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Statistical Shape Modeling Predicts Patellar Bone Geometry to Enable Stereo-Radiographic Kinematic Tracking

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Abstract

Complications in the patellofemoral (PF) joint of patients with total knee replacements include patellar subluxation and dislocation, and remain a cause for revision. Kinematic measurements to assess these complications and evaluate implant designs require the accuracy of dynamic stereo-radiographic systems with 3D-2D registration techniques. While tibiofemoral kinematics are typically derived by tracking metallic implants, PF kinematic measurements are difficult as the patellar implant is radiotransparent and a representation of the resected patella bone requires either pre-surgical imaging and precise implant placement or post-surgical imaging. Statistical shape models (SSMs), used to characterize anatomic variation, provide an alternative means to obtain the representation of the resected patella for use in kinematic tracking. Using a virtual platform of a stereo-radiographic system, the objectives of this study were to evaluate the ability of an SSM to predict subject-specific 3D implanted patellar geometries from simulated 2D image profiles, and to formulate an effective data collection methodology for PF kinematics by considering accuracy for a variety of patient pose scenarios. An SSM of the patella was developed for 50 subjects and a leave-one-out approach compared SSM-predicted and actual geometries; average 3D errors were 0.45 ± 0.07 mm (mean $\pm$ standard deviation), which is comparable to the accuracy of traditional segmentation. Further, initial imaging of the patella in five unique stereo radiographic perspectives yielded the most accurate representation. The ability to predict the remaining patellar geometry of the implanted PF joint with radiographic images and SSM, instead of CT, can reduce radiation exposure and streamline *in vivo* kinematic evaluations.

Keywords

statistical shape modeling; patella; stereo radiography; fluoroscopy; kinematics; subject-specific geometry; total knee replacement

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CONFLICT OF INTEREST STATEMENT

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INTRODUCTION

Patellofemoral (PF) joint complications continue to account for a significant percentage (up to 10%) of total knee replacement (TKR) revisions (Healy et al., 1995; Dalury et al., 2003; Rhee et al., 2008). Pathologies such as patellar subluxation and maltracking are characterized by medial-lateral (ML) translation and internal-external rotation of the patella (Singer et al., 1997; Kawano et al., 2002) and remain a common cause of TKR revisions (D'Lima et al., 2003). Measuring dynamic *in vivo* PF joint kinematics to understand these complications and interactions between PF and tibiofemoral (TF) joints continues to be a challenge for patients with TKR because of the need for accuracy in multiple degrees of freedom (DOF) and difficulties in tracking the implanted patella. Current *in vivo* kinematic measurement techniques for the knee include static (Fellows, 2005; Pal et al., 2011) and dynamic (Powers et al., 2003; von Eisenhart-Rothe et al., 2004; von Eisenhart-Rothe et al., 2007; Sheehan et al., 2009; Carpenter et al., 2009) magnetic resonance (MR) imaging. Static MR images have been unable to reproduce dynamic joint motion and dynamic MR studies have relied on combining scans from multiple cycles (Powers et al., 2003; Bey et al., 2008). Dynamic single-plane fluoroscopy has been used to evaluate primarily sagittal plane *in vivo* PF kinematics for TKR patients during flexion-dominated activities (Komistek et al., 2000, Stiehl et al., 2001, Argenson et al., 2005, Leszko et al., 2010). In particular, the sagittal angle of the patellar bone resection surface was measured relative to the tibial long axis and an estimate of the PF contact location was defined by the shortest perpendicular distance between the femoral component and patellar mass center.

