Published in final edited form as:

J Orthop Res. 2015 November ; 33(11): 1620–1630. doi:10.1002/jor.22948.

Statistical Modeling to Characterize Relationships between Knee Anatomy and Kinematics

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Abstract

The mechanics of the knee are complex and dependent on the shape of the articular surfaces and their relative alignment. Insight into how anatomy relates to kinematics can establish biomechanical norms, support the diagnosis and treatment of various pathologies (e.g. patellar maltracking) and inform implant design. Prior studies have used correlations to identify anatomical measures related to specific motions. The objective of this study was to describe relationships between knee anatomy and tibiofemoral (TF) and patellofemoral (PF) kinematics using a *statistical shape and function modeling* approach. A principal component (PC) analysis was performed on a 20-specimen dataset consisting of shape of the bone and cartilage for the femur, tibia and patella derived from imaging and six-degree-of-freedom TF and PF kinematics from cadaveric testing during a simulated squat. The PC modes characterized links between anatomy and kinematics; the first mode captured scaling and shape changes in the condylar radii and their influence on TF anterior-posterior translation, internal-external rotation, and the location of the femoral lowest point. Subsequent modes described relations in patella shape and alta/baja alignment impacting PF kinematics. The complex interactions described with the data-driven statistical approach provide insight into knee mechanics that is useful clinically and in implant design.

Keywords

statistical shape modeling; knee anatomy; kinematics; joint mechanics; principal component analysis

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Declarations: The study was performed using data collected on cadaveric specimens. The statistical shape modeling research was reviewed and approved for exemption by the Institutional Review Board at the University of Denver (Protocol 2010-1368).

Disclosures: Chadd Clary is an employee of DePuy Synthes. The University of Denver and University of Kansas received institutional support from DePuy Synthes. Peter Laz and Paul Rullkoetter are consultants for DePuy Synthes.

Introduction

“Form ever follows function” is the credo in design attributed to architect Louis Sullivan. Shape of the articular geometry is known to influence the mechanics of the knee^{1–4} and differences in knee morphology have been shown to exist across the population.^{5,6} Accordingly, the anatomy and function of the structures of the knee have been well studied to establish biomechanical norms, diagnose pathology, guide surgical treatments, and inform implant design.^{7–9}

Knee kinematics are measured using in-vitro cadaveric experiments^{10–13} and in-vivo data collections with fluoroscopy^{14,15} and open magnetic resonance (MR) imaging.^{8,16} Subject-specific kinematics are described after registering representations of the anatomy derived from MR or computed tomography (CT) imaging. The kinematics of the knee are characterized by a combination of sliding and rotation marked by anterior translation and internal rotation of the tibia relative to the femur combined with lateral translation and external rotation of the patella relative to the femur during flexion.^{17,18}

Prior research has investigated relationships between anatomy and functional behavior. Freeman et al. described sagittal plane condylar geometry as a sequence of arcs with different radii that interact with a flat medial and convex lateral tibial plateau.¹⁷ Due to varying radii, tibiofemoral (TF) conformity changes during flexion and the medial condyle remains relatively motionless while the lateral condyle translates posteriorly on the tibia, contributing to characteristic tibial rotation. In the natural knee, Hoshino et al. identified correlations between the condylar offset ratio and anterior-posterior (AP) translation and between condylar twist angle and internal-external (IE) rotation.¹⁹ While considering total knee replacement (TKR) implants, Clary et al. investigated abrupt versus gradually reducing changes in the femoral sagittal radius of curvature and their impact on TF AP motion, particularly at mid-flexion.⁹

Incorporating the patellofemoral (PF) joint, Li et al. described the effects of TF rotation on PF contact mechanics.¹⁵ Other studies have used image-based measurements of the articular surface of the patella and femoral trochlear groove to investigate differences between patellar pain or maltracking groups and normal subjects. Correlations have been identified between patellar kinematics and depth of the trochlea or sulcus angle.²⁰ Considering lateral and nonlateral maltrackers and controls, Harbaugh et al.¹⁶ found morphological differences in sulcus angle, patellar height, articular cartilage depth, and lateral trochlear inclination (the angle between the tangent to the lateral trochlear edge and posterior condylar line) between groups.¹⁶ Recently, Pal et al. showed that patellar maltracking in early flexion was more prevalent in patellar pain subjects than in pain-free subjects, with the patellar pain cohort having more alta patellar alignment.⁸ While correlations importantly identify influential parameters, they do not quantitatively define the relationships or provide thresholds for diagnosis. Freedman and Sheehan applied a regression-based approach to predict three-dimensional (3D) PF kinematics from two-dimensional (2D) static measures of geometry and alignment, although the technique was not able to fully predict the kinematic results.²¹

