New Registration Algorithm for Determining 3D Knee Kinematics Using CT and Single-Plane Fluoroscopy with Improved Out-of-Plane Translation Accuracy

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ABSTRACT: To understand the kinematic effects of surgery, arthroplasty or conservative treatments, a noninvasive system to capture accurate 3D imaging of functional activities in prospective, controlled studies is required. To provide such a technique, a new algorithm was developed to register 3D CT data of normal bones to the same bones in a 2D fluoroscopy frame. The algorithm produces a digitally reconstructed radiograph (DRR) from the CT data and then filters this to produce an edge-enhanced image. The resulting image is then registered with an edge-enhanced version of the fluoroscopy frame using a new similarity measure called Cross-Correlation Residual Entropy (CCRE). The system was evaluated by implanting tantalum beads into three cadaveric knees to act as fiducial markers. The knees were flexed between 0° and 70°, and single-plane fluoroscopy data of the knees were acquired. CT data of the femur and tibia were then individually registered to the fluoroscopy images. No significant measurement bias was observed, and the standard deviation of the error in bead positions was 0.38 mm for in-plane translation and 0.42 degrees for rotation. To determine the accuracy of the registration algorithm for out-of-plane translations, fluoroscopy frames were scaled in size by fixed increments; the average standard deviation of the errors for out-of-plane translation was 0.65 mm. The ability to obtain such accurate 3D motion data from a noninvasive technique will enable prospective, longitudinal, and controlled studies of reconstruction surgery, and conservative management of joint pathologies. J Orthop Res 28:334–340, 2010

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The function of the knee relies on balancing the demands for stability required for weight bearing and protection of the articular tissues, and flexibility required for function in a range of activities.[1] This balance is sought between stability and flexibility for rehabilitation after injury, and for total knee replacement kinematics, where improved function and flexibility cannot be at the cost of increased wear. The bony architecture of the knee means it is an inherently unstable joint that relies on soft tissue constraints provided by ligaments, muscle forces, menisci, capsule, and retinaculum to provide guidance to planes of motion and restraints to motion beyond the physiological optimum.[2] Indeed, aberrant knee kinematics are responsible for accelerated wear in knee prostheses and in the natural knee, stressing articular cartilage beyond its physiological capacity to respond.[3,4]

For the past 20 years, roentgen stereophotogrammetry (RSA) has been the gold standard for mapping 3D kinematics.[5] RSA was slow at 2 frames/s; however, dynamic RSA is now capable of acquisition rates of up to 250 frames/s. The major limitation of RSA is that it is invasive, requiring prior implantation of tantalum beads as fiducial markers into the bone.[6] Consequently, it has been mainly used for postoperative studies. Prospective or controlled studies have been rare.[7]

Recent advances in digital medical imaging and computational software have enabled new noninvasive imaging techniques for kinematic modeling. Of particular interest are those studies founded on single-plane fluoroscopy, as bi-planar fluoroscopy has the operational disadvantages of providing only a limited field of view and requiring a double radiation exposure for each second of imaging.[8–10] Previously reported single-plane techniques can be divided into those that explored 3D kinematics of knees postarthroplasty[11–13] and normal knees.[12,14] The postarthroplasty techniques involved registering a 3D computer-aided-design (CAD) model of the tibial and femoral components of the knee implant with the associated 2D projection of the implant depicted in each fluoroscopy frame. The approach adopted for normal knees was also to create 3D models of the normal tibia and femur using techniques similar to those defined for postarthroplasty knees. The reported limitation for these methods is their lack of precision for out-of-plane translation.[15]

We aimed to develop an algorithm for image registration from 3D CT data to single-plane fluoroscopic images to facilitate the construction of accurate 3D models of knee kinematics. Our new approach to this problem was to generate a digitally reconstructed radiograph (DRR) from the CT data (rather than a 3D model) and then filter this to produce an edge-enhanced image. The resulting image was then registered with an edge-enhanced version of the fluoroscopy frame. The registration algorithm also uses a new similarity measure called Cross-Correlation Residual Entropy (CCRE) that has not previously been applied in such applications. The algorithm was required to be accurate to less than 1 mm and 1°, in six degrees of freedom, and hence permit

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kinematic analysis with strong external validity. The major clinical application of our proposed algorithm is the prospective analysis of knee ligament reconstruction surgery and total knee arthroplasty.