Stereo radiography or biplane fluoroscopy has been utilized for *in vivo* kinematic measurements of the natural TF (You et al., 2001; Li et al., 2008; Bey et al., 2008; Baka et al., 2014) and PF (Nha et al., 2008; Bey et al., 2008), and implanted TF (Bingham 2006; Hanson 2006) joints. Traditionally, the process to obtain *in vivo* kinematics from stereo radiography involves two steps: representation of the bone, typically from segmentation of a CT or MRI scan, or metallic implant from a computer-aided-design (CAD) model, followed by manual or semi-automated tracking of the bone or implant in the image sequence of the dynamic activity. Studies have used automated three-dimensional to two-dimensional (3D-2D) registration of virtual bone or implant models to fluoroscopic images for 3D pose extraction of the joint (Mahfouz et al., 2003; Bingham et al., 2006). Feature-based methods work to align silhouette contours of the model projections to the x-ray images, while intensity-based methods work to align digitally reconstructed radiographs (DRRs) considering density (You et al., 2001; Yao et al., 2003; Dworzak et al., 2010). Comparing a 3D-2D registration method between dual-plane and single-plane systems, Zhu et al. (2012) reported sub-millimeter and sub-degree accuracies for TF joint tracking with a dual-plane system; this amounted to 2x in-plane and 10x out-of-plane improvements in accuracy over their single-plane system. Recently, a high-speed stereo-radiographic imaging system was developed at the University of Denver to measure *in vivo* joint kinematics (Ivester et al., 2015). Using a semi-automated 3D-2D registration method of DRRs from a CT-based model with XROMM (Brown University; Brainerd et al., 2010), 3D tracking errors of 0.15 mm (S.D. 0.13 mm) in translation and 0.41° (S.D. 0.30°) in rotation were reported for motions of a human knee phantom containing cadaver bones (Ivester et al., 2015).

Stereo radiographic tracking of the implanted patella is challenging because the patellar implant is radiotransparent. While tracking embedded beads is accurate to <0.1 mm and 0.1° (Bingham et al., 2006), this method would require custom modified implants and does not translate easily to larger, population studies. Without beaded implants, the resected patella requires either pre-surgical imaging of bones and precise implant placement or post-surgical imaging, which can be challenging due to imaging artifacts, expense, and increased patient exposure to radiation with computed tomography (CT).

Statistical shape models (SSMs) have become an accepted tool to describe anatomical variation and have characterized the morphology of individual bones and whole joints (Bryan et al., 2010; Yang et al., 2008; Rao et al., 2013; Smoger et al., 2015). SSMs offer an alternative means to obtain the 3D subject-specific bone geometry from stereo-radiographic images, thus avoiding the need for segmentation. Recently, automated approaches have utilized SSMs to *simultaneously* estimate 3D shape and 6-DOF pose from 2D image sets (Laporte et al., 2003; Zheng et al., 2009; Kurazume et al., 2009; Zhu et al., 2011, Baka et al., 2011). Feature-based methods, utilizing contours of bone shape, have been applied to the rib cage (Dworzack et al., 2010), the distal femur using a deformable Kriging algorithm (LaPorte et al., 2003) or Canny edge detection (Zheng et al., 2006), and the proximal femur using a level set method (Kurazume et al., 2009). Intensity-based methods, which are computationally expensive and benefit from more complex shapes to generate unique DRRs, have been previously applied in the pelvis (Yao et al., 2003), as well as the femur and tibia (You et al., 2001). Recently, using an automated feature-based approach with multi-stage optimization, Baka et al. (2012, 2014) predicted both shape and pose of the tibia and femur from biplane fluoroscopic images of a gait cycle. Valenti et al. (2016) and Yu et al. (2017) applied similar techniques to obtain 3D representations of the distal femur with Gaussian Mixture Models and proximal femur with B-spline based statistical models, respectively. Schumann et al. (2016) simulated the proximal and distal portions of the femur as fragments; their fully automated image processing technique assessed bone alignment with applications to deformity correction and fracture repair. While these methods are advancing the state of the art and show promise, 3D-2D registration is challenging and in our experience, often requires manual intervention, leading to a semi-automated kinematic extraction process.

Limited work has been performed to generate subject-specific patellae or track implanted PF joint motions from fluoroscopic images, likely due to the issues with radiotransparency of the implant. Further, the resected patella may be occluded by the metallic femoral component depending on the activity, patient positioning and imaging planes. Most prior automated studies used images from an activity to develop the shape representation. Accordingly, the objectives of this study were to use a virtual platform of a stereo-radiographic imaging system to evaluate the ability of an SSM to predict subject-specific 3D patellar geometries from simulated target profiles, and to formulate an effective data collection methodology for *in vivo* PF kinematics by considering accuracy in a variety of patient pose scenarios. The emphasis of this study was to develop an approach to initially image the knee in controlled positions where the patella was clearly visible. Further, simulating the radiographic images, 2D profiles were used with the SSM to create the 3D representation of the resected patella, which can be used for subsequent semi-automated

kinematic tracking. This approach is in contrast to more automated, combined (shape and pose) optimization studies, which have been applied in the femur and tibia (e.g. Baka et al., 2014; Yu et al., 2017). By utilizing a cohort of segmented patellar geometries and a computational platform, this study establishes a workflow for future experimental protocols without risk to patients.

