. The speed of these data trials were kept constant to ensure similar levels of muscle activity, and these speeds are shown in figure 1. The data collections were performed at Spencer Laboratory at the University of Delaware (Newark, DE, USA). The lab is equipped with an 20 meter isolated walkway with surface level force plates (Advanced Mechanical Technology Inc., Watertown, MA, USA) and an eight camera motion capture infrared system (Qualysis, Gothenburg, Sweden). Subjects were instructed to walk, run, or run and cut with having their involved limb land on the force plate. Practice trials were taken before each task so the subject could practice striking the force plate with the involved limb. MVIC and isokinetic trials were collected on a Biodex 3 dynamometer (Biodex Medical Systems, Shirley, New York) with MVIC trials taken for both the knee joint (knee at 90 degrees flexion) and the ankle (knee at full extension, ankle at 90 degrees). Isokinetic trials were taken at a constant speed of 60 degrees per second and included a range of motion (ROM) from 90 degrees flexion to full extension for the knee and the subjects' maximum angles of dorsiflexion and plantar flexion for the ankle.



**Figure 1: Average speed of running and cutting trials for this study.**

### 2.3 EMG Processing

Post collection, the EMG signals were high pass filtered using a bi-directional fourth-order Butterworth filter at 50 Hz to remove DC bias, subtracted by its average value to account for DC shift, full wave rectified and lowpass filtered using a bi-directional fourth-order Butterworth filter at 6 Hz to create a linear envelope, then normalized by dividing each signal by the muscle's global maximum value for a particular subject, ensuring all values of normalized EMG input to the model would lie between 0.0 and 1.0, inclusive. The EMG of the vastus intermedius (VI) was

approximated at each data point as the average of the EMG of the VM and VL. The EMG of the biceps femoris short head (BFS) and the semitendinosus (ST) were approximated as the EMG of the BFL and the SM, respectively [Lloyd and Buchanan, 1996]. Motion capture data were processed using Qualisys Track Manager (Qualysis, Gothenburg, Sweden) and were analyzed with Visual 3D (C-Motion Inc., Bethesda, MD).

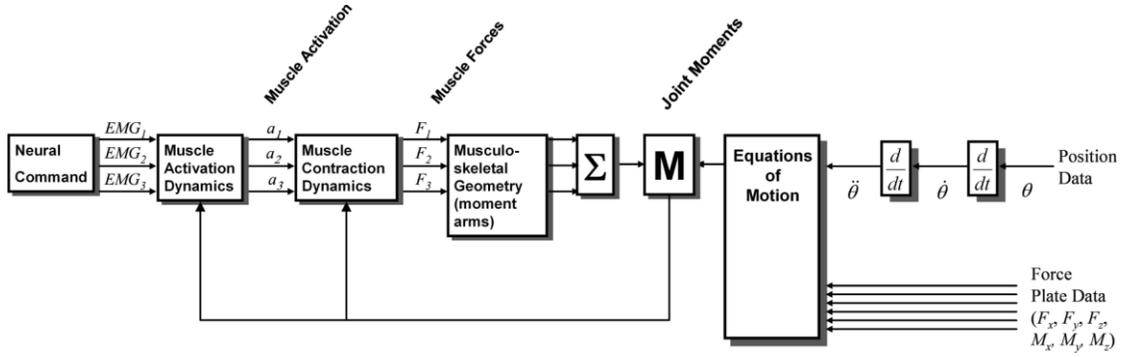
## 2.4 EMG-Driven Model

The EMG-Driven Model [Buchanan 2005, Lloyd and Besier 2003] combines using processed EMG and joint moments calculated by inverse dynamics in conjunction with a musculoskeletal model to estimate muscle tendon lengths and muscle moment arms, muscle activation dynamics, muscle contraction dynamics, and an optimization routine in order to estimate muscle forces and joint moments, as is summarized in figure 2.

### 2.4.1 Modeling Neural Activation

Processed EMG was input into a recursive filter which accounts for electromechanical delay (EMD), the time-varying nature of EMG, and activation dynamics to calculate neural activation,  $u(t)$ ,:

$$\mathbf{u}(t) = \alpha * e(t - d) - \beta_1 * \mathbf{u}(t - 1) - \beta_2 * \mathbf{u}(t - 2) \quad (1)$$



**Figure 2: Flow chart diagram describing the EMG-Driven Model.**

where  $d$  is the EMD,  $e(t)$  is the processed EMG, and  $\alpha$ ,  $\beta_1$ , and  $\beta_2$  are coefficients that define the second-order dynamics. For equation (1) to reach a positive stable solution, the following must be true:

$$\beta_1 = \gamma_1 + \gamma_2 \quad \beta_2 = \gamma_1 * \gamma_2 \quad \alpha - \beta_1 - \beta_2 = 1 \quad (2)$$

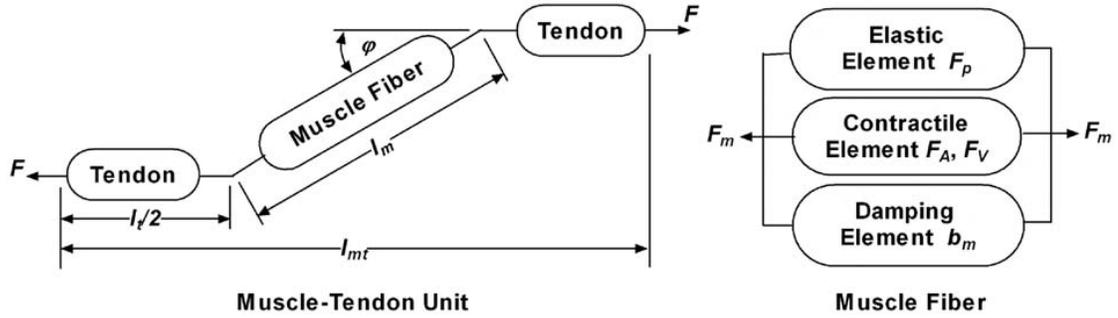
where  $|\gamma_1| < 1$  and  $|\gamma_2| < 1$  [Buchanan 2004].

Neural activation was estimated using a single-parameter model [Manal and Buchanan 2003]. This model accounts for the potential nonlinear relationship between EMG and muscle force at low levels of force [Woods 1983]. A logarithmic function was used to calculate muscle activation from neural activation for low levels of EMG and a linear function was used for high values:

$$a(t) = d * \ln(c * u(t) + 1) \quad 0 \leq u(t) < 0.3 \quad (3)$$

$$a(t) = m * u(t) + b \quad 0.3 \leq u(t) < 1 \quad (4)$$

where  $u(t)$  is the neural activation and  $a(t)$  is the muscle activation. The coefficients  $c$ ,  $d$ ,  $m$ , and  $b$  can be reduced to a single parameter,  $A$ , to characterize the curvature of the relationship [Manal and Buchanan, 2003]. This parameter is called the shape factor and its range of values are listed in Table 3.



**Figure 3: Representation of the modified Hill muscle model. The muscle-tendon unit consists of a muscle fiber in series with the tendon. Total muscle-tendon force ( $F$ ) passes through all components.**

#### 2.4.2 Modeling Muscle and Tendon Force

Activations and muscle tendon lengths were input into a modified Hill model to estimate muscle force [Lloyd and Besier 2003] (figure 3). The muscle-tendon unit consists of a muscle fiber in series with a tendon. The muscle fiber is comprised of three elements in parallel: an elastic element, a contractile element, and a damping element. The fiber length is  $l_m$ ,  $l_t$  is the total tendon length, and  $l_{mt}$  is the total muscle-tendon unit length. The pennation angle,  $\varphi$ , is the angle between the lines of action of the muscle fiber and the tendon and was calculated as shown in equation 5, where  $l_m^o$  is the optimal fiber length (OFL) and  $\varphi_o$  is the pennation angle at  $l_m^o$  [Scott 1991].

$$\varphi(t) = \sin^{-1} \left( \frac{l_m^o \sin \varphi_o}{l_m(t)} \right) \quad (5)$$

$$F = F^t = F_{max} * [\tilde{F}_A(\tilde{l}_m) * \tilde{F}_v(\tilde{v}_m) * a(t) + \tilde{F}_p(\tilde{l}_m) + b_m * \tilde{v}_m] \cos \varphi \quad (6)$$

Muscle-tendon force ( $F$ ) was calculated as a combination of passive and active components,

where  $F^t$  is tendon force,  $F_m$  is muscle fiber force,  $F_{max}$  is the maximum isometric muscle force,  $\tilde{l}_m$  is normalized muscle fiber length,  $\tilde{v}_m$  is normalized muscle fiber velocity,  $a(t)$  is muscle activation,  $\tilde{F}_A(\tilde{l}_m)$  represents the active force-length relationship [Gordon 1966],  $\tilde{F}_v(\tilde{v}_m)$  represents the force-velocity relationship [Hill 1938, Zajac 1989, Epstein 1998],  $\tilde{F}_p(\tilde{l}_m)$  represents the passive elastic force-length relationship [Schutte 1992], and  $b_m$  is the damping factor [Schutte 1993]. These parameters are normalized to  $F_{max}$ , optimal fiber length ( $l_m^o$ ), and maximum muscle contraction velocity ( $v_{max}$ ).

It has been previously reported that  $l_m^o$  increases with decreased muscle activation [Guimaraes 1994, Huijing 1996]. This is accounted for in the EMG-Driven Model by the following equation [Lloyd and Besier 2003]:

$$l_m^o(t) = l_m^o(\lambda(1 - a(t)) + 1) \quad (7)$$

where  $\lambda$  is the percent change in optimal fiber length.

Tendon force will only manifest when the tendon length,  $l_t$ , is greater than tendon slack length (RTL),  $l_t^s$ . Tendon force also varies with tendon strain,  $\varepsilon$  [Schutte 1993]. This is accounted for in the model via the following equations:

$$\tilde{F}_t = 0 \quad \varepsilon \leq 0 \quad (8)$$

$$\tilde{F}_t = 1480.3\varepsilon^2 \quad 0 < \varepsilon < 0.0127 \quad (9)$$

$$\tilde{F}_t = 37.5\varepsilon - 0.2375 \quad \varepsilon \geq 0.0127 \quad (10)$$

$$\varepsilon = (l_t - l_t^s)/l_t^s \quad (11)$$

### 2.4.3 Musculoskeletal Model

A generic lower limb musculoskeletal model [Delp 2007] was scaled from measurements obtained from a static standing trial for each subject and used to obtain subject-scaled muscle moment arms and muscle tendon lengths. The generic model was modified for use in this study. The following muscle-tendon actuators were modeled: BFL, BFS, SM, ST, VL, VI, VM, RF, MG, LG, SL, TA, tensor fascia latae, gracilis, and sartorius. The tensor fascia latae, gracilis, and sartorius were not used due to their small physiological cross sectional areas and relatively small impact on joint moments in the sagittal plane. This study focused only on kinetics and kinematics in the sagittal plane as the flexion-extension knee moment and the plantar flexion-dorsiflexion ankle moments are, for the most part, provided entirely by muscle forces. By focusing on the sagittal plane, we are able to isolate the relationship between muscle forces and inverse dynamics calculated joint moments. A motion file containing hip, knee, and ankle kinematics was used as an input to the musculoskeletal model and to compute time-varying muscle tendon lengths, flexion-extension knee moment arms, and plantar flexion-dorsiflexion ankle moment arms.

#### 2.4.4 Calculating Joint Moments

The muscle tendon lengths provided by the musculoskeletal model and the muscle activation were input to the muscle model, with muscle fiber lengths calculated by numerical integration of the muscle fiber velocities using a Runge-Kutta-Fehlberg algorithm (with Adams-Bashforth and isometric algorithms available if the Runge-Kutta-Fehlberg algorithm failed). These muscle fiber lengths and velocities were input to equation (6) to determine muscle forces. Muscle forces,  $F_i$ , were multiplied by their respective strength parameters,  $g_i$ , and muscle moment arms,  $r_i$ . These products, the individual muscle moments,  $M_i$ , were summed for all muscles acting at the joint to obtain net joint moments approximated by forward dynamics,  $M_j$ , as seen in equation (12).

$$\mathbf{M}_j = \sum_i \mathbf{M}_i = \sum_i [g_i * F_i * r_i] \quad (12)$$

#### 2.4.5 Optimization and Model Parameters

The EMG-Driven Model uses a Simulated Annealing numerical optimization in order to minimize the square difference between joint moments calculated from inverse dynamics and the joint moments estimated from the EMG-Driven Model at both the ankle and knee joints [Kirkpatrick 1983, Higginson 2005]. This process is summarized in equation (13):

$$\min \left[ \sum_j \left\{ \sum_k (I_{j,k} - M_{j,k})^2 \right\} + P \right] \quad (13)$$

where  $P$  is the penalty (described in 2.4.7),  $I_{j,k}$  is the inverse dynamics calculated moment for joint  $j$  at data point  $k$  and  $M_{j,k}$  is the forward dynamics calculated moment for joint  $j$  at data point  $k$ . This routine is performed on both the knee and ankle joints whereas previous iterations of the EMG-Driven Model only performed optimization on one joint.

The optimization was applied to a single running trial for each subject in order to obtain a calibrated model parameter set. Muscle and activation parameters ( $n=52$ ) were allowed to vary between predetermined ranges. Optimal fiber length and tendon slack length for each muscle were initialized to an average value based on cadaver studies and were allowed to vary  $\pm 20\%$  (Table 2) [Yamaguchi, 1990]. Four strength parameters were allowed to vary between 0.5 and 2.0. These were used to scale the  $F_{max}$  of different muscle groups: dorsiflexors (TA), plantar flexors (MG, LG, SL), knee extensors (VL, VI, VM, RF), and knee flexors (BFL, BFS, SM, ST). The shape factor was allowed to vary between 0.0 and 0.12 for each muscle [Manal and Buchanan 2003]. EMD was allowed to vary between 40ms and 100ms for each muscle. Tables 2 and 3 summarize initial parameters and ranges.

Muslce	OFL	RTL
BFL	0.109	0.341
BFS	0.173	0.100
SM	0.080	0.349
ST	0.201	0.262
VI	0.087	0.136
VL	0.084	0.157
VM	0.089	0.126
RF	0.084	0.346
LG	0.064	0.385
MG	0.045	0.402
SL	0.030	0.268
TA	0.098	0.223

**Table 2: Initial values of muscle optimal fiber lengths and resting tendon lengths used for the EMG-Driven Model. All measurements are in meters.**

Parameter	Starting Value	Upper Limit	Lower Limit
EMD	40ms	100ms	40ms
OFL	Varies	120%	80%
RTL	Varies	120%	80%
C1	0.00	0.90	-0.90
C2	0.00	0.90	-0.90
A	0.01	0.12	0.01
Strength Gain	1.00	0.50	2.00

**Table 3: Variable parameters with initial values and ranges for the EMG-Driven Model. C1 and C2 are coefficients that define second order dynamics of the muscle activation model.**

#### 2.4.6 Model Predictions

The calibrated model parameter sets were used to predict muscle forces during running and cutting tasks. Comparisons between estimated joint moments joint moments calculated with inverse dynamics were performed using a squared Pearson product-moment correlation ( $R^2$ ) and normalized root mean square error,  $nRMSE$ , as is shown:

$$nRMSE = \frac{RMSE}{|M_{max} - M_{min}|} \quad (12)$$

with  $RMSE$  being root mean square error and  $M_{max}$  and  $M_{min}$  being maximum and minimum joint moments calculated by inverse dynamics, respectively. Using  $nRMSE$  is practical as it explains the error in terms of the range of calculated joint moment values. In order to make comparisons between subjects, muscle forces were normalized to their theoretical maximum value as calculated by the model, i.e. the product of  $F_{max}$  and the strength parameter. Moment data was normalized to the product of a subject's mass and height. The average values of three predicted trials each of running and cutting are presented.