As an alternative to 2D measurements, Fitzpatrick et al. applied principal component analysis (PCA) to investigate relationships between shape morphology represented by a statistical shape model (SSM) and PF kinematics predicted by finite element analysis.²² Statistical or active shape models quantify the variation between members of a population^{23,24}, and have been used previously to characterize variability in bone morphology and density.^{25,26} Prior SSM studies have focused on individual bones with applications to fracture risk^{27,28} and sizing lines for implants.^{29,30} Recently, statistical models have been applied to consider multiple structures of a joint.^{31–33}

As imaging is typically part of the kinematic measurement process, there is an opportunity to use the combination of anatomy and kinematic data to gain insight into the functional interactions of the knee joint in a more holistic way. Accordingly, the objectives of this study were to characterize relationships between knee anatomy and kinematics using a *statistical shape and function modeling* approach. The current approach is unique in that the PC modes defining the relationships between anatomy and kinematics are determined from the data, in contrast to other studies which required *a priori* identification of the anatomical measure and kinematic output to be investigated. Further, the statistical shape model more fully accounts for the geometry and alignment by using the complete articular geometry instead of a limited set of linear measurements. Lastly, the approach enables the prediction of kinematics for a knee's geometry, which is not possible with correlation-based evaluations.

Methods

This study utilized a combination of imaging and in-vitro kinematic data from a cohort of 20 cadaveric specimens to develop a statistical shape and kinematics model of the knee joint (Figure 1). The specimens were male with an average age of 64 years (range: 44 to 80), average weight of 78 kg (range: 60 to 127 kg) and average body mass index (BMI) of 25 (range: 19 to 41). The specimens were all considered healthy normal with no signs of osteoarthritis. Each knee was imaged approximately 75 mm above and below the joint line using MR (Siemens Avanto 1.5T, 3D balanced gradient echo sequence, in-plane resolution of 0.35 mm, axial slice thickness of 1 mm). Subsequently, the knees in their intact condition were tested in the Kansas knee simulator under a squat activity.¹² The loading condition used a combination of load-controlled actuators at a simulated hip and ankle and a position-controlled quadriceps actuator to simulate a 90° knee bend.¹³ An Optotrak motion capture system (Northern Digital Inc., Waterloo, CA) recorded the movement of rigid body markers attached to the femur, tibia and patella.

Bone and cartilage geometry for the femur, tibia and patella were reconstructed from the MR images using ScanIP (Simpleware, Exeter, UK). A template mesh, including local anatomic coordinate systems, was developed for the median subject.³³ An iterative closest point (ICP) algorithm registered each knee to the template mesh resulting in a consistent mesh and coordinate system for all specimens (Figure 2). The bone template mesh contained 2384, 1101 and 472 nodes for the femur, tibia and patella, respectively. A template hexahedral mesh of each cartilage structure was morphed for each specimen using a mesh-morphing approach with Hyperworks (Altair, Troy, MI).³¹

The local anatomic coordinate systems for each bone were established based on the articular surface geometry and anatomical landmarks.³³ The femoral coordinate system was defined by the axis of a cylinder fit through the flexion facet of the medial and lateral condyles of the femur and the anatomic axis as defined by the centroids of three equally-spaced slices through the proximal portion of the transected femur (midshaft).^{34,35} The origin was located at the midpoint between the medial and lateral epicondylar points. The tibial coordinate system was constructed with the origin at the medial tibial eminence, using axes defined by the centroids of three equally-spaced slices through the distal portion of the transected tibia (midshaft), and through the centers of the tibial condyles.³⁵ The patellar coordinate system was developed using the proximal, distal, and lateral points around the articular periphery with the origin located at the geometric centroid³⁵. Experimentally-measured kinematics from the squat cycle were converted to six degree-of-freedom (DOF) TF and PF kinematics using a three-cylindrical open-chain description of motion.³⁶ Joint kinematic data for all specimens were normalized from 0 (0° flexion) to 100% (90° flexion) of the cycle and discretized at 1% intervals for each degree of freedom.

