Normative Three-Dimensional Patellofemoral and Tibiofemoral Kinematics: A Dynamic, *in Vivo* Study

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***Abstract*—In order to advance biomechanical modeling, knee joint implant design and clinical treatment of knee joint pathology, accurate *in vivo* kinematic data of the combined patellofemoral and tibiofemoral joint during volitional activity are critical. For example, one cause of the increased prevalence of anterior knee pain in the female population is hypothesized to be altered tibiofemoral kinematics, resulting in pathological patellofemoral kinematics. Thus, the objectives of this paper were to test the hypothesis that knee joint kinematics vary based on gender and to explore the correlation between the 3-D kinematics of the patellofemoral and tibiofemoral joints. In order to accomplish these goals, a large ($n = 34$) normative database of combined six degree of freedom patellofemoral and tibiofemoral kinematics, acquired noninvasively during volitional knee extension-flexion using fast-PC (dynamic) magnetic resonance imaging, was established. In this normative database, few correlations between tibiofemoral and patellofemoral kinematics were found. Specifically, tibial external rotation did not predict lateral patellar tilt, as has been stated in previous studies. In general, significant differences could not be found based on gender. Further investigation into these relationships in the presence of pathology is warranted.**

***Index Terms*—Femur, gender, healthy, kinematics, knee, patella, patellofemoral, tibia, tibiofemoral.**

I. Introduction

THE quality of biomechanical models, accuracy of clinical diagnosis, and fidelity of joint implants are all dependent upon the quality of the *in vivo* experimental data used in their creation. For this reason there have been a host of experimental and modeling studies focused on knee joint dynamics. Yet, current knowledge of this joint is limited by the fact that complete six degree of freedom kinematics of this joint, inclusive of both the patellofemoral (PF) and tibiofemoral (TF) joints, have been presented for only 5 knee joints [1], [2]. Since alterations in TF kinematics have been hypothesized to result in pathological PF kinematics [3], a combined PF and TF joint study is key to understanding pathologies such as anterior knee pain and patellar maltracking.

For the TF joint, past research has focused on the finite helical axis direction and location [4], [5], the existence of the screw home mechanism [6], [7], the effects of ligament loss [5], [6], total knee arthroplasty [8], [9] and cartilage contact patterns [10]. Gait analysis is commonly used to collect *in vivo* data, but is susceptible to errors based on well documented skin motion artifacts [11]. Biplane radiography [12] has proven to be an accurate tool for studying TF kinematics. However, the required tracking of beads, implanted within the bone, narrows the populations that can be studied. Further, accurate *in vivo* patellar tracking is difficult with this technique. Single-plane fluoroscopy has provided strong results for sagittal-plane kinematics of total TF knee replacement [8], but tracking motion in the other cardinal planes is not as accurate [13]. Recent studies registering fluoroscopic images to CT bone models were able to eliminate the need for directly tracking metallic beads or implants, but out-of-plane tibiofemoral accuracies (translational: 8.4 mm to 22.3 mm; rotational: 1.3° to 3.1°; assuming all data fell within 2 standard deviations of the reported mean error) are poor [14].

Numerous PF disorders are thought to arise from abnormal PF kinematics, yet defining these kinematics has been quite difficult. An excellent review [15] summarized 15 *in vitro* and 13 *in vivo* (primarily 2-D) PF kinematic studies with population sizes ranging from 2–32 specimens and 1–20 knees, respectively. Katchburian *et al.* [15] noted that the interstudy variability stemmed from various study design differences, including TF orientation, coordinate system definitions, muscular loading conditions, study type (static, quasi-static or dynamic), and motion direction (flexion versus extension). The potential variability resulting from the gender of the subjects studied was not discussed.

Potential gender-based kinematics differences [3], [16], [17] may be key to understanding the higher prevalence of anterior knee pain in the female population. Structural differences, muscular strength differences, sociologic factors, and hormonal factors have all been hypothesized to lead to increased anterior knee pain among women [3]. Yet, a direct link between these factors and anterior knee pain or altered PF kinematics has not been shown and few studies have investigated gender-based differences in PF kinematics. A single study did directly investigate gender differences at the PF joint and found increased static PF contact pressures in female cadaver specimens [10]. TF kinematic and kinetic gender-based differences have been shown using gait analysis during various tasks [18]–[21]. Such TF kinematic variations have been hypothesized to lead to PF instability and anterior knee pain [3]. Unfortunately, few studies have quantified the kinematic relationships between these two joints. Li *et al.* [22] found that hamstrings co-contraction under static loading conditions increased tibial posterior translation and external rotation relative to the femur with an accompanying

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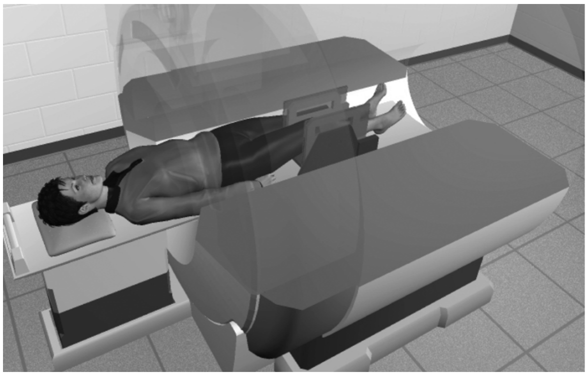

Fig. 1. Subject position within the imager.

increase in PF contact. In addition, anterior cruciate ligament loss has been shown to alter both TF and PF kinematics [23].

Thus, the purpose of this paper was to explore whether excessive tibial external rotation results in PF maltracking by correlating PF and TF kinematics and to test the hypothesis that knee joint kinematics vary with gender. In order to accomplish these goals, a large $(n = 34)$ normative database of combined six degree of freedom patellofemoral and tibiofemoral kinematics, acquired noninvasively during volitional knee extension-flexion using fast-PC (dynamic) magnetic resonance imaging (MRI), was established.

II. Methods

Twenty-five healthy subjects (14 female, 11 male; $\mathrm{age} = 26.7 \pm 8.8$ years; $\mathrm{weight} = 67.5 \pm 12.7$ kg, $\mathrm{height} = 172.3 \pm 7.5$ cm) participated in this Institutional Review Board-approved study and gave informed consent upon entering the study. If time allowed, both knees were imaged resulting in data from 34 knees (9 knee pairs, 18 Left, and 16 Right; 20 female and 14 male). Subjects were excluded from this study if they had a history of knee problems or pain, were diagnosed with any knee pathology, had previous knee joint surgery or had any contraindications to having an MRI scan.