#### 2.4.7 Modifications to EMG-Driven Model

The most significant modification of the EMG-Driven Model was expanding the model from one to two joints. In order to accommodate two joints, the model was edited to tune to two joint moment predictions, requiring the input of joint moments calculated with inverse dynamics for both the ankle and knee and muscle moment arms for all included muscles about the ankle and knee joints.

The numerical integration method previously used for calculating fiber lengths was a Runge-Kutta-Fehlberg method that would exit its execution if the step size were to drop below a minimum value. This was implemented in order to have a low processing time for tuning trials. However, when predicting trials, if the Runge-Kutta-Fehlberg algorithm were to exit, no prediction would be made despite acceptable model parameter values. In order to prevent this, the model was edited to use an Adams-Bashforth numerical integration method if the Runge-Kutta-Fehlberg algorithm exited, providing a more robust multistep alternative. If the Adams-Bashforth algorithm also exited, an isometric approximation was used.

An observation from initial testing of the two-joint EMG-Driven Model revealed the model was prone to selecting parameter sets that did not make physiological sense in order to provide better joint moment predictions. Specifically, the values of normalized muscle fiber lengths were much higher than observed physiological limits, taken from a conservative interpretation of limits reported by Burkholder and Lieber [2001]. To correct for this, a penalty function was added to the model, which is summarized in equation (13):

$$p_{i,k} = \begin{cases} 0.6 - l_m/l_m^o & \text{for } l_m/l_m^o < 0.6 \\ 0 & \text{for } 0.6 \leq l_m/l_m^o \leq 1.2 \\ l_m/l_m^o - 1.2 & \text{for } 1.2 < l_m/l_m^o \end{cases} \quad (13)$$

where  $p_{i,k}$  is the amount the normalized fiber length is out of bounds of physiological limits [0.6,1.2] for muscle  $i$  at data point  $k$ . These were summed over all muscles and

data points and multiplied by a constant,  $c$ , to produce the overall penalty,  $P$ , as described in equation (14):

$$P = c * \sum_i [\sum_k p_{i,k}] \quad (14)$$

The benefit of including the penalty function resulted in more physiologically accurate estimates of muscle fiber lengths; however, it was observed that  $R^2$  values of joint moment predictions decreased as a result.

## 2.5 Statistical Analysis

Due to the small number of recorded data collections in each group, no statistical analysis was performed on this data set. This study is to be considered as motivation for further exploration using the two-joint EMG-Driven Model for comparisons of estimated muscle forces between healthy, ACL-D, and ACL-R groups.

## Chapter 3

### RESULTS

#### 3.1 Joint Moment Predictions

Before examining the predicted muscle forces from the EMG-Driven Model, it is important to explore the differences in joint moment predictions given by the EMG-Driven Model among groups. Figure 4 shows estimated and calculated knee joint moments from a calibration trial and a predicted trial for a healthy subject, and figure 5 shows the same for ankle joint moments. Figures 6 and 7 display the average  $R^2$  and  $nRMSE$  values for joint moment predictions with standard error. As it's shown in the figures, there is little change between  $R^2$  and  $nRMSE$  values between the different groups and both running and cutting tasks, as the two-joint EMG-Driven Model is able to predict all with similar accuracy.

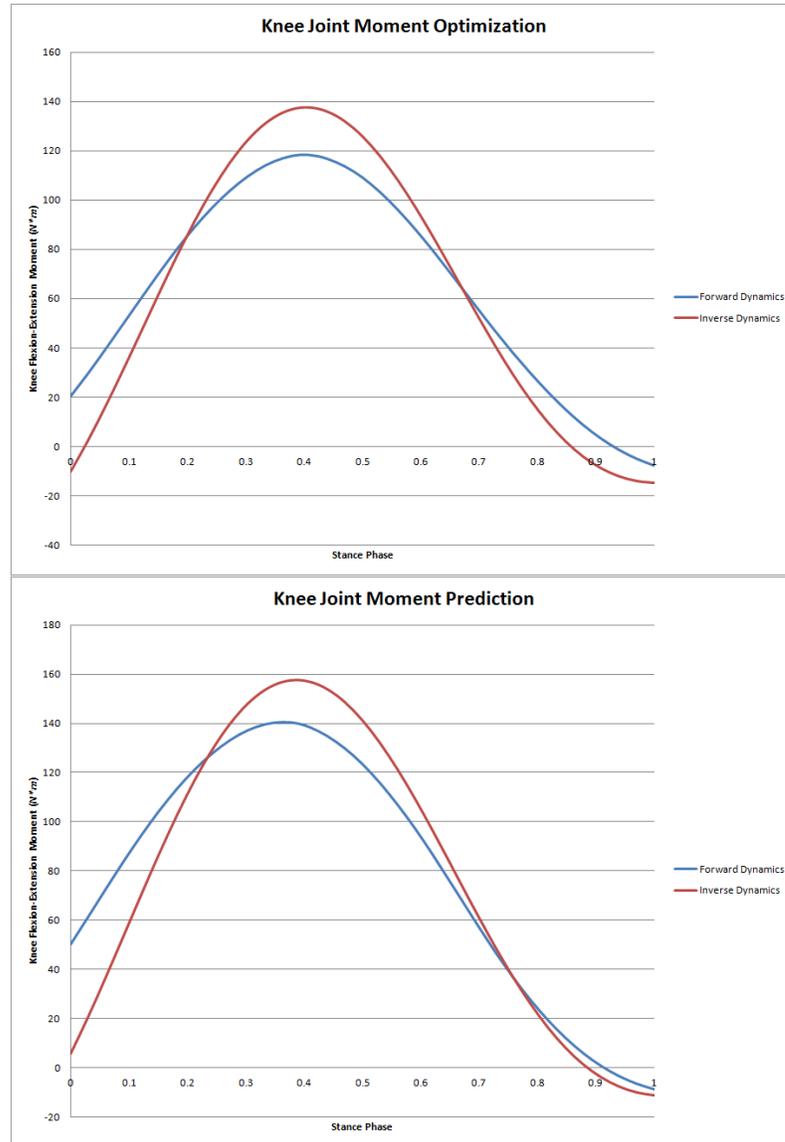
#### 3.2 Normalized Peak Joint Moments

Normalized peak flexion-extension knee joint moments for the ACL-D and ACL-R subject groups were found to be less than that of the healthy subjects, for both running and cutting tasks. These results are summarized in figure 8. For each subject group, knee moments were greater for the cutting task when compared to the running task. Figure 9 shows the effect of the average peak flexion-extension knee

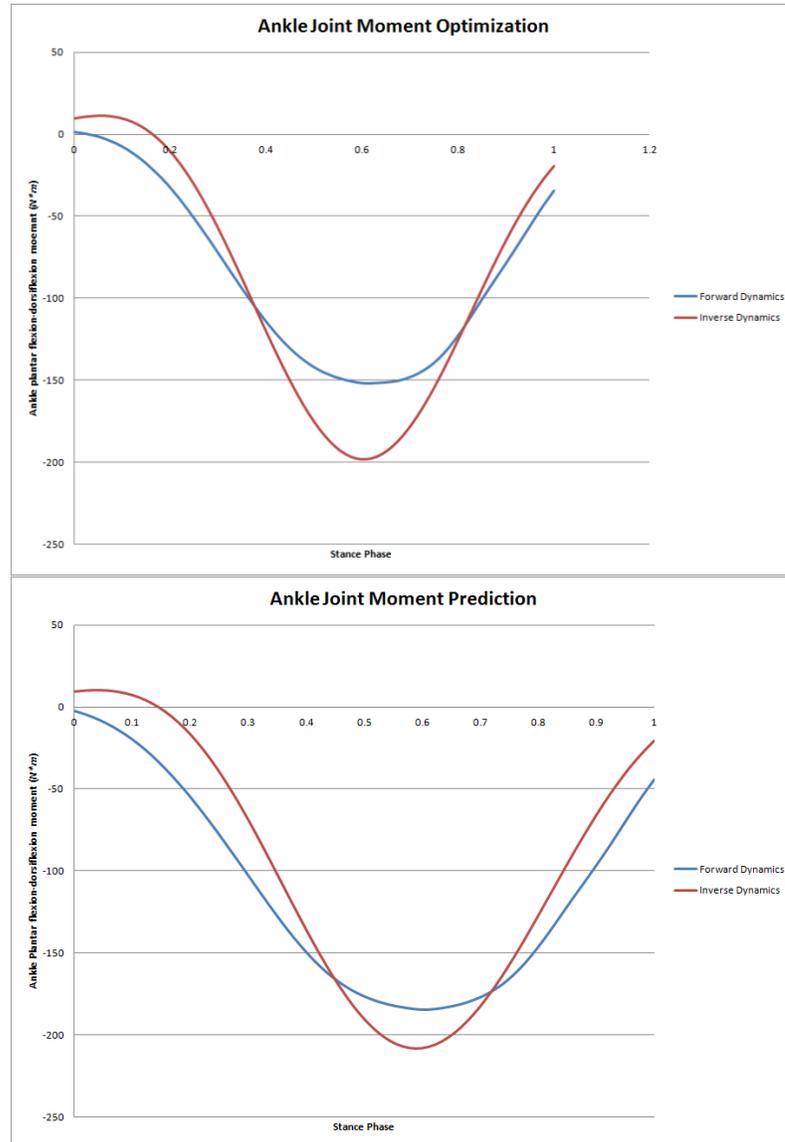
joint moment of the ACL-R group when excluding the subject who underwent a BPTB graft procedure. More on this in the discussion section.

### **3.3 Normalized Muscle Forces**

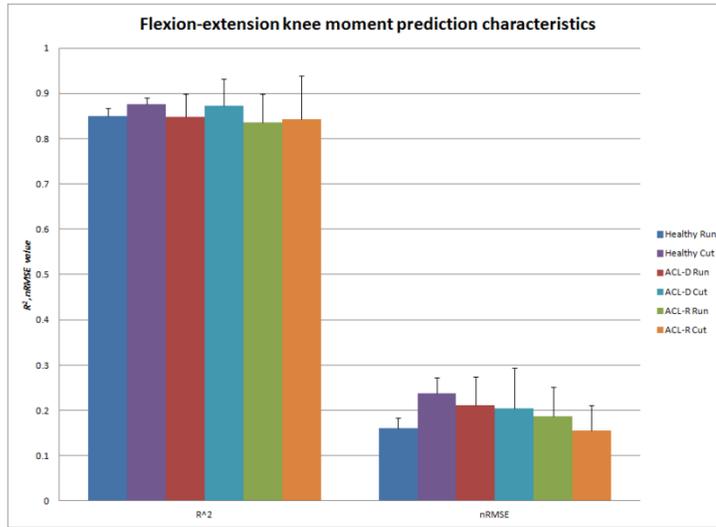
Average normalized muscle forces and standard deviations for all subject groups and tasks are depicted in figures 10, 11, and 12 for the hamstrings, quadriceps, and muscles acting at the ankle, respectively. As was shown in previous studies for healthy running [Besier 2009], the BFL, BFS, SM, and ST exert a peak force during heel strike and are mostly active during weight acceptance, the vastii exert their maximum force during peak flexion-extension moment, and the gastrocnemius muscles exert peak force late in the running cycle near toe off. Slight differences in force patterns and magnitudes between this study and the previous study can be explained partially by the two-joint approach taken by this study. The previous study focused only on forces about the knee, thus not having a true estimation of the function of the gastrocnemius muscles. This has a cascade effect on the hamstrings as well, as the model will balance contributions by both groups of knee flexors in order to match the calculated knee joint moments. Muscle patterns for the ACL-D and ACL-R subjects vary slightly, which is reflective of altered EMG patterns, figure 13 shows an example of EMG of the BFL from one representative subject in each group. The healthy subject shows a much greater magnitude of neural activation of the BFL during running. This motivates examining the difference of peak muscle forces of these different subject groups at their previously described maximum exertion in order to observe the manifestation of possible changes in muscle function.



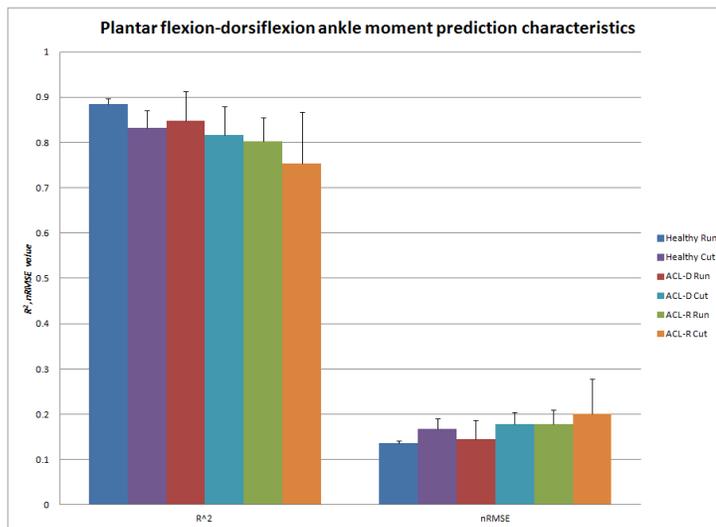
**Figure 4: Knee joint moments for a healthy subject. Top is for a running calibration trial ( $R^2 = 0.92$ ;  $nRMSE = 0.09$ ), bottom for a predicted run and cut trial ( $R^2 = 0.87$ ;  $nRMSE = 0.10$ ).**



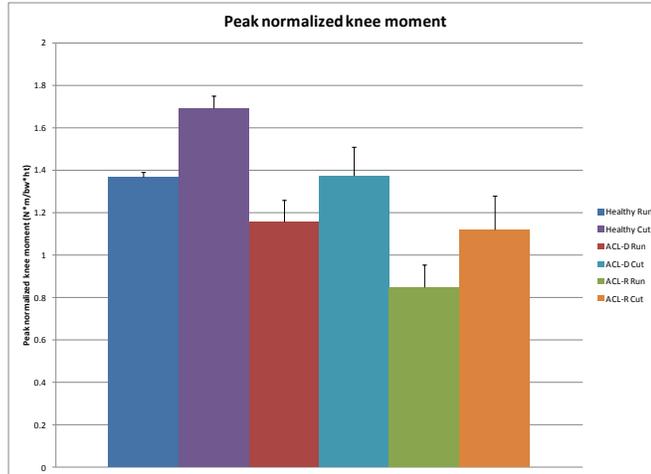
**Figure 5: Ankle joint moments for a healthy subject. Top is for a running calibration trial ( $R^2 = 0.93$ ;  $nRMSE = 0.11$ ), bottom for a predicted running trial ( $R^2 = 0.93$ ;  $nRMSE = 0.11$ ).**



