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Dynamic *in vivo* quadriceps lines-of-action

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ABSTRACT

Tissue stresses and quadriceps forces are crucial factors when considering knee joint biomechanics. However, it is difficult to obtain direct, *in vivo*, measurements of these quantities. The primary purpose of this study was to provide the first complete description of quadriceps geometry (force directions and moment arms) of individual quadriceps components using *in vivo*, 3D data collected during volitional knee extension. A secondary purpose was to determine if 3D quadriceps geometry is altered in patients with patellofemoral pain and maltracking. After obtaining informed consent, cine-phase contrast (PC) MRI sets (*x*,*y*,*z* velocity and anatomic images) were acquired from 25 asymptomatic knees and 15 knees with patellofemoral pain during active knee extension. Using a sagittal-oblique and two coronal-oblique imaging planes, the origins and insertions of each quadriceps line-of-action were identified and tracked throughout the motion by integrating the cine-PC velocity data. The force direction and relative moment (**RM**) were calculated for each line-of-action. All quadriceps lines-of-action were oriented primarily in the superior direction. There were no significant differences in quadriceps geometry between asymptomatic and subjects with patellofemoral pain. However, patellofemoral kinematics were significantly different between the two populations. This study will improve the ability of musculoskeletal models to closely match *in vivo* human performance by providing accurate 3D quadriceps geometry and associated patellofemoral kinematics during dynamic knee motion. Furthermore, determination that quadriceps geometry is not altered in patellofemoral pain supports the use of generalized a knee model based on asymptomatic quadriceps architecture.

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1. Introduction

Quadriceps forces and tissues stresses are crucial factors when considering knee joint biomechanics, as they are controlling factors in tibiofemoral kinematics, patellofemoral kinematics, and cartilage contact forces. Since it is difficult to obtain direct measurements of *in vivo* forces and stresses, musculoskeletal models are increasingly used to provide estimates of these quantities. However the accuracy of model-based analyses is highly dependent on the quality of input data used to create the model. For example, both the 3D force vectors (Elias and Cosgarea, 2007) and moment arms (Hunter et al., 2009) used to represent the quadriceps muscles can dramatically influence computational output (Delp et al., 1990). Musculoskeletal models have also been used in the study of pathology (Arnold et al., 2001; Besier et al., 2009; Hunter et al., 2009). However, musculoskeletal parameters may be altered in pathology, making it difficult to create accurate models of specific pathologies without *a priori* knowledge of musculoskeletal geometry under pathologic conditions (Elias et al., 2006).

Musculoskeletal models of the knee typically rely on input data from a variety of sources. Quadriceps force directions or lines-of-action have been derived primarily from static, cadaver-based studies (Delp et al., 1990; Farahmand et al., 1998; Garg and Walker, 1990; Herzog and Read, 1993; Powers et al., 1998; van Eijden et al., 1986). While these studies provide valuable data, cadaver-based studies cannot reproduce the complex loading patterns applied to the knee *in vivo*. Quadriceps moment arms are often derived based on skeletal geometry (Besier et al., 2009; Delp et al., 1990; Elias et al., 2006; Shelburne et al., 2004) or obtained from other studies that reduce the quadriceps muscles to a single tendon (Powers et al., 2004; Spoor and van Leeuwen, 1992; van Eijden et al., 1986). Reducing the quadriceps to a single tendon eliminates the ability to assess the 3D dynamics of the knee extensor mechanism. Model performance can be enhanced by incorporating a more complete description of 3D quadriceps geometry (force direction and moment arm of individual quadriceps components) quantified *in vivo*, during volitional knee extension.

