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**Article Title:** Quantification of Tibiofemoral Shear and Compressive Loads Using a MRI-Based EMG-driven Knee Model

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QUANTIFICATION OF TIBIOFEMORAL SHEAR AND COMPRESSION LOADS USING A MRI-BASED EMG-DRIVEN KNEE MODEL

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Abstract

The purpose of this study is to describe a MRI-based EMG-driven knee model to quantify tibiofemoral compressive and shear forces. Twelve healthy females participated. Subjects underwent 2 phases of data collection: 1) MRI assessment of lower-extremity to quantify muscle volumes and patella tendon orientation; and 2) biomechanical evaluation of a drop-jump task. A subject-specific EMG-driven knee model that incorporated lower extremity kinematics, EMG, and muscle volumes and patella tendon orientation estimated from MRI was developed to quantify tibiofemoral shear and compressive forces. A resultant anterior tibial shear force generated from the ground reaction force (GRF) and muscle forces was observed during the first 30% of the stance phase of the drop-jump task. All of the muscle forces and GRF resulted in tibiofemoral compression with the quadriceps force being the primary contributor. Acquiring subject-specific muscle volumes and patella tendon orientation for use in an EMG-driven knee model may be useful to quantify tibiofemoral forces in persons with altered patella position or muscle atrophy following knee injury or pathology.

Keywords: MRI; muscle volume; patella tendon orientation
Introduction

Excessive joint loading may increase the risk of developing musculoskeletal disorders at the knee. Specifically, excessive anterior tibial shear force has been hypothesized as one possible cause of anterior cruciate ligament (ACL) injury,\(^1\) and excessive compressive loading may be associated with the development of knee osteoarthritis.\(^2,3\) As such, accurate assessment of knee loading is essential in understanding pathological joint function.

Assessment of joint loading cannot be valid without considering the contributions of individual muscle forces acting across the joint. The development of electromyography (EMG) driven modeling provides one possible solution for estimating muscle forces.\(^4\) However, a challenge in creating a valid EMG-driven musculoskeletal model is the need to obtain accurate muscle anatomic parameters.

An EMG-driven model that utilizes direct measurements of muscle anatomic parameters (i.e., muscle volume, tendon orientation) may improve estimates of the individual muscle force vectors across the knee, thereby leading to more accurate assessment of joint loading. The purpose of this study was to describe an EMG-driven knee model using subject-specific individual muscle volumes and patella tendon orientation from magnetic resonance imaging (MRI) to quantify tibiofemoral compressive and shear forces.

Methods

Subjects

Twelve healthy females (25.9 ± 3.5 y; 59.2 ± 5.8 kg; 166.3 ± 4.5cm) participated. Subjects were excluded from participation if they had previous knee surgery, or any medical condition that would impair their ability to perform the tasks described below. Prior to participation, all procedures were explained to each subject and informed consent was obtained.
as approved by the Institutional Review Board of the University of Southern California Health Sciences Campus.

**Procedures**

Subjects participated in 2 data collection sessions: 1) MRI scan to obtain subject-specific muscle anatomic parameters; and 2) biomechanical testing. First, each subject’s dominant leg (the leg used to kick a ball) was imaged using a 3.0 Tesla MRI system (GE Signa HDx 3.0 T). Axial images of the leg (ankle mortise to the iliac crest) were acquired using a spin-echo pulse sequence (TR: 2600-3700 ms, TE: 11.3 ms, slice thickness: 10 mm). Sagittal-plane knee images were obtained using a spin-echo pulse sequence (TR: 1100 ms, TE: 37 ms, slice thickness: 3 mm) at 0, 15, 30, 45, and 60 degrees of knee flexion during static partial weight-bearing by having subjects push against a load of 111 N provided by a custom-made loading device.

For biomechanical testing, each subject performed a drop-jump task by starting from a standing position on a platform (height: 35 cm) in front of 2 force plates. Subjects were instructed to land with one foot on each plate and then jump upward as high as possible. Three trials were collected for each subject. Kinematic data during the drop-jump task were recorded at 250 Hz using an 8-camera motion analysis system (Vicon 612; Oxford Metrics, Oxford, UK). Ground reaction forces (GRF) were collected at 1500 Hz using a force plate (AMTI, Newton, USA).

Muscle activation levels during drop-jump were recorded from the vastus lateralis (VL), vastus medialis (VM), rectus femoris (RF), semitendinosis (ST), biceps femoris long head (BFL), medial gastrocnemius (MG), and lateral gastrocnemius (LG) using surface electrodes (MA300 EMG system, Motion Lab Systems, Baton Rouge, USA). Prior to the drop-jump task,
EMG signals from each muscle were obtained while performing 3 maximum voluntary isometric contractions (MVIC). These tests were conducted for normalization purposes.