METHODS

Development of the Registration Algorithm

Image registration is the process of spatially aligning one image to another. Registration algorithms consist of two main components: a similarity measure and an optimization technique. When the similarity measure is maximized, the difference between the two images is minimized, and hence the two images are aligned spatially or “registered.” The optimization technique is a procedure for determining the transformation parameters that, when applied to one of the images, will result in a global maximum for the similarity measure, and hence spatial alignment with the second image. For 2D to 3D registration, this problem is more difficult as there are six parameters that can be varied to describe the 3D rigid-body motion of the object depicted by the 3D data. These six parameters control translation in the x, y, and z directions (denoted by Tx, Ty, and Tz, respectively) and rotation about the x, y, and z axes (denoted by Rx, Ry, and Rz, respectively).

The first step of the proposed image registration procedure is to capture CT and fluoroscopy data (Fig. 1). CT data was captured at 0.5 mm slice intervals on an Aquilion 16 (Toshiba, Tokyo, Japan) using bone and soft tissue algorithms, for 150 mm above and below the knee joint. The voxels belonging to the femur and tibia were then segmented out using a simple thresholding algorithm on the voxel values. Single-plane fluoroscopy was used to capture 2D video frames using an Axiom Artis MP fluoroscopy unit (Siemens, Munich, Germany). Fluoroscopy was performed in the sagittal plane, so the y axis refers to superior–inferior translation; x axis, anterior–posterior translation; and the z axis, medial–lateral translation.

Fluoroscopy images undergo pincushion distortion due to the projection of the X-ray beam onto the curved input surface of the X-ray image intensifier.[16,17] The pincushion distortion produced by the fluoroscopy unit was estimated by imaging a 400 mm $\times$ 400 mm perspex calibration cube, implanted with a grid of tantalum beads. This distortion was then corrected by applying a cubic warping function to the image of the cube so that the beads appeared in their correct alignment. The same distortion correction algorithm was applied to each frame of the fluoroscopy video before the registration algorithm was applied.

For each iteration of the registration algorithm, a 3D rigid-body geometric transform was applied to the CT volume containing the voxels corresponding to the tibia or femur to produce a change in the 3D position of the bone. To account for the conical spread of the X-ray beams from the fluoroscopy source, a 3D projective transform was also applied to the CT volume at this stage. This transform was used to expand parts of the bone that were closer to the X-ray source and shrink parts of the bone that were further away from the source. After applying this transform, a simple sum in the z direction simulated the projection onto a 2D plane of the conical X-ray beam of the fluoroscopy unit.

The 3D volume was then reduced to a 2D DRR by summing the voxel values of the transformed CT volume in the z direction. Both the DRR and corrected fluoroscopy frames were then filtered using a Laplacian-of-Gaussian (LoG) filter to highlight the edges of the objects (Fig. 2). An advantage of this new approach is that the algorithm is able to use edge information from the internal structure of the bones during