The *statistical shape-function model* was established by applying PCA to the training set data consisting of nodal coordinates (3D Cartesian coordinates for each node in its local coordinate system) for each bone and cartilage structure and discretized TF and PF kinematics (101 points for each degree of freedom). The array of raw data, V , consisted of an $n \times N$ matrix with n corresponding variables (18153 describing shape and 1212 describing kinematics) for N specimens. PCA was performed on the covariance matrix of V ; PCA is a widely-used statistical technique to decompose a large data set into its primary modes of variation or principal components. The analyses resulted in a series of non-zero eigenvalues characterizing the amount of variability explained and associated eigenvector matrix, E . Each of the subjects was represented by a series of PC scores, P . Note: ' corresponds to transpose, which is equivalent to the inverse for an orthogonal matrix.

V = raw data containing subjects ($n \times N$)

where $v = (n \times 1) = \{v_{\text{shape}} \ v_{\text{kinematics}}\}'$ for one subject

C = covariance matrix ($n \times n$) of V

E = eigenvector matrix ($n \times N-1$) from PCA on C for non-zero eigenvalues

$$E = \begin{Bmatrix} E_{\text{shape}} \\ E_{\text{kinematics}} \end{Bmatrix}$$

$P = v' * E$ = PC scores ($1 \times N-1$) for one subject

$v' = P * E' =$ subject representation ($1 \times n$)

The modes of variation were perturbed by ± 1.5 standard deviations (denoted hereafter simply by + or -) from the mean to visualize the changes in size, shape, alignment or initial position, and kinematics through the squat cycle. This level was set to balance emphasizing the geometric differences, while maintaining realistic instances given the size of the training set. As a composite of the AP and IE kinematics and to enable comparisons to in-vivo studies, the location of TF contact was estimated using the lowest point on the medial and

lateral femoral condyles relative to the tibial SI axis.^{37,38} To describe variations in shape and alignment, a series of measurements used commonly in clinical and radiographic assessments were automatedly performed on the 3D representation; measurements included epicondylar width, Insall-Salvati index, sulcus angle, and bisect offset³⁹⁻⁴² (Table 1^{*}). Pearson's correlation coefficients were computed between the measurements and PC scores representing each specimen, as well as between the measurements, initial alignment and range of motion (ROM) kinematics.

Lastly, a leave-one-out (LOO) evaluation was performed to assess the ability of the model to predict the shape and kinematics of a new subject from outside of the training set. To perform kinematic predictions from the geometry of a new or left-out subject, an approach similar to Fitzpatrick et al.²² was applied. The geometry was described as the shape-only variables of the raw subject vector (v_{shape}). Using the shape portion of the eigenvectors, the shape representation was transformed into PC scores (P_{new}) corresponding to each mode. Then, the PC scores and full eigenvector matrix were used to predict the shape and kinematic vector for the new specimen. All PC modes were utilized in these predictions.

$$P_{\text{new}} = v_{\text{new shape}}' * E_{\text{shape}} = (1 \times n_{\text{shape}}) * (n_{\text{shape}} \times N - 1) = (1 \times N - 1)$$

$$v_{\text{new}}' = \{v_{\text{new shape}} \ v_{\text{new kinematics}}\}' = P_{\text{new}} * E' = (1 \times N - 1) * (N - 1 \times n) = (1 \times n)$$

Mean absolute errors were calculated between actual and model-estimated shape and location of the lowest contact point.

Results

The statistical model identified relationships between shape and kinematic variation in the training set as a series of modes of variation. As the earliest modes captured the largest amount of variability in the data, they are emphasized here. For example, the first 3 modes of variation explained 49.0% of the variability, with 6, 15 and 19 modes capturing 69.4%, 95.1% and 100%, respectively (Table S1). By perturbing individual modes, the corresponding changes in anatomy and TF and PF kinematics are shown in Figures 3, 4 and 5[†] for Modes 1–3 and in the Supplement for Modes 4–6. Additionally, correlations described the shape and kinematic parameters captured in each mode (Table 2).