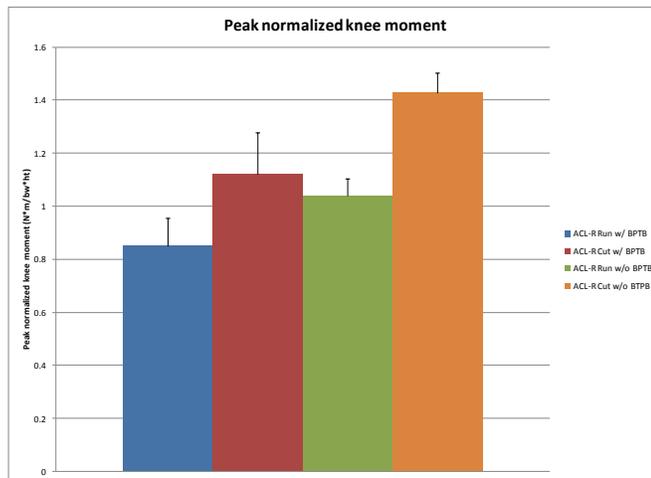
**Figure 6:  $R^2$  and  $nRMSE$  average values for flexion-extension knee joint moment predictions, positive error bars show standard error.**



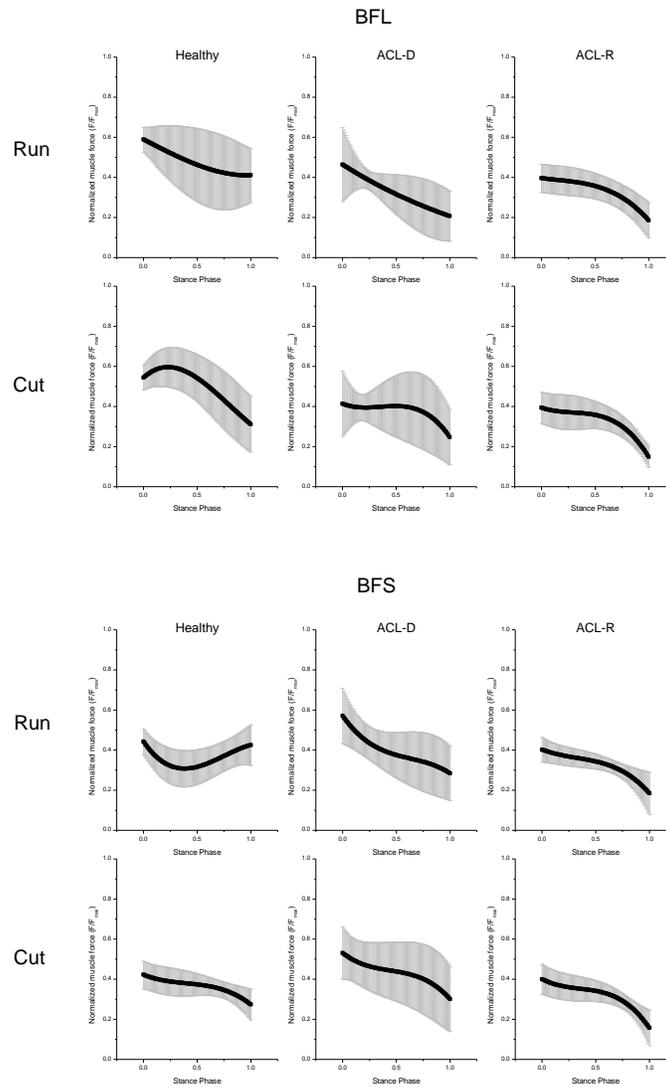
**Figure 7:  $R^2$  and  $nRMSE$  average values for plantar flexion-dorsiflexion ankle joint moment predictions, positive error bars show standard error.**



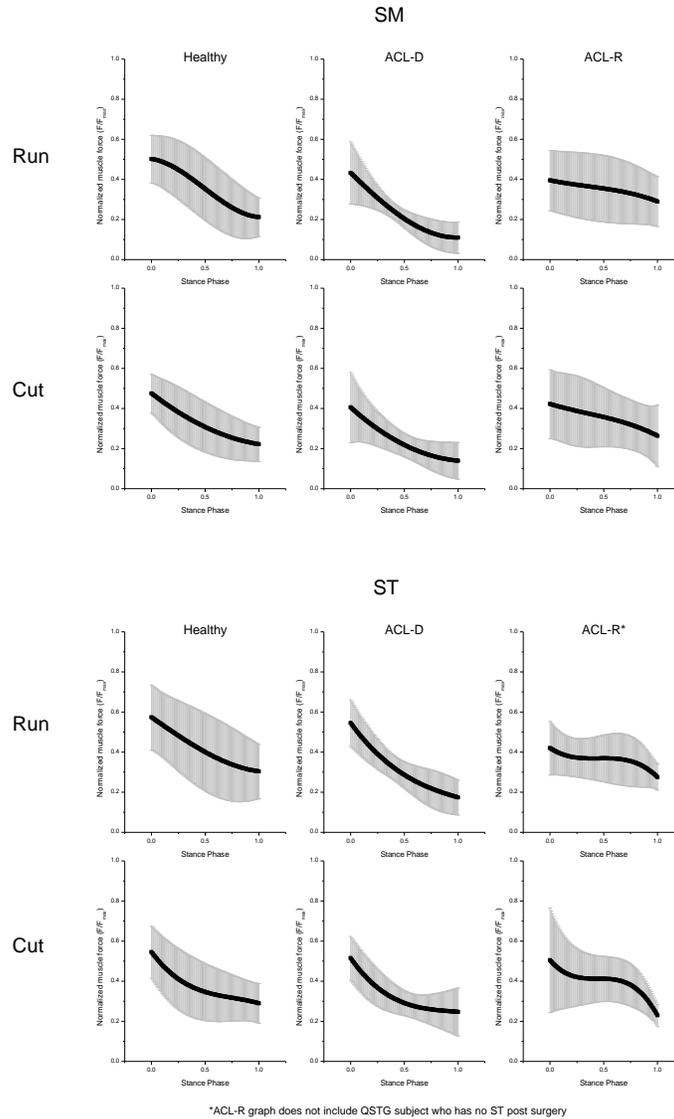
**Figure 8: Average peak normalized knee joint moment during running and cutting. Moments are normalized to the product of a subject's height and mass, positive error bars represent standard error.**



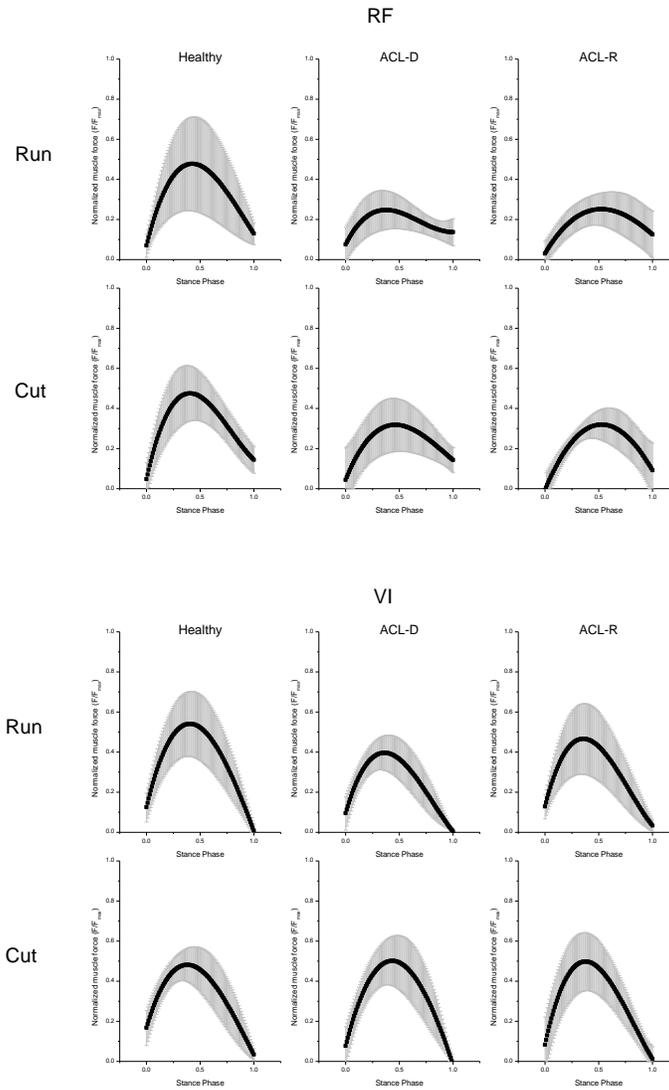
**Figure 9: Average peak normalized knee joint moment for ACL-R subjects during running and cutting with and without the BPTB subject. Moments are normalized to the product of a subject's height and mass, positive error bars represent standard error.**



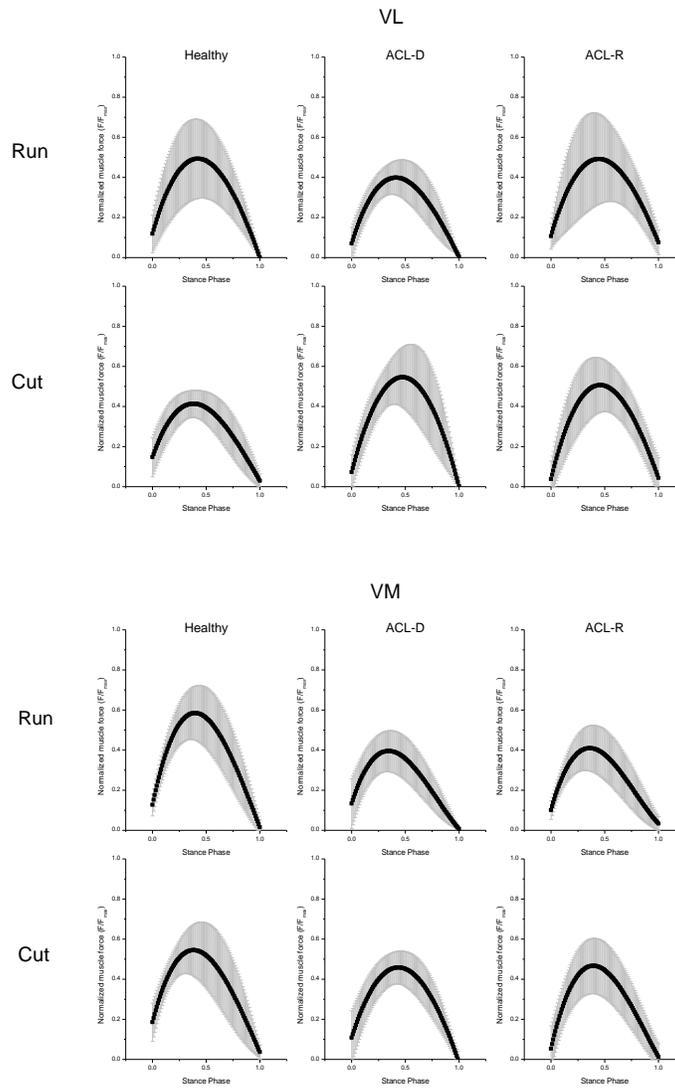
**Figure 10a: Normalized muscle force distribution average and standard deviations for the BFL and BFS. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



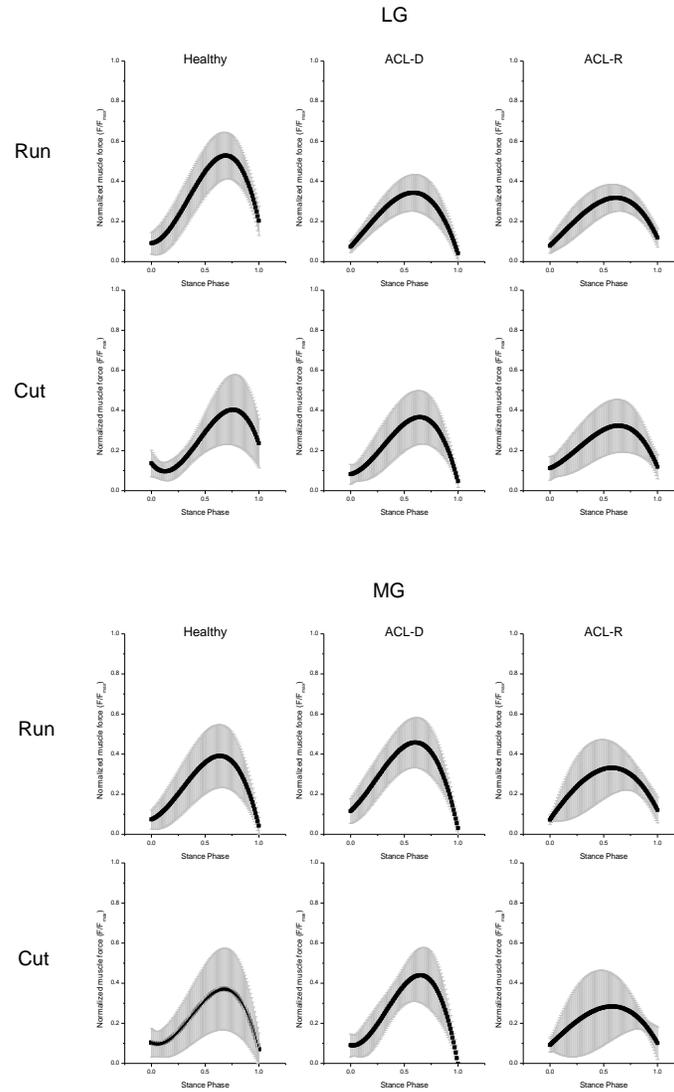
**Figure 10b: Normalized muscle force distribution average and standard deviations for the SM and ST. Plots were interpolated to 101 points using a cubic b-spline method before being averaged. ACL-R shows large standard deviations due to one QSTG subject having no ST in the modeling.**



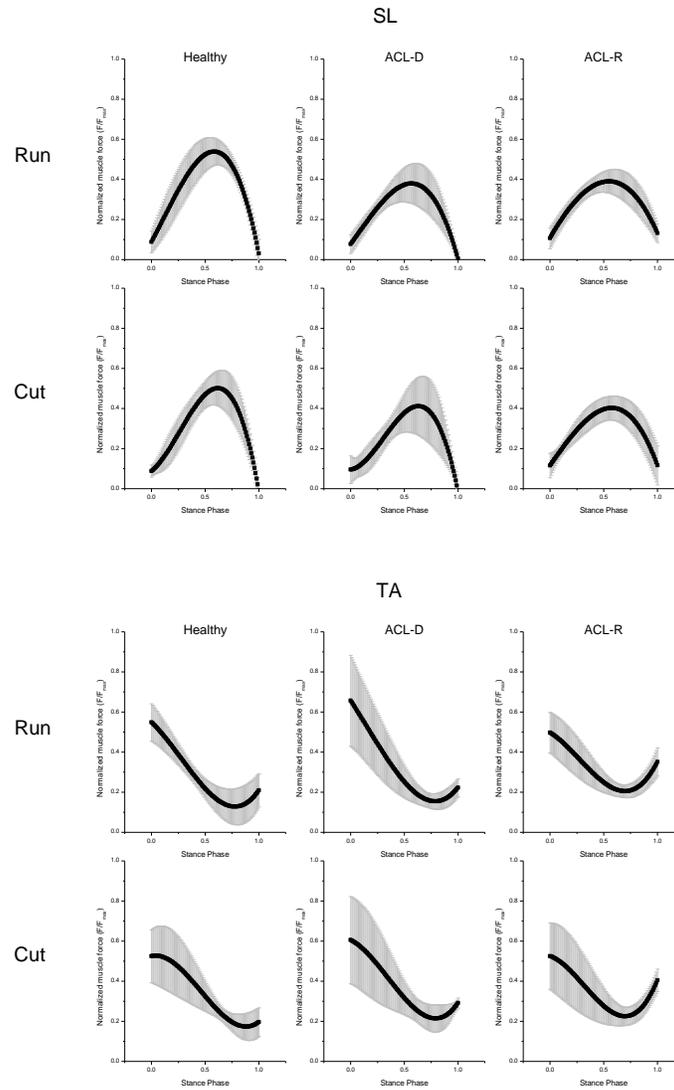
**Figure 11a: Normalized muscle force distribution average and standard deviations for the RF and VI. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