Subjects were placed supine in a 1.5-Tesla MR imager (CV-9.1M4 or LX-9.1M4; GE Medical Systems, Milwaukee, WI) (Fig. 1). A cushioned wedge was placed under the thigh without contacting the posterior knee such that full extension of the leg was attainable. Cushioning supported the head, neck and lower back. A custom-designed coil holder [24] was used to stabilize two phased array torso coils medial and lateral to the knee. An optical trigger, placed on the imaging bed beneath the ankle, was used to synchronize data collection to the motion cycle.

Dynamic MRI sequences acquired the data from which the 3-D kinematics were determined. During dynamic imaging, subjects were asked to extend and flex their knee from maximum attainable flexion to full extension and back at 35 cycles/minute to the beat of an auditory metronome. Prior to data collection, subjects practiced the task until they could comfortably repeat the motion. A dynamic exam involved three movement trials. During the first trial, anatomic images at the mid-patellar level were collected using an axial fast cine gradient echo sequence (Fastcard-1, Table I). This image set was used to select the proper sagittal-oblique plane for the next dynamic trial. A full fast-PC (Fast-PC, Table I) data set was collected using a single sagittal-oblique plane that was perpendicular to the posterior femoral condyles, bisected the patella and did not pass through the popliteal artery (Fig. 2). The last dynamic trial was a fastcard acquisition (Fastcard-3, Table I) at three axial slice locations (**patellar image**: through the mid-patella, **femoral image**: through the femoral epicondyles, and **tibial image**: through the tibia, approximately 2 cm below the tibial insertion of the patellar tendon). For each subject, these slice locations were selected from the fast-PC anatomic image representing full extension (the most extended position).

TABLE I
KEY IMAGING PARAMETERS FOR THE DYNAMIC SEQUENCE

	Fastcard-1 [41]	Fast-PC [42]	Fastcard-3
Imaging Plane	Axial	Sag. or Sag.-Obl	Axial
Number of Slices	1	1	3
Repetition Time (ms)	5.0	9.2	5.0
Echo Time (ms)	1.3	4.4	1.3
Flip Angle (deg)	20	20	20
Field of View (cm)	24×24	30 × 22.5	24 × 24
Slice Thickness (mm)	10	10	10
Matrix (pixels)	256×128	256×128	256×128
Number of Excitations	1	2	1
Phase Encoding Acquisitions	16	2	16
Temporal Resolution (ms)	80	73.6	80
Imaging Time (m:s)	0:17	2:48	0:51

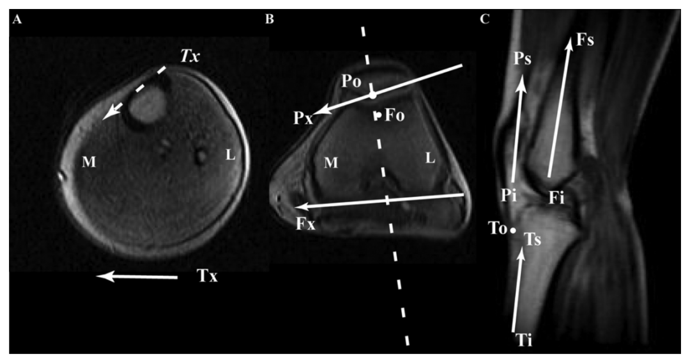

Fig. 2. Anatomically based coordinate systems. All axes were defined in the image representing the fully extended position for that subject. For all three bones, the x-axis was defined first. (A) Tibial image at full extension (20 mm below the tibial patellar tendon insertion). The anterior edge of the tibia defined $\boldsymbol{Tx}$. The average angle between this line and the femoral x-axis ($\mathbf{Fx}$) in the normative population was 41°. To ensure a medial-lateral x-axis, the tibial x-axis ($\mathbf{Tx}$) was defined as a unit vector rotated 41° away from $\boldsymbol{Tx}$ in the axial plane for all subjects. (B) Femoral image (at the level of the epicondyles). Note, this image was intentionally selected not to be in full extension so that the mid-patella could be visualized. The posterior edge of the femoral condyles in the femoral image defined $\mathbf{Fx}$. The patellar lateral posterior edge in the patellar image defined $\mathbf{Px}$. The dashed line represents the plane used for the fast-PC data collection. (C) Sagittal-oblique fast-PC anatomical images at full extension. A temporary y-axis ($\mathbf{Ty_{temp}}$, $\mathbf{Py_{temp}}$, $\mathbf{Fy_{temp}}$) was established for each bone in the sagittal image. $\mathbf{Ty_{temp}}$ was defined as a unit vector directed along the tibial anterior edge (TiTs), $\mathbf{Py_{temp}}$ was defined as a unit vector directed along the patellar posterior edge (PiPs), and $\mathbf{Fy_{temp}}$ was defined as a unit vector that bisected the angle subtended by the femoral anterior and posterior edges (FiFs). The z-axis for each bone was then defined as the unit cross-product between that bone's x-axis and its temporary y-axis. Finally, the y-axis was defined as the unit cross-product between that bone's z-axis and x-axis.

As previously described [24], 3-D rigid body rotations and translations (jointly referred to as attitudes) of the femur, tibia and patella were quantified through integration of the fast-PC velocity data. It is important to note that although the fast-PC acquisition was based on a single imaging plane (due to time constraints), the 3-D velocity data allowed the attitude of all three bones to be accurately tracked three-dimensionally throughout the movement. To account for differences in translation, due skeletal size variations across subjects, the translations were scaled by the ratio of the average epicondylar width for all 34 knees (77.3 mm) to the epicondylar width for that individual knee. Unlike earlier cine MRI experiments [25], these kinematics were based on an anatomical coordinate system (Fig. 2), which was identified in a single time frame only. The fast-PC data were then used to track the bones' changes in attitude through all time frames. To reduce variability within the kinematics due to the sensitivity of bone shape to the bone's attitude within the imaging plane [26], clear rules were created for identifying imaging planes and establishing coordinate systems. As part of this study, improvements to the original coordinate systems [24] were made. The original tibial anatomical landmarks were selected in the sagittal plane only. As this may cause imprecision in quantifying out-of-plane motion, this coordinate system is now selected in two orthogonal planes (Fig. 2). In addition, the patellar superior-inferior was previously defined by its most superior and inferior points. The posterior flat edge of the patella is more consistent to define and is now used for the patellar superior-inferior line.

The interdependence of the 12 parameters defining the PF and TF kinematics (6 translations and 6 rotations) was evaluated using a Pearson's linear correlation

To create population averages and analyze differences between groups, each kinematic variable was interpolated to single degree knee angle increments. Certain subjects achieved greater or lesser than the full 40° of range of motion, depending on their individual leg length. Thus, not all subjects are represented in the average kinematics at all knee angles. Data representing 3 or fewer subjects were eliminated from the total group average. In order to explore the relationship between knee joint kinematics and gender, the kinematic database was divided into two subpopulations (male and female) and the value of each kinematic parameter for both joints was compared. Statistical differences were investigated at each knee angle using a 2-tailed student t-test, assuming unequal population sizes and unequal variances. Prior to any statistical analysis, the data at each knee angle were checked for normality. All data are presented for the extension portion of the movement only.

