Quantification of Patellofemoral Joint Reaction Forces During Functional Activities Using a Subject-Specific Three-Dimensional Model

Yu-Jen Chen, Irving Scher, and Christopher M. Powers

The purpose of this study was to describe an imaging based, subject specific model that was developed to quantify patellofemoral joint reaction forces (PFJRF’s). The secondary purpose was to test the model in a group of healthy individuals while performing various functional tasks. Twenty healthy subjects (10 males, 10 females) were recruited. All participants underwent two phases of data collection: 1) magnetic resonance imaging of the knee, patellofemoral joint, and thigh, and 2) kinematic, kinetic and EMG analysis during walking, running, stair ascent, and stair descent. Using data obtained from MRI, a subject specific representation of the extensor mechanism was created. Individual gait data were used to drive the model (via an optimization routine) and three-dimensional vasti muscle forces and subsequent three-dimensional PFJRF’s were computed. The average peak PFJRF was found to be highest during running (58.2 N/kg-bwt), followed by stair ascent (33.9 N/kg-bwt), stair descent (27.9 N/kg-bwt), and walking (10.1 N/kg-bwt). No differences were found between males and females. For all conditions, the direction of the PFJRF was always in the posterior, superior, and lateral directions. The posterior component of the PFJRF always had the greatest magnitude, followed by superior and lateral components. Our results indicate that estimates of the magnitude and direction of the PFJRF during functional tasks can be obtained using a 3D-imaging based model.

Keywords: Patellofemoral joint, muscle forces, modeling

Patellofemoral pain affects as many as 22% of the general population and is more commonly observed in females when compared with males (Fulkerson, 2002; Malek & Mangine, 1981; Thomee et al., 1999). Although patellofemoral pain is one of the most common disorders of the knee, the underlying pathomechanics remain unclear. A commonly cited theory states that patellofemoral pain is the result of elevated patellofemoral joint reaction forces (PFJRF’s) which contribute to excessive joint stress and subsequent cartilage wear (Goodfellow et al., 1976).

Given the inherent difficulties in quantifying joint loading in-vivo, mathematical modeling has been employed to estimate the forces experienced at the patellofemoral joint. Although modeling approaches have varied, most authors have used a two-dimensional, planar representation (Yamaguchi & Zajac, 1989; van Eijden et al., 1986; Simpson et al., 1996; Cohen et al., 2001; Steinkamp et al., 1993; Heino-Brechter & Powers, 2002; Wallace et al., 2002; Salem & Powers, 2001). In contrast, only a few studies have used three-dimensional approaches to estimate PFJRF’s (Neptune et al., 2000; Kwak et al., 2000; Hirokawa, 1991). A limitation of previous modeling approaches is that each has lacked subject specific parameters, making extrapolation of information to a patient population difficult. Another disadvantage has been that estimations of PFJRF’s have been limited to specific joint angles instead of during dynamic tasks.

A subject specific, three-dimensional model would have advantages over a two-dimensional or a generic three-dimensional representation by providing a more physiologic representation of the forces experienced at the patellofemoral joint. The purpose of this paper was to describe an imaging based, subject specific model that was developed to quantify PFJRF’s. A secondary purpose of this study was to compare PFJRF’s between healthy males and females while performing various functional tasks. We hypothesized that the patellofemoral joint reaction forces would be higher in females when compared with males.

Methods

Subjects
Twenty healthy pain-free individuals (10 male, 10 female, age: 26 ± 4.3 years, height: 165 ± 7.9 cm, weight: 68 ±
8.6 kg) between the ages of 18 and 45 were recruited. Specific exclusion criteria included: 1) history or diagnosis of knee pathology or trauma, 2) current knee pain or effusion, 3) knee pain with any recreational activities or activities of daily living, 4) limitations that would influence gait, and 5) implanted biological devices, such as pacemakers, cochlear implants, or clips that are contraindicated for MRI.

