Kinematic Measurement of Knee Prosthesis from Single-Plane Projection Images*

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In this paper, the measurement of 3D motion from 2D perspective projections of knee prosthesis is described. The technique reported by Banks and Hodge was further developed in this study. The estimation was performed in two steps. The first-step estimation was performed on the assumption of orthogonal projection. Then, the second-step estimation was subsequently carried out based upon the perspective projection to accomplish more accurate estimation. The simulation results have demonstrated that the technique archived sufficient accuracies of position/orientation estimation for prosthetic kinematics. Then we applied our algorithm to the CCD images, thereby examining the influences of various artifacts, possibly incorporated through an imaging process, on the estimation accuracies. We found that accuracies in the experiment were influenced mainly by the geometric discrepancies between the prosthesis component and computer generated model and by the spacial inconsistencies between the coordinate axes of the positioner and that of the computer model. However, we verified that our algorithm could achieve proper and consistent estimation even for the CCD images.

Key Words: Knee Prosthesis, Single-Plane Projection, Fourier Transform, Image-Based Matching, Kinematic Measurement, CCD Images

1. Introduction

An increase in the demand for total knee replacement (TKR) has necessitated the design of improved TKR from the aspects of durability and endurance. In-vivo measurement of prosthetic kinematics will yield significant information on the TKR design, and therefore many different techniques for knee kinematics analysis have been reported. In the past, several prosthesis parts such as heads, stems, and edges have been used as landmarks for Roentgen stereo-photogrammetry proposed by Selvik⁽¹⁾. The image-based technique⁽²⁾⁻⁽¹³⁾ may be highly feasible for the measurement of complex shaped implants.

In the image-based technique, the 2D perspective images of the object with unknown position and orien-

tation are compared with pre-computed 2D images generated using a precise geometrical model of the object, then the object's position and orientation are estimated based on a matching criterion. Several matching methods have been proposed for measurement of TKR kinematics. The method proposed by Banks and Hodge⁽²⁾ was an extension of a previously reported object recognition technique⁽³⁾. A library of contours of the CAD model in its various poses was prepared. Fourier descriptors (FDs) were used for compact representation of library templates and contours extracted from single projective X-ray images of prosthesis. Hoff et al.⁽⁴⁾ used the same method as used by Banks and Hodge to determine femoro-tibial contact positions. However, they used multi plane images to improve the accuracy of the algorithm. Another method⁽⁵⁾⁻⁽⁷⁾ involved using areas of X-ray images instead of the contours. The position and orientation estimation were performed by reducing the image projection to the fixed number of pixels of library templates. In this method, the library was not

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used, however a pair of X-ray images was necessary for accurate estimation. Zuffi et al.⁽⁸⁾ applied the method developed by Lavalée and Szeliski⁽⁹⁾ for measurement of TKR kinematics, in which single projection X-ray images were used instead of multiple projections for simplification. In this method, matching was performed by minimizing the Euclidean distance of the projection rays from the model surface. The projection rays were drawn from the projection center to the detected contour of the prosthetic X-ray images. This method did not require the library either, and the initial estimation of position and orientation was performed manually. Besides the measurement of TKR kinematics, the abovementioned image-based techniques have been widely applied to in-vivo measurement of human knee kinematics⁽¹⁰⁾ and assessment of wear of rate of the polyethylene insert^{(11)–(13)} as well.

Despite the conduction of extensive studies for the measurement of TKR kinematics, none of them have been put into practice for daily clinical usage to date. Possible reasons for this are that the accuracies of position and orientation estimation were still not sufficiently high for a clinical assessment, and estimation algorithms were sophisticated in addition to being time consuming and were likely affected by the image quality. More or less, the involvement of manual operation was indispensable.