As a follow-up to previous investigations into the accuracy of fast-PC MRI [27], [28], the PF attitude in the axial plane was measured based on the visual identification of landmarks in the fastcard images [29]. This was then compared to the results derived from the fast-PC data for two separate time points. Two subjects, K#13 and K#24, were selected for this analysis based on the fact that they had the most lateral and medial, respectively, absolute value of patellar tilt (at full extension) and slope of patellar tilt versus knee angle. At full extension the patellar and femoral images were used to define the patellar and femoral 2-D attitudes, respectively. These were then combined to define the PF attitude. The second time point analyzed was early extension, selected such that the view of the patella in the femoral image visually matched the patellar image at full extension as closely as possible. The PF attitude was then defined.

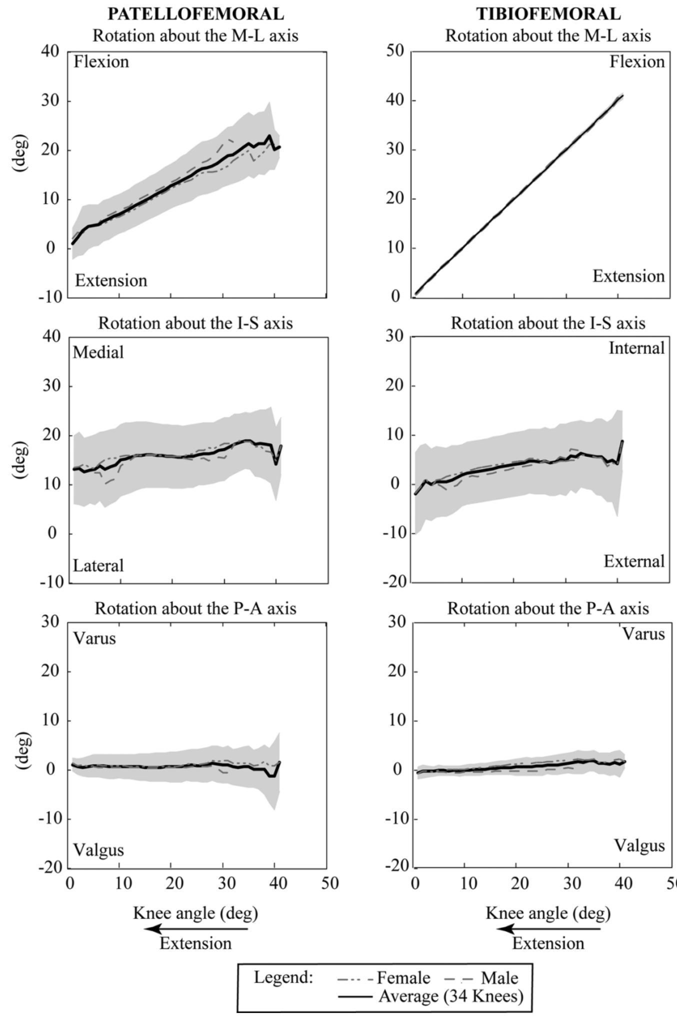

Fig. 3. Orientation of the patella and tibia relative to the femoral coordinate system. The average of all subjects is shown as a black line with $\pm 1\mathrm{SD}$ shown on either side in grey shading. The average kinematics for the male and female populations are shown with a grey dashed line and grey dashed-dotted line, respectively (SD not given). All axes are drawn with the same range for the y-axis and for the x-axis. The knee angle and rotation about the tibial $\mathrm{M\text{-}L}$ axis are approximately the same, so a very small standard deviation and a 1:1 slope are expected.

III. Results

The kinematics of the PF and TF joints (Tables V and VI) were similar to those previously measured using cine-PC MRI [24], but patellar rotation was more varus and the translation of the patellar origin was more superior and posterior. In comparing the average range of kinematic values to the range of values obtained during the swing phase of Lafortune's [1] study (one of only two studies that have reported combined 3-D PF and TF kinematics) the rotational data were similar. The translational data had different ranges (Table II).

All data were determined to be normally distributed $(P < 0.001)$. Based on the Pearson correlation, none of the kinematic variables could be significantly predicted

TABLE II
TABULATED RANGES OF MOTION FOR THE AVERAGE PF AND TF KINEMATICS DURING TF EXTENSION FOR THE CURRENT STUDY AND THE SWING PHASE FROM THE WORK OF LAFORTUNE [1]

	Current (34 subjects)	**Lafortune (5 subjects)**
Rotations		
TF Flexion-Extension	43° Extension	60° Extension
TF Internal-External	11° External	6° External
TF Varus-Valgus	1° Valgus	5° Varus
PF Flexion-Extension	20° Extension	17° Extension
PF Lateral-Medial Tilt	6° Medial	Inconsistent (3 subjects)
PF Varus-Valgus	4° Valgus	6° Varus
Translations		
TF Medial-Lateral	11 mm Lateral	4 mm Lateral
TF Superior-Inferior	16 mm Superior	7 mm Superior
TF Anterior-Posterior	37 mm Anterior	14 mm Anterior
PF Medial-Lateral	8 mm Lateral	1 mm Medial
PF Superior-Inferior	23 mm Superior	40 mm Superior
PF Anterior-Posterior	3 mm Posterior	19 mm Anterior