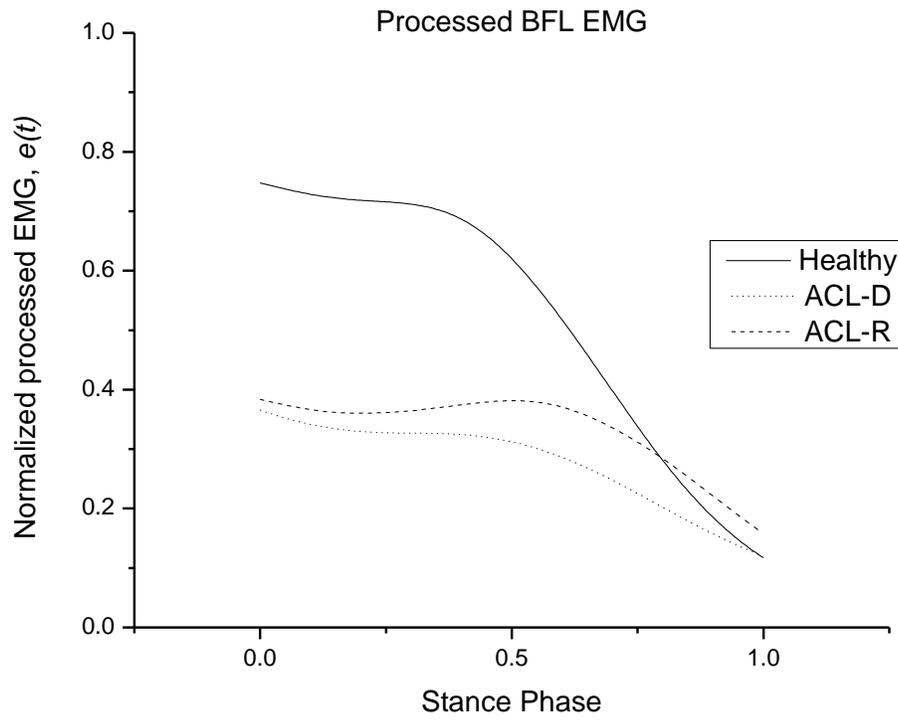
**Figure 11b: Normalized muscle force distribution average and standard deviations for the VL and VM. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



**Figure 12a: Normalized muscle force distribution average and standard deviations for the LG and MG. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



**Figure 12b: Normalized muscle force distribution average and standard deviations for the SL and TA. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



**Figure 13: Processed EMG of the BFL for running trials of healthy, ACL-D, and ACL-R subjects.**

### **3.2.1 Average Peak Forces for Muscle Groups**

Before exploring differences between peak individual muscle forces, it's useful to look at average peak forces for each muscle group in order to determine if there are any overall trends. Figures 14 and 15 show the average peak vastii and hamstrings forces, respectively (peak hamstrings force assumed as the average force during weight acceptance). The average peak vastii force drops considerably for the ACL-D group during running, while during cutting there is little difference from other groups. The average peak hamstrings force is highest with healthy subjects and trends downward as time after ACL rupture increases.

### **3.2.2 Peak Forces for Individual Muscles**

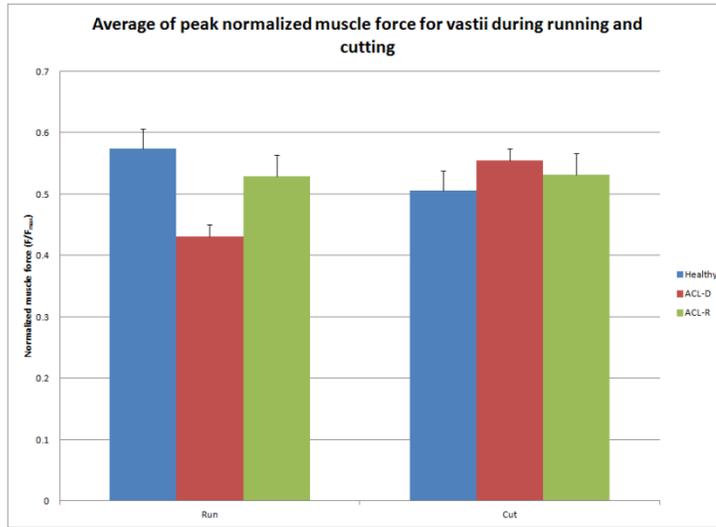
Looking at the change in individual peak forces across groups will give some insight into the mechanisms that cause the behind the previous observations. The average peak forces for the vastii and hamstrings are shown in figures 16 and 17, respectively. The decrease in peak vastii force for ACL-D subjects during running is consistent across the board for the VL, VI, and VM. The VM is the only vastii to exhibit a decrease in force for both tasks for the ACL-D and ACL-R subject groups. Peak forces for the BFL for the ACL-D and ACL-R groups are decreased compared to the healthy group, while peak forces of the BFS for the ACL-D group are increased over the healthy and ACL-R groups. The difference in peak force of the SM and ST between groups is not as pronounced as the differences found in the peak forces of the BFL and BFS, as figure 17 shows error bars overlapping for the different subject

groups. More data would be needed in order to see if any trends form with a broader subject pool.

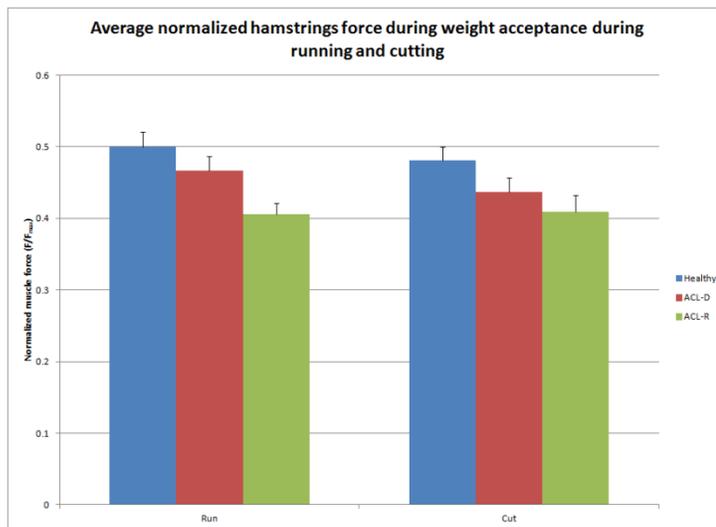
### **3.3 QSTG Subject, Pre-Surgery and Post-Surgery**

Figures 18 and 19 show the average normalized muscle forces of the hamstrings and quadriceps, respectively, for the QSTG subject both pre-surgery and post-surgery. The vastii show the same decrease in quadriceps force exhibited for the ACL-D subjects during running. This is further illustrated by figure 20, which compares peak vastii forces pre-surgery and post-surgery.

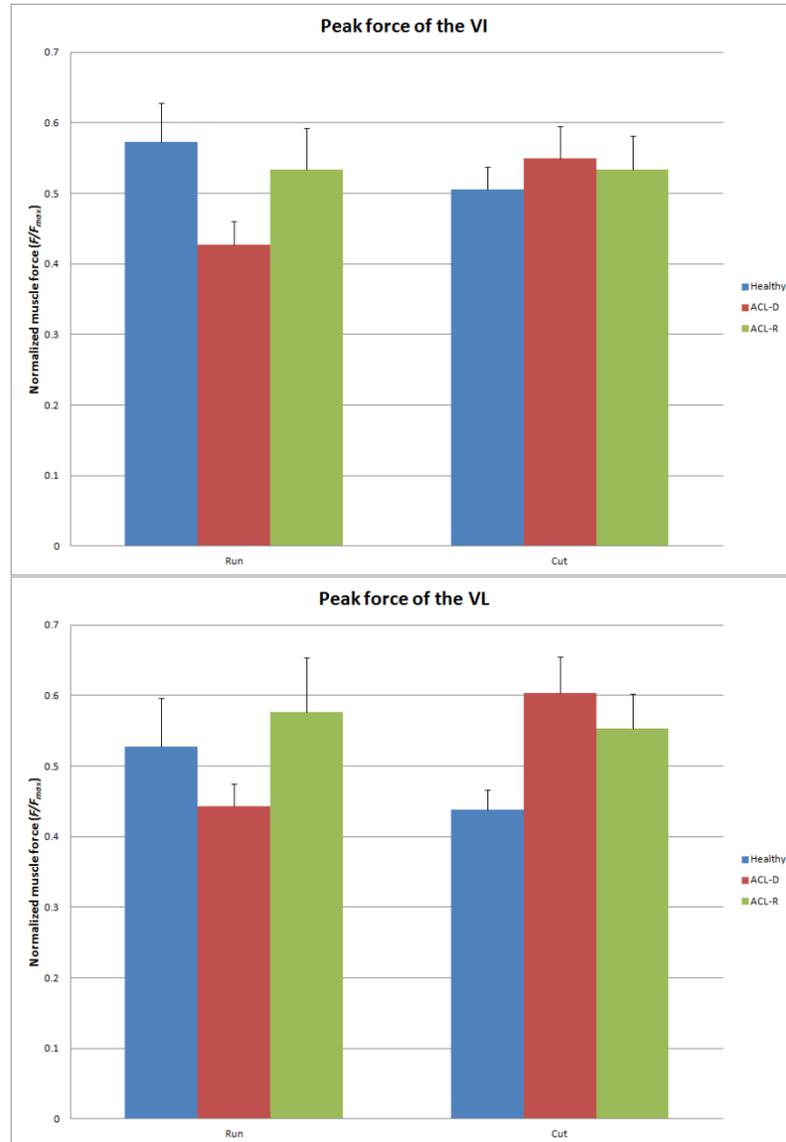
The force pattern of the BFL pre-surgery deviates from what was seen in figure 6a where the maximum force is occurring around 50% of stance phase and not during weight acceptance. Post-surgery the force pattern of the BFL matches those found in figure 11a. The force of the SM increases greatly post-surgery compared to pre-surgery, while the ST is sacrificed during surgery so no comparisons can be made. Figure 21 compares the average normalized force during weight acceptance further highlighting the previously mentioned differences.



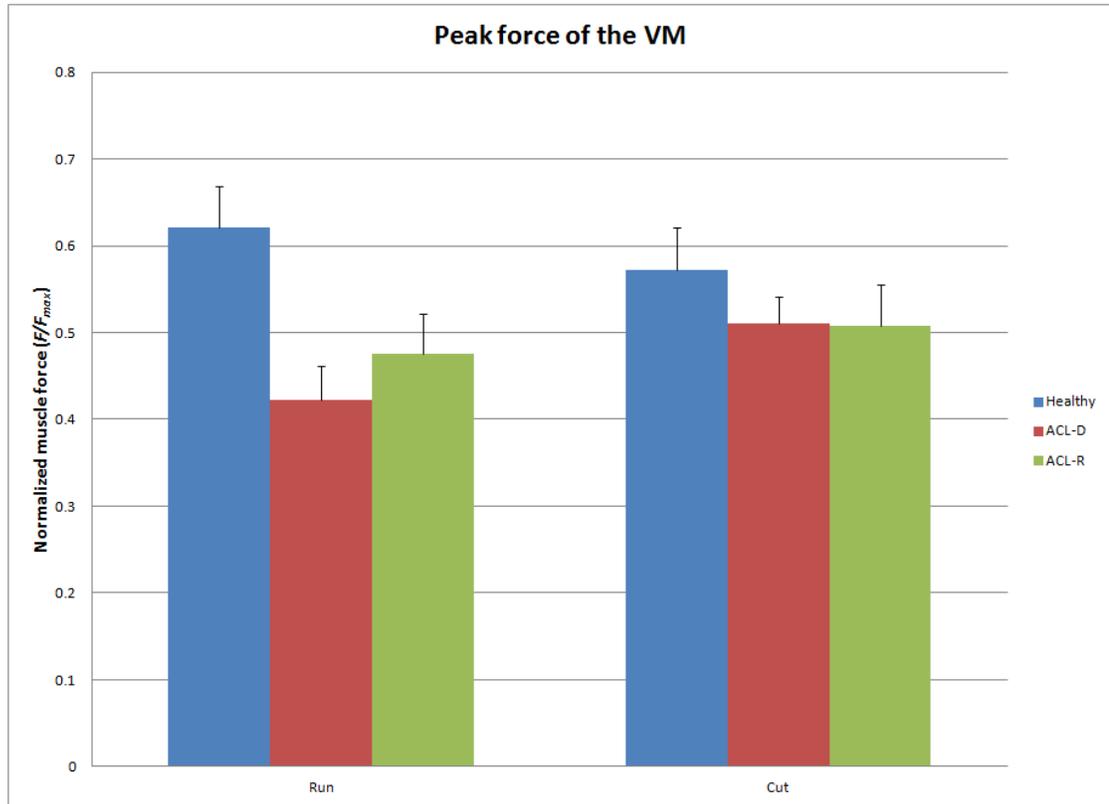
**Figure 14: Average peak normalized vastii force for all groups, positive error bars represent standard error.**



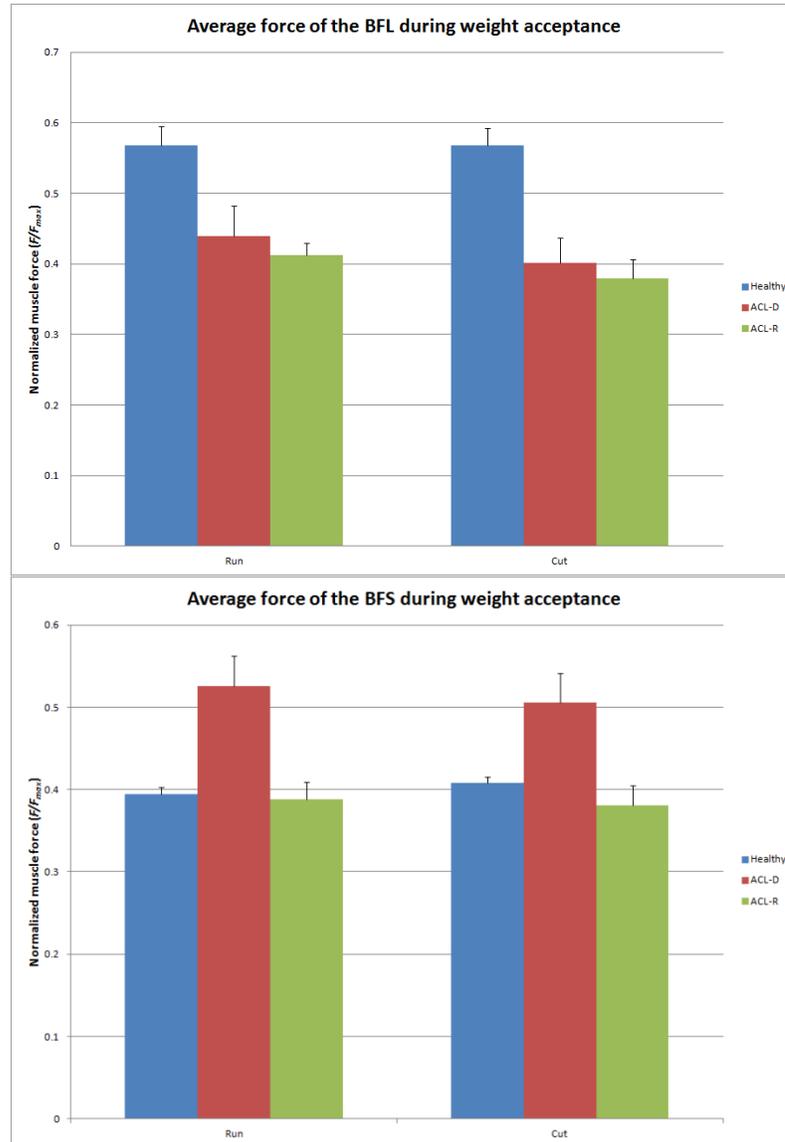
**Figure 15: Average normalized hamstrings force during weight acceptance, positive error bars represent standard error.**