**Data Analysis**

Muscle volumes of the quadriceps, hamstrings, and gastrocnemius muscles were measured by multiplying the cross sectional area from each axial MR image by the slice thickness. The total muscle volume measured from all slices was combined with the fiber length and pennation angle\(^5,6\) to calculate the physiological cross sectional area (PCSA) for each muscle. The PCSA was then multiplied by a specific tension of 23 (N·cm\(^{-2}\))\(^7\) to approximate the maximum isometric muscle force. The orientation of the patella tendon relative to the tibia was quantified by measuring the angle formed by the patella tendon and the medial tibia plateau from the sagittal-plane MR images obtained at each of the 5 knee flexion angles. A linear line was fit to the 5 data points to estimate the orientation angle from 0 to 150 degrees of knee flexion.

Visual3D (C-motion, Rockville, USA) was used to compute the segmental kinematics of the lower-extremity. Raw trajectory data were filtered using a 4\(^{th}\)-order Butterworth low-pass filter at 6 Hz. Segment mass/weight and center of mass location were approximated from the data of Dempster\(^8\). Raw EMG signals were band-pass filtered (35-500 Hz), rectified, and low-pass filtered at 6 Hz. The low-passed EMG data were normalized to the highest value recorded from either the MVIC or drop-jump task.

**Subject-specific EMG-driven knee model to quantify knee loading**

SIMM software (MusculoGraphics Inc., Chicago, USA) was used to create a generic knee model that included 10 musculotendon actuators: VL, VM, vastus intermedius (VI), RF, ST, semimembranosus (SM), BFL, biceps femoris short head (BFS), MG, and LG. The anatomic parameters of the 10 muscles were based on the values reported by Friederich and Brand\(^5\) and
Wickiewicz et al. Tendon slack lengths were based on the average values reported by Delp and Lloyd and Buchanan.

Normalized EMG and lower extremity kinematics were used as input variables for the EMG-driven model. Lower extremity kinematic data were used to determine muscle tendon lengths and contraction velocities for the Hill-type muscle model built in SIMM. Normalized EMG data were used to represent muscle activation. Muscle activation of the VI was estimated as the average of the VM and VL activation. SM and BFS were assumed to have the same activation as ST and BFL, respectively. A 40 ms electromechanical delay was used.

The maximum isometric muscle forces and patella tendon orientation derived from MRI were incorporated into the generic SIMM knee model to calculate individual muscle force vectors for each subject. The force of each muscle \( F^M \) was calculated using Equation 1:

\[
F^M = F^{\text{Max}} [f(l)f(v)\text{act} + f_p(l)]\cos(\theta)
\] (1)

where \( F^{\text{Max}} \) is the maximum isometric muscle force; \( f(l) \) and \( f(v) \) are the muscle length-tension and force-velocity relationships, respectively; \( \text{act} \) is muscle activation; \( f_p(l) \) is the length-tension of the parallel elastic element; and \( \theta \) is the pennation angle.

Using equation 2, the calculated muscle force vectors were combined with the anthropometry, linear accelerations of the lower leg (foot and shank), and the GRF vector to calculate the tibiofemoral joint compressive and shear forces (i.e., the forces acting on the tibia from the femur).

\[
\sum \text{Forces} = \text{muscle forces} + \text{GRF} + \text{weight of the leg} + \text{tibiofemoral joint compressive and shear forces} = m \cdot a
\] (2)

where \( m \) is the mass of the lower leg and \( a \) is the linear acceleration of the center of mass of the lower leg (Figure 1). All force data were time normalized to represent 100% of the stance phase.
of the drop-jump task (defined as the time from initial contact to toe off). On average, the stance duration was 641 ± 128 ms.

The force and acceleration vectors were referenced to the 3-dimensional reference frame located in the tibia (Figure 1). The longitudinal axis (y-axis) was defined by the line connecting the knee and ankle joint centers, and the anterior-posterior axis (x-axis) was defined as the line perpendicular to the plane formed by the longitudinal axis and lateral femoral epicondyle. Compressive and shear forces were defined as the components of the force vector along the longitudinal and anterior-posterior axes, respectively. The tibiofemoral joint forces calculated from equation 2 represented the internal joint forces required to maintain the kinematic status of the lower leg given the muscle forces and GRF. For interpretation purposes, the signs of the internal joint forces calculated from equation 2 were reversed to represent the total loading applied to the tibiofemoral joint via muscle forces and GRF. Therefore, positive values represent forces that push the tibia upward and anteriorly while negative values are the forces that push the tibia downward and posteriorly.