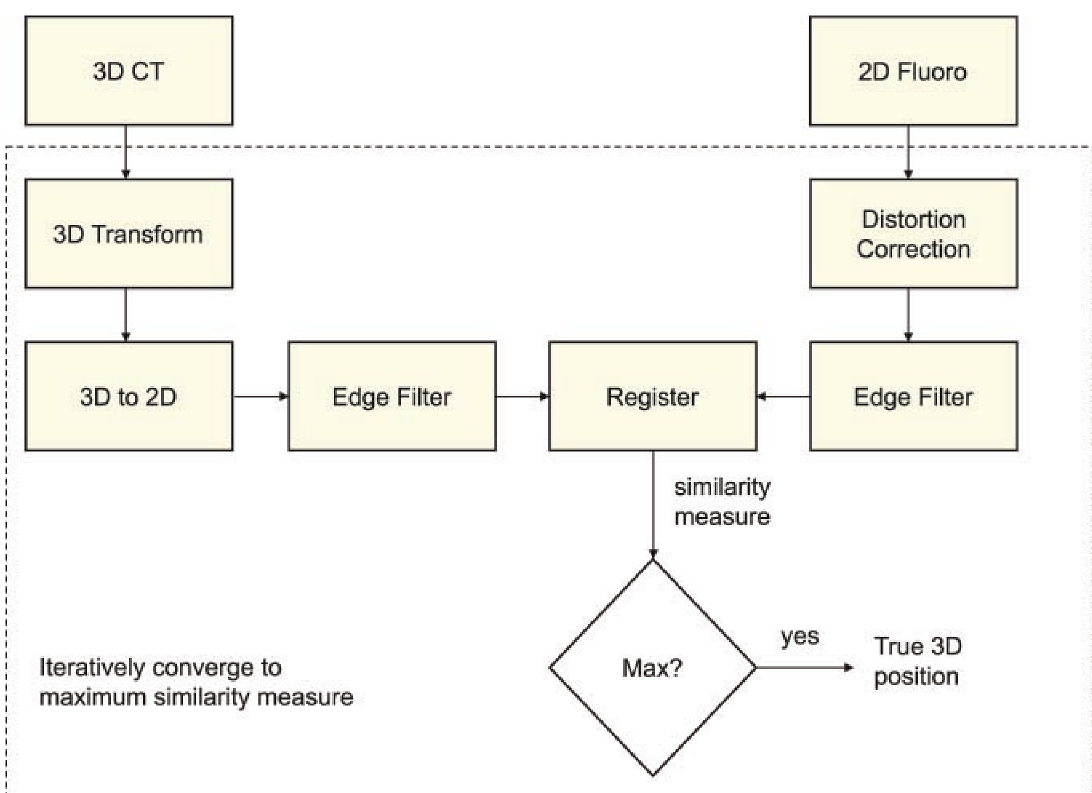

Figure 1. A block diagram of the registration procedure commencing with 3D CT and single-plane fluoroscopy.