Of the methods reported in the past, Banks and Hodge's method should be the most acknowledged because their simple algorithm is suitable for personal-computer-based analysis, measurement from single plane fluoroscopy is acceptable in routine clinical settings, and no manual operations/adjustments are required throughout the entire estimation process. The abovementioned FDs, permitting compact representation of shape information, are known to be robust against noise containing images. Yet some assumptions have been incorporated in their method, and therefore accuracies of position and orientation estimation of prosthesis remained eventually insufficient. Thus we attempted to extend Banks and Hodge's algorithm, by which the estimation accuracy was significantly improved while the feasibility of the estimation process was maintained. In this study, we propose a two-step estimation algorithm and the performance of the algorithm is fully investigated.

2. Theory

2.1 Principle of Banks and Hodge's algorithm⁽²⁾

Figure 1 shows the coordinate conventions for perspective imaging. In Fig. 1, the objective coordinates $\Sigma x-y-z$ and the screen coordinates $\Sigma X-Y-Z$ are set mutually parallel, and the Z and z axes are

Fig. 1 Coordinates of perspective transformation

aligned perpendicular to the image plane. We denote the distance between the projection center and the image plane, i.e. the principal distance as C_{pd} , and the distance between the objective center and the image plane as Z_{lib} . In Fig. 1, we call the object's translations along the x - and the y -axes as in-plane translations, and along the z -axis as out-of-plane translation. Also we call the object's rotations about the x - and the y -axes as out-of-plane rotations, and about the z -axis as in-plane rotation. The processes of position/orientation estimation for TKR motion are itemized as follows.

2.1.1 Shape representation

a) Calculation of the Fourier transform of the silhouette contour

Given the 2D coordinates of a closed contour in an image, it is possible to produce a periodic function by repeatedly tracing the contour in equal-length increments. Since the contour is described in terms of discrete picture coordinates, the discrete Fourier transform, $S(k)$ as shown in Eq. (1), is used to represent the silhouette's shape.

$$S(k) = \sum_{n=-N/2+1}^{N/2} a(n) e^{-j(2\pi kn/N)} \quad (1)$$

for

$-N/2+1 \leq n \leq N/2$, N : the total number of samples

b) Normalization of contour position and the object size/scale

The coefficient of the constant term in the discrete Fourier transform, $S(0)$ represents the geometric centroid of the contour. Setting $S(0)=0$ results in normalizing the contour position. The coefficient of largest magnitude, $|S(1)|$ represents the size of the contour, and setting all $S(k)=S(k)/|S(1)|$ results in normalizing the size of the contour. Through these normalizing processes, the values of translations along x, y and z axes have been calculated (estimated).

c) Normalization of the in-plane rotation and contour starting point (Estimation of an in-plane rotation)

The in-plane rotation about the z -axis is normalized by rotating the contour so that the phases of the largest and the second largest $S(k)$ s will respectively coincide with the predetermined phases, and the value of in-plane rotation has been calculated (estimated).

2.1.2 Shape library generation Through the abovementioned processes, the contours are represented by normalized Fourier descriptors (NFD) where each contour is normalized for an in-plane rotation and all translations. The contour shapes arranged with respect to the in-plane rotations about the x and y axes are stored in the library. Each library entry includes the NFD and all of the normalization parameters for that specific contour.

2.1.3 Matching/Pose estimation (Estimation of out-of-plane rotations) Pose estimation in terms of out-of-plane rotations about the x - and the y -axes should be performed by library matching. Library matching is the process by which the input NFD is compared with all the library NFDs in order to determine which library entries most closely resemble the input NFD. The measure of similarity between NFDs is simply the Euclidean distance between the coefficient vectors. Instead of increasing the density of library entries, the pose estimate by interpolating between several of the closest matching library entries is used.

2.2 Extended algorithm

For the sake of explanation, we will designate the algorithm stated in sec. 2.1 as the first-step estimation, whereas our extended algorithm is designated as the second-step estimation.