In addition to describing scaling of the knee (Figure 3), Mode 1 captured changes in the medial and lateral AP condylar geometry, AP and IE kinematics and the location of the lowest contact point (Figures 4 and 6). The sagittal radius of curvature for the medial condyle scaled uniformly between Mode 1+ and Mode 1– (Figure 6). The sagittal radius for the lateral condyle was relatively constant between 20° and 50° flexion, albeit with offset centers, but varied in deeper flexion (Figure 6); between 60° and 90° flexion, the radius of curvature for the lateral condyle was up to 1.45X larger for Mode 1+, while it remained relatively constant (within 0.96X) over the same range for Mode 1–. The initial TF position

^{*}See supplement for measurement descriptions.

[†]Absolute and relative kinematics for all degrees of freedom are presented in supplement Figures S1 and S2.

of the Mode 1+ (larger) knee was more flexed, external, and posterior compared to the Mode 1- (smaller) knee (Figure 4). During the flexion cycle, both knees rotated internally a similar amount; however, the majority of the motion occurred in early flexion for the Mode 1- knee. The Mode 1+ knee exhibited a steady anterior tibial motion with flexion, while the tibia initially moved anteriorly, followed by posterior translation after 20° flexion for the Mode 1- knee. The lowest point data showed little motion of the medial contact point, while the lateral contact point moved posteriorly, capturing the differences in the amount of IE rotation in early flexion with Mode 1 (Figure 6). In the PF joint, Mode 1 described variation in ML translation (Figure 5), which was linked to differences in the anterior aspects of the condylar geometry. PC scores for Mode 1 were strongly correlated to anatomic size measurements, including femoral epicondylar width (correlation $r = 0.91$) (Table 2).

Mode 2 described anatomical shape changes in the bone and cartilage, alta-baja (SI) alignment of the patella relative to the femur and PF kinematic changes in AP and SI translation and IE rotation (Figures 3 and 5). Illustrating alta-baja alignment of the patella, PC scores for Mode 2 were correlated to the Insall-Salvati index ($r = -0.61$, Table 2). The alta patella for Mode 2- had consistently larger AP and SI kinematics than the baja patella of Mode 2+, which affects the moment arm of the quadriceps. Further, the Mode 2- geometry with the alta patella had a shallower trochlear angle of 2.5° and exhibited internal patellar rotation during early flexion, compared to the baja patella in Mode 2+ with a steeper trochlear angle of 4.9° and external patellar rotation during flexion (Figure 5). Both models (Mode 2+ and 2-) realized a similar IE position at roughly 25° flexion when the patella engaged the trochlear groove. Differences in cartilage coverage on the bone were also noted in Mode 2 (Merchant view of Figure 3).

Mode 3 accounted for further anatomic shape changes in the femur and patella, and patellar alignment, including some alta-baja variability ($r = 0.54$) and initial PF IE and VV position. Differences were observed in the anterior-lateral aspect of the femur and trochlear groove; correlations between the antero-inferior femoral sulcus angle, measured in a 45° merchant view, and PC score for Mode 3 were -0.52 . The prominence of the anterior-lateral facet influenced both the initial PF IE (Figure 5) and VV (Figure S2) alignment, although motions through the flexion cycle were similar (Figure 5). Mode 3 also resulted in the largest differences in the AP location of the lateral contact point (Figure 6) and ROM for TF IE rotation with 8.8° and 11.5° for Mode 3+ and 3-, respectively (Figure 4, S1). While not significantly correlated ($r < 0.5$), Mode 3 described differences in the distal sulcus angle of 7° between the +/- models, versus 3° and 4° in Modes 1 and 2, respectively.