METHODS

An SSM of the healthy normal patellar bone was developed from a training set of 50 knees, derived from imaging on cadavers and the Osteoarthritis Initiative (OAI) database. The knees were from 25 males and 25 females whose average age was 64 years (range: 44 to 87), average weight was 73 kg (range: 43 to 127), and average body mass index (BMI) was 25.2 (range: 19.0 to 41.3). The OAI database is available for public access at <http://www.oai.ucsf.edu/>. The specific dataset used was 0.E.1, which consisted of baseline (initial healthy) scans. Subjects were included if they were deemed healthy and normal with no signs of osteoarthritis. The patellae were reconstructed from MR scans using ScanIP (Simpleware, Exeter, UK) and registered to a template mesh (Rao et al., 2013; Smoger et al., 2015). Registration was performed through an iterative closest point algorithm and provided nodal correspondence between all subjects in a common local coordinate system. The template mesh was developed for the median subject and consisted of 472 nodes with an average element edge length of approximately 3 mm (Fitzpatrick et al., 2011). The local coordinate system for the template patella was developed with the origin located at the geometric centroid and axes constructed from the proximal, distal and lateral points around the articular periphery (Rao et al., 2013).

The SSM was established by applying principal component analysis (PCA) to the training set of 50 subjects. Each subject's shape was represented by a 1416×1 vector of its registered 3D nodal coordinates (x, y, z). PCA is a statistical technique to decompose a large data set into its primary modes of variation or principal components (PCs). After applying PCA to the covariance matrix of the training set data, the modes of variation described the anatomic variation present in the dataset and enabled both training set subjects and new instances to be represented by the SSM with a series of PC scores. By reducing the dimensionality of a dataset, the subsequent optimization of patellar shape can be performed with a small set of PC scores.

A virtual model of the stereo-radiography system was developed to simulate an experimental data collection on a population of subjects. Leveraging prior data collections in the knee, the model was based on experimental images and projection parameters for a single subject participating in a larger IRB-approved study (Kefala et al., 2015). To capture the resected bone of the implanted construct and avoid occlusion from the metallic implants and surrounding tissue, the patella was imaged in specific positions to capture the shape and resection plane. With the subject supine and knee flexed at 45°, images were collected in the sagittal and transverse directions with respect to cameras A and B, respectively (Figure 1). Radiopaque markers were placed on the patella to locate the patella in the imaging volume of the system. A direct linear transform (DLT), which maps 3D objects onto the camera's 2D image plane, was calculated in a calibration process (Brainerd et al., 2010). Briefly, a

DLT is estimated by identifying, in each camera image, points on a cube at known relative positions in the 3D object space. The markers were digitized on the radiographs and the DLT transformed the digitized points into the 3D object space to locate the virtual imaging volume. Marker digitization and camera calibration was conducted in XRayProject (XROMM, Brown University).

For each patella in the training set, an appropriately-sized patellar implant was selected and located, the bone was virtually resected (Figure 2) (Ali et al., 2015), and simulated 2D images representative of the stereo-radiography system were generated. The virtual implantation optimized the size and placement of a commercially available patellar implant to reproduce the original thickness and maximize coverage of the implant with minimal overhang. Within the virtual stereo-radiography platform, training set instances of the SSM were transformed to the imaging volume based on experimental marker data, thus simulating a data collection for a subject. Using the DLT, a rotation was applied to each resected patella to simulate the virtual 2D images of a subject in 5 positions; the varying poses resulted in 10 unique views of the patella (Figure 3). Positioning actual knee joints for a cohort in the imaging volume would likely introduce some random alignment variability between the patella and the image planes. To account for this, each patella was randomly perturbed in the imaging volume (prior to applying the DLT) by up to ± 50 mm and $\pm 20^\circ$. An alpha-shapes algorithm outlined 2D profiles of the projected vertices of the intact patella, which serves as a surrogate for manually outlining the bone contour on fluoroscopic images in an experimental study. Briefly, alpha-shapes are a generalization of the convex hull and compute the external shape of a point cloud (Edelsbrunner et al., 1983). The intact profile was separated into anterior and posterior regions by locating the intersection between the intact and resection profiles. The anterior region, representing the remaining bone's surface projections, was used to simulate each implanted patella's target profile.