**Results**

The ensemble average curves of the tibiofemoral shear and compression calculated from the MRI-based EMG-driven knee model are shown in Figure 2. An anterior tibial shear force was observed during initial landing and reached its peak within the first 10-15% of the stance phase during the drop-jump task. After the first 30% of the stance phase, a posterior shear force was observed. In addition, a compressive load was observed at the tibiofemoral joint through the entire stance phase. The tibiofemoral compressive force reached its peak within the first 20% of the stance phase (Figure 2).
Discussion

This study described an EMG-driven knee model that incorporated MRI-measured muscle volumes and patella tendon orientation to quantify tibiofemoral compressive and shear forces. As noted in Figure 2, a resultant anterior tibial shear load was observed during the first 30% of the stance phase of the drop-jump task. In the knee, the ACL has been shown to provide about 85% of the posterior shear force to resist an anterior shear load applied to the tibia. The tibiofemoral shear loading profile estimated using our model is in agreement with the results by other investigators who reported that the ACL experienced a loading during the first 25-30% of the landing phase of a drop-land task.

As illustrated in Figure 2, the peak anterior tibial shear force was primarily the result of forces generated from the quadriceps with a very minor contribution provided by the gastrocnemius. In contrast, the GRF and hamstring muscle forces produced a posterior shear force. These findings are consistent with previous investigations demonstrating that contraction of quadriceps and gastrocnemius muscles increases ACL strain and contraction of the hamstring muscles reduces ACL tension.

Although all muscle forces compressed the tibiofemoral joint, the primary contributor to the peak compressive force was the compressive force from the quadriceps (Figure 2). In contrast, the hamstring muscles provided minimal joint compression. The large compressive force observed by the quadriceps muscles reflects the large PCSA and high activation of the quadriceps muscles as well as a more parallel orientation of the patella tendon relative to the tibia with the knee flexed when compared to the hamstrings and gastrocnemius muscles. As such, incorporating subject-specific muscle volumes and patella tendon orientation estimated from
MRI may be useful to quantify tibiofemoral joint forces in persons with altered patella position\textsuperscript{17} or muscle atrophy following knee injury or pathology\textsuperscript{18}.

Using the maximum isometric muscle forces derived from direct measurements of muscle volumes has been shown to reduce joint moment prediction errors of an EMG-driven model\textsuperscript{19,20}. On average, the estimated maximal isometric forces of the gastrocnemius muscles were quite similar between the generic SIMM and MRI-based models (Figure 3). However, the estimated maximal isometric forces of the quadriceps and hamstrings muscles of the generic SIMM model were greater than that of the MRI-based model (Figure 3), indicating the muscle forces and joint compressive loading might be overestimated if direct measurements of muscle volumes were not acquired for the subjects in the current study. To better understand the influence of incorporating MRI measurements into the EMG-driven model, we performed a post-hoc analysis in which we compared the tibiofemoral shear and compressive forces calculated by the generic SIMM and the MRI-based model (Figure 4). As expected, the generic SIMM model overestimated the compressive force and underestimated the anterior shear force when compared to the MRI-based model. The higher compressive force from the generic model was primarily the result of higher muscle forces (mostly quadriceps and hamstring muscles). In turn, the lower anterior tibial shear force from the generic model was primarily due to a more vertically oriented patella tendon orientation (Figure 5) and greater posterior shear forces from the hamstring muscles.

Several limitations need to be acknowledged. While muscle volumes were measured from MRI, PCSA calculations incorporated in-vitro pennation angles and muscle fiber lengths\textsuperscript{5,6}. This may have limited the accuracy of calculating a subject-specific PCSA of each muscle. Another limitation of our model is that tibial slope was not included in the model. This may lead
to underestimation of the shear forces and overestimation of the compressive forces. Lastly, while the EMG-driven model incorporated MRI-estimated muscle volumes and patella tendon orientation, generic values were used for muscle contractile parameters (e.g., tendon slack length). Muscle contractile parameters are difficult to measure in-vivo, and are often estimated through optimization procedures.\textsuperscript{4} Future modeling approaches that combine direct measurements of muscle anatomic parameters and optimization procedures for muscle contractile parameters may further advance the application of an EMG-driven knee model.
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Figure 1 - The free body diagram showing the tibia reference frame and the forces used to calculate the tibiofemoral shear and compressive forces.
Figure 2 - The tibiofemoral shear (A) and compressive (B) forces as well as the shear and compressive components of the GRF and muscle forces calculated from the MRI-based EMG-driven knee model during drop-jump. Positive values indicate an anterior shear force and upward compressive force pushing the lower leg toward the femur.
Figure 3 - Generic SIMM and MRI-derived maximum isometric muscle forces of the quadriceps (VL, VI, VM, RF), hamstrings (BFS, BFL, ST, SM), and gastrocnemius muscles (MG, LG).
Figure 4 - The tibiofemoral shear (A) and compressive (B) forces estimated from the generic SIMM and MRI-based EMG-driven knee model during drop-jump.
Figure 5 - Generic SIMM and MRI-estimated patella tendon orientation angles (the angle between the patella tendon and the anterior-posterior axis of the tibia reference frame). Values are mean ± SD. An angle of 90 degrees indicates that the patella tendon is perpendicular to the anterior-posterior axis.