Using data for the specimens in the training set, significant correlations were identified between 2D anatomical measurements and kinematics (Table 3, Figure S3). The size measures, largely captured in Mode 1, were strongly correlated to many of the initial alignment degrees of freedom, particularly SI, AP and ML translation. A greater distal sulcus angle corresponded to greater tibial IE ROM through the flexion cycle ($r = 0.83$), while a greater distal condylar angle was correlated to a more valgus TF VV alignment ($r = 0.73$). As described in Modes 2 and 3, the Insall-Salvati ratio was correlated to initial PF FE ($r = -0.54$), SI ($r = 0.46$) and AP ($r = 0.50$) alignment. Bisect offset, a measure of patellar tracking, increased with more laterally-aligned PF joints ($r = 0.67$). Further, strong

correlations were identified between TF and PF kinematics through the cycle, specifically between PF FE and TF FE, between PF VV and TF VV, and between PF ML and TF IE (Table S2).

Results of the LOO evaluation characterized the predictive ability of the model with errors computed between the actual and model-predicted geometry and lowest point locations for the left-out knee. The absolute geometric error averaged across all nodes and for all specimens was 1.90 mm with a standard deviation of 0.39 mm. Differences between predicted and actual femoral lowest point were typically smaller in the medial compartment than the lateral and smaller in the ML direction compared to the AP direction (Figure 7, Table 4). Averaged over the flexion range and for all specimens, the mean absolute errors were 2.11 mm and 2.87 mm for the ML and AP directions on the medial condyle and 2.22 mm and 4.53 mm on the lateral condyle, respectively.

Discussion

The data-driven statistical modeling approach developed in this study demonstrated the ability to capture the role of complex anatomic and kinematic interactions and present them in a way that provides design and surgical insights. Relationships described between anatomy, initial alignment, and TF and PF motions through a squat cycle confirmed findings from several other studies, with the current approach enabling a more holistic consideration of the interactions. The benefits of the PCA-based approach were threefold: the PC modes describing the relationships between anatomy and kinematics were elucidated from the data without requiring *a priori* identification of the input and output measures, the evaluations were performed with the full articular geometry rather than a limited set of linear measures, and lastly, the resulting model enabled the prediction of kinematics for a new subject's geometry.

Traditionally, studies have identified sets of measurements and investigated relationships with correlations between measures of interest. For instance, Harbaugh et al.¹⁶ focused on lateral trochlear inclination angle (LTI) and patellar height, and reported a correlation between LTI and medial patellar tracking in their healthy control group ($r = 0.35$). Investigating the patellar anatomy of 907 subjects, Stefanik et al.⁴² reported the highest correlation between bisect offset and LTI ($r = -0.38$). Further, Powers et al.²⁰ reported correlations between bisect offset and sulcus angle ($r = 0.74$). The approach taken in the current study can be used to report similar correlations to these prior studies; for example, this data elucidated links between bisect offset and LTI ($r = -0.62$), between LTI and PF ML alignment (-0.46), between bisect offset and PF ML alignment (0.67), and between bisect offset and sulcus angle ($r=0.50$). However, the PCA-based approach implemented in the current work utilizes three-dimensional representations of the bones and cartilage, which enabled a more comprehensive analysis, including the potential to discover unanticipated links between anatomy and kinematics and the ability to investigate interrelationships between measurements when interpreting findings, which may not be possible with the traditional approach. This approach does not require a priori knowledge of factors which are anticipated to be linked; instead, the entire shape and kinematic database is interrogated and relationships within that dataset emerge within each of the modes of variation. This

extended beyond correlation of a single shape metric with a single kinematic metric to a more holistic interpretation of the data and the relationships within. For example, Mode 2 showed a relationship between an alta patella with shallow trochlear angle and internal patellar rotation.

The first 6 PC modes of variation were investigated to describe associations between changes in anatomy and kinematics. Emphasis was placed on the early modes which captured the largest amount of variance in the data. Further, a parallel analysis, which involved randomizing the variables within each observation and performing PCA on this new dataset in an effort to quantify the inherent noise in the data⁴³, found the first 6 modes were significant.

Described by Mode 1 with the most variation explained, changes in the sagittal femoral condyle geometry or J-curve were directly linked to the AP and IE kinematics and, ultimately, the location of the lowest point. The finding that more gradual radius changes through the flexion facet of Mode 1- led to reduced posterior tibial translation (or anterior femoral translation) (Figures 4 and 6) agreed with Clary et al.,⁹ which observed that an increasing "braking" radius results in less anterior motion and more rollback of the femur with respect to the tibia. Early flexion differences in the lowest point representation were affected by shape of the distal region of the femoral condyles with Mode 1- having a more flattened profile. Coupled with a steeper slope for tibial IE rotation, the lowest-point behavior near extension was characteristic of the screw-home mechanism.⁴⁴ The lowest point location provides a surrogate measure of contact and enables comparisons to prior fluoroscopic studies.^{37,38} Alternatively, a contact or patch-based analysis of the kinematics could provide additional information, but was not considered in the current study.