Instrumentation

MRI. Images were obtained from a 1.5T MR system (General Electric Medical Systems, Waukesha, WI). Sagittal plane images of the knee were acquired with two 5-inch receive only coils placing on each side of the knee joint and secured with tape. T1-weighted imaging was obtained with the following parameters: TR: 300 ms, TE: 10 ms, field of view (FOV): 20 cm x 20 cm, slice thickness: 2 mm, matrix: 256 x 256. Frontal plane images of the thigh were acquired using T1-weighted coronal and sagittal imaging with the following parameters: TR: 450 ms, TE: 12.6 ms, field of view (FOV): 48 cm x 48 cm, slice thickness: 10 mm, matrix: 512 x 512. Axial images of the thigh were acquired using T1-weighted, fast spin echo images with the following parameters: TR: 400 ms, TE: 22 ms, FOV: 24 cm x 24 cm, slice thickness: 7 mm, matrix: 256 x 256.

Gait Analysis. Three-dimensional motion analysis was performed using a computer aided video (VICON) motion analysis system (Oxford Metrics LTD. Oxford, England.) Kinematic data were sampled at 120 Hz. Reflective markers (9 mm spheres) placed at specific anatomical landmarks (see procedure section for details) were used to determine sagittal plane motion of the pelvis, hip, knee, and ankle. As described in previous publications (Powers et al., 2004a; Powers et al., 2004b) data for walking and running were acquired along a 10-m walkway. Analysis of stair ascent and descent was made using a portable 4-step staircase which had a slope of 33.7 degrees, a step height of 20.3 cm, and a tread depth of 30.5 cm. Ground reaction forces were collected at a rate of 2500 Hz using two AMTI force plates (Model #OR6–6–1, Newton, Mass). The force plates were situated within the middle of the 10-m walkway with the pattern of tile flooring camouflaging their location. For assessment of ground reaction forces during stair ambulation, the raised floor tiling was removed and portable steps were positioned such that the force plate became the second step of the staircase. The portable steps were designed to avoid contact with the force plate.

EMG activity of the flexor muscles crossing the knee was recorded at 2500 Hz, using preamplified bipolar, grounded, surface electrodes (Motion Laboratory, Salt Lake City, UT). EMG signals were telemetered to an analog to digital converter using a 14 channel unit. Differential amplifiers were used to reject the common noise and amplify the remaining signal (gain = 1000). EMG signals were band pass filtered (20–500 Hz) and a 60 Hz notch filter was applied.

Procedures

Before participation, the purpose of the study, procedures, and risks were explained to each subject. Informed consent as approved by the Institutional Review Board for the University of Southern California Health Science Campus was obtained.

Subjects underwent two phases of testing. Phase one consisted of MRI assessment of the knee, and thigh, while phase two consisted of instrumented gait analysis. Only the dominant limb of each subject was tested (as determined by the preferred limb used to kick a ball). Before MRI and gait testing, the following anthropometric measurements were obtained: height, weight, leg length (from ASIS to the medial malleolus), and frontal plane patellar ligament orientation (Q-angle).

Following relaxed supine imaging of the thigh, loaded sagittal plane images of the knee and patellofemoral joint were obtained using a custom made non-ferromagnetic loading apparatus. Subjects were asked to lie supine on the loading device with their knee extended and foot placed on the footplate of the loading device. A 25% body weight resistance was provided through the pulley system to the footplate. Sagittal plane images were then obtained. This procedure was repeated at 20, 40, 60 degrees of knee flexion. Loaded images were used to obtain structural measures of the vasti (fiber orientations, PCSA, etc.).