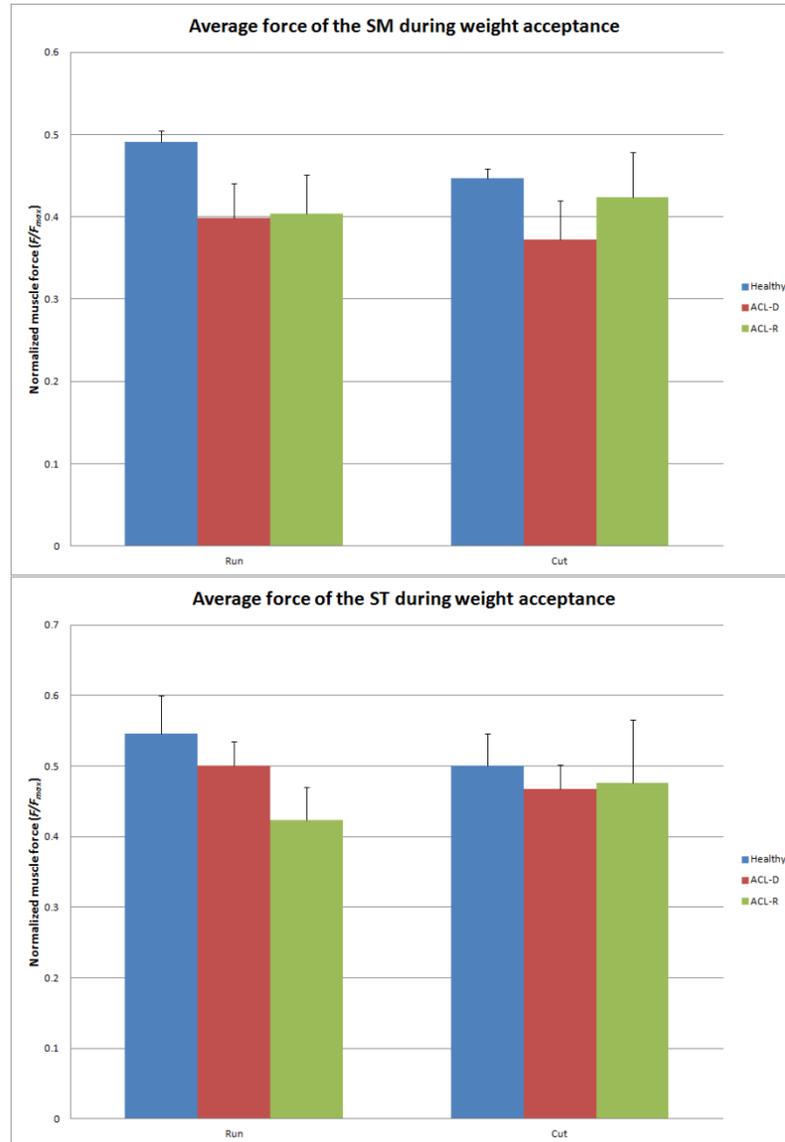
**Figure 16a: Peak normalized VI and VL force, positive error bars represent standard error.**



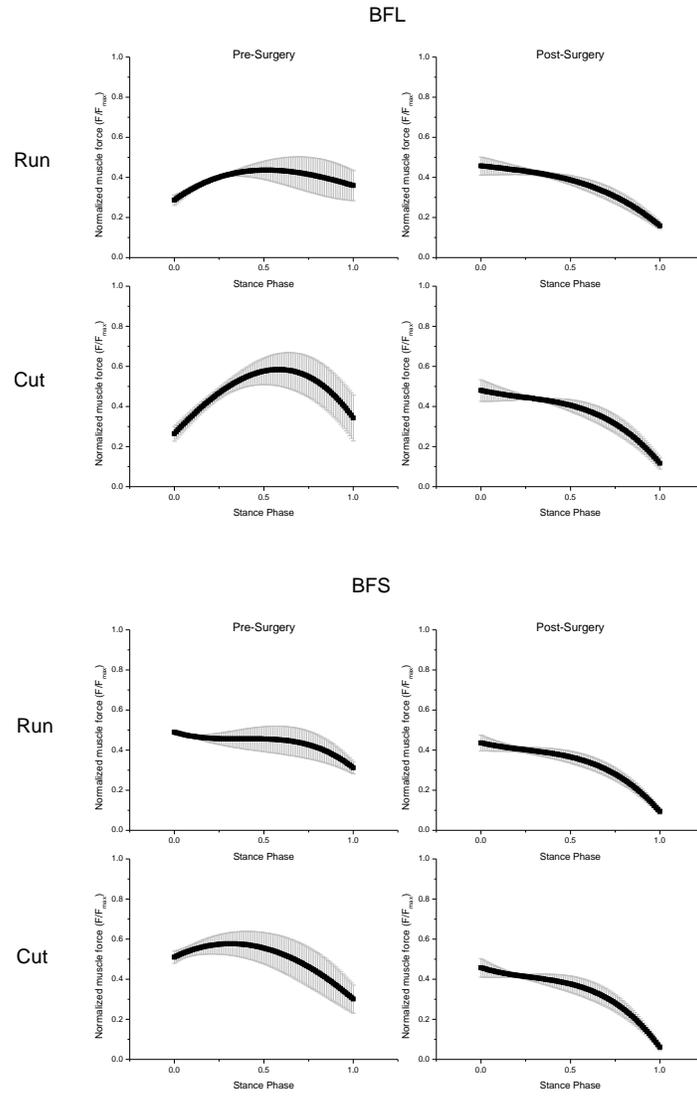
**Figure 16b: Peak normalized VM force, positive error bars represent standard error.**



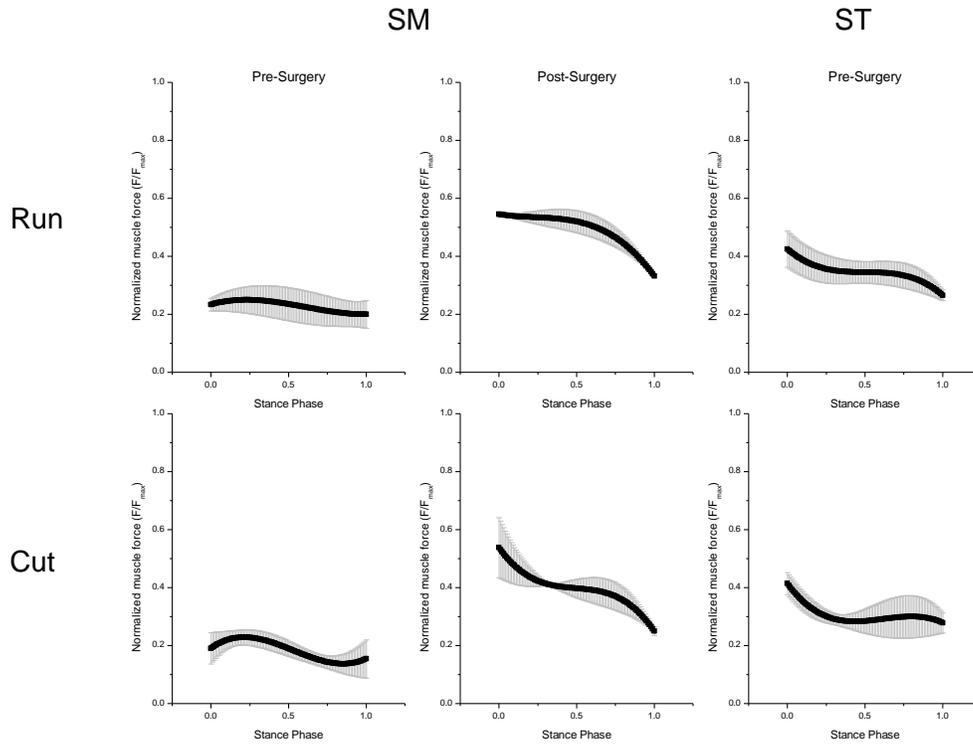
**Figure 17a: Average normalized BFL and BFS force during weight acceptance, positive error bars represent standard error.**



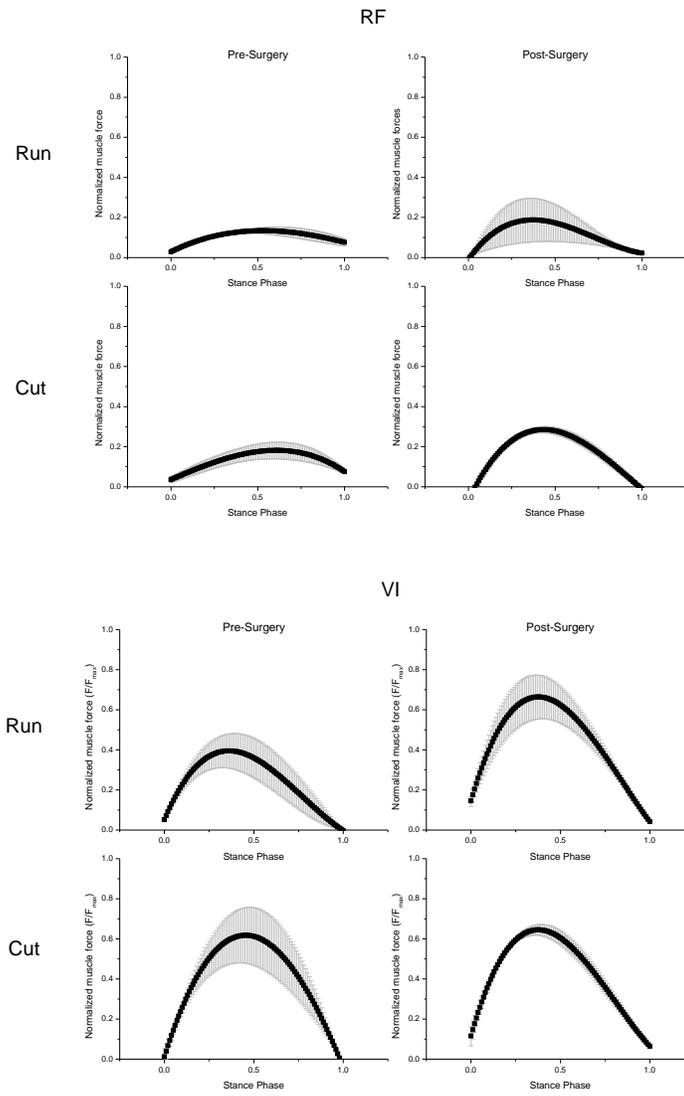
**Figure 17b: Average normalized SM and ST force during weight acceptance, positive error bars represent standard error. ST values for ACL-R subjects do not include subject who underwent QSTG procedure.**



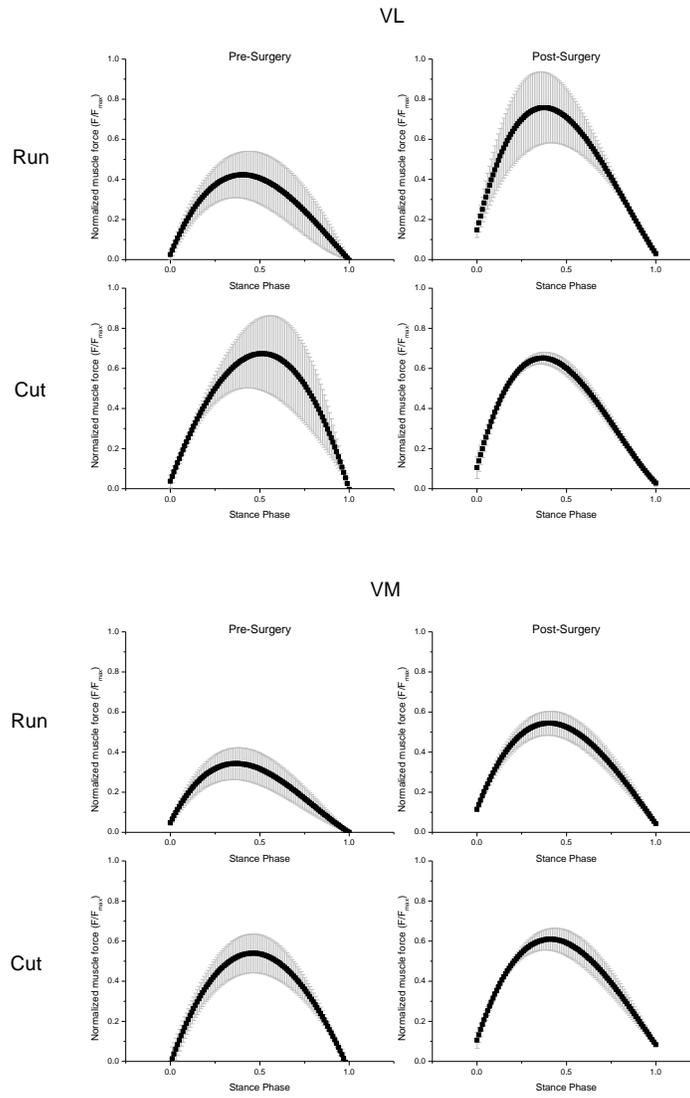
**Figure 18a: Normalized muscle force distribution average and standard deviations for the BFL and BFS of the QSTG subject pre-surgery and post-surgery. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



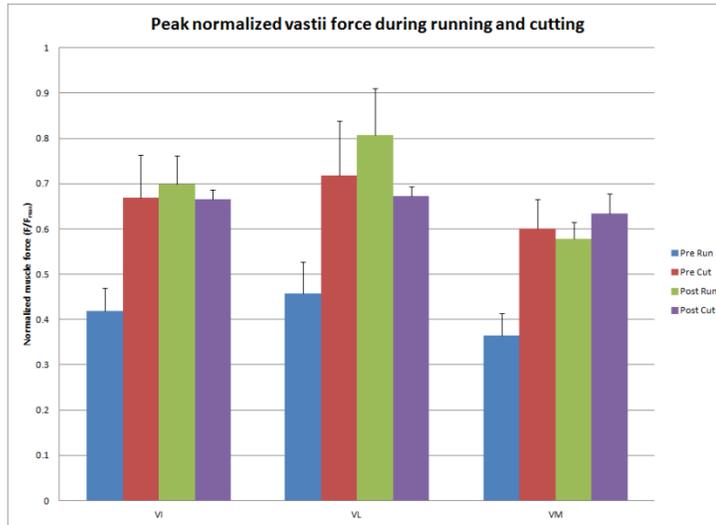
**Figure 18b: Normalized muscle force distribution average and standard deviations for the SM and ST of the QSTG subject pre-surgery and post-surgery. There are no data for the ST post-surgery since the tendon of the ST was sacrificed during surgery. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