A leave-one-out evaluation was performed on three different imaging scenarios consisting of the 1st position alone (2 images), the 1st, 3rd, and 5th positions (6 images), and all five positions (10 images). Each of the 50 subjects were separately left out of the training set and new SSMs consisting of the remaining 49 subjects were created. An optimization algorithm was designed to change shape and alignment of a new instance of the SSM to match the 2D anterior target profiles of the left-out patella (Figure 4). The objective function minimized the difference between the 2D anterior target profiles and the projected profile of the 3D SSM. 2D image plane error, E , was calculated as the average of the root-mean-square (RMS) errors, e_i , across the images:

$$E = F(X) = \frac{1}{n} \sum_{i=1}^n e_i \quad (1)$$

$$e_i = \sqrt{\frac{1}{m} \sum_{j=1}^m d_{ij}^2} \quad (2)$$

RMS error was calculated using the 2D Euclidean distances, d_{ij} , measured between pairs of matched points in each image:

$$d_{ij} = \sqrt{\left\{ (x_{ij} - \hat{x}_{ij})^2 + (y_{ij} - \hat{y}_{ij})^2 + (z_{ij} - \hat{z}_{ij})^2 \right\}} \quad (3)$$

where n equals the number of images (or twice the number of positions) and m equals the number of matched points in an image. Design variables included 6 pose (3 rotation, 3 translation) parameters and 18 shape parameters (PC scores), which accounted for 95% of the variability in the training set. Using a subset of PC scores with this criteria eliminated features that did not vary across the population. Side constraints were placed on the design variables in an effort to limit the model to realistic geometries; the shape parameters were bounded by ± 3 standard deviations.

Given a manually-aligned starting pose of the mean SSM instance of the patella, its nodes were projected onto the image planes and the alpha-shapes profiles were matched to the simulated anterior target profiles using a perpendicular nearest neighbor search algorithm. Since imaged positions were linked by the known transformation (DLT) between them, only one set of 3D pose parameters was optimized to simultaneously align projections in the multiple image planes. To ultimately assess accuracy, 3D error on only the anterior region (as defined by the cut plane) was calculated as the Euclidean distance in mm between matched nodes for the SSM prediction and original patellar geometry. Errors were averaged across all subject predictions.

In addition to the leave-one-out evaluation, a k -fold cross-validation was performed to further assess accuracy and robustness. The population was split into k groups with each group tested against a model developed from the remaining $k-1$ groups (Hastie, 2009). A k value of 5 was chosen to balance bias and variance, and the dataset was randomly divided into 5 equally-sized groups with equal numbers of males and females. Errors were computed with patellae for each group tested against a model developed from the remaining 4 groups.

RESULTS

The SSM characterized anatomic variation in the training set with a small number of shape parameters; to visualize the shape changes described in the most significant modes, the mean geometry was perturbed by ± 1.5 standard deviations for each PC (Figure 5). The first PC mode, accounting for 65.0% of the total variation, was strongly correlated to scaling ($r = -0.95$ between PC1 score and patellar ML width). The second PC mode, accounting for 6.5% of the total variation, primarily described shape changes including the ML/AP aspect ratio and concavity of the articular surface at -1.5 standard deviations compared to a flatter articular surface at -1.5 standard deviations. The third PC mode, accounting for 4.6% of the total variation, described changes in shape of the medial articular facet and shifting of the mediolateral location of the apex. 18 PCs accounted for 95% of the total variation and were used in the optimization.

Simulating three pose scenarios, the virtual platform evaluated the accuracy of the predicted 3D patellar geometry considering 1, 3, and 5 imaging positions for each left-out subject (shown for all 5 imaging positions in Figure 6). Using only the 1st position, composed of a sagittal and axial (merchant) view, the 2D image plane error on the anterior region averaged 0.48 ± 0.15 pixels (mean $\pm$ std. dev.) across subjects (Figure 7). The associated average 3D geometric error was 0.58 ± 0.12 mm with noticeable differences where the patellar profile was not well defined, along nominally $\pm 45^\circ$ (Figure 7, left). Image sets using positions 1, 3 and 5 together achieved average 2D image plane errors of 1.05 ± 0.61 pixels and 3D geometric errors of 0.48 ± 0.10 mm. Using all 5 positions resulted in the largest average 2D image plane error (1.33 ± 0.50 pixels), but the lowest 3D geometric error (0.45 ± 0.07 mm) and overall better representation of the actual geometry across the anterior surface (Figures 7 and 8). On average, a single subject optimization completed in 58, 60, and 29 iterations spanning 10, 22, and 19 minutes for 1, 3 and 5 positions, respectively.