Gait Analysis. Gait analysis was performed at the Musculoskeletal Biomechanics Research Laboratory at the University of Southern California. Subjects were appropriately clothed, allowing the pelvis, hip, thigh and lower leg of the involved limb to be exposed. A triad of rigid reflective tracking markers was securely placed on the lateral surface of the subject’s right and left thigh, leg, and heel counter of the shoe. Tracking markers were placed on each anterior superior iliac spine and iliac crest, as well as on the L5/S1 junction. In addition to the tracking markers, calibration markers were placed bilaterally on the greater trochanters, medial and lateral femoral epicondyles, medial and lateral malleoli, and 1st and 5th metatarsal heads. Surface EMG electrodes were taped to the skin over the medial and lateral hamstrings, and the medial head of the gastrocnemius. Electrodes were connected to the hardwire unit, which was strapped around the subjects’ back.

To allow for comparison of EMG intensity between subjects and muscles, and to control for variability
induced by electrode placement, EMG data were normalized to the EMG acquired during a maximum voluntary isometric contraction (MVIC). The MVIC of the hamstrings was performed with subject lying supine on treatment table with a custom-made box placed under the tibia to keep both the hip and knee flexed to 90 degrees. A strap was used to secure the hip to the table. Subjects were asked to maximally flex their knee and raise their hip, and hold for 6 s. The gastrocnemius MVIC was performed with the subject standing with an adjustable strap placed from the bottom of the tested foot to both sides of the subjects’ shoulders. Subjects were asked to perform a maximal heel raise for 6 s with the strap providing resistance.

Following MVIC testing, a standing calibration trial was obtained. Practice trials of walking and running allowed subjects to become familiar with the instrumentation. Kinematic, kinetic and EMG data were collected simultaneously (and synchronized) during walking (80 m/min), running (200 m/minute) and ascending/descending stairs (50 steps/minute). These velocities are considered to be normative values for these tasks (Powers et al., 2004a, 2004b).

A trial was considered successful if the subject’s instrumented foot landed within either force plate (without targeting) at the required task velocity. Task velocity was calculated immediately following each trial, and trials were repeated as necessary. Three acceptable trials of data were obtained and averaged for analysis.

Model Description. Using data obtained from magnetic resonance imaging (MRI) and clinical measurements, a subject specific representation of the extensor mechanism was created using SIMM modeling software (MusculosGraphics, Santa Rosa, CA). Individual gait data were used to drive the model (via an optimization routine) and three-dimensional vasti muscle forces and subsequent three-dimensional PFJRF’s were computed (Figure 1). The following subject specific input variables were used in the model: 1) three-dimensional kinematics of the lower extremity, 2) net knee joint moment in the sagittal plane, 3) hamstring and gastrocnemius electromyography (EMG), 4) extensor mechanism lever arm, 5) vasti muscle orientation, 6) vasti muscle physiological cross-sectional area proportions, 7) patella flexion angle, and 8) patellar ligament orientation. Each of these variables was quantified as described below.

Gait Input Variables

Visual3D software (C-Motion, Inc., Rockville, MD, USA) was used to quantify 3D motion of the hip, knee, and ankle motion. In Visual3D, all lower extremity segments are modeled as frusta of cones while the pelvis is modeled as an ellipsoid. VICON Workstation software was used to digitize the kinematic data. The local coordinate systems of the pelvis, thigh, leg and foot were derived from the standing calibration trial. In addition, the segment ends were identified from the standing calibration trial to locate the segment origins. Coordinate data were low-pass filtered using a fourth-order Butterworth filter with an 6 Hz cutoff frequency for the gait and ascend/descend stairs data and an 12 Hz cutoff frequency for the running data. Visual3D software was used to calculate 3D net joint moments using inverse dynamics equations. To facilitate comparisons between groups, kinetic data were normalized to body mass.

Determination of the presence or absence of muscle activity was made using the EMG data. Initially, the root mean square values were calculated for each EMG channel sample-by-sample. This process is the mathematical equivalent to full wave rectification. Next, a moving average was calculated to generate a linear envelope. The EMG data were then integrated over time periods corresponding to 2% of the activity cycle. EMG intensities were expressed as a percentage of the EMG obtained during the MVIC.