The need for high-quality data describing musculoskeletal geometry is magnified when modeling pathology, where generalized assumptions regarding musculoskeletal geometry may lead

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to large errors. Hunter et al. (2009) demonstrated that improvements in 3D moment arm estimates were essential for investigating pathology, particularly when predicting joint angular accelerations. Arnold et al. (2001), in a study of cerebral palsy, noted that variations in muscle attachment locations and changes in musculotendon paths with joint motion influenced the calculation of muscle–tendon lengths. Besier et al. (2009) used a general model to estimate sagittal-plane quadriceps muscle forces and joint moments in patellofemoral pain (PFP) syndrome. While this model used individualized EMG data to account for variations in muscle recruitment patterns, it did not consider the secondary planes of motion, which play a large role in PFP syndrome and many other pathologies (Elias et al., 2006; Hunter et al., 2009; Makhsous et al., 2004; Sheehan et al., 2009; Wilson et al., 2009), nor did it incorporate the potential for variations in quadriceps geometry associated with PFP (Lin et al., 2004; Jan et al., 2009). One way of addressing these limitations has been through the use of subject-specific modeling (Elias et al., 2006). However, subject-specific modeling is time-consuming and computationally intensive.

The primary purpose of this study was to provide the first description of quadriceps geometry using *in vivo*, 3D data collected during volitional knee extension. A secondary purpose was to determine if the 3D quadriceps force directions and moment arms were altered in pathology, with specific application to subjects with PFP and maltracking. Determination that the quadriceps geometry is not altered in PFP syndrome would support the use of a generalized knee model based on asymptomatic quadriceps architecture.

2. Methods

All participants gave informed consent upon entering this IRB-approved study, followed by a complete history and physical. Asymptomatic subjects were excluded if they had any current or past history of knee pain (regardless of etiology), any history of lower leg abnormality, surgery, or major injury. Subjects with PFP syndrome had a clinical diagnosis of idiopathic anterior knee pain present for at least one year and were included based on previously published inclusion/exclusion criteria (Sheehan et al., 2009). In total 23 asymptomatic volunteers and 11 subjects with PFP syndrome were included in this study. If both knees from a single subject fit the criteria for inclusion into a single group and time permitted, both knees were evaluated (Sheehan et al., 2009), resulting in 25 asymptomatic knees and 15 knees with PFP syndrome being included in the study. Demographic characteristics from the two cohorts were similar, except for gender (Table 1).

Using a previously published imaging protocol (Wilson and Sheehan, 2009), a sagittal-oblique and two coronal-oblique dynamic cine phase contrast (PC) MR image sets (x, y, z velocity and anatomic images over 24 time frames) along with a dynamic cine MR image set (anatomic images in three axial planes over 24 time frames) were acquired while each subject, laying supine, performed cyclic knee flexion/extension movements from maximum attainable flexion ($\sim 50^\circ$) to full extension (0°). The standard MR image planes (axial, sagittal, and coronal) were defined relative to a fixed coordinate system within the MRI. Using an oblique imaging plane (a plane rotated away from the true plane) allowed these planes to be aligned with anatomical features, enhancing consistency across subjects (Fig. 1 in Wilson and Sheehan, 2009 provides a visual definition of the imaging planes). 3D rigid body rotations and translations (kinematics) of the femur, tibia, and patella were quantified through integration of the sagittal-oblique cine-PC velocity

Table 1
Subject demographics.

	Asymptomatic	Patellofemoral pain	*p*-value
Gender	13 male, 12 female	2 male, 13 female	0.020
Age (years)	25.1 ± 4.9	27.1 ± 11.8	0.456
Height (cm)	171.2 ± 7.4	178.9 ± 5.8	0.312
Weight (kg)	67.5 ± 12.6	65.6 ± 12.9	0.651
Epicondylar width (mm)	76.9 ± 6.5	73.9 ± 4.5	0.117