For the 5-fold cross-validation using the best performing imaging scenario (5 positions), the average 2D image plane error was 2.16 ± 1.20 pixels and the corresponding average 3D geometric error was 0.57 ± 0.18 mm for all resected patellae. In both measurements, the error magnitude was slightly larger in the 5-fold cross validation than for the leave-one-out evaluation and likely influenced by the size of the training sets (40 versus 50 patellae).

DISCUSSION

An approach was developed to predict a 3D geometric representation of the anterior patellar bone geometry using an SSM and sets of simulated stereo-radiographic images from initial posed positions, which enables PF kinematic tracking for TKR subjects without requiring specialized beaded implants or additional imaging. The controlled positioning of the subject to collect initial images is important to avoid occlusion by other metallic components, which is different from other SSM and fluoroscopy workflows that utilize images from the activity (Baka et al., 2014). By utilizing a leave-one-out approach with a virtual stereo radiography platform, the study robustly evaluated the predictive capability of the approach by using 50 subject-specific geometries to generate simulated target profiles. In practice, target profiles of the resected patella would be obtained from the radiographic images via manual digitization or edge detection methods, which was not included in the current study. With the current SSM-enabled approach, the need for post-surgical imaging can be eliminated, which may significantly reduce radiation exposure in some protocols. Radiation dose level for a knee CT was reported to be 0.16 mSv (Biswas et al., 2009) whereas the dose in the current stereo radiography protocol was less than 0.006 mSv (PCXMC STUK, Radiation and Nuclear Safety Authority, Helsinki, Finland), or about 4% of the radiation a person would receive from CT and equivalent to approximately one day of natural background radiation in the US. The virtual stereo-radiography platform was well suited to assess accuracy for the various pose scenarios without impact on patients; the best performing scenario has been implemented in *in vivo* data collections (Ali et al., 2016).

The SSM developed in this study described the main modes of variation in a population of 50 subjects. As in other studies (Bryan et al., 2010, Rao et al., 2013, Smoger et al., 2015), scaling was the primary mode of variation. In structures, like the patella, whose geometry is

relatively convex compared to the neighboring femur and tibia, it is understandable that scaling accounted for a large amount (65%) of the overall variability. Variations in the concavity of the articular surface described in Mode 2 were comparable to a prior study, which developed an SSM of the cartilage in the PF joint (Fitzpatrick et al., 2011) and illustrates the broader capability of SSMs to consistently characterize shape variation in related structures (i.e. patellar bone and cartilage).

The ability of the algorithm to predict 3D geometries from 2D radiographic profiles, simulating the stereo radiographic images that would be obtained from a data collection, was evaluated with a leave-one-out approach. Differences between the actual and predicted patellae occurred primarily in regions where the geometry was not well captured by the profiles associated with the imaging orientation. When more positions were incorporated into the optimization, differences in the average 3D shape (mean $\pm$ std. dev.) decreased from 0.58 ± 0.12 mm with 1 position to 0.45 ± 0.07 mm with all 5 positions; incidentally, average 2D errors increased from 0.48 ± 0.15 pixels to 1.33 ± 0.50 pixels in these scenarios as the optimization required balancing the 3D shape over a larger number of target profiles (Figure 7). The leave-one-out approach, which made geometric predictions on each subject excluded from the SSM training set, illustrated the ability to describe new patellar geometries with the SSM parameters, as well as the robustness of the SSM in characterizing the population. A subsequent k-fold cross validation with training sets of 40 patellae and testing sets of 10 patellae resulted in slightly larger errors, likely attributed to differences in the size of the training set.

In the virtual stereo radiography system, an alpha-shapes algorithm used to outline the dense 3D point projections provided similar results to edge-detection techniques (Canny et al., 1986) and served as a surrogate for digitally reconstructed radiographs in 3D-2D registration. The rounded shape of the patella was an ideal geometry for the alpha-shapes algorithm with a perpendicular nearest neighbor search, because interior and hidden edges are uncommon and the radial nature of the projected image reduces the number of possible nearest neighbors, thus improving optimization speed. This method was similar to the 2D distance maps developed by Kurazume et al. (2009), but simpler to implement.