MRI Input Variables

Extensor Mechanism Lever Arm. The lever arm of the extensor mechanism was measured on the sagittal MR image containing the intersection of the ACL and PCL. The lever arm was defined as the perpendicular distance from the axis of rotation of the knee joint to the line of action of quadriceps muscle (Figure 2). The axis of rotation of the knee was identified as the intersection of the anterior cruciate ligament and the posterior cruciate ligament, and the line of action of the quadriceps muscle was identified from the insertion of the quadriceps muscle on the superior aspect of the base of patella to the midpoint of the quadriceps muscle tendon (Ward & Powers, 2004). The lever arm of the quadriceps muscle was measured at 0, 20, 40, and 60 degrees of knee flexion. A second order
polynomial curve fitting procedure was used to estimate lever arm data for every degree of knee flexion from 0 to 90 degrees.

**Vasti Muscle Fiber Orientation.** MR images were analyzed to define the fiber orientation of each of the vasti relative to the femur in both sagittal and frontal planes. For the sagittal plane orientation of the vastus lateralis (VL) and vastus medialis (VM), a reference line for the femur was first defined from the centroid of the femoral head to the centroid of the lateral epicondyle of the femur (Figure 3a). These reference lines were identified on the image that contained the largest section of femur. The sagittal plane vasti orientation was measured on the image containing the largest cross-sectional area of the VL and VM, and the femur reference line was superimposed on these sagittal images. The upper and lower border of the VL and VM was identified following the borders of their fascial planes. The muscle was then divided into thirds and the fiber orientation of each third (relative to femur) was measured (Figure 3a). The average value among these three measurements was used to determine the sagittal orientation for both VL and VM.

For the frontal plane orientation of the VL and VM, the reference line for the femur was defined on the frontal image containing the largest section of the femur. More specifically, the reference line of the femur was defined by the bisection of two points used to delineate the medial and lateral border of proximal femur and two points delineating the medial and lateral border of the distal femur (Figure 3b). The fiber orientation was measured on the image containing the largest cross-sectional area of the VL and VM, and the reference line of the femur was superimposed on these frontal plane images. The medial and lateral border of the VL and VM was then identified following the fascial planes. Each muscle was divided into thirds and the fiber orientation of each third (relative to femur) was measured (Figure 3b). The average value among these three measurements was used to determine the frontal orientation for both VL and VM.

The frontal plane orientation of the RF was obtained from the image containing the largest section of the femur. The reference line of the femur was defined by the bisection of two points used to delineate the medial and lateral border of proximal femur and two points delineating the medial and lateral border of distal femur. The RF orientation was measured on the image containing the largest cross-sectional area of this muscle, and the reference line of the femur was superimposed on this frontal plane image. The medial and lateral border of the RF was then identified following the fascial plane and a line that bisects the RF was drawn. The frontal plane orientation of the VI was defined by a line parallel to femur.

**Quadriceps Muscle Physiological Cross-Sectional Area Proportions.** To estimate individual quadriceps muscle forces, the physiological cross-sectional areas of each of the vasti was measured from axial images of the thigh (Figure 4). The individual vasti and RF muscle cross sectional areas (CSA) was measured on each image based on the fascial planes separating each muscle. The muscle volume was then calculated from each slice by multiplying the CSA by the slice thickness. The sum
of the volumes of each axial slice was multiplied by the cosine of the muscle pennation angle (as described above) and divided by the fiber length (Equation 1) (Brand et al., 1986). This yielded the physiological cross-sectional area (PCSA) for each muscle. The PCSA proportion for each muscle was calculated as the PCSA of each vasti divided by the sum of the PCSA’s. For all the quadriceps muscles, PCSA measurements were made using SliceOmatic imaging software (TomoVision, Montreal, Quebec, Canada).