Through the first-step estimation, we could not accomplish highly accurate position and orientation estimation due to the inclusion of several assumptions. First of all, for the preparation of the library, we had to introduce the NFDs for the whole range of out-of-plane rotations of the implant. The input NFD had to be compared with all the library NFDs to determine the best match. Due to computation cost, we could not increase the density of library entries, which directly reflected the estimation accuracy. When the library NFDs were introduced, an assumption was made that the shape of an object's silhouette did not vary for translations or for in-plane rotation, even though the size of the silhouette would vary due to a translation along the z -axis. Consequently, we compared the input image with the so-called scaled orthographic images⁽¹⁴⁾ in the library. Apparently, we could not perform highly accurate estimation unless the accurate contour shape of the implant was represented. To overcome the problems in association with the first-step estimation, we applied the second-step estimation after the first-step estimation.

Fig. 2 Libraries in the first- and second-step estimation processes

Figure 2(a) and (b) explain the difference of the libraries used in the first- and the second-step estimations. In the second-step estimation (Fig. 2(b)), the computer-generated model is placed at the position and orientation introduced from the first-step estimation and the NFD corresponding to its projection image is obtained. Then, small angles of out-of-plane rotations about x and y axes are given to the model in both positive and negative directions respectively, thereby obtaining eight NFDs corresponding to eight images surrounding the first image. A new library is prepared, whose elements are the above nine NFDs, and the same process as the first-step estimation is repeated using the new library. Since we have arrived close to the actual position and orientation of the implant through the first-step estimation, we can prepare the new library with a small range of out-of-rotations, which enables us to decrease the intervals between elements. The model is already located and posed very close to the actual state of the implant such that the discrepancy in shapes between the model image and the actual image is small. The second-step estimation, including a renewal of the library with narrower elements' intervals should be repeated if higher estimation accuracy is required.

3. Simulation Study

3.1 Method

Simulation studies were conducted to assess the accuracy of the pose estimation technique. First, the 3D computer models of the femoral and tibial components of an implant (Type TCK, Nakashima Propel Co., Japan) were created from the data obtained by the 3D laser scanner (TDS-0530, Pulstec Co., Japan) in addition to their CAD data. The projection views of the models were generated by applying the x - and y -axes rotations to the model, silhouette contours

were drawn, and NFD were computed and stored as the library entries. Figure 3 illustrates a sequential image process for the implants. In Fig. 3(b) and (c), the polyethylene insert is not seen because of its fluoroscopic transparency.

Table 1 shows various parameters for position/orientation of the tibial and femoral components, the projection view system and their image processes used in each simulation. In Table 1, row (A) indicates the conditions of rotation and translation applied to the model to prepare the library to be used in the first-step estimation: The rotation about the z -axis and all translations were set at zero. The angles were set over a range of -12° to 12° about the x -axis and -24 to 24 about the y -axis, in 2 increments each, resulting in 325 elements in all.

Row (B) shows the conditions of the simulation used to examine the influence of the successive x and/or y axes rotations on the accuracies of estimation for all kinds of translations and rotations. The rotation about the z -axis was set at the values of rotation by $\pm 30^\circ$ from zero, and all translations were set at the values deviated by ± 5 mm from zero. The x and y axes rotation angles to be estimated were set over a range of -10° to 10° about the x -axis and -20° to 20° about the y -axis, in 2° increments each, and position

Fig. 3 Illustrations of a sequential image process for the implant

and orientation estimations were performed. The second-step estimations were repeated three times; a more precise library was prepared specifically around the position and orientation introduced through the previous trial, by halving the library entry intervals of the previous one.

Row (C) shows the reference conditions of the simulation; the results obtained from these conditions are to be compared with those from other conditions, (D) through (I), thereby examining the accuracies due to parametric changes applied to each condition. The rotation angles and translations to be estimated were set randomly over a range of expected joint motion during activity, and estimations were performed for 100 sets of position/orientation in all.