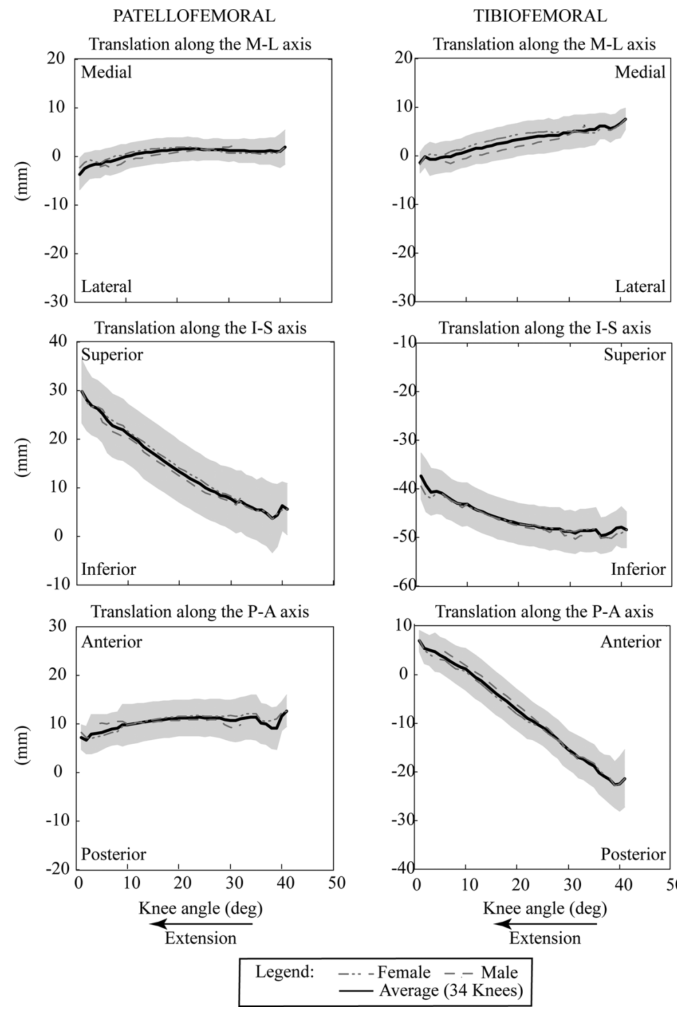

Fig. 4. Normalized patellar and tibial displacement relative to the femoral coordinate system. The average of all subjects is shown as a black line with $\pm 1\mathrm{SD}$ shown on either side in grey shading. The average kinematics for the male and female populations are shown with a grey dashed line and grey dashed-dotted line, respectively (SD not given). All data are normalized by the ratio of the average normative epicondylar width (77.3 mm) to the individual subject's epicondylar width. All axes are drawn with the same range for the y- and x-axes.

by TF internal-external rotation (Table III, data column 5: "Rotation-Tibiofemoral-I/S" has no value over 0.500). Over 50% of the variability in PF extension-flexion along with PF and TF inferior-superior displacement could be explained by TF extension-flexion (Table III, data column 4: "Rotation-Tibiofemoral-L/M). PF lateral-medial tilt and TF lateral-medial along with inferior-superior displacement had weaker correlations to TF extension-flexion (predictability ranged from 20% to 50%).

Few significant differences were found between the male and female populations (Table IV). Small, but significant differences were found in TF lateral-medial translations (for knee angles ranging from 13° to 23°) and TF varus-valgus rotation (for knee angles ranging from 12° to 28°). The latter agreed with previous gait analysis studies [18]–[20]. The average absolute differences across all knee angles between the populations ranged from 0.2°-2. 5° and 1.3-1.9 mm (Table IV) and were more pronounced in the PF kinematics.

The accuracy of tracking out-of-plane motion using fast-PC MRI was supported by the current results. The 26.9° and -3.6 mm difference in patellar tilt and shift, respectively, between K#13 and K#24, derived from the fastcard full-extension images matched the 27.1° and -3.6 mm difference determined by the fast-PC data (Fig. 5). The errors in the full-extension image tended to be lower, as compared to the early-extension image.

IV. DISCUSSION

One of the more interesting outcomes of the present study was the fact that only 28% of the variability in PF lateral-medial tilt and only 12% of the variability in PF inferior-superior translation could be explained by TF internal-external rotation. These findings do not support earlier clinical findings that external tibial rotation leads to PF instability [30], which is typically associated with excessive patellar lateral tilt and translation relative to the femur. Yet, in pathology, excessive tibial external rotation may apply larger forces on the patella, resulting in abnormal lateral tilt or translation. The locations of the patellar tendon insertions and the tendon elasticity, both of which could be affected by pathology, would dictate the amount of tibial external rotation that would be transferred to the patella. Future studies are planned to further explore these relationships in pathology.

Unlike the ankle, where rotations are reported to be coupled [31], the knee appears not to have coupled rotations at either joint. The fact that only 60% of the variability in PF extension could be explained by TF extension supports the fact that the patellar tendon is a nonrigid link between the two, which allows the quadriceps to extend both joints at different rates. The strong correlation between PF extension-flexion and PF inferior-superior displacement is due to the quadriceps extending and translating the patella. The higher correlation between TF superior

TABLE III

PEARSON CORRELATION R VALUES ($N = 392$). FOR EACH OF THE 34 SUBJECTS THERE WERE ~ 12 EVENLY (TEMPORALLY) SPACED TIME FRAMES DURING EXTENSION, WHICH IN TOTAL PROVIDED 392 DATA POINTS FOR THIS ANALYSIS. IF MORE THAN 50% OF THE VARIABILITY IN ONE KINEMATIC PARAMETER COULD BE EXPLAINED BY ANOTHER, THE VALUE WAS BOLDED. IF THE PREDICTABILITY WAS MORE THAN 20%, BUT LESS THAN 50%, THE VALUE WAS BOLDED AND ITALICIZED. ALL OTHERS WERE GREYED. **ROTATION** L/M; I/S; P/A ROTATIONS ABOUT THE LATERAL-MEDIAL, INFERIOR-SUPERIOR AND POSTERIOR-ANTERIOR AXES, RESPECTIVELY**TRANSLATION** L/M; I/S; P/A TRANSLATIONS ALONG THE LATERAL-MEDIAL, INFERIOR-SUPERIOR POSTERIOR-ANTERIOR AXES, RESPECTIVELY

			Rotation						Translation					
			Patellofemoral			Tibiofemoral			Patellofemoral			Tibiofemoral		
			L/M	I/S	P/A	L/M	I/S	P/A	L/M	I/S	P/A	L/M	I/S	P/A
Rotation	PF	L/M		0.017	0.034	**0.633**	0.025	0.180	0.127	**0.533**	0.016	0.141	0.197	**0.617**
		I/S	0.017		0.023	0.037	***0.283***	0.043	0.028	0.064	0.000	0.074	0.017	0.013
		P/A	0.034	0.023		0.001	0.013	0.000	0.017	0.023	0.045	0.053	0.002	0.037
	TF	L/M	**0.633**	0.037	0.001		0.086	0.120	0.166	**0.606**	0.034	***0.330***	***0.209***	**0.811**
		I/S	0.025	***0.283***	0.013	0.086		0.001	0.140	0.033	0.007	0.145	0.017	0.034
		P/A	0.180	0.043	0.000	0.120	0.001		0.001	0.086	0.013	0.074	0.002	0.128
Translation	PF	L/M	0.127	0.028	0.017	0.166	0.140	0.001		0.153	0.025	0.127	0.113	0.143
		I/S	**0.533**	0.064	0.023	**0.606**	0.033	0.086	0.153		0.045	0.105	***0.337***	***0.411***
		P/A	0.016	0.000	0.045	0.034	0.007	0.013	0.025	0.045		0.025	0.016	0.003
	TF	L/M	0.141	0.074	0.053	***0.330***	0.145	0.074	0.127	0.105	0.025		0.041	***0.390***
		I/S	0.197	0.017	0.002	***0.209***	0.017	0.002	0.113	***0.337***	0.016	0.041		0.184
		P/A	**0.617**	0.013	0.037	**0.811**	0.034	0.128	0.143	***0.411***	0.003	***0.390***	0.184	