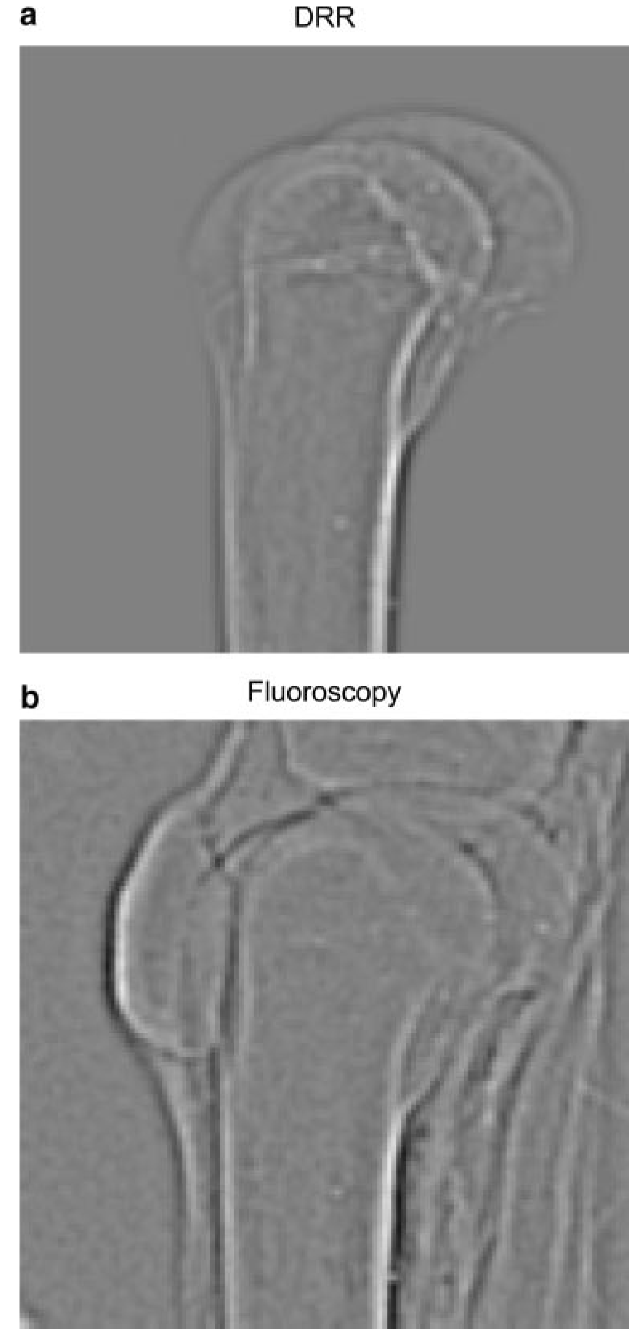

Figure 2. Edge-enhanced versions of (a) the 2D digitally reconstructed radiograph (DRR) produced from the CT volume of the femur and (b) the distortion-corrected fluoroscopy frame.

the registration process. This information is preserved during the creation of the DRR directly from the CT data. In previous approaches that require the creation of 3D surface models from the CT data, information about the internal structure is not required to calculate the surface model and is therefore discarded.

The similarity measure used in the registration algorithm is a relatively new measure called CCRE.[18] CCRE is similar to mutual information that has been used as a similarity measure in many medical imaging registration applications.[19] When compared to mutual information, CCRE provides faster convergence to the global maximum and a greater range of initial disparities for which successful convergence can be obtained[18,20] (Fig. 3). Curves were obtained by first registering a DRR containing a segmented tibia to a fluoroscopy frame of the same knee using the registration algorithm. Then each translational and rotational parameter was varied individually around this global maximum (keeping the values of the other parameters constant), and the value of the global similarity measure was recorded. No local maxima occurred for a range of about 2 mm and 2° around the global maximum for the translational and rotational parameters, respectively.

To increase this range of initial displacements for which the algorithm will converge to the global maximum, a coarse-to-fine approach was adopted. This approach is commonly used to increase the convergence range of gradient-descent optimization algorithms[21] and was achieved by applying the algorithm using progressively narrower LoG filter kernels. The wider filter kernels produce a much smoother edge image that allows the algorithm to avoid being trapped in a local maximum. The narrower filter kernels have a more limited range for which the algorithm will converge to the global maximum but provide greater accuracy. The overall accuracy of the algorithm is dependent on the accuracy of the finest stage of the coarse-to-fine approach and so curves were produced using the narrowest filter kernel (Fig. 3). Similar curves produced using wider filter kernels exhibited a much wider, flatter peak and hence had an increased range of initial displacements for which the algorithm converges to the global maximum. The disadvantage of using wider filter kernels is that the global maximum may not correspond as accurately to the true registered position as when using narrow kernels.

The parameters produced by the smoother edge image were used as the initial parameters in the following stage that uses a narrower filter kernel. This approach enables the algorithm to initially converge to an approximate global maximum from a wide range of initial displacements by using a wide LoG filter kernel, and then successively produce more accurate registration results by using further stages with narrower filter kernels.

A standard Gauss-Newton gradient-based optimization algorithm was used[22] with gradients calculated analytically using the method defined by Wang and Vemuri.[18]

Experiment 1

The aim was to evaluate the accuracy of the registration algorithm for in-plane translations and rotations around all three axes. Three lower limb specimens from embalmed cadavers were obtained (denoted by knees 101, 71, and 109). Two were from female donors; all were over 70 years of age. The specimens were implanted with 1 mm tantalum beads to act as gold standard fiducial markers. Beads in groups of three were placed into the femoral shaft, medial femoral epicondyle, lateral femoral epicondyle, medial tibia, lateral tibia, and tibial shaft.

The knees were passively flexed between 0° and 70° while fluoroscopy data were captured at 25 frames/s and a resolution of 0.5 mm/pixel. However, during the registration procedure, the resolution of the fluoroscopy frame and the CT volume was reduced to about 1 mm/pixel to reduce the time required to produce the DRR at each iteration. For simplicity, the resolution was reduced using nearest-neighbor interpolation. For every frame of data, the 3D femur and tibia data were individually registered to the fluoroscopy images using the proposed algorithm.