Row (D) shows the condition of simulation, in which the distance between the projection center and the screen, C_{pd} was set to be twofold that of (C), thereby examining the accuracy for more telescopic condition. Row (E) shows the condition of simulation, in which the library entry interval was set constant in all three trials in the second step. Rows (F) and (G) show the condition of simulation at various sampling numbers along the contour of the projection image to assess the effective sampling numbers for estimation. Rows (H) and (I) show the conditions of simulation at various sizes to assess the relationship between the image size and the estimation accuracy.

The standard variation values of errors were used to assess the estimation accuracies.

3.2 Results

Table 2(a) and (b) show the results from each simulation condition for the femoral and tibial components. In Table 2, the values in rows (B) through (I) are the results from simulation (B) through (I) in Table 1. The results from the simulation of the reference condition demonstrated that errors in rotations were smaller than 0.12° , and those in in-plane translations were smaller than 0.04 mm. The results also showed errors in the z direction to be about five

Table 1 Simulation conditions

	C_{pd} (mm)	Z_{lb} (mm)	rotation (deg)			translation (mm)			L.E.I. (deg)			sampling numbers	comments on the condition
			x	y	z	x	y	z	1st	2nd	3rd		
(A)	2025	1800	$-12 \sim 12^*$	$-24 \sim 24^*$	0	0	0	0	—	—	—	128	for the library elements
(B)	2025	1800	$-10 \sim 10^*$	$-20 \sim 20^*$	± 30	± 5	± 5	± 5	2	1	0.5	128	effects of successive x, y rotations
(C)	2025	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	2	1	0.5	128	reference condition
(D)	4050	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	2	1	0.5	128	telescopic projection
(E)	2025	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	2	2	2	128	no change of L.E.I
(F)	2025	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	2	1	0.5	16	decreasing sampling numbers
(G)	2025	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	2	1	0.5	1024	increasing sampling numbers
(H)	2025	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	2	1	0.5	128	halve the size of image
(I)	2025	1800	$-10 \sim 10$	$-20 \sim 20$	$-90 \sim 90$	$-10 \sim 10$	$-10 \sim 10$	$-10 \sim 10$	0.5	1	0.5	128	double the size of image

* 2deg increment, L.E.I.: Library Elements' Interval

Table 2 Simulation results
(a) Femoral component

	rotation (deg)			translation (mm)			comments on the condition
	x	y	z	x	y	z	
(B)	0.00±0.10	0.00±0.11	-0.01±0.08	-0.00±0.03	-0.00±0.03	-0.06±0.20	effects of successive x, y rotations
(C)	0.00±0.12	0.00±0.11	-0.01±0.08	-0.00±0.03	-0.00±0.04	-0.06±0.21	reference condition
(D)	0.00±0.11	0.01±0.13	0.00±0.12	-0.00±0.04	-0.00±0.04	0.13±0.74	telescopic projection
(E)	0.01±0.10	-0.01±0.11	-0.01±0.10	-0.00±0.04	-0.00±0.04	-0.07±0.63	no change of L.E.I
(F)	0.01±0.23	-0.01±0.13	-0.01±0.11	-0.00±0.06	-0.00±0.04	0.07±0.29	decreasing sampling numbers
(G)	0.00±0.13	0.00±0.10	-0.01±0.08	-0.00±0.03	-0.00±0.03	-0.06±0.20	increasing sampling numbers
(H)	0.01±0.21	-0.01±0.17	-0.01±0.13	-0.00±0.05	-0.00±0.04	0.07±0.38	halve the size of image
(I)	0.01±0.07	-0.01±0.06	-0.01±0.04	-0.00±0.02	-0.00±0.02	0.07±0.11	double the size of image