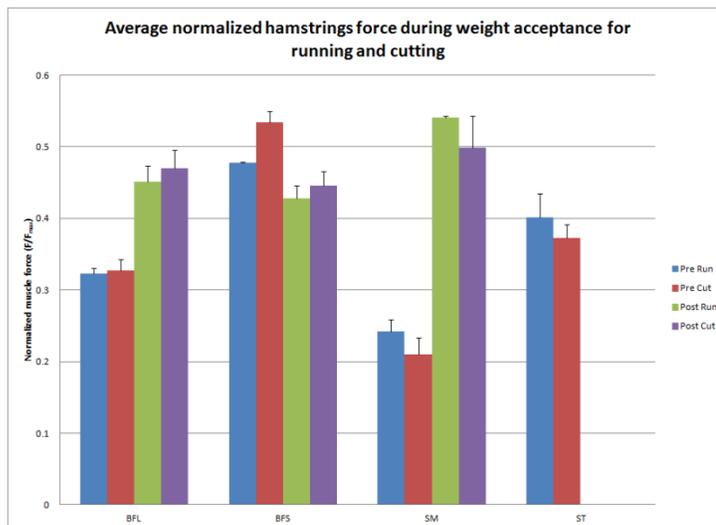
**Figure 19a: Normalized muscle force distribution average and standard deviations for the RF and VI of the QSTG subject pre-surgery and post-surgery. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



**Figure 19b: Normalized muscle force distribution average and standard deviations for the VL and VM of the QSTG subject pre-surgery and post-surgery. Plots were interpolated to 101 points using a cubic b-spline method before being averaged.**



**Figure 20: Average peak normalized vastii forces for subject with pre and post QSTG procedure data, positive error bars represent standard error.**



**Figure 21: Average normalized hamstrings forces during weight acceptance for subject with pre and post QSTG procedure data, positive error bars represent standard error.**

## Chapter 4

### DISCUSSION

#### 4.1 Predicted Joint Moments

The predicted joint moments did not show differences in  $R^2$  and  $nRMSE$  between subject groups and task types for both the ankle and knee joints, as shown in figures 3 and 4. Average values of  $R^2$  and  $nRMSE$  were typically greater than 0.8 and lower than 0.2, respectively. These values confirm that the model was also accurate in matching the forward dynamics calculated joint moments to the joint moment calculated by inverse dynamics. The implementation of constraints on normalized muscle fiber lengths, as is described in section 2.4.7, had a negative effect on  $R^2$  and  $nRMSE$  values; however, the small decrease in accuracy of the predicted joint moment is outweighed by gaining higher confidence in the model's ability to produce results that make physiological sense. This had a direct effect on muscle forces and, as a result, joint moments since muscle fiber length is a determining factor for muscle force [Gordon 1966]. A greater range of values for the EMG-Driven Model's strength parameter would also allow for a greater range of muscle forces, which, in turn, could yield better joint moment predictions. This study chose to be consistent with past studies concerning the limits of the strength parameter [Buchanan 2004, 2005; Lloyd and Besier 2003; Besier 2009; Shao 2009].

The similar accuracy of joint moment predictions for all subject groups illustrates the two-joint EMG-Driven Model's potential applications. Interesting future

applications include subjects with OA, stroke subjects, subjects that have had total knee arthroplasty, subjects suffering achilles tendon rupture, subjects post microfracture surgery, etc.

#### **4.2 Peak Flexion-Extension Knee Joint Moment**

Figure 8 displays average peak knee joint moments normalized to the product of a subject's height and mass. Healthy subjects have a larger peak normalized knee joint moment than the ACL-D and ACL-R subjects, which further supporting quadriceps weakness for ACL-D subjects. The low peak normalized knee joint moments for ACL-R subjects can be partially explained by the BPTB subject, who is a forefoot striking runner. Figure 9 compares the average peak normalized knee joint moment for the ACL-R group when including and excluding the BPTB subject. Forefoot strikers begin each stride while running by landing with their forefoot, as opposed to a more traditional heel strike. To illustrate this point, figure 22 compares the ankle joint angle during running between the BPTB subject and a healthy subject. The altered kinematics cause increased activation to the soleus and gastrocnemius muscles at foot strike and result in a decreased knee joint moment when compared to heel strikers, as can be seen in figure 23 [Williams 2000]. Previous studies have observed that ACL-R subjects can exhibit quadriceps weakness after successful surgery [Konishi 2002], which is supported by this data even after accounting for the BTPB subject.

Quadriceps weakness exhibited by the ACL-D patients is possibly in response to preventing excessive pain, a lack of confidence in the injured knee joint,

or abnormal joint proprioception, as will be discussed in section 4.4. It is more difficult to explain the decreased knee joint moment of the ACL-R group. Even excluding the BPTB subject, the ACL-R group's average peak normalized knee joint moment is decreased from the average peak normalized knee joint moment for healthy subjects. Since subjects for this study were, on average, observed 28.5 weeks after ACL reconstructive surgery there could still be lingering knee joint pain during running that is inhibiting quadriceps force production. Confidence in the knee joint should be corrected for after ACL reconstruction surgery, ruling out that possibility for reduced joint moment. Konishi et al [2002] attributed quadriceps weakness to abnormal proprioception in the knee, as the afferent receptors in the ACL are not reconstructed during surgery, thus not providing necessary feedback to the central nervous system. Of the three possible explanations of quadriceps weakness, this is the most likely; however, Konishi observed this result for ACL-R subjects, while our results do not show quadriceps weakness for the ACL-R subjects across the board; only the VM exhibits quadriceps weakness. The decreased average peak normalized knee joint moment for ACL-R subjects cannot be described by peak forces of the vastii and hamstrings alone. Since it's seen that the vastii peak forces return to healthy values and the BFL actually has a decrease in average force during weight acceptance, weakening the knee flexor mechanism, one would assume an increased knee joint moment given those factors.

Until this point the focus of this study has been on the vastii and the hamstrings since they are the primary knee extensors and knee flexors, respectively. It was sensible to exclude other muscles used in the model for this analysis since the

MG, LG, and SL are primarily plantar flexors, the TA is primarily a dorsiflexor, and the RF is a hip flexor and does not contribute as much to the knee extension moment during running compared to the vastii; yet the LG and MG act as knee flexors and RF acts as a knee extensor since these muscles span two joints. From figure 26 it is seen that the peak force of the RF for ACL-D and ACL-R subjects during both tasks is lower compared to the values for healthy subjects. This decrease in RF force could be the determining factor in observed decrease in knee joint moment in the ACL-R group.

### **4.3 Estimated Hamstrings Forces**

Altered muscle force patterns were observed for the hamstrings for the ACL-D and ACL-R groups, notably that the average force for the BFL during weight acceptance for healthy subjects was found to be much greater than that of the ACL-D and ACL-R subjects while the average force for the BFS during weight acceptance for the ACL-D subjects was greater than that of the healthy and ACL-R subjects. Both the BFL and BFS are lateral hamstrings, thus it would seem that the increase in force of the BFS and the decrease in force of the BFL are related in order to maintain the same overall force or moment contribution from the lateral hamstrings. In the case of the ACL-R subjects, we see that the BFS peak force is similar to healthy levels, indicating that the BFS may have increased force in ACL-D subjects to counteract ATT and medial rotation of the tibia that would normally be resisted by the ACL.

It is possible that following ACL rupture, the lateral hamstrings alter their force patterns where the BFS increases its force to account for the absence of the ACL

in the knee during dynamic tasks. The force of the BFL is then decreased to provide the necessary knee joint moment contribution that is normally expected of the lateral hamstrings. The troubling result is that the peak force in the BFL does not equal healthy levels for the ACL-R subjects. It is possible that this altered force pattern coincides with muscle atrophy of the BFL, which can have detrimental effects on loading patterns of the knee joint, as it is speculated that altered force patterns post ACL rupture contribute to the development of OA. Observing subjects at longer time intervals after ACL reconstructive surgery would indicate if the loss in the force of the BFL is a long term effect of ACL rupture, as this would be a key indicator of the onset of OA.

#### **4.4 Estimated Quadriceps Forces**

The data shows decreased peak quadriceps force for ACL-D subjects in the running case for all vastii. The VM and the RF do exhibit a decrease in peak force for both tasks for the ACL-D and ACL-R subject groups. This result can be related to previous findings of quadriceps weakness in ACL-D and ACL-R subjects [Konishi 2002, Lewek 2002, Williams 2005a]. A mechanism is possibly employed by the ACL-D subjects in order to prevent large knee joint moments, but cannot be controlled by complex dynamic tasks such as running and cutting. The causes for this mechanism seem threefold: pain avoidance, lack of confidence in the injured leg, and decreased proprioception. Pain in the knee joint is a common symptom of ACL injury [Daniel 1994], especially for dynamic tasks which require increased joint moments, range of motion, muscle forces, and ligament loading, all of which contribute to exacerbating

joint pain. The lack of confidence in the knee joint is due to the greater laxity and decreased proprioception in the joint after ACL rupture. Reduced confidence is reflected in decreased kinetic patterns, as was seen in both decreased quadriceps force and decreased flexion-extension knee joint moments for the ACL-D subjects during running. Since cutting is a more kinetically demanding task, it is possible that the same strategies used during straight running are not physically possible to be used.

As discussed in section 4.2, previous studies have investigated the relationship between decreased proprioception in the knee and quadriceps weakness for ACL-R subjects [Konishi 2002]. Undoubtedly, proprioception in the knee is affected by ACL rupture as the ACL contains many afferent mechanoreceptors. The result for ACL-R subjects is easily translated to ACL-D subjects as they both lack the original ligament and its mechanoreceptors, providing the best explanation for quadriceps weakness since it is only observed during running and not cutting, as other mechanoreceptors of the knee could provide the necessary afferent response regarding quadriceps force production during cutting only. It is difficult to determine if the quadriceps weakness and decreased joint proprioception are linked or merely two different responses to ACL rupture. The results of this study help support the observation of quadriceps weakness for ACL-D subjects; however, further research is required to investigate the potential link between quadriceps weakness and reduced joint proprioception.

#### **4.5 QSTG Subject, Pre-Surgery and Post-Surgery**

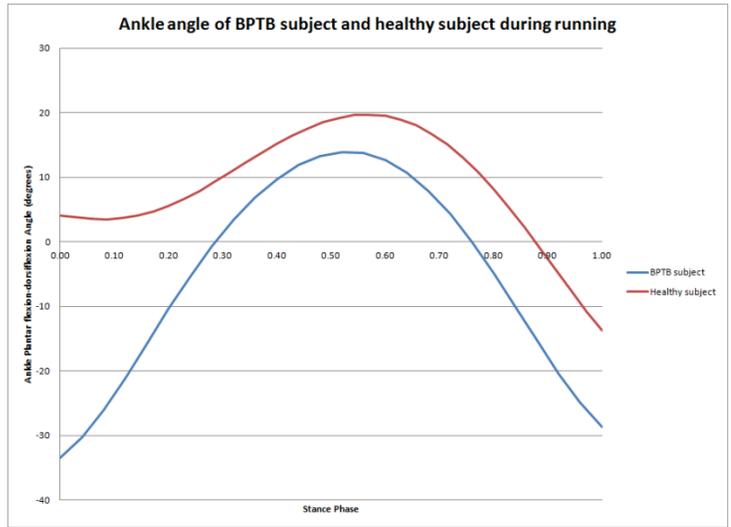
The results found by comparing muscle force patterns and peak normalized forces for the QSTG subject differ from the overall trends in two significant ways. First, the force pattern of the BFL changes drastically, as the BFL exerts its maximum force at about 50% of stance phase and not during weight acceptance. Looking at the processed EMG of the BFL it appears this change is the result of altered muscle activation patterns, as magnitude of the BFL EMG and timing of the peak EMG both change post-surgery, as shown in figure 24. This also explains the result in figure 21 of the average BFL force during weight acceptance increasing post-surgery. Figure 19a shows the average peak force does not change pre-surgery and post-surgery. The second difference from the overall results exhibited by the QSTG subject is the large increase in force of the SM post-surgery. This is a direct response to the harvesting of the ST tendon for the QSTG procedure, as the SM increases its force production in order to provide the same level of force and moment contribution from the medial hamstrings as a whole.

Figure 25 compares the average peak muscle force for all muscles acting at the knee during running for the QSTG subject pre-surgery and post-surgery. Note that for 5 of the 8 muscles present in this graph, the peak force increases post-surgery. That graph assumes the peak hamstrings force occurs during weight acceptance, which was found not to be the case for the BFL pre-surgery; the ratio is actually 4 of 8. Only one muscle, the ST, had a decrease in peak force post-surgery, due to its tendon being harvested. Peak muscle force is a good indicator of the magnitude of muscle force patterns. This implies an increase in joint contact force at the knee during running

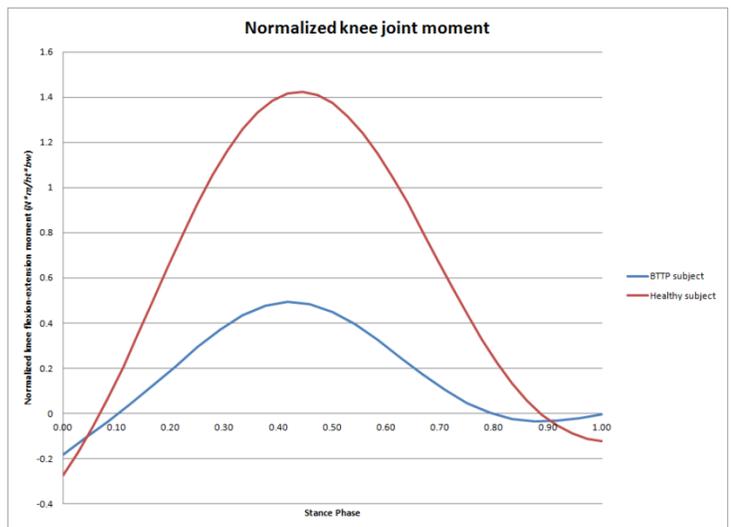
post-surgery for the QSTG subject. This result could be unique to this particular subject or a reflection of an overall trend. Regardless, it requires further investigation.

#### **4.6 Other Observations**

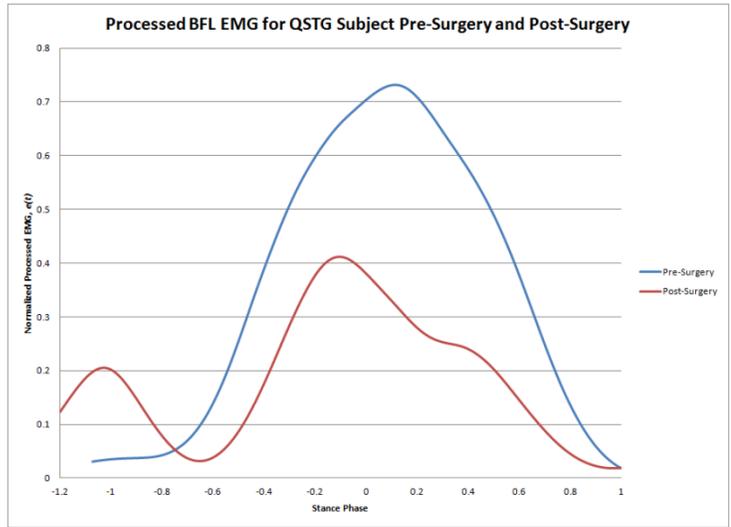
Other observations from figure 26 include decreased normalized peak forces for the LG and SL for the ACL-D and ACL-R subjects compared to the healthy subjects, and an increase in normalized peak force for the MG and the TA for the ACL-D subjects when compared to healthy and ACL-R subjects. These changes reflect the chain reaction of altered muscle force patterns in the human body that are required in order to maintain homeostasis after traumatic injury and surgical repair.