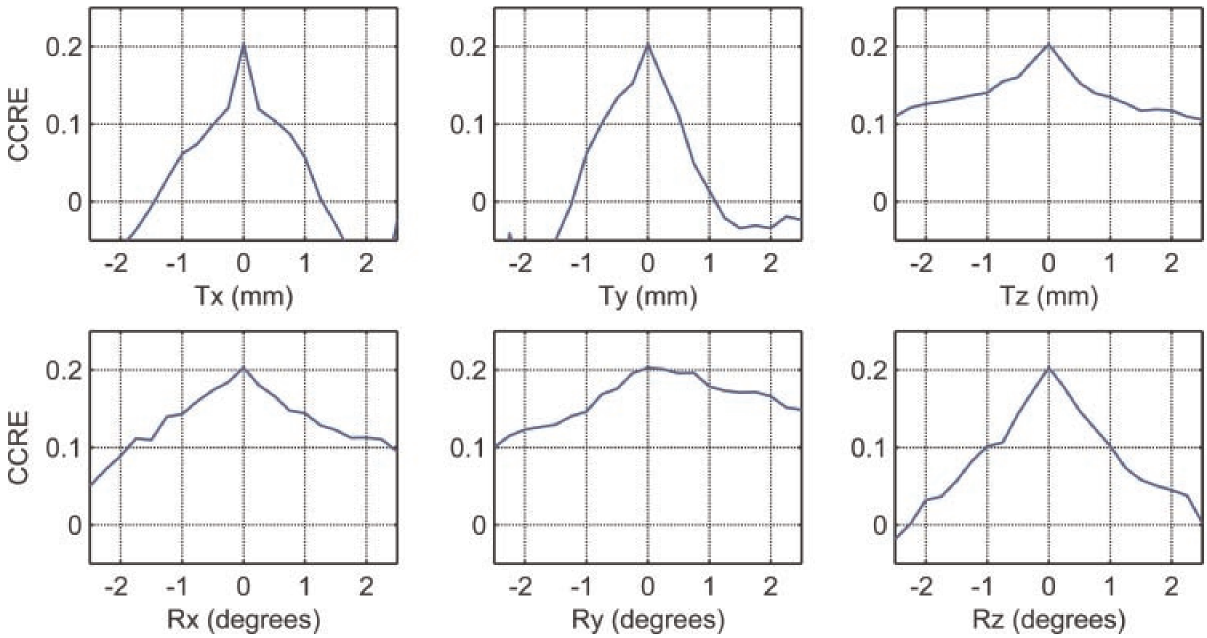

Figure 3. Cross-Correlation Residual Entropy (CCRE) values obtained by varying each rigid-body transformation parameter individually around the optimal set of values.

Experiment 2

The aim was to evaluate the accuracy of the registration algorithm for out-of-plane translation. Since the X-ray beams in the fluoroscopy unit diverge from a common point as they pass through the knee and arrive at the image intensifier, as an object moves closer to the X-ray source it will produce a larger image at the image intensifier (Fig. 4). Hence relative translation in the z direction is estimated by a change in the size of the object in the fluoroscopy frame. However, this change in size is typically difficult to measure accurately since a large movement in the z direction results in a relatively small change in the size of the object. For our fluoroscopy unit, a scale change of 0.0014 corresponds to a movement of 1 pixel closer to the X-ray source.

Changes in size by a factor of 0.0014 result in only small changes in the position of the beads in the fluoroscopy frame. It is therefore impractical to use the position of the tantalum beads to measure the accuracy of the registration algorithm for