TABLE IV

AVERAGE OF AVERAGE ABSOLUTE DIFFERENCE BETWEEN MALE AND FEMALE FOR EXTENSION

	Patellofemoral	Tibiofemoral
	Average (SD)	*Average (SD)*
Rotations in (deg)		
***lateral-medial tilt* (θ1)**	2.5 (1.9)	0.2 (0.4)
***PF lateral-medial tilt* (θ2) *or* *TF external-internal rotation* (θ2)**	1.5 (1.5)	1.2 (0.7)
***extension-flexion* (θ3)**	1.1 (1.0)	1.1 (0.6)
Translations in (mm)		
***lateral-medial* (X)**	1.5 (1.1)	1.9 (0.7)
***inferior-superior* (Y)**	1.7 (1.0)	1.3 (2.0)
***posterior-anterior* (Z)**	1.7 (1.2)	1.5 (0.7)

displacement (as compared to anterior displacement) with TF extension indicated that the axis of rotation for TF joint was more superior rather than anterior relative to the tibial origin.

The current study could not support the hypothesis that PF and TF joint kinematics vary with gender. Potentially, a larger population could allow significant differences to be found. Since the pattern of the kinematic data demonstrated that the male and female kinematic profiles were intertwined, this is unlikely. Also, the similarity between men and women does not support the concept that women are more prone to anterior PF pain syndrome due to altered TF kinematics [3], [10]. It is possible that altered bone shape could predispose women to greater contact pressures [10], which could lead to anterior knee pain syndrome, but that is beyond the scope of the current work.

One major outcome of this study was the formation of a large normative database for the six degree of freedom kinematics of the combined patellofemoral and tibiofemoral joints. To date this is the largest database of its type. The low intrasubject and intersubject variability, high accuracy, and clearly defined coordinate systems make these data (Tables V and VI) readily available for experimental comparison, modeling inputs and device design. The discontinuity seen in many of the kinematic plots at $\sim 38^\circ$ knee flexion was not seen in any individual subject, but was due to the fact that not all subjects reached this flexion angle. This was a follow-up to the studies that were the first to report 3-D PF kinematics quantified noninvasively and *in vivo* using cine-PC MRI [24], [27]. These earlier studies measured TF kinematics, but did not fully report them. The small differences in kinematics between the two studies are likely due to an increased extensor load required for the present study. The larger population size did allow for clearer insights into a much argued variable, patellar tilt. Three patterns of patellar tilt (no tilt, slight medial tilt and lateral tilt) were found, although the average tilt was lateral during extension. In the current study, the patella is primarily outside the femoral groove, thus, the distinct PF kinematic patterns across subjects are most likely due to alterations in muscular control. Therefore, the emulated muscle force in cadaver studies [22] and the decision to exclude active muscle contraction in static studies [25] may force the knee joint to follow a movement pattern inconsistent with that of active muscular control and prevent true joint motion from being studied. The smaller populations of the previous investigations may also have prevented the distinct patterns from coming to light, causing some studies to report lateral and others to report medial PF tilt in terminal extension.

In general, the largest differences between the current study and past studies were in the translational degrees of freedom, which is likely due to the variation in the physical locations of the origins. For example, in the Lafortune study [1], the tibial origin was at the tibial spine, placing it much further posterior (and closer the TF finite helical axis) than the origin used in the current study. Thus, the comparatively smaller translations in the Lafortune study were as expected. In a similar manner, the patellar origin used in the Lafortune study appeared to be much further anterior and slightly inferior than that used in the current study. This placed the origin further from the PF finite helical axis. Thus, the comparatively larger superior and ante-

TABLE V
AVERAGE RAW DATA FOR KINEMATIC ROTATIONS AND THE NUMBER OF SUBJECTS INCLUDED IN THE AVERAGE AT EACH KNEE ANGLE INCREMENT. THETA1 – ROTATION ABOUT THE L-M AXIS, THETA2 – ROTATION ABOUT THE I-S AXIS, AND THETA3- ROTATION ABOUT THE P-A AXIS. ALL ANGLES ARE IN DEGREES. THE STANDARD DEVIATION IS IN PARENTHESES