\[
\text{PCSA} = \sum \left( \text{CSA} \times \text{thickness of each slice [mm]} \right) \times \cos(\text{pennation angle}) / \text{fiber length (mm)}
\]

(1)

**Patella Flexion Angle.** The patella flexion angle was measured on the sagittal image containing the largest length of the patella and was defined as the angle between the long axis of patella and a line parallel to the long axis of femur (Figure 5a). The long axis of the patella was defined by a line defining the posterior superior point of the patella to the tip of the apex. The long axis of femur was defined as the line that bisected the femur. A second order polynomial curve fitting procedure was used to estimate lever arm data for every degree of knee flexion from 0 to 90 degrees.

**Patellar Ligament Orientation.** Sagittal plane orientation of the patellar ligament was measured from sagittal MR images containing the largest length of the patella. Sagittal plane orientation was defined as the angle between long axis of the tibia and the line defining the patellar ligament (Figure 5b). The long axis of the tibia was defined as the line parallel to the tibial shaft at a level below the tibial plateau.

**Clinical Measurement Input Variable**

The frontal plane orientation of the patellar ligament was defined by the quadriceps angle (Q-angle) and was measured directly from each subject using a standard goniometer. The Q-angle was measured with subjects standing and was quantified as the angle between the line from the mid point of the patella to the anterior superior iliac spine and the line from the midpoint of the patella to the tibial tuberosity.

**Reliability of Model Input Measurements**

All MRI measurements described above were made by the same investigator. To establish the intrarater reliability of all MRI model input measurements, data from five subjects were obtained on two separate occasions at least one week apart. Using the Intraclass Correlation Coefficients (model 2,3), acceptable reliability was established for all variables (reliability coefficient ranged from 0.85–0.97) (Shrout, 1998).

**Model Algorithm**

Once the individual input parameters had been obtained, each was used as an input variable for the patellofemoral joint model algorithm outlined in Figure 1. The specific steps of the algorithm are described below.

**Step 1: Creating a Subject Specific Representation of the Extensor Mechanism.** Using the subject specific MRI and clinical measurements described above, individual subject models were created using SIMM modeling software. The knee extensor model was created in reference to the mid point of patella (i.e., patella convention system: X-axis= anterior/posterior, Y-axis= superior/inferior, and Z-axis= medial/lateral) (Figure 6).

**Step 2: Estimation of Knee Extensor Moment.** While the net knee extensor moment gives a reasonable estimate of quadriceps demand, this value would be underestimated in the presence of muscle co-contraction. To
account for co-contraction during the functional tasks evaluated, an estimate of the knee flexor moment was required. The knee flexor moment (KFM) was obtained from SIMM modeling software. The SIMM lower limb model contains musculo-tendon actuators with information about peak isometric muscle force, optimal muscle-fiber length, pennation angle, and tendon slack length for the muscles of the lower extremity (Besier et al., 2005). In the SIMM software, muscles are represented as a series of three-dimensional vectors that are constrained to wrap over underlying structures. Using a Hill-based model (Besier et al., 2005), the SIMM software estimated the KFM based on the individual’s lower extremity kinematics, velocity of movement, and flexor muscle EMG. The adjusted knee extensor moment (KEMadj) was calculated by adding the net knee joint moment (NJM) and the KFM (Equation 2).

\[ \text{KEM}_{\text{adj}} = \text{NJM} + \text{KFM} \]  

**Step 3: Calculating Resultant Quadriceps Force (Sagittal Plane).** Using Equation 3, the sagittal quadriceps force \( F_Q \) was calculated as the adjusted knee extensor moment \( \text{KEM}_{\text{adj}} \) divided by the lever arm of the quadriceps \( \text{LA} \):

\[ F_Q = \frac{\text{KEM}_{\text{adj}}}{\text{LA}} \]  

**Step 4: Optimization.** The individual sagittal plane vasti forces were estimated in MATLAB (Mathworks, Natik, MA) using a static optimization routine with the criteria of minimizing muscle stress cubed for each muscle (Crowninshield & Brand, 1981; Equation 4).