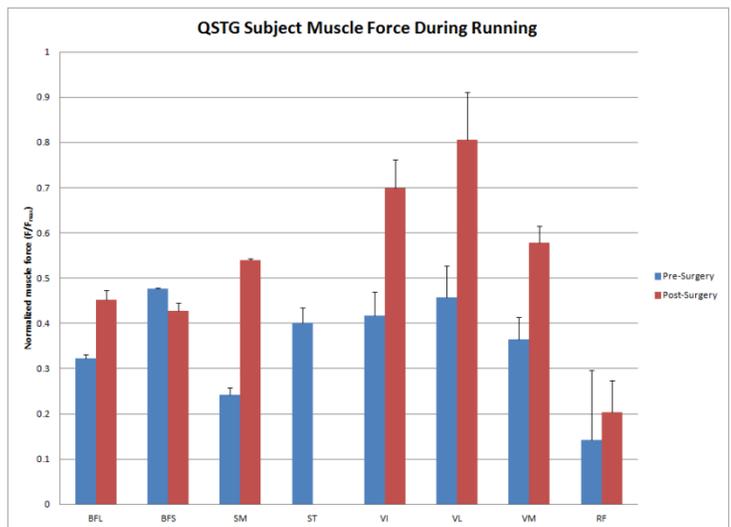
**Figure 22: Ankle joint angle during running for the BPTB subject (forefoot striker) and a healthy subject (heel striker).**



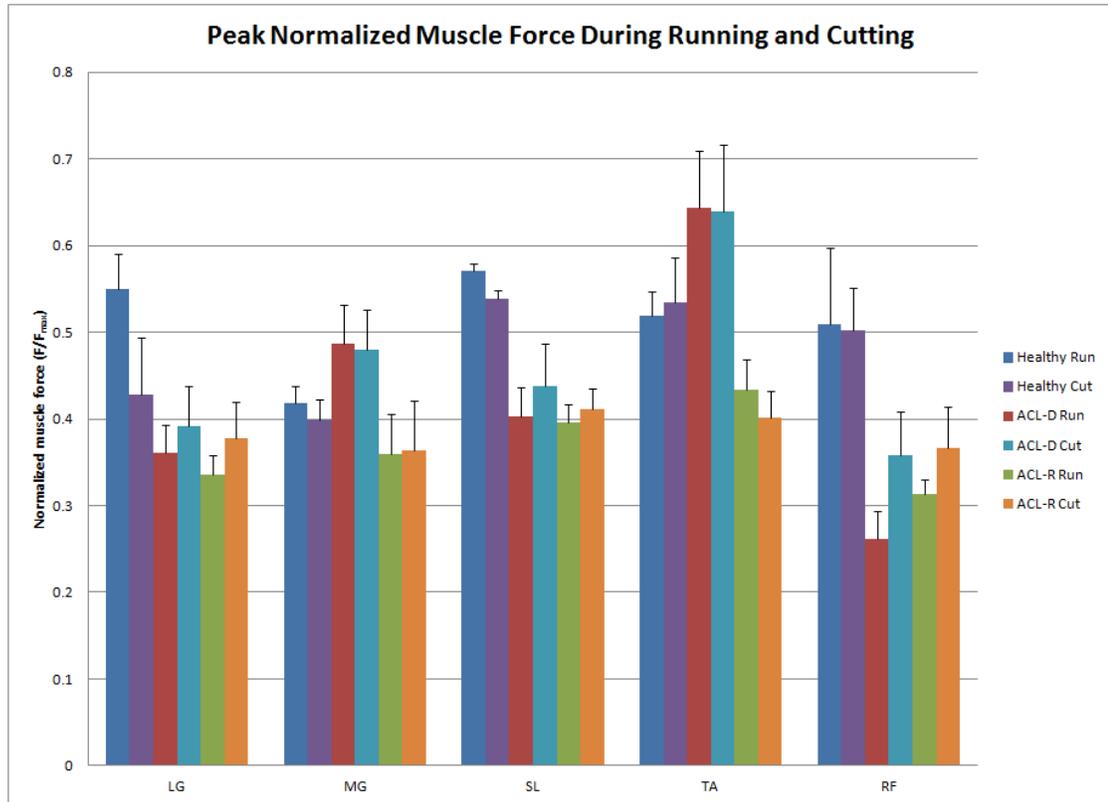
**Figure 23: Knee joint moment normalized to the product of height and mass for the BPTP subject (forefoot striker) and a healthy subject (heel striker).**



**Figure 24: EMG of the BFL during running and cutting trials for the QSTG subject pre-surgery and post-surgery.**



**Figure 25: Average peak normalized muscle force during running for the QSTG subject pre-surgery and post-surgery. Graph only shows hamstrings and vastii.**



**Figure 26: Average peak normalized muscle force for plantar flexors, dorsiflexors, and the rectus femoris during running and cutting for healthy, ACL-D, and ACL-R groups. Positive error bars represent standard error.**

## **Chapter 5**

### **LIMITATIONS**

#### **5.1 Exclusion of Hip Joint Moment**

The version of the EMG-Driven Model used for this study included the ankle in order to account for the bi-articulating nature of the LG and MG; however, it ignored the bi-articulating nature of the hamstrings and RF. The hamstrings are both knee flexors and hip extensors and the RF, as previously stated, is a hip flexor in addition to being a knee extensor. A three joint model would be able to account for the bi-articulating nature of these muscles, but would also require EMG collection of the gluteus maximus, psoas major, tensor fasciae latae, and other muscles crossing the hip joint that contribute to its flexion-extension joint moment. This provides additional complexity to the feasibility of data collections where more channels of EMG need to be monitored during the collection process.

#### **5.2 Frontal and Transverse Plane Joint Moments**

This study only considered joint moments in the sagittal plane for its analysis. Including joint moments in the frontal and transverse planes in the model would give a truer estimate of muscle forces yet would increase the complexity of the model greatly, requiring adding ligaments to the model that resist motions in the

frontal and transverse planes. The computations required to match three joint moments for each joint used in the model would affect the run time of optimization trials.

### **5.3 Model Parameters**

Pennation angle at optimal fiber length and percent change of optimal fiber length was not allowed to vary for the model and physiological cross sectional area (PSCA) was not accounted for. The pennation angle at optimal fiber length and percent change were based on literature reported values [Yamaguchi 1990] and other parameters were allowed to vary such as the strength parameter,  $l_m^o$ , and  $l_t^s$  to account for not including PCSA. The strength parameter range could have been expanded to obtain better joint moment predictions; however, it was determined best to be consistent with previous studies [Buchanan 2004, 2005; Lloyd and Besier 2003; Besier 2009; Shao 2009].

### **5.4 ACL-D Copers**

There was an inherent difficulty in finding ACL-D subjects who felt comfortable running and cutting. Typically, dynamic tasks produce great knee joint pain for ACL-D subjects. Additionally, dynamic tasks allow for a greater chance of giving way episodes. It is possible that the ACL-D subjects that were recruited for this study were all copers, and exhibit different muscle force patterns than non-copers. For this study, it would be fruitless to designate between copers and non-copers due to the small number of subjects recruited for this study. Further studies should designate

which subjects are copers and non-copers, exploring differences between the estimated muscle forces of these groups, if they exist.

## **Chapter 6**

### **CONCLUSION**

In summary, we have presented an extension of the EMG-Driven Model to two joints which was used to predict joint moments and estimate muscle forces for healthy, ACL-D, and ACL-R subjects. Post calibration, this model is able to accurately predict joint moments of the ankle and knee during dynamic tasks. Altered muscle force patterns were observed in the ACL-D and ACL-R groups. Altered lateral hamstrings force patterns, decrease in BFL force and increase in BFS force, for ACL-D subjects are thought to prevent anterior tibial translation in the ACL-D subjects. This changed muscle force strategy could have adverse long term effects. For ACL-R subjects, the BFL force is similar to the decreased force of ACL-D subjects, indicating possible muscle atrophy, while the BFS force is similar to the value for healthy subjects. Decreased flexion-extension knee joint moments and vastii forces were observed for ACL-D subjects during running, indicating knee joint pain, a lack of confidence in the injured knee joint, and decreased joint proprioception. One subject had data collected pre and post a QSTG surgery and was found to have increased BFL, SM, VI, VL, and VM forces during running post-surgery, indicating an increase in joint contact forces. This result was not observed during cutting and further investigation is required. It is evident that the two-joint EMG-Driven Model has useful application outside of ACL-D and ACL-R subjects and can be applied to other subject

populations (OA, stroke, total knee arthroplasty, etc) in order to gain insight to potentially altered muscle force patterns due to the results of other traumas.

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## APPENDIX A - SUBJECT CONSENT FORM



RESEARCH OFFICE

210 Hulihan Hall  
University of Delaware  
Newark, Delaware 19716-1551  
Ph: 302/831-2136  
Fax: 302/831-2828

DATE: October 26, 2011

TO: Toran MacLeod, PT, PhD  
FROM: University of Delaware IRB

STUDY TITLE: [142589-3] The Muscle and Tendon Morphology after Reconstruction of the Anterior Cruciate Ligament with an Autologous Semitendinosus-Gracilis Graft: A Five Year Follow-up

SUBMISSION TYPE: Continuing Review/Progress Report

ACTION: APPROVED  
APPROVAL DATE: October 26, 2011  
EXPIRATION DATE: November 23, 2012  
REVIEW TYPE: Expedited Review

REVIEW CATEGORY: Expedited review category # 4

Thank you for your submission of Continuing Review/Progress Report materials for this research study. The University of Delaware IRB has APPROVED your submission. This approval is based on an appropriate risk/benefit ratio and a study design wherein the risks have been minimized. All research must be conducted in accordance with this approved submission.

This submission has received Expedited Review based on the applicable federal regulation.

Please remember that informed consent is a process beginning with a description of the study and insurance of participant understanding followed by a signed consent form. Informed consent must continue throughout the study via a dialogue between the researcher and research participant. Federal regulations require each participant receive a copy of the signed consent document.

Please note that any revision to previously approved materials must be approved by this office prior to initiation. Please use the appropriate revision forms for this procedure.

All SERIOUS and UNEXPECTED adverse events must be reported to this office. Please use the appropriate adverse event forms for this procedure. All sponsor reporting requirements should also be followed.

Please report all NON-COMPLIANCE issues or COMPLAINTS regarding this study to this office.

Please note that all research records must be retained for a minimum of three years.

Based on the risks, this project requires Continuing Review by this office on an annual basis. Please use the appropriate renewal forms for this procedure.

If you have any questions, please contact Jody-Lynn Berg at (302) 831-1119 or [jlberg@udel.edu](mailto:jlberg@udel.edu). Please include your study title and reference number in all correspondence with this office.

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

--- Consent Form 1: MRI ---

### PURPOSE/DESCRIPTION

In this research project, we are studying how people who have complete anterior cruciate ligament (ACL) injuries stabilize their knees before and after reconstructive surgery.

You have been referred to us by your orthopaedic surgeon as having a complete ACL rupture, in which case you have already been determined to be eligible to participate in this study, or you are in the control group. Sixty-nine subjects (age 14-50) will be involved in this study. Testing will be performed, two times, approximately six months apart.

The experiments require that we obtain Magnetic Resonance Imaging (MRI) scans of your knee MRI images will be taken while you remain still. You will lie on your back in the MRI machine. Typical imaging times range from 3 - 7 minutes. This test should take approximately 1 hour.

### CONFIDENTIALITY

Your participation in this study is voluntarily and you can withdraw at any time without penalty. You should not participate if you are currently pregnant, have had previous surgery to your other knee, or have any other nerve or musculoskeletal disorder such as cerebral palsy, multiple sclerosis, or arthritis. The findings of our studies will remain confidential and personal information will not be released without your consent. Data obtained from this study will be recorded on a computer and archived for a period of up to fifteen years. You are free to withdraw at any time. Your participation in this study will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

### RISKS AND BENEFITS

There are no benefits to you for participation in this study. Risks to subjects are very low. All procedures involve standard imaging practices. MRI does not involve exposure to radiation.

Claustrophobia, often experienced by patients while in the imager, is not a problem for this study because the subject's head remains outside the confines of the inner bore of the imager. In the event of physical injury as a direct result of these research procedures, you will receive first aid. If you require additional medical treatment, you will be responsible for the costs.

### FINANCIAL CONSIDERATIONS

To compensate for your time and travel expenses, you will be reimbursed \$50 for participating in this research study.

CONTACTS

Further information regarding this study may be obtained from the project director, Dr. Thomas Buchanan, at telephone number (302) 831-2410. Other questions about your rights as a research subject can be directed to Chair, Human Subject Review Board Office of the Research Office, (302) 831-2137.

Subject's initials: \_\_\_\_\_

Page 1 of 2

SUBJECT ASSURANCES

Participation in this research study is voluntary. Your participation in this study (or refusal to participate in this study) will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

*I have had the procedures explained, have had all my questions answered. I agree to participate in the research study described above.*

Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
Parent or Guardian Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
(if subject is under 18 years of age)  
Witness Signature: \_\_\_\_\_ Date: \_\_\_\_\_

Page 2 of 2

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

### - Consent Form 2: Knee Extensor Strength-

#### PURPOSE/DESCRIPTION

In this research project, we are studying how people who have complete anterior cruciate ligament, (ACL) injuries stabilize their knees before and after reconstructive surgery-

You have been referred to us by your orthopaedic surgeon as having a complete ACL rupture, in-which case you have already been determined to be eligible to participate in this study, or you are in the control group. Sixty-nine subjects will be involved in this study. Testing will be performed two times, approximately six months apart.

In this particular study, we are examining how knee strength differs with different treatments. A force measuring device will be carefully fixed to your leg just above your ankle. Your leg will be positioned with your knee bent at a right angle. Two self-adhesive electrodes, 3" x 5", will be placed over your thigh muscles. You will be instructed to contract your thigh muscles as hard as you can. During the contraction, a very brief (10 millisecond) burst of electrical stimulation will be delivered by an electrical stimulator. This will tingle a bit. We will measure your joint strength before and during the electrical stimulation. These tests should take approximately 1 hour.

#### CONFIDENTIALITY

Your participation in this study is voluntarily and you can withdraw at any time without penalty. You should not participate if you are currently pregnant, have had previous surgery to your other knee, or have any other nerve or musculoskeletal disorder such as cerebral palsy, multiple sclerosis, or arthritis. The findings of our studies will remain confidential and personal information will not be released without your consent. Data obtained from this study will be recorded on a computer and archived for a period of up to fifteen years. You are free to withdraw at any time. Your participation in this study will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

#### RISKS AND BENEFITS

The risks to the subjects are very low. The electrical stimulator delivers an electrical current that should not cause any damage to your muscles. The electrodes also pose little risk. There may be some minor irritation of the skin around the site of the electrode following the experiment. This is most likely due to the mild adhesive and electrolytes in the electrode. Your muscles may be a little sore after the maximal contractions, but this should not have lingering effects.