Further, differences in TF IE rotation during the cycle were described in Modes 3, both in ROM (Figures 4 and S1) and the location of the femoral lowest point (Figure 6). Hoshino et al. noted the importance of distal femur morphology, particularly a correlation between condylar twist angle and internal tibial rotation.¹⁹ Changes in the morphology of the distal femur were also described in Mode 3 with differences reported in distal sulcus angle. According to Freeman et al.¹⁷, the inner facets of the condyles and tibial eminence interact to guide tibial rotation. Lastly, the more general correlation between tibial IE rotation and patellar ML translation is consistent with Sheehan et al.⁴⁵ and underscores the shape-driven interactions between TF and PF joints.

PF kinematics were dependent on the anatomy of the patella and trochlear groove, and initial patellar alignment. A deeper trochlear groove or smaller sulcus angle led to more PF external rotation (Mode 2), while the anatomy of the anterior-lateral facet and trochlear groove influenced the initial PF IE alignment. Regarding patellar alta-baja, the Insall-Salvati index, PC score for Mode 2 and PF AP alignment all shared strong correlations (Tables 2 and 3). Fitzpatrick et al. showed that quadriceps efficiency during a deep knee bend was affected by patellar resection thickness⁴⁶, highlighting that the AP position of the patella serves as an effective moment arm. Accordingly, the current model may be useful in developing subject-specific representations for musculoskeletal simulations considering shape and alignment variability in the population.

Numerous studies have used shape of the femur, patella and relative alignment as measures to differentiate healthy normal and pathologic groups. This study confirmed the importance of alta-baja, bisect offset and sulcus angle in PF mechanics, and notes their established links to PF pain and maltracking.^{8,16} This study also demonstrated the ability to efficiently measure these important parameters within the SSM and evaluate them with respect to dynamic motions, rather than static poses used in prior MR based studies. Both of these considerations are important in enabling evaluations of larger-scale populations under clinically relevant conditions.

As subject-specific measurement of kinematics is time consuming and expensive, it is rare to have a dataset of 20 natural knees for the same activity. Many studies have presented kinematics for datasets with smaller numbers of subjects. However, the size of the dataset remains relatively small when compared to the overall population. The all-male training set is not representative of the overall population and is a limitation of the study; however, the group does represent a subset of the potential total knee replacement population. A further limitation is that the kinematic data were measured from cadaveric specimens in a simulator. The simulator applied the same loading condition to all of the specimens and thereby, did not capture the loading variability that is present across the population. However, the controlled simulator data allowed isolation of the effect of knee anatomy/shape on kinematics, which was the objective of this study. Recognizing the impact of weight bearing and muscle loading on kinematics shown in the literature^{47,48}, the shape-function approach could be similarly implemented using in-vivo data from biplane fluoroscopy, which potentially allows for consideration of larger numbers of subjects and loading variability for a variety of activities. As the data required to represent the subject's shape is part of the workflow, implementation can be performed with minimal additional processing. Interestingly, SSM has recently been used to represent the subject's geometry using an optimization to the fluoroscopy data alone, alleviating the need for additional imaging and segmentation.⁴⁹

This study did not directly investigate the role of soft tissue structures, which are known to provide constraint and impact knee mechanics. The cadaveric simulator data notably captured intersubject variability in anatomy, alignment and soft tissue constraint, and thereby considered factors not included in Fitzpatrick et al.²², which used kinematics derived from finite element analyses with a constant soft tissue representation.

The predictive capability of the SSM was evaluated using the LOO test; results demonstrated the ability of the model to accurately recreate the shape and kinematics of the left-out specimen. The errors in shape representation were comparable to those reported in Rao et al.³³ and other SSM models. Errors in the lowest point predictions were dependent on the condyle and DOF, but less than 3 mm on average for all DOF. The ability to represent new subjects accurately provides confidence in using the approach in larger population studies which require the generation of virtual instances.