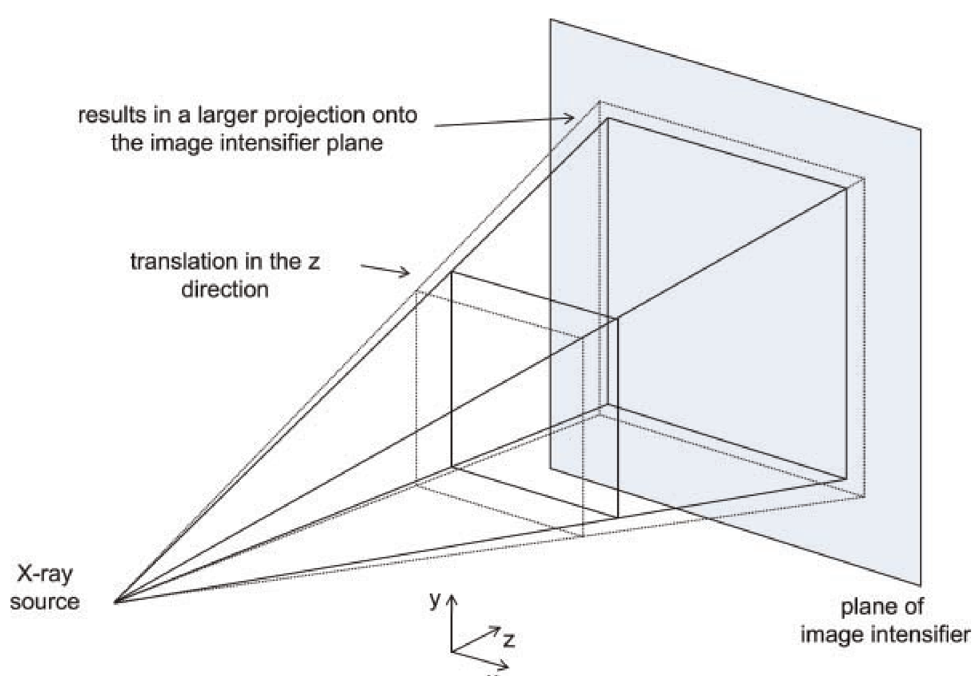

Figure 4. As an object moves closer to the X-ray source, it will produce a larger image at the image intensifier of the fluoroscopy unit.

out-of-plane translations. Instead, we used the approach of simulating a known movement in the z direction by changing the size of the object in the fluoroscopy frame with a 2D geometric transform. This transform effectively applies a zooming operation to the object in the fluoroscopy frame without changing the resolution of the image. Applying this transform produces a new fluoroscopy frame that is exactly as it would appear for an actual movement in the z direction by the object. The advantage of performing the experiment in this way is that the amount of scale change and hence the amount of out-of-plane translation is known exactly and can be used as the ground truth for evaluating the accuracy of the registration algorithm. This approach was used to simulate z direction translation of −4 pixels to +4 pixels in increments of 1 pixel for a representative frame of the tibia and femur for the three cadaveric specimens. The proposed registration algorithm was then used to determine the translation in the z direction. The amount of out-of-plane translation was converted from pixels to milliliters using the exact resolution of the fluoroscopy frame.

RESULTS

Experiment 1

The position of the implanted tantalum beads in the fluoroscopy frame and the position of the corresponding beads in the DRR were determined manually for 100 fluoroscopy frames depicting the tibia and femur of the three cadaveric knee specimens. For all six bones, no significant mean bias was found in the registration errors for in-plane translation or rotation. The standard deviation of the registration errors for the six bones is shown in Table 1. The last row shows the standard deviation of the error between bead positions averaged over all six bones. For in-plane translation, this average standard deviation was 0.39 mm and 0.36 mm for Tx and Ty, respectively, and for the three rotation angles, the average standard deviation was 0.46°, 0.40°, and 0.40° for Rx, Ry, and Rz, respectively.

Experiment 2

No significant mean bias in the registration errors for out-of-plane translation was found for any of the six bones. The standard deviation of the registration errors for the set of simulated out-of-plane translations is given for the six bones in Table 2. The average of this error standard deviation over the six bones tested was 0.65 mm.

DISCUSSION

Other proposed approaches for registering 3D CT data with single-plane fluoroscopy of normal knee bones have adopted the approach of creating 3D models of the normal tibia and femur, and registering 2D projections of these models with the fluoroscopy frame.[12,14] Fregly et. al. conducted a theoretical analysis of the accuracy of this approach.[15] The standard deviation of the errors in registration parameters in their study were 0.42 mm for in-plane translation, 5.6 mm for out-of-plane translation, and 1.3° for all rotations. Fregly et. al. also found that these theoretical levels of accuracy were consistent with the accuracy of similar approaches reported in the literature (but substantially less than was achieved with our proposed algorithm). Note that all errors reported in Experiment 1 are based on comparison with an experimental gold standard that includes all sources of error (experimental, modeling), whereas Fregly et al. reported “best case” theoretical errors based primarily on modeling considerations.