(B) Tibial component

	rotation (deg)			translation (mm)			comments on the condition
	x	y	z	x	y	z	
(B)	-0.04±0.21	0.01±0.15	-0.00±0.07	-0.00±0.02	-0.01±0.04	-0.05±0.24	effects of successive x, y rotations
(C)	-0.02±0.07	0.01±0.12	-0.01±0.04	0.00±0.02	-0.00±0.03	-0.07±0.18	reference condition
(D)	-0.05±0.68	0.01±0.70	0.00±0.08	-0.00±0.03	-0.00±0.04	0.28±1.72	telescopic projection
(E)	-0.02±0.30	-0.01±0.24	-0.00±0.06	-0.00±0.03	-0.00±0.03	0.17±1.14	no change of L.E.I
(F)	0.01±0.16	-0.01±0.32	-0.01±0.16	-0.00±0.05	-0.00±0.07	0.07±0.53	decreasing sampling numbers
(G)	-0.02±0.08	0.01±0.11	-0.01±0.04	0.00±0.02	-0.00±0.02	-0.07±0.19	increasing sampling numbers
(H)	0.01±0.17	-0.01±0.29	-0.01±0.09	-0.00±0.04	-0.00±0.04	0.07±0.46	halve the size of image
(I)	0.01±0.03	-0.01±0.08	-0.01±0.03	-0.00±0.02	-0.00±0.02	0.07±0.11	double the size of image

times larger than the errors for the x and the y directions. This was not only because the translation in the z direction resulted in changes of the shape, but also because significant changes of the size, namely the value of $|S(1)|$, were caused due to the out-of-plane rotations.

3.2.1 Relationship between the successive x , y rotations and accuracies (row (B)) We did not find any correlation between incremental changes in x and y rotation angles and the estimation accuracies. The maximum deviation of errors for the femoral component was found to be 0.11 for all rotations and 0.20 mm for all translations, and those for the tibial component were 0.21 and 0.24 mm, respectively. Errors for the tibial component were larger than those for the femoral component. The maximum error in z translation was large compared to those in x and y translations.

3.2.2 Influence of distance between the projection center and the screen (rows (C), (D)) When the projection distance C_{pd} was set large to obtain a telescopic viewing, errors increased not only for rotations but also for translations. Theoretically, in the case of telescopic viewing, the projection becomes more orthogonal, and therefore the individual rotations and translations are to be distinctly separated from each other. In an actual case, however, the error in the z -direction became so large that errors of other rotations and translations increased as

(a) Femoral component (b) Tibial component

Fig. 4 Decrease in errors according to the estimation processes

well.

3.2.3 Influence of library entries' interval in the second-step estimation (rows (C), (E)) By comparing between the results from (C) and those from (E), we found that the accuracies were not improved when the library entry intervals were kept constant, especially for the tibial component. Figure 4(a) and (b) show how the standard deviations of errors decreased according to the estimation process in the case of (C). Rapid decreases in errors between the first-step and the second-step are marked. We may conclude that the second-step estimation proposed in this study functions reasonably well.

3.2.4 Influence of sampling number (rows (F), (G)) Figure 5(a) and (b) show graphs of errors at three different sampling numbers. The

Fig. 5 Estimation errors at three different sampling numbers

Fig. 6 Estimation errors at three different image sizes

figures show large errors when the sampling number, N was set to 16, especially for the tibial component. No large differences are seen between $N=128$ and 1024, indicating that an increase in sampling number does not necessarily decrease error values.

3.2.5 Influence of size of image (rows (H), (I)) Figure 6(a) and (b) show graphs of errors for three different image sizes. Errors decreased as the image size increased. This is more typical for the tibial component than for the femoral component.

4. Experimental Study

4.1 Material and methods

Figure 7 shows the experimental setup. Each of the femoral and tibial components of the same prosthesis as used for the computer model were fixed on

Fig. 7 Illustration of experimental setup

the table of the apparatus (Six-Axis Positioner, Melles Griot Co., USA), that allows motion of 6 degree of freedom within the errors of $\pm 0.01^\circ$ for rotations and ± 0.01 mm for translations. A CCD camera (S602, Fuji Film Co., Japan) was fixed on the stand so that the optical axis was aligned with the z axis of the positioner. The table of the apparatus was moved to set the femoral or tibial component of the prosthesis at the desired position/orientation, then locked to maintain the chosen state and a photograph was taken by the CCD camera. The focus of the camera was adjusted manually but its auto focus function was not used. All the photographs were converted into binary cord images, the contours were drawn, and the position/orientation estimations were processed.