Insight into relationships between knee anatomy and kinematics has broad reaching impact in biomechanics. Relationships for the healthy normal dataset can address current areas of interest in knee mechanics and the design of total knee replacement implants, particularly

regarding the impact of shape of the condylar geometry or j-curve and identifying anatomical features that drive motion (e.g. rotation or rollback). Additionally, insight into the kinematics associated with patients with certain characteristics (e.g. patella alta or a narrow trochlear groove) may lead to altered surgical decision-making related to implant selection, sizing and placement to avoid overloading regions of bone, crepitus and other complications.^{7,46} The approach can be extended to further investigate differences between healthy normal and pathologic groups, especially when shape and alignment are contributing factors, as in patellar maltracking, PF pain and varus/valgus deformities.

Supplementary Material

Refer to Web version on PubMed Central for supplementary material.

Acknowledgments

This research was supported in part by the National Science Foundation (CBET-1034251), National Institutes of Health (Grant number: 1R01EB015497-01) and DePuy Synthes, a Johnson & Johnson Company. The authors also acknowledge the assistance of Chandreshwar Rao, Laura Hendrickson, and Arianna Vierczhalek in developing the statistical shape model and Sami Shalhoub and Fallon Fitzwater in experimental testing.

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Figure 1.

Development of a statistical shape and function model. The shape representation was derived from image data by segmenting and establishing correspondence to a template mesh, and tibiofemoral and patellofemoral kinematics were obtained from cadaveric testing and registration of the anatomy and local coordinate systems (CS) to experimentally probed points.

Figure 2.
Training set of 20 specimens used to create the statistical shape-kinematics model.
Specimens are represented with the template mesh in the initial kinematic position.

Figure 3.

Representations of bone and cartilage for the first three principal component modes. Knees are shown at ± 1.5 standard deviations. Coronal and sagittal views at initial alignment with merchant view at 45° TF flexion.

Figure 4.

Tibiofemoral kinematics for the first 3 principal component modes and all specimens (gray lines). Clockwise from top: Tibial flexion-extension, anterior-posterior (AP) translation and internal-external (IE) rotation. Inset bar charts show relative contribution of each mode.

Figure 5.

Patellofemoral kinematics for the first 3 principal component modes and all specimens (gray lines). Clockwise from top-left: Patellar internal-external (IE) rotation, medial-lateral (ML) translation, superior-inferior (SI) translation and anterior-posterior (AP) translation. Inset bar charts show relative contribution of each mode. ISI = Insall-Salvati Index.

Figure 6. Sagittal condylar geometry (with radius of curvature lines) for the mean and first three modes (left). Femoral lowest point representation of the TF contact points for the first three modes at ± 1.5 standard deviations (right).

Figure 7.

Comparison of actual (solid) and predicted (dashed) lowest contact point for a leave-one-out evaluation with varying flexion. Predictions were made with each member of the training set left out of the analysis. Error bars shown for anterior-posterior (AP) and medial-lateral (ML) degrees of freedom on each condyle. Full extension results reported as 0-10° flexion as not all specimens achieved 0°.

Table 1

Descriptive statistics and anatomical measures for the training set.

Anatomic Measure	Mean	Standard Deviation	Min.	Max.
Age [yr]	64	10.8	44	80
Weight [kg]	78	16.2	61	127
Body Mass Index (BMI)	25	5.2	19	41
Epicondylar Width [mm]	87.4	4.0	81.0	98.3
Femur AP Width [mm]	66.4	2.8	62.1	75.2
Tibia ML Width [mm]	80.3	3.5	73.8	89.4
Patella AP Thickness [mm]	19.0	2.3	14.9	22.8
Patella Angle [°]	146.5	6.4	136.2	155.7
Insall-Salvati Index	1.2	0.2	0.8	1.5
Trochlear Angle [°]	4.2	2.3	0.3	8.7
Anterior Sulcus Angle [°]	144.3	8.8	131.6	171.5
Medial Trochlear Inclination [°]	163.1	6.0	152.5	179.8
Lateral Trochlear Inclination [°]	18.8	3.7	8.4	24.9
Antero-inferior Sulcus Angle [°]	72.2	13.8	56.3	118.1
Distal Sulcus Angle [°]	132.4	6.2	122.4	145.0
Distal Condylar Angle [°]	6.7	2.0	3.6	10.9
Bisect Offset [%]	60.1	6.7	50.2	76.1

Pearson’s correlation coefficients between the first six principal components and anatomical and kinematic measures. Initial alignment correlations were calculated at approximately 10 degrees TF flexion. Range-of-motion (ROM) was defined as the difference between the minimum and maximum values in a kinematic measure. Correlations are presented as absolute values. Anatomical measures with no significant correlations were omitted. White cells indicate no correlation.