		Patellofemoral			Tibiofemoral		
# Subjects	Knee Angle	Theta1	Theta2	Theta3	Theta1	Theta2	Theta3
7	1	0.8 (3.2)	13.1 (6.9)	1.0 (1.4)	0.5 (0.7)	-2.0 (8.2)	-0.5 (1.3)
9	2	2.0 (3.9)	13.3 (7.4)	0.6 (1.7)	1.4 (1.3)	-0.9 (8.6)	-0.2 (1.3)
12	3	3.5 (5.0)	12.5 (7.0)	0.5 (1.9)	2.6 (1.2)	0.4 (7.5)	-0.2 (1.1)
16	4	4.3 (4.5)	13.0 (6.8)	0.4 (1.9)	3.8 (0.6)	-0.2 (7.3)	-0.3 (1.0)
22	5	4.6 (4.2)	13.0 (7.2)	0.7 (2.1)	5.0 (0.2)	0.5 (7.2)	-0.2 (1.1)
24	6	4.7 (4.2)	13.7 (6.9)	0.7 (2.2)	6.0 (0.2)	0.5 (7.0)	-0.2 (1.0)
26	7	5.5 (4.2)	13.1 (7.7)	0.6 (2.2)	7.0 (0.2)	0.4 (7.0)	-0.2 (1.0)
26	8	6.0 (4.1)	13.6 (7.5)	0.5 (2.3)	8.0 (0.2)	0.8 (7.0)	-0.2 (1.1)
29	9	6.6 (4.0)	13.9 (7.3)	0.5 (2.3)	9.0 (0.2)	1.2 (6.6)	-0.2 (1.1)
33	10	7.0 (4.0)	14.9 (7.3)	0.5 (2.3)	10.0 (0.2)	1.8 (6.3)	-0.1 (1.0)
33	11	7.4 (4.0)	15.3 (7.3)	0.5 (2.4)	11.0 (0.2)	2.1 (6.3)	0.0 (1.1)
33	12	7.9 (4.0)	15.5 (7.1)	0.5 (2.4)	12.0 (0.2)	2.4 (6.3)	0.0 (1.1)
34	13	8.6 (4.2)	15.8 (6.9)	0.4 (2.5)	13.0 (0.2)	2.4 (6.4)	0.1 (1.2)
34	14	9.1 (4.2)	15.9 (6.8)	0.4 (2.6)	14.0 (0.2)	2.6 (6.4)	0.2 (1.2)
34	15	9.7 (4.2)	16.0 (6.7)	0.4 (2.6)	15.0 (0.2)	2.8 (6.4)	0.3 (1.3)
34	16	10.2 (4.3)	16.0 (6.5)	0.4 (2.7)	16.0 (0.2)	3.0 (6.4)	0.3 (1.3)
34	17	10.8 (4.3)	15.9 (6.5)	0.4 (2.8)	17.0 (0.2)	3.2 (6.4)	0.4 (1.4)
34	18	11.4 (4.4)	15.8 (6.4)	0.4 (2.8)	18.0 (0.2)	3.4 (6.4)	0.5 (1.5)
34	19	11.9 (4.4)	15.7 (6.4)	0.4 (2.9)	19.0 (0.2)	3.6 (6.4)	0.5 (1.5)
34	20	12.5 (4.5)	15.7 (6.4)	0.4 (2.9)	20.0 (0.3)	3.8 (6.5)	0.6 (1.6)
34	21	13.1 (4.6)	15.6 (6.4)	0.5 (2.9)	20.9 (0.3)	4.0 (6.5)	0.7 (1.7)
34	22	13.7 (4.7)	15.5 (6.4)	0.5 (3.0)	21.9 (0.3)	4.1 (6.5)	0.8 (1.7)
33	23	14.2 (4.8)	15.7 (6.4)	0.7 (3.0)	22.9 (0.3)	4.5 (6.5)	0.7 (1.7)
32	24	14.7 (5.0)	15.7 (6.4)	0.7 (3.1)	23.9 (0.3)	4.5 (6.6)	0.7 (1.8)
31	25	15.6 (5.1)	16.1 (6.2)	0.7 (3.1)	24.9 (0.3)	4.5 (6.6)	0.7 (1.9)
30	26	16.1 (5.2)	16.2 (6.3)	0.7 (3.2)	25.9 (0.3)	4.7 (6.7)	0.9 (1.8)
28	27	16.4 (5.3)	16.3 (6.2)	0.9 (3.3)	27.0 (0.3)	4.4 (6.8)	0.9 (1.9)
28	28	17.0 (5.3)	16.4 (6.1)	1.0 (3.4)	28.0 (0.4)	4.5 (6.7)	1.0 (2.0)
25	29	17.5 (5.5)	16.9 (6.2)	0.8 (3.6)	29.0 (0.4)	5.1 (7.0)	1.2 (2.0)
23	30	18.1 (5.6)	17.1 (6.4)	0.7 (3.6)	29.9 (0.4)	4.9 (7.3)	1.4 (2.1)
22	31	18.7 (5.9)	17.6 (6.0)	0.8 (3.8)	30.9 (0.4)	5.6 (6.9)	1.4 (2.2)
19	32	19.0 (5.8)	18.2 (6.1)	0.4 (3.6)	31.9 (0.4)	5.5 (7.3)	1.6 (2.3)
18	33	19.7 (6.1)	18.5 (5.8)	0.3 (3.8)	32.9 (0.5)	6.2 (6.7)	1.5 (2.3)
17	34	20.5 (6.4)	18.9 (5.7)	0.4 (4.0)	33.9 (0.5)	5.8 (6.8)	1.8 (2.1)
17	35	21.2 (6.4)	18.8 (5.8)	0.4 (4.2)	34.9 (0.5)	5.7 (6.8)	1.8 (2.1)
13	36	20.6 (6.3)	18.2 (6.6)	-0.1 (3.8)	36.0 (0.5)	5.5 (7.8)	1.3 (2.1)
13	37	21.3 (6.2)	18.4 (6.9)	-0.1 (4.1)	37.0 (0.5)	5.5 (7.8)	1.3 (2.1)
10	38	21.4 (6.3)	18.1 (7.2)	-0.2 (5.0)	38.0 (0.5)	4.5 (8.1)	1.2 (2.3)
9	39	23.0 (6.7)	18.0 (7.7)	-1.4 (5.3)	39.0 (0.5)	4.8 (8.5)	1.4 (2.3)
6	40	20.1 (3.7)	14.1 (7.3)	-1.6 (6.7)	40.1 (0.6)	4.2 (10.7)	1.2 (2.7)
4	41	20.9 (1.8)	17.8 (5.8)	1.2 (5.8)	40.9 (0.5)	8.8 (6.0)	1.7 (1.3)
2	42	19.8 (0.3)	15.2 (7.2)	4.1 (1.8)	42.2 (0.7)	13.0 (4.0)	0.7 (0.2)
2	43	20.0 (0.2)	15.5 (6.4)	4.5 (1.4)	43.1 (0.6)	12.8 (3.9)	0.8 (0.0)
2	44	21.1 (0.0)	19.2 (0.0)	4.9 (0.0)	43.7 (0.0)	9.4 (0.0)	0.8 (0.0)

rior translations were as expected. Some of the differences in the ranges of translation and rotation between studies could be accounted for by the differences in overall knee extension excursion. Unlike numerous previous knee joint studies [32], the PF and TF joints were not assumed to be aligned at one specific reference position (e.g., full extension). Although such an assumption might reduce intersubject variability, it would make the transition to the study of pathology much more difficult since all initial alignment differences would be lost. Lastly, kinematic differences between past static studies and the current dynamic study under volitional control is expected based on past cine MRI experiments [25].

Since the original study, three other techniques (magnetic tracking [33], [34], fluoroscopy [14], [35], and 3-D static MRI [36], [37]) have been demonstrated to be potentially useful in the noninvasive, *in vivo* study of combined 3-D PF and TF kinematics. The use of magnetic tracking is susceptible to the same soft tissue and skin motion errors that have been reported for gait analysis [11]. Although the PF and TF joints can be studied under static weight bearing, dynamic studies are nearly impossible with the clamping technique and less accurate with the plastic mold. Further, rigidly attaching a marker to the patella likely alters the PF kinematics due to soft-tissue impingement. Fluoroscopy, both single-plane [13], [14], [38] and biplane [12], [35], exposes subjects to ionizing radiation, but biplane radiography provides more accurate results for TF kinematics. Fregly *et al.* [14] demonstrated that single-plane fluoroscopy images registered to CT bone models could not track out-of-plane PF and TF motion accurately. Biplane radiography registered to CT knee models was implemented for

TABLE VI
AVERAGE RAW DATA FOR KINEMATIC TRANSLATIONS AND THE NUMBER OF SUBJECTS INCLUDED IN THE AVERAGE AT EACH KNEE ANGLE INCREMENT. KNEE ANGLE IS IN DEGREES AND TRANSLATIONS ARE IN MILLIMETERS. THE STANDARD DEVIATION IS IN PARENTHESES