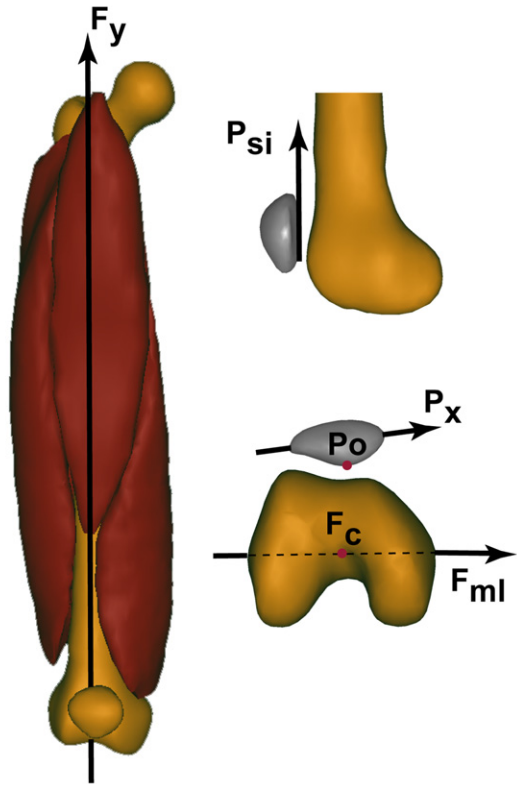

Fig. 1. Femoral and patellar coordinate systems . All coordinate systems were defined in the full extension time frame. The femoral *y*-axis (**Fy**), or superior/inferior axis, was defined as the unit vector in the direction from the center of the femoral epicondyles (point Fc, selected on the axial cine image at the level of the femoral epicondyles) to the rectus femoris musculotendon junction (selected in the cine-PC coronal-oblique image of the rectus femoris), such that **Fy** was aligned with the anatomical axis of the femur. **Fml** was defined as the unit vector in the direction from the lateral to the medial femoral epicondyles (selected on the axial cine image at the level of the femoral epicondyles). The anterior/posterior axis (**Fz**) was defined by the cross product of **Fml** with **Fy**. The femoral *x*-axis (**Fx**), or medial/lateral axis was defined by the cross product of **Fy** with **Fz**. Fc was defined as the femoral origin. The patellar *x*-axis (**Px**) was defined as the unit vector connecting the most lateral (PL) and medial (PM) points on the patella (selected on the axial cine image at the mid-patella level—Fig. 2). **Psi** was defined as the unit vector along the most posterior edge of the patella (selected on the sagittal-oblique cine-PC image). The anterior/posterior axis (**Pz**), was defined by the cross product of **Px** with **Psi.** The superior/inferior axis (**Py**) was defined as the cross product between **Pz** and **Px.** The patellar origin (Po) was defined as the most posterior patellar point (selected on the axial cine image at the mid-patella level).

data (Sheehan et al., 1999), with an accuracy of <0.5 mm (Sheehan et al., 1998). In a similar manner, the origin of each musculotendon line-of-action (vastus intermedius (VI), rectus femoris (RF), vastus medialis (VM), and vastus lateralis (VL)) was tracked through integration of the coronal-oblique cine-PC velocity data.

Dynamic anatomic cine images were used to establish the patellar and femoral anatomical coordinate systems (Fig. 1) and define bony points of interest, including the patellar center of mass (defined as the centroid of the patella) and the insertion of each musculotendon unit onto the patella, as in a previous study (Wilson and Sheehan, 2009). The femoral coordinate system definition was modified from the previous study (Wilson and Sheehan, 2009) in that the femoral superior/inferior axis was aligned with the femoral anatomical axis, as opposed to the femoral mechanical axis. All lines-of-action, relative moments, and patellar kinematics were defined relative to this femoral reference system. The coordinate system and all points of interest were identified in a single time-frame (full extension) of the dynamic images and integration of the cine-PC velocity data was used to track kinematic changes throughout the motion cycle (Sheehan et al., 1999).

The quadriceps muscles have myotendinous junctions which approach the quadriceps tendons through a range of angles (Buford Jr. et al., 1997), such that the tendon line-of-action (defined as the unit vector from tendon insertion on the patella to its muscular origin) varies across the width of the tendon. Therefore, six lines-of-action were used to characterize the geometry of the four quadriceps tendons (Fig. 2). The central quadriceps components, VI and RF, were each