Table 2

	0.4 to 0.6	0.6 to 0.8	0.8 to 1.0	PC 1	PC 2	PC 3	PC 4	PC 5	PC 6
Epicondylar Width				0.91					
Femur AP Width				0.82	0.45				
Tibia ML Width				0.82	0.40				
Patella AP Width				0.59					0.46
Insall-Salvati Index					0.61	0.54			
Trochlear Angle					0.44			0.44	
Lateral Troch. Incl.						0.46			
Antero-inferior Sulcus Angle						0.52			
Bisect Offset							0.52		
Distal Sulcus Angle									0.40
TF FE	Alignment			0.70					
	ROM			0.42					
TF VV	Alignment						0.48	0.44	
	ROM			0.55					
TF IE	Alignment			0.63					
	ROM								0.51

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		PC 1	PC 2	PC 3	PC 4	PC 5	PC 6
		0.4 to 0.6					
		0.6 to 0.8					
		0.8 to 1.0					
TF ML	Alignment						
	ROM					0.77	
TF AP	Alignment				0.43		
	ROM						0.55
TF SI	Alignment	0.71				0.40	
	ROM	0.47			0.55		
PF FE	Alignment	0.55				0.47	
	ROM	0.51					
PF VV	Alignment			0.80			
	ROM	0.67					
PF IE	Alignment			0.44			
	ROM		0.56				
PF ML	Alignment				0.64		
	ROM						0.45
PF AP	Alignment		0.62				
	ROM	0.61			0.42		
PF SI	Alignment					0.58	
	ROM						

Table 3

Pearson’s correlation coefficients between anatomical and kinematic measures. Initial alignment correlations were calculated at 10 degrees TF flexion. Range-of-motion (ROM) was defined as the difference between the minimum and maximum values in a kinematic measure. Correlations are presented as absolute values. Anatomical measures with no significant correlations were omitted. White cells indicate no correlation.

	0.4 to 0.6	0.6 to 0.8	0.8 to 1.0	Epicondylar Width	Femur AP Width	Tibia ML Width	Patella AP Width	Insall-Salvati Index	Trochlear Angle	Anterior Sulcus Angle	Lateral Troch. Incl.	Antero-inferior Sulcus Angle	Distal Sulcus Angle	Distal Condylar Angle	Bisect Offset
TF				0.56	0.77	0.54	0.51								
FE				0.51						0.50	0.62				
TF														0.73	0.43
VV				0.53	0.53										
TF				0.60		0.58								0.40	
IE												0.83			
TF				0.41									0.49		
ML															
TF				0.50	0.46										
AP															
TF				0.69	0.64	0.62						0.40			
SI															
TF				0.42		0.41		0.46	0.42	0.47					
FE				0.44	0.53			0.41							
TF															
VV				0.65	0.64	0.67					0.41	0.42			
PF															
IE															

Table 4

Mean absolute error between experimental and predicted lowest point results averaged across all specimens. All values are in millimeters.

Flexion Angle (°)	Medial						Lateral					
	ML			AP			ML			AP		
	μ		σ	μ		σ	μ		σ	μ		σ
	μ	σ		μ	σ		μ	σ		μ	σ	
0	1.98	1.34		3.76	2.81		2.20	1.56		4.53	3.35	
30	1.99	1.46		2.77	2.56		2.49	1.76		4.03	3.69	
60	1.83	1.56		2.56	1.75		2.46	1.52		4.49	3.23	
80	2.65	1.91		2.39	1.69		1.74	1.51		5.06	3.20	
Average	2.11	1.57		2.87	2.20		2.22	1.59		4.53	3.37	