Our approach was to generate a DRR from the CT data (rather than a 3D model) and then filter this image to produce an edge-enhanced version that was then registered with an edge-enhanced version of the fluoroscopy frame. The use of edge-enhanced images via the LoG filtering stage adds to the accuracy of the registration procedure, but has the disadvantage of reducing the range of initial displacements for which the optimization algorithm will converge to the global maximum of the similarity measure. This problem was overcome by adopting a coarse-to-fine approach using progressively narrower filter kernels. The similarity measure used in the procedure was the newly-proposed CCRE measure, which provides superior performance to the more commonly used mutual information measure.[18]

The strengths of this image registration approach are that it provides a noninvasive procedure and that good accuracy is obtained, especially for out-of-plane translation. The registration errors obtained in Experiments 1 and 2 provide evidence of the superior performance of our proposed approach. The errors for in-plane translation and rotation are consistent, if not slightly better, than for other approaches, but the registration errors for the out-of-plane translation are much smaller than for existing techniques. The accuracy of this method of image

Table 1. Standard Deviation of Registration Errors for In-Plane Translation and Rotation of the Three Cadaveric Knee Specimens

		Tx (mm)	Ty (mm)	Rx (°)	Ry (°)	Rz (°)
Femur	101	0.31	0.22	0.32	0.40	0.36
	71	0.39	0.35	0.61	0.40	0.64
	109	0.49	0.42	0.29	0.29	0.24
Tibia	101	0.39	0.33	0.61	0.48	0.34
	71	0.38	0.41	0.49	0.43	0.54
	109	0.36	0.43	0.45	0.38	0.52
Average		0.39	0.36	0.46	0.40	0.44

Table 2. Standard Deviation of Registration Errors in Tz (mm) for the Femur and Tibia of the Three Cadaveric Knee Specimens

Femur			Tibia			
101	71	109	101	71	109	Average
0.56	0.86	0.80	0.27	0.62	0.77	0.65

registration also compares favorably with RSA. For rotations about the longitudinal and sagittal axes, the reported error standard deviations for RSA when measured at 10° intervals of knee flexion are 0.8° and 0.2°, respectively. For translations along the transverse, longitudinal, and sagittal axes, the error standard deviations for RSA are 0.2, 0.1, and 0.5 mm, respectively.[5] Our proposed method demonstrated accuracy of $<$0.9 mm in all three translations, and $<$0.6° in rotations. It remains to be seen whether this accuracy will be better still for image registration of postoperative knee implants, since object silhouettes will be clearer.

A disadvantage of this technique is the extra radiation exposure required to capture the CT data. An alternative to this approach is the use of MRI. However, Moro-oka et al. found that the fidelity of MRI-derived bone models is degraded by a variety of technical and practical factors, and shape-matching results typically will be inferior to those obtained with CT-derived bone models.[14]

Some possible sources of registration error that remain in the procedure include deficiencies in the removal of the pincushion distortion from the fluoroscopy frames and the inability of the optimization algorithm to converge to the exact global maximum. There is also inherent inaccuracy in the manual determination of the bead positions when evaluating the accuracy of the procedure. The most promising avenue for further improvement in accuracy stems from the fact that the resolution of the fluoroscopy and CT data were reduced from 0.5 mm/pixel to 1 mm/pixel to facilitate faster registration times. At this resolution, it is unlikely that any further significant improvements in accuracy could be obtained. If computational speed can be improved, using the data at its original resolution should allow further improvement in the accuracy of the procedure.

In summary, the new registration algorithm developed for the project provided superior accuracy for out-of-plane measurements. This new level of accuracy opens the way for sophisticated kinematic analysis of knee joints that has not been previously achievable when using single-plane fluoroscopy. If it can be established in vivo that this image registration technique has the accuracy of RSA, then this method will permit real-time, noninvasive kinematic studies using only single-plane fluoroscopy. This capability would enable prospective, longitudinal, and controlled studies of reconstruction surgery, and conservative management of joint pathologies.

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