Compared to shape prediction techniques with other intact bones (Baka et al., 2011; Li et al., 2008), the current methodology was applied to the resected patella in an implanted construct and thus required imaging in specific positions to capture the shape and resection plane. Current natural knee studies have used fluoroscopic images taken directly from the activities they seek to track and the natural femur and tibia provide sufficient geometric information for registration in a single set of biplane images (Baka et al., 2011; Li et al., 2008). The implanted patella presented a challenge in that a portion of the bone was removed and the surrounding metallic implants may occlude some of the remaining patellar bone in dynamic 6-DOF activities. As a result, prescribed stereo images in specific positions (subject supine with the knee flexed) were proposed in this methodology. These additional images contributed minimal additional radiation exposure during the stereo-radiographic procedure, and a reduction overall by replacing the need for a CT scan.

The ability to track *in vivo* PF kinematics with a 3D-2D registration method depends on the geometric accuracy of the 3D model, the registration algorithm, and the accuracy of the 2D image edge detection or manual digitization protocol. The current study focused on recreating accurate 3D geometric models by optimizing pose of a deformable SSM to simulated target profiles, and reported 3D errors of 0.45 ± 0.07 mm for the patella using 5 imaging positions. These errors are based on the virtual simulation and are thereby not a direct comparison to other automated x-ray based techniques; however, comparisons to the accuracy of other studies provide some context. Baka et al. (2012) reported a 0.78 mm point-to-surface difference between their SSM-based femur bone prediction derived from x-rays and ground truth models. Baka et al. (2014) applied their shape prediction method to simultaneously track the femur in a biplane fluoroscopic setup and achieved average tracking errors less than 0.07 mm translation and less than 0.2° rotation except for longitudinal rotation, which was reported at 0.32° . Similarly, Zhu et al. (2011) reported an absolute average geometric error of 0.9 mm for femur predictions. Geometric errors with the current approach were comparable to traditional segmentation, which is influenced by scan resolution, slice thickness (typically 1.0 to 1.5 mm) and inter-/intra-rater variability. Based on the small geometric errors and similar magnitudes in other fluoroscopy studies, the SSM-based predictions are sufficient for tracking in single and dual-plane studies.

In summary, this study evaluated the ability of an SSM to predict the 3D natural geometry of the implanted patellar bone from simulated 2D stereo-radiographic images of the knee joint. By establishing the resected patella geometry with this approach, PF kinematic tracking can be performed without additional pre-op or post-op CT imaging. The level of accuracy achieved was comparable to existing 3D geometric predictions for other bones and traditional segmentation considering inter-rater variability. The demonstrated approach streamlines the workflow for kinematic analysis of the TKR-implanted knee and provides the foundation for *in vivo* evaluations of implanted PF kinematics.

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Figure 1. Imaged position of the flexed left knee in the high speed stereo radiography (HSSR) system, top view (left). Representation of the imaged knee as seen on camera planes A and B (right). Note: Subject is supine.

Figure 2.
A representative patella, shown intact (left) and resected after automated implantation (right).

Figure 3.
Proposed imaging positions with varying orientation in the horizontal plane of the stereo-radiography system.

Figure 4.

Schematic of the optimization algorithm to reconstruct the full 3D patellar geometry from simulated sparse data. PC scores of the patellar bone SSM are optimized until the profile (black) of the projected point cloud (green) matches the simulated target profiles (red). The magenta lines represent corresponding points on the target and predicted profiles. RMS error calculated between these corresponding point sets is minimized by the optimization algorithm.

Figure 5. Representations of patellar bone for the first three principal component modes. Patellae are shown at ± 1.5 standard deviations. Views include coronal (top), sagittal (middle), and transverse (bottom).

Figure 6.

Nodes of an instance of the patellar bone SSM (green) projected by the DLT onto the camera image planes and its profile (black). The anterior region of the target profiles for a representative left-out subject (red). Initial rigid body alignment of the mean SSM instance is shown for position 1 (left) and then for all five positions of the optimized instance.

Figure 7.

Comparison of the three imaging scenarios for a representative subject: the 1st position (left), the 1st, 3rd, and 5th positions (center), and all five positions (right). Bar charts show the 2D image plane error (dark gray) in pixels and anterior surface 3D geometric error (light gray) in millimeters for each scenario. Contours represent absolute difference between the actual and predicted shape for the resected case. Contour units are in millimeters.

Figure 8. Contours of absolute difference between the actual and predicted shape for the resected case using all 5 positions. Four representative subjects are shown. Individual RMS errors are reported next to each subject. Units are in millimeters.