# Subjects	Knee Angle	Patellofemoral			Tibiofemoral		
		X(L/M)	Y(I/S)	Z(P/A)	X(L/M)	Y(I/S)	Z(P/A)
7	1	-3.6 (3.6)	28.3 (5.7)	6.8 (2.3)	-1.2 (2.0)	-35.8 (6.1)	6.6 (2.0)
9	2	-2.6 (3.5)	26.5 (5.8)	6.3 (2.5)	-0.1 (2.0)	-37.5 (5.4)	5.2 (3.0)
12	3	-2.0 (2.9)	25.7 (5.4)	7.7 (4.3)	-0.8 (3.7)	-39.5 (5.9)	4.9 (3.2)
16	4	-1.6 (2.7)	24.8 (5.0)	7.8 (4.1)	-0.8 (3.3)	-39.0 (6.8)	4.6 (4.1)
22	5	-1.6 (2.7)	24.1 (5.5)	8.0 (3.8)	-0.5 (3.2)	-39.8 (7.0)	4.0 (3.7)
24	6	-1.2 (2.6)	23.0 (6.0)	8.4 (3.6)	-0.3 (3.2)	-41.1 (7.1)	3.5 (3.9)
26	7	-1.3 (2.9)	22.4 (5.8)	9.0 (3.7)	-0.3 (3.3)	-42.4 (7.3)	2.9 (3.9)
26	8	-0.9 (2.9)	21.8 (5.8)	9.1 (3.6)	0.1 (3.2)	-43.0 (7.4)	2.1 (3.9)
29	9	-0.6 (2.7)	21.2 (5.8)	9.5 (4.0)	0.3 (3.0)	-42.7 (7.9)	1.5 (3.9)
33	10	-0.2 (2.5)	20.7 (6.0)	9.7 (3.9)	0.6 (3.2)	-43.3 (7.6)	1.2 (4.2)
33	11	0.0 (2.4)	20.0 (6.0)	9.9 (3.8)	1.0 (3.1)	-43.9 (7.7)	0.4 (4.2)
33	12	0.3 (2.4)	19.3 (6.0)	10.1 (3.8)	1.3 (3.1)	-44.4 (7.8)	-0.4 (4.2)
34	13	0.4 (2.4)	18.2 (6.3)	10.2 (3.7)	1.4 (3.2)	-45.1 (7.8)	-1.2 (4.2)
34	14	0.6 (2.4)	17.4 (6.3)	10.3 (3.6)	1.7 (3.2)	-45.6 (7.9)	-2.0 (4.2)
34	15	0.7 (2.4)	16.7 (6.3)	10.5 (3.5)	1.9 (3.2)	-46.0 (8.0)	-2.9 (4.2)
34	16	0.9 (2.3)	15.9 (6.3)	10.6 (3.5)	2.1 (3.2)	-46.4 (8.1)	-3.8 (4.1)
34	17	1.0 (2.3)	15.2 (6.2)	10.7 (3.5)	2.4 (3.2)	-46.7 (8.1)	-4.6 (4.1)
34	18	1.1 (2.3)	14.5 (6.2)	10.9 (3.5)	2.6 (3.2)	-47.0 (8.2)	-5.5 (4.1)
34	19	1.2 (2.3)	13.8 (6.2)	10.9 (3.5)	2.8 (3.2)	-47.3 (8.3)	-6.3 (4.0)
34	20	1.3(2.2)	13.1 (6.1)	11.0 (3.5)	3.1 (3.2)	-47.5 (8.3)	-7.2 (4.0)
34	21	1.3 (2.2)	12.4 (6.1)	11.1 (3.4)	3.3 (3.2)	-47.8 (8.4)	-8.1 (4.0)
34	22	1.4 (2.2)	11.7 (6.0)	11.1 (3.4)	3.5 (3.2)	-47.9 (8.4)	-8.9 (4.0)
33	23	1.4 (2.2)	11.1 (6.0)	11.2 (3.5)	3.6 (3.1)	-48.2 (8.6)	-9.5 (3.8)
32	24	1.4 (2.2)	10.5 (6.0)	11.2 (3.5)	3.7 (3.2)	-48.6 (8.6)	-10.2 (3.8)
31	25	1.3 (2.0)	9.8 (6.0)	11.2 (3.6)	3.8 (3.2)	-49.1 (8.5)	-11.1 (3.8)
30	26	1.3 (2.0)	9.2 (6.0)	11.2 (3.7)	4.0 (3.2)	-48.9 (8.6)	-11.7 (3.8)
28	27	1.3 (2.1)	8.9 (5.9)	11.1 (3.7)	3.9 (3.2)	-49.0 (8.4)	-12.6 (4.0)
28	28	1.4 (2.1)	8.2 (5.9)	11.1 (3.6)	4.1 (3.1)	-49.0 (8.4)	-13.5 (4.1)
25	29	1.3 (2.1)	7.9 (6.1)	10.6 (3.7)	4.5 (3.2)	-49.3 (8.6)	-14.5 (4.3)
23	30	1.2 (2.2)	7.4 (6.3)	10.4 (3.8)	4.6 (3.2)	-49.0 (8.6)	-15.6 (4.3)
22	31	1.3 (2.3)	6.6 (6.3)	10.2 (3.8)	4.8 (3.3)	-48.9 (8.8)	-16.1 (4.2)
19	32	1.1 (2.4)	6.8 (6.2)	10.4 (3.7)	4.8 (3.1)	-47.5 (8.2)	-16.6 (4.4)
18	33	1.2 (2.5)	6.1 (6.3)	10.7 (3.8)	4.8 (3.2)	-47.9 (8.2)	-17.1 (4.5)
17	34	0.9 (2.2)	5.5 (6.3)	10.6 (3.9)	5.1 (3.1)	-46.9 (7.5)	-17.6 (4.6)
17	35	0.9 (2.2)	5.0 (6.2)	10.6 (3.8)	5.2 (3.1)	-46.8 (7.5)	-18.2 (4.7)
13	36	1.0 (2.4)	5.0 (6.8)	9.8 (3.8)	5.9 (2.6)	-49.0 (6.3)	-19.9 (4.3)
13	37	0.9 (2.4)	4.3 (6.6)	9.6 (3.8)	5.9 (2.6)	-49.0 (6.3)	-20.7 (4.4)
10	38	1.0 (2.5)	3.2 (7.1)	8.5 (3.4)	5.3 (2.6)	-47.6 (6.3)	-20.9 (4.7)
9	39	0.8 (2.6)	3.8 (6.5)	8.4 (3.5)	5.6 (2.3)	-46.4 (5.6)	-21.8 (4.8)
6	40	0.9 (2.9)	5.8 (5.0)	10.6 (1.7)	6.1 (2.7)	-44.8 (4.6)	-21.1 (5.5)
4	41	1.6 (3.1)	4.9 (5.0)	11.3 (1.8)	6.8 (2.0)	-44.1 (5.9)	-19.4 (5.6)
2	42	4.2 (0.8)	7.9 (2.3)	10.9 (1.6)	7.7 (2.7)	-46.1 (7.7)	-22.5 (7.2)
2	43	4.1 (0.5)	7.1 (2.3)	10.8 (1.5)	7.7 (2.9)	-46.0 (8.0)	-22.9 (7.4)
2	44	4.0 (0.0)	7.0 (0.0)	9.7 (0.0)	9.6 (0.0)	-51.6 (0.0)	-28.8 (0.0)