Based on the three-dimensional orientation and geometry of quadriceps elements from the SIMM model and the sagittal plane vasti forces, the three-dimensional vasti forces with the sum of the projections of the forces at the sagittal plane equal to the resultant sagittal quad force \( F_Q \) were obtained (Equation 5).

\[
\text{Minimize} \left[ \sum (F_{\text{vasti}}/\text{PCSA}_{\text{vasti}})^3 \right]
\]  

\[
F_Q = F_{\text{VL}} + F_{\text{VM}} + F_{\text{RF}} + F_{\text{VI}}
\]  

**Step 5: Calculating Three-Dimensional Resultant Quadriceps Force.** Using Equation 6, the three-dimensional vasti force magnitudes multiplied by the unit vector of each muscle were used to provide the 3D vasti forces (magnitudes and direction).

\[ F_{\text{vasti3D}} = F_{\text{vasti3D magnitude}} \times \text{unit vector } [i \ j \ k] \]

**Step 6: Calculation of 3D Patellar Ligament Force.** Because of the lever action of the patella, the force in the patellar ligament \( F_{\text{PL}} \) cannot be assumed to be equal to the force in the quadriceps tendon \( F_Q \) (Yamaguchi & Zajac, 1989). Therefore the patellar ligament force magnitude was estimated based on data obtained from a cadaveric study using a three-dimensional muscle force application similar to what was employed in the current model (Powers et al., 2006a). Using Equation 8, the magnitude of the patellar ligament force was obtained by multiplying the resultant quadriceps muscle force by the \( F_{\text{PL}}/F_Q \) ratio.

\[ F_{\text{PL3D}} = F_{\text{Q3D}} \times \frac{F_{\text{PL}}}{F_Q} \]

**Step 7: Calculation of Three-Dimensional PFJRF’s.** Before calculating the sum of 3D quadriceps muscle force vectors and patella ligament force, frontal and transverse plane motions were modeled in Matlab software based on the frontal and transverse plane knee angles during the various functional tasks performed (see below for details). All 3D quadriceps force/patellar ligament vectors were rotated in the frontal plane based on instantaneous angles between femur/ tibia and vertical, to account for frontal plane motion of the knee. For transverse plane motion, the 3D patellar ligament force vector was rotated in the transverse plane, based on instantaneous rotation angle, to account for transverse plane motion of the knee.

Using Equation 9, the 3D PFJRF was expressed as the resultant of the 3D quadriceps force vector \( F_{\text{Q3D}} \) and the 3D patellar ligament force vector \( F_{\text{PL3D}} \). The three-dimensional orientation of the patellar ligament force vector was obtained from the SIMM lower extremity model and adjusted based on knee kinematics of each subject.

\[ 3D \text{PFJRF} = F_{\text{Q3D}} + F_{\text{PL3D}} \]
Data Analysis

Calculation of PFJRF’s. Calculation of the PFJRF was made using the model algorithm described above. The reference point (origin) of the PFJRF was defined as the mid point of patella and the resultant PFJRF was resolved into components along 3 orthogonal axes as illustrated in Figure 6 (anterior/posterior, medial/lateral, superior/inferior). Peak forces during each task were identified and used for statistical analysis.

Statistical Analysis

To evaluate gender differences in PFJRF’s across tasks, $2 \times 4$ (gender $\times$ task) ANOVAs were performed. Separate ANOVAs were employed for each dependent variable of interest (peak resultant PFJRF, peak anterior/posterior force, peak medial/lateral force, peak superior/inferior force). For all ANOVA tests, significant main effects were reported if there were no significant interactions.