Three sets of experiments were performed. First, the experiments were performed under the same conditions as for simulation (B) in Table 1. For the sake of experimental convenience, we applied condition (B) instead of reference condition (C).

Second, in order to examine how out-of-focus images decrease in estimation accuracy, the focus of the camera was intentionally misadjusted; the focus position was displaced by 100 mm (slightly out-of-focus condition) and 200 mm (greatly out-of-focus condition) from the in-focus position, thereby obtaining blurry images for the estimation. Other parameters were fixed to those in condition (B), and the same estimation processes as the first experiment were repeated.

Third, photographs with known position and orientation of each of the femoral and tibial components were taken, then this shape information was employed to prepare a new type of library instead of using the shape information of the computer model.

We named it the real image library. To prepare the real image library, the rotations and translations applied to the component were set; the rotations about the z -axis and all translations were set at zero. The angles were set over a range of -9° to 9° about the x and y axes, in 3° increments each, resulting in 49 elements in all. The x and y axes rotation angles to be estimated were set at random over a range of -9° to 9° . The z axis rotation angles were set at random over a range of 0° to 90° . The translations along the x and y axes were set at random over a range of -10 mm to 10 mm. The translation along the z axis was set at zero in order to take in-focus photographs. Then, only the first step estimations were performed using the real image library. Also the estimations using the computer model library were performed for comparison.

4.2 Results

Figure 8 shows the comparison between the results obtained from the simulation (B) and the experiment for the femoral component. The experimental results indicated that errors in rotations were as large as 1° , and those in in-plane translations were up to 0.5 mm. The error in the z direction was more than three times larger than the errors for the x and the y directions.

Figure 9 shows the photographs of the femoral component taken at various focus conditions, and Figure 10 shows the estimation errors for the photographs of the femoral component taken at various focus conditions. As compared with the in-focus

Fig. 8 Comparison of estimation errors for the simulation and the experiment

Fig. 9 Photographs of femoral component taken at various focus conditions

images, the slightly out-of-focus images did not decrease the accuracy significantly.

Figure 11 shows the comparison between the results obtained using the real image library and the computer model library. Figure 11 demonstrates that accuracies were improved when the real image library was used; the error values were almost comparable to those from the simulation after the first-step estimation.

From the results shown in Fig. 8 through 11, we may conclude that the main sources contributing to the errors in the experiment are the misalignment between the coordinates of the computer model and experimental setup, and the geometrical mismatch between the prosthesis components and the corresponding computer model.

5. Discussion

5.1 Simulation

The results from the simulation studies demonstrated high accuracies in rotations and in in-plane translations. Poor accuracies along the z -axis, the axis vertical to the image plane, are inevitable as long as estimations are performed using a single plane projection. Although several researchers^{(2),(7),(8)} have commented that the displacement of the knee prosthesis components along the medio-lateral (z) axis is unlikely, these comments cannot justify the large errors of z -translation because errors of any kind of rotation/translation result in errors of other kinds of

Fig. 10 Estimation errors for the photographs taken at various focus conditions

Fig. 11 Comparison of estimation errors in the cases of using the computer model library and the real image library

rotation/translation as well. For the sake of simplicity, we used the assumption that for z -translation, the shape of an object's silhouette does not vary, while the size of the silhouette will vary. This assumption, so-called weak perspective (central) projection⁽¹⁴⁾ does not hold in the actual perspective projection; the shape does vary. For example, a projection image of a cube becomes more distorted according as a view distance becomes smaller.

When an object motion is small, an object's rotation and translation have the same effects on an image⁽¹⁴⁾. In Fig. 1, the component's small rotation about the y axis and its small translation along the x axis can generate similar changes in the silhouette contour. Possible sources contributing to the errors were found to have interrelations among them.