While there is no benefit to you for participating in this study, the chance of learning how to best treat people with knee ligament injuries in the future is high. Research projects such as this are the only ways to ascertain how the body naturally functions.

In the event of physical injury as a direct result of these research procedures, you

will receive first aid. If you require additional medical treatment, you will be responsible for the costs.

Subject's initials: \_\_\_\_\_

Page 1 of 2

#### FINANCIAL CONSIDERATIONS

To compensate for your time and travel expenses, you will be reimbursed \$50 for participating in this research study.

#### CONTACTS

Further information regarding this study may be obtained from the project director, Dr. Thomas Buchanan, at telephone number (302) 831-2410. Other questions about your rights as a research subject can be directed to Chair, Human Subject Review Board Office of the Research Office, (302) 831-2137.

#### SUBJECT ASSURANCES

Participation in this research study is voluntary. Your participation in this study (or refusal to participate in this study) will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

*I have had the procedures explained, have had all my questions answered. I agree to participate in the research study described above.*

Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
Parent or Guardian Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
(if subject is under 18 years of age)  
Witness Signature: \_\_\_\_\_ Date: \_\_\_\_\_

Page 2 of 2

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

-Consent Form 3: EMG-

### PURPOSE/DESCRIPTION

In this research project; we are studying how people who have complete anterior cruciate ligament (ACL) injuries stabilize their knees before and after reconstructive surgery. You have been referred to us by your orthopaedic surgeon as having a complete ACL rupture, in which case you have already been determined to be eligible to participate in this study, or you are in the control group. Sixty-nine subjects will be involved in this study. Testing will be performed two times, approximately six months apart.

For this research on knee muscle coordination, the experiments require that we record the electrical activity from leg muscles during controlled contractions while seated and standing. For the seated task, you will be securely seated in a chair with your knee flexed at a specific flexion angle and a force measuring device will be carefully fixed to your leg just above your ankle. For the standing task, you will stand with bare feet on two force measuring devices (one for each foot).

During both the seated and standing tasks you will be instructed to push your leg in different directions (forward, backward, right and left) while given visual feedback on a computer monitor about how much force you are producing and in which direction. We will ask you to try to produce force in specific directions by displaying targets on the monitor and you will be asked to match those targets and maintain that force for less than a second. You will then be given a short rest before another target appears.

Muscle activity will be recorded by the use of surface electrodes and/or fine wire electrodes inserted inside your muscle. Surface electrodes will be taped to your skin. For recordings inside the muscle, fine needles will be used to introduce wires into your muscles. These wires will be used to record the electrical activity of muscle fibers allowing us to make precise measurements of the signals being sent to your muscles. While there is some discomfort during passage of the needles, this usually passes quickly, and no local anesthesia is required. At the end of the experiment, all electrodes will be removed. These tests should take approximately 3 1/2 hours.

### CONFIDENTIALITY

Your participation in this study is voluntarily and you can withdraw at any time without penalty. You should not participate if you are currently pregnant, have had previous surgery to your other knee, or have any other nerve or musculoskeletal disorder such as cerebral palsy, multiple sclerosis, or arthritis. The findings of our studies will remain confidential and personal information will not be released without your consent. Data obtained from this study will be recorded on a computer and archived for a period of up to fifteen years. It may be used for additional research studies in the future. You are free to withdraw at any time. Your participation in this study will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

### RISKS AND BENEFITS

Although the use of recordings with wire electrodes is a standard clinical procedure for

diagnosis of neural and muscular problems, there are a number of risks involved- There is some discomfort: you will feel a brief prick to your skin as if receiving a shot with a very small needle. There is also a slight risk of bleeding and of damage to nerves and vessels, and a possibility of delayed infection. Infection can be identified as a small area of redness around the site. If it persists for more than 24 hours, a physician should be consulted- These various risks are minimal when the recordings are practiced by an experienced electromyographer, as will be done here.

Subject's initials: \_\_\_\_\_

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Surface EMG recordings pose little risk. There may be some minor irritation of the skin around the site of the electrode following the experiment. This is most likely due to the mild adhesive and electrolytes in the electrode. These procedures are entirely experimental and they are not intended to provide any specific medical diagnosis or treatment. It is hoped that such studies will eventually provide considerable help in our understanding of how to best treat people with knee ligament injuries, but no immediate benefit may result.

In the event of physical injury as a direct result of these research procedures, you will receive first aid. If you require additional medical treatment, you will be responsible for the cost.

#### FINANCIAL CONSIDERATIONS

To compensate for your time and travel expenses, you will be reimbursed \$50 for participating in this research study.

#### CONTACTS

Further information regarding this study may be obtained from the project director, Dr. Thomas Buchanan, at telephone number (302) 831-2410. Other questions about your rights as a research subject can be directed to Chair, Human Subject Review Board Office of the Research Office, (302) 831-2137.

#### SUBJECT ASSURANCES

Participation in this research study is voluntary. Your participation in this study (or refusal to participate in this study) will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

*I have had the procedures explained, have had all my questions answered. I agree to participate in the research study described above.*

Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
Parent or Guardian Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
(if subject is under 18 years of age)  
Witness Signature: \_\_\_\_\_ Date: \_\_\_\_\_

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

-Consent Form 4: Gait Analysis-

### PURPOSE/DESCRIPTION

In this research project, we are studying how people who have complete anterior cruciate ligament (ACL) injuries stabilize their knees before and after reconstructive surgery.

You have been referred to us by your orthopaedic surgeon as having a complete ACL rupture, in which case you have already been determined to be eligible to participate in this study, or you are in the control group. Sixty-nine subjects will be involved in this study. Testing will be performed two times, approximately six months apart.

For this research on knee muscle coordination, the experiments require that we record the electrical activity from leg muscles during different types of movements. You will be asked to wear your own running shoes and run at a consistent speed down a runway in our lab- As you reach the end of the runway, you will be asked to perform one of two tasks: a straight run or a sidestep to approximately 45.

While you are running, we will record your limb positions using markers taped to your skin or clothing. We will also record the muscle activity (EMG) by the use of surface electrodes and/or fine wire electrodes inserted inside your muscle. Surface electrodes will be taped to your skin. For recordings inside the muscle, fine needles will be used to introduce wires into your muscles. These wires will be used to record the electrical activity of muscle fibers allowing us to make precise measurements of the signals being sent to your muscles. While there is some discomfort during passage of the needles, this usually passes quickly, and no local anesthesia is required. At the end of the experiment, all electrodes will be removed. These tests should take approximately 2 hours. Testing will be performed twice, shortly before your surgery and six months after your surgery.

### CONFIDENTIALITY

Your participation in this study is voluntarily and you can withdraw at any time without penalty. You should not participate if you are currently pregnant, have had previous surgery to your other knee, or have any other nerve or musculoskeletal disorder such as cerebral palsy, multiple sclerosis, or arthritis. The findings of our studies will remain confidential and personal information will not be released without your consent. Data obtained from this study will be recorded on a computer and archived for a period of up to fifteen years. You are free to withdraw at any time. Your participation in this study will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

### RISKS AND BENEFITS

Although the use of recordings with wire electrodes is a standard clinical procedure for diagnosis of neural and muscular problems, there are a number of risks involved. There is some discomfort: you will feel a brief prick to your skin as if receiving a shot with a very small needle. There is also a slight risk of bleeding and of damage to nerves and vessels, and a

possibility of delayed infection. Infection can be identified as a small area of redness around the site. If it persists for more than 24 hours, a physician should be consulted- These various risks are minimal when the recordings are practiced by an experienced electromyographer, as will be done here.

Subject's initials: \_\_\_\_\_

Page 1 of 2

Surface EMG recordings pose little risk. There may be some minor irritation of the skin around the site of the electrode following the experiment. This is most likely due to the mild adhesive and electrolytes in the electrode. These procedures are entirely experimental and they are not intended to provide any specific medical diagnosis or treatment. It is hoped that such studies will eventually provide considerable help in our understanding of how to best treat people with knee ligament injuries, but no immediate benefit may result.

In the event of physical injury as a direct result of these research procedures, you will receive first aid. If you require additional medical treatment, you will be responsible for the cost.

#### FINANCIAL CONSIDERATIONS

To compensate for your time and travel expenses, you will be reimbursed \$50 for participating in this research study.

#### CONTACTS

Further information regarding this study may be obtained from the project director, Dr. Thomas Buchanan, at telephone number (302) 831-2410. Other questions about your rights as a research subject can be directed to Chair, Human Subject Review Board Office of the Research Office, (302) 831-2137.

#### SUBJECT ASSURANCES

Participation in this research study is voluntary. Your participation in this study (or refusal to participate in this study) will not affect current or future Physical Therapy treatment at the University of Delaware in any way.

*I have had the procedures explained, have had all my questions answered. I agree to participate in the research study described above.*

Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_

Parent or Guardian Signature: \_\_\_\_\_ Date: \_\_\_\_\_

*(if subject is under 18 years of age)*

Witness Signature: \_\_\_\_\_ Date: \_\_\_\_\_

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

- Assent Form 1 (for subjects age 14-17): MRI-

### EXPLANATION OF PROCEDURES

We will take Magnetic Resonance Imaging (MRI) scans of your knee. These scans will be taken while you remain still. You will lie on your back in the MRI machine. Typical imaging times range from 3 - 7 minutes. This test should take approximately 1 hour.

### PURPOSE OF THE STUDY

In this research project, we are studying how people who have completely torn one of their knee ligaments still manage to stabilize their knees. Either your doctor referred you to us because of your injury or you are in the control group. If you are in the injured group, we will be studying your knee before and after your surgery-

### DISCOMFORTS

There are no benefits to you for participation in this study and all risks to you are very low. MRI scanning is a standard procedure and does not involve exposure to radiation.

Some people get uncomfortable being in the close confines of a scanner, but we will minimize this problem by having your head outside the scanner since we are only interested in scanning your knee

Participation in this research study is voluntary and if for any reason you become uncomfortable and would like to stop, please let us know because we can stop at any time.

*I have had the procedures explained, have had all my questions answered. I assent to participate in the research study described above.*

Subject Name: \_\_\_\_\_ Age: \_\_\_\_\_  
Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
Name of Person obtaining assent: \_\_\_\_\_  
Signature of Person obtaining assent: \_\_\_\_\_ Date: \_\_\_\_\_

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

- Assent Form 2 (for subjects age 14-17): Knee Extensor Strength---

### EXPLANATION OF PROCEDURES

We will measure the strength of your knee. You will sit in a special device with your knee bent at 90° and a force measuring device will be put on your leg just above your ankle. We will stick electrodes over your thigh muscles. You will be instructed to contract your muscles as hard as you can. During the contraction, a very brief burst of electrical stimulation will be sent to the electrodes. This will tingle a bit. We will measure your joint strength before and during the electrical stimulation- These tests should take approximately 1 hour.

### PURPOSE OF THE STUDY

In this research project, we are studying how people who have completely torn one of their knee ligaments still manage to stabilize their knees, Either your doctor referred you to us because of your injury or you are in the control group. If you are in the injured group, we will be studying your knee before and after your surgery.

### DISCOMFORTS

There are no benefits to you for participation in this study and all risks to you are very low. The electrical stimulator should not cause any damage to your muscles. Some people have a little irritation of the skin around the site of the electrode following the experiment. This is most likely due to the mild adhesive in the electrodes. Your muscles may be a little sore after the maximal contractions, but this should not last long, maybe a day or so.

Participation in this research study is voluntary and if for any reason you become uncomfortable and would like to stop, please let us know because we can stop at any time.

*I have had the procedures explained, have had all my questions answered. I assent to participate in the research study described above.*

Subject Name: \_\_\_\_\_ Age: \_\_\_\_\_  
Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
Name of Person obtaining assent: \_\_\_\_\_  
Signature of Person obtaining assent: \_\_\_\_\_ Date: \_\_\_\_\_

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

- Assent Form 3 (for subjects age 14-17): EMG-

### EXPLANATION OF PROCEDURES

We will measure the way muscles are used in your knee while seated and standing. During the seated task you will sit in a special chair with your knee bent and a force measuring device will be put on your leg just above your ankle. During the standing task you will stand with bare feet on force measuring devices. We will stick electrodes on some of your leg muscles. You will be instructed to try to push your leg in different directions. To help you know if you are pushing in the right direction, a computer will display how well you are pushing, making this task like a video game played with your leg. Once you get it to work in one direction, you will get a short rest and then will get to try another.

Muscle activity will be recorded using two types of electrodes. For some muscles, electrodes will be taped to your skin. For others, we will have to put the electrode inside your muscles. To do that, very small needles will be used to help put the wires into your muscles. At the end of the experiment, all electrodes will be removed. These tests should take approximately 3 1/2 hours.

### PURPOSE OF THE STUDY

In this research project, we are studying how people who have completely torn one of their knee ligaments still manage to stabilize their knees. Either your doctor referred you to us because of your injury or you are in the control group. If you are in the injured group, we will be studying your knee before and after your surgery.

### DISCOMFORTS

There are no benefits to you for participation in this study and all risks to you are low. Some people have a little irritation of the skin around the site of the electrodes following the experiment. This is most likely due to the mild adhesive in the electrodes. For the electrodes put inside your muscles, you will feel a brief prick to your skin, just like when you get a shot with a very small needle. Also just like with a shot, sometimes this causes a little bleeding.

Participation in this research study is voluntary and if for any reason you become uncomfortable and would like to stop, please let us know because we can stop at anytime.

*I have had the procedures explained, have had all my questions answered. I assent to participate in the research study described above.*

Subject Name: \_\_\_\_\_ Age: \_\_\_\_\_

Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_

Name of Person obtaining assent: \_\_\_\_\_

Signature of Person obtaining assent: \_\_\_\_\_ Date: \_\_\_\_\_

## ACL Injured Knee: MRI and Biomechanical Modeling

Investigators: Thomas Buchanan, PhD, Lynn Snyder-Mackler, PT, ScD,  
Michael Axe, MD and Larry Miller, MD.

-Assent Form 4 (for subjects age 14-17): Gait Analysis -

### EXPLANATION OF PROCEDURES

We will measure the way muscles in your knee are used when you run. You will be asked to run on a long runway in our lab and when you reach the end of it, you will be asked to do one of two things: straight run or sidestep to approximately 45°.

To measure what your muscles are doing, we will stick electrodes on some of your leg muscles. We will also put markers on your skin so our cameras can measure how you run.

Two types of electrodes will be used. For some muscles, electrodes will be taped to your skin. For others, we will have to put the electrode inside your muscles. To do that, very small needles will be used to help put the wires into your muscles. At the end of the experiment, all electrodes will be removed. These tests, should take approximately 2 hours.

### PURPOSE OF THE STUDY

In this research project, we are studying how people who have completely torn one of their knee ligaments still manage to stabilize their knees. Either your doctor referred you to us because of your injury or you are in the control group. If you are in the injured group, we will be studying your knee before and after your surgery.

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Subject Name: \_\_\_\_\_ Age: \_\_\_\_\_  
Subject Signature: \_\_\_\_\_ Date: \_\_\_\_\_  
Name of Person obtaining assent: \_\_\_\_\_  
Signature of Person obtaining assent: \_\_\_\_\_ Date: \_\_\_\_\_