Results

The ANOVA results revealed no statistically significant differences (no gender effect, no gender $\times$ task interaction) between males and females for any of the variables of interest (peak resultant PFJRF, peak anterior/posterior force, peak medial/lateral force, peak superior/inferior force). When averaged across genders, the peak resultant PFJRF was the highest during running ($58.2 \pm 2.7$ N/kg-bwt), followed by stair ascent ($33.9 \pm 2.3$ N/kg-bwt), stair descent ($27.9 \pm 1.9$ N/kg-bwt), and walking ($10.1 \pm 1.2$ N/kg-bwt) (Figure 7).

For all conditions tested, the direction of the PFJRF was always in the posterior, superior, and lateral directions. The posterior component of the PFJRF always had the greatest magnitude, followed by superior and lateral components (Table 1).

Discussion

The purpose of this paper was to describe an imaging based, subject specific model to quantify PFJRF’s, and to evaluate the model in a group of healthy individuals while performing various functional tasks. The peak resultant PFJRF’s observed during the functional tasks evaluated ranged from 10.1 N/kg-bwt for walking to 58.2 N/kg-bwt for running. On average, the PFJRF’s exhibited by the female subjects were very similar to those PFJRF’s exhibited by the male subjects (Figure 7, Table 1).

The resultant PFJRF’s during both walking and running demonstrated a biphasic pattern (peaks in early stance and early swing; Figure 7). In both cases, the forces during initial stance were always greater than early swing which reflects the higher quadriceps demand during weight acceptance. The peak PFJRF’s during running were 5.8 times greater than walking. Our resultant PFJRF’s obtained for walking and running are similar.

Figure 7 — Averaged resultant PFJRF’s for 10 males and 10 females during walking (top-left), running (top-right), ascend (bottom-left), and descend stairs (bottom-right; male-solid, female—dash line).
to values reported previously using two dimensional modeling approaches. For example, our average PFJRF value of 10.1 N/kg-bwt for walking is similar to that reported by Brechter and Powers (11.9 N/kg-bwt) and Ward and Powers (10.8 N/kg-bwt) (Heino-Brechter & Powers, 2002; Ward & Powers, 2004). For running, Scott et al. reported peak PFJRF’s of 54.3 N/kg-bwt which was similar to our average value of 58.2 N/kg-bwt (Scott & Winter, 1990).

For stair ascent, the peak PFJRF occurred at approximately 30% of the stance phase. This peak corresponds to the point at which bodyweight is being transferred to the supporting limb. For stair descent, a biphasic pattern was observed during the stance phase (Figure 7). The first peak corresponded to initial loading while the second peak reflects the lowering of the center of mass before initial contact of the opposite limb.

In comparing our stair data to previous two-dimensional modeling studies, Heino-Brecher and Powers reported an average peak PFJRF of 36.2 N/kg-bwt during stair ascent which was slightly higher than our average value of 33.9 N/kg-bwt. Heino-Brecher and Powers also reported an average peak PFJRF of 34.0 N/kg-bwt during stair descent which also is higher than our average value of 27.9 N/kg-bwt (Brechter & Powers, 2002). Despite the observed differences in peak forces between studies, the overall patterns of the PFJRF during stair ascent and descent were similar.

A unique contribution of the current modeling approach was that the direction of the resultant PFJRF could be evaluated. Consistent with previously reported cadaver simulations (Singerman et al., 1994), the PFJRF components were always in the posterior, superior, and lateral directions. Our results demonstrated that a majority of the PFJRF is directed in the posterior direction (i.e., compressive force), when compared with the superior and the lateral components. The high compressive force reflects the posteriorly directed line of pull of the quadriceps muscles (particularly the vastus medialis and lateralis) with respect to the femur (Figure 3a).

A superior component of the PFJRF would be expected based on the fact that the quadriceps muscles insert into the proximal quadriceps tendon. Given as such, quadriceps muscle contraction would create a superior pull on the patella during dynamic tasks (Lin et al., 2003). Although the superior pull of the patella would be counterbalanced by the inferior pull of the patella ligament, the ratio of quadriceps force to patella ligament force used as a part of our model was always greater than 1.0 (Powers et al., 2006a). As a result, a net superior force would be expected.