The second-step estimation proved to work reasonably well, since many of the influences mentioned above were eliminated or suppressed through the second-step estimation. We found that the estimation accuracy did not increase after sampling number N was larger than 128, while it did in proportion to the size of the image. This accounts for the so-called quantization error of the image. As the sampling number increases, the NFDs tend to contain higher frequency terms. Yet, the NFDs' higher frequency terms do not necessarily represent precise shape information due to the quantization errors. On the other hand, the quantization errors do decrease as the image size increases.

5.2 Experiment

We found that the errors in each rotation and translation appeared in the same manner as in the simulation, but the values were large in the experiment. Initially, we believed that the most critical point in the experiment was how to extract the contour in a low quality image. However, we found that the slightly out-of-focus images were not decreased in accuracy as much as we had expected, since the blurry images have an anti-aliasing effect, i.e. the smoothing of contours. We also found that relatively highly accurate estimations were achieved when we used the real image library.

Thus, we came to the conclusion that the most critical points were the mismatch of geometries between the prosthesis components and the corresponding computer generated model, and the misalignment between the coordinate system of the computer model and that of the prosthesis in the experimental system.

Figure 12(a) and (b) show the Euclidian distance maps for the femoral and tibial components as functions of x and y rotations; The Euclidian distances were calculated by the difference between the

(a) Femoral component (b) Tibial component

Fig. 12 A map of Euclidian distance of NFDs between the model and the component as functions of x and y rotation angles

(a) Femoral component (b) Tibial component

Fig. 13 Superimposition of silhouette shapes of the model and the component (The silhouette of the computer model is drawn with a closed line, and the silhouette of the prosthetic component is painted gray)

NFDs of the prosthesis component and the corresponding computer model when both are at the same position/orientation with respect to each other. Figure 12(a) and (b) indicate that the Euclidian distances are larger than 0.3 for the whole range of x and y rotations, and reach up to 0.8 for the femoral component, and 0.6 for the tibial component at the x and y rotation angles indicated by arrows, respectively. The values are pretty large considering the fact that the maximum value of Euclidian distance was less than 0.1 in the simulation.

Figure 13(a) and (b) demonstrate the superimpositions of the silhouette's contour shapes of the prosthesis and model at the x and y rotation angles indicated by an arrow in Fig. 12. As already mentioned by other researchers^{(2),(7),(8)}, we may conclude that the accuracy of the current model-based method is sensitive to dimensional tolerances of the implant. In the manufacturing process, the articulating surfaces are usually obtained with a very high level of reproducibility, whereas the shapes of other parts such as stems, pegs and lateral edges vary from product to product. If such is the case, we will have to reshape the computer model so as to eliminate geometrical mismatch with the implant as much as possible. Research is now being conducting how to reshape our computer model made by using CAD data,

in order to automatically eliminate mismatch with the individual implant.

As already mentioned in the introduction, extensive studies have been performed on kinematic measurements of TKR using the image-based technique. In the literature, comparisons among the previous studies were made with respect to the accuracies and computation cost. However, we cannot simply compare them because the conditions were diverse among them: multi plane projection images were used in some studies, while the involvements of manual operations were crucial during the estimation process in other studies. However, we verified that our algorithm could achieve suitable and consistent estimation, and is promising for a clinical application by the creation of a computer model which precisely duplicates the actual implant.

6. Conclusion

In this study, the image-based technique was developed and computer model studies and in-vitro measurements were performed to assess the feasibility of using single plane fluoroscopy for the measurement of 3D kinematics of TKR. We developed a two-step estimation technique to accomplish accurate position and orientation estimations for the prosthetic components.

The simulation results demonstrated that the technique achieved sufficient accuracies for the TKR kinematics measurement. While the in-vitro studies demonstrated that our algorithm could achieve proper and consistent estimation, the accuracies were not sufficient as compared with the simulation. We found that accuracies in the experiment were influenced more by the geometrical discrepancies between the prosthesis component and computer model rather than the image qualities due to focuses, resolutions and other artifacts.

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