the quantification of combined PF and TF kinematics [35]. Accuracies were not reported, but the 2 mm slice thickness of the CT images used to create the reference model would likely be a limiting factor for the accuracy. In addition, the effects of motion artifacts, present during dynamic testing, were not explored. The last, and most promising, technique is the use of 3-D static MRI to measure the PF joint while partial body-weight is simulated with an axial load [36], [37]. Although these studies have good accuracy (1.75° and 0.88 mm [36]), they are static and have long imaging times. For example, the acquisition time for a single image set at one knee angle ranged between 41 and 57 s, which would require approximately 8–9 min of contraction time to acquire the 8–9 knee flexion positions. The TF joint could be studied in addition to the PF joint with this technique, but that would double the imaging time. The required contraction time limits the range of pathologies that can be studied and would likely add motion artifacts, degrading accuracy.

Earlier accuracy studies demonstrated that fast-PC MRI has a system bias of 0.06 mm (SD 0.35), an average absolute error of 0.35–0.50 mm and a precision of 1.2°, 1.5°, and 0.7° for TF extension-flexion, internal-external rotation and varus-valgus, respectively [27], [28]. PF precision was similar [39]. The comparison of manual measurements from the axial images to the fast-PC results further supports the accuracy of this technique. The differences between the fast-PC measures and the visual measures are most likely due to the sensitivity of bone shape to the bone's attitude within the imaging plane [26] and not errors inherent within the data. For example, the patellar tilt relative to the image edge in the 3-D static image set for K#24 changed 3.7° when measured in an axial plane 3 mm above, versus 3 mm below, the mid-patellar image.

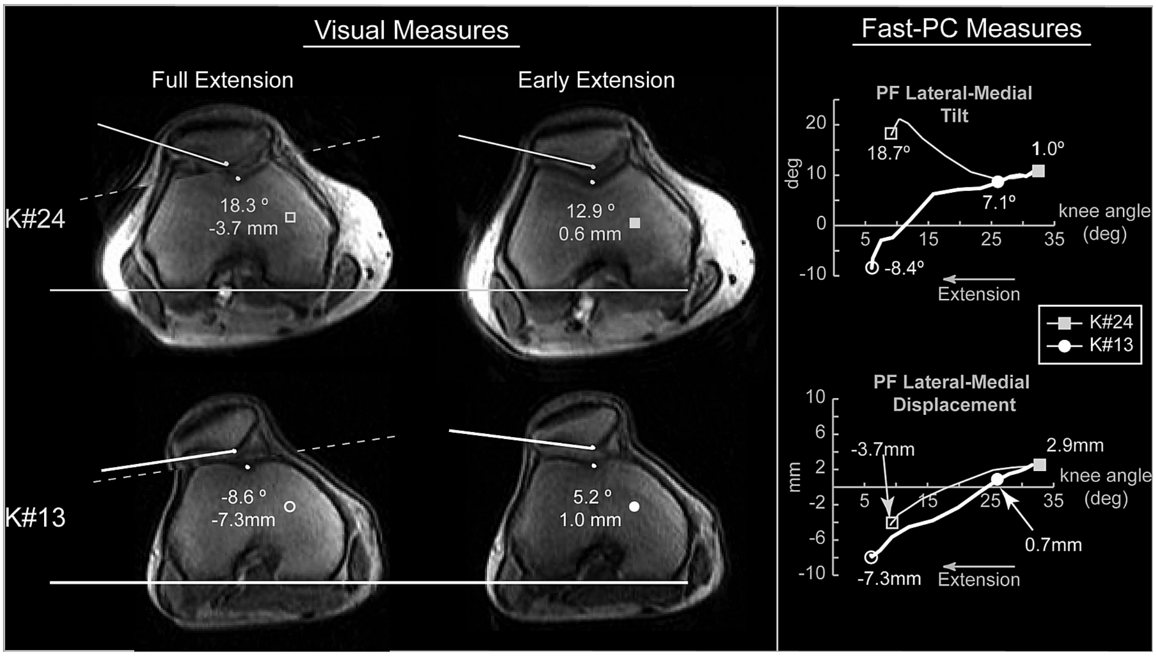

Fig. 5. Comparison between analytical (fast-PC MRI) and visual determination of PF tilt and PF medial-lateral displacement. The data given on the anatomic images represent the measures determined visually from the images and the graphs represent the data acquired through the use of fast-PC MRI. K#24 (squares – top row images) had the most medial tilt and K#13 (circles – bottom row images) had the most lateral tilt of all 34 subjects. Measuring the patellar tilt angle required a view of the patella at the mid-patellar level and a view of the femur at the femoral epicondylar level. Thus, for display, the full extension images (hollow symbols) were concatenated. The white dotted line separates the patellar from the femoral image. The early extension femoral images (filled symbols) were selected so that the view of the patella in this image appeared similar to the mid-patellar image at full extension. Values for PF tilt and PF lateral-medial displacement were derived from direct visual inspection (listed on the images) and compared to those derived based on the integration of the fast-PC data (shown on the graph). The lines and point used to define the PF tilt and displacement are shown on the figure.

The study was delimited to a nonweight bearing partial range of anatomically available flexion. The noninvasive nature of this experiment and its high accuracy and precision justifies these delimitations. Unlike other *in vivo* techniques, fast-PC MRI is able to track the combined 3-D PF and TF kinematics along with the knee musculature (outside the scope of this paper), so that the entire joint can be studied at once. As open MRI technology improves, experiments looking at the full range of motion will become possible. The similarities with past work [40], which demonstrated a lack of significant differences in PF attitude between weight bearing and nonweight bearing exercises, and Lafortune's data [40] demonstrates that the current data are likely applicable to more functional tasks.

In summary, this study accomplished its stated goals of providing a large normative database in order to explore gender-based kinematic differences and the correlation between TF and PF kinematics. In general, significant differences could not be found based on gender. Thus, gender-based kinematics differences are not likely a key factor in the prevalence of anterior knee pain in females and likely play only a minimal role in inter-study variability. Lastly, tibial external rotation did not predict lateral patellar tilt. Further investigation into this relationship in the presence of pathology is warranted.

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Authors' photographs and biographies not available at the time of publication.