The laterally directed force was the smallest of the PFJRF components. The lateral force is partly dictated by the angle between the quadriceps and patellar ligament force vector in the frontal plane (i.e., Q-angle). In the current study, this angle was relatively small (average of 14 degrees) across all subjects. Given as such, one would not expect large laterally directed forces. The laterally directed force also is influenced by the larger physiological cross-sectional area proportion of the VL with increases in the Q-angle, a larger discrepancy in volume). An increase in lateral force would be expected with increases in the Q-angle, a larger discrepancy in cross-sectional area between the VL and VM, and/or any increase in frontal plane motion at the knee during functional tasks.

In light of the findings reported in the current study, there are limitations that need to be acknowledged. Although most of the parameters used in our model were subject specific, the knee flexor moment was estimated using nonspecific data nested within the SIMM modeling software. However, a post hoc analysis revealed that the contribution of the knee flexor moment to the total knee extensor moment was relatively low and equal for both groups of subjects (approximately 10% of the total joint moment). As such, any error in PFJRFs associated with the estimation of the knee flexor moment would be expected to be relatively small. Second, our model did not consider the influence of contact geometry in the modeling PFJRFs. Although it has been reported that reasonable estimates of PFJRFs can be obtained using models that do not consider patellofemoral joint contact geometry (Powers et al. 2006b), some error would be expected. Third, we did not perform a sensitivity analysis are part of the current study. As such, it is not known

Table 1: Peak patellofemoral forces and peak PFJRF components during functional activities (N/kg-bwt), mean (SD)

<table>
<thead>
<tr>
<th>Activity</th>
<th>Resultant PFJRF</th>
<th>Posterior force</th>
<th>Superior force</th>
<th>Lateral force</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Male</td>
<td>Female</td>
<td>Ave</td>
<td>Male</td>
</tr>
<tr>
<td>Walking</td>
<td>10.2</td>
<td>10.0</td>
<td>10.1</td>
<td>9.4</td>
</tr>
<tr>
<td>(1.4)</td>
<td>(1.0)</td>
<td>(1.2)</td>
<td>(0.8)</td>
<td>(0.4)</td>
</tr>
<tr>
<td>Descend stairs</td>
<td>29.4</td>
<td>26.4</td>
<td>27.9</td>
<td>21.7</td>
</tr>
<tr>
<td>(2.3)</td>
<td>(1.5)</td>
<td>(1.9)</td>
<td>(1.8)</td>
<td>(2.0)</td>
</tr>
<tr>
<td>Ascend stairs</td>
<td>36.2</td>
<td>31.6</td>
<td>33.9</td>
<td>26.7</td>
</tr>
<tr>
<td>(2.4)</td>
<td>(2.2)</td>
<td>(2.3)</td>
<td>(2.1)</td>
<td>(0.9)</td>
</tr>
<tr>
<td>Running</td>
<td>60.1</td>
<td>57.3</td>
<td>58.2</td>
<td>50.0</td>
</tr>
<tr>
<td>(2.6)</td>
<td>(2.8)</td>
<td>(2.7)</td>
<td>(1.8)</td>
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</table>
how inaccuracies in measuring or estimating various modeling parameters would affect model predictions of PFJRFs.

An imaging-based, subject specific model to quantify PFJRF’s has been presented. Model testing revealed that PFJ loading varies considerably based on the task evaluated. No differences in PFJ loading were found between healthy males and females. When broken down into orthogonal components, the PFJRF’s were always in the posterior, superior, and lateral directions. Future studies will be directed toward evaluating differences in PFJRF’s between individuals with patellofemoral pain and pain-free controls to determine if loading differences exist in these populations.

References


