Repeat validation of a method to measure in vivo three dimensional hip kinematics using computed tomography and fluoroscopy

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***Abstract*— Total hip arthroplasty or THA is a surgical procedure for the relief of significant disabling pain caused by osteoarthritis or hip fracture. Knowledge of the 3D kinematics of the hip during specific functional activities is important for THA component design. In this paper we compare kinematic measurements obtained by a new 2D-3D registration algorithm with measurements provided by the gold standard roentgen stereo analysis (RSA). The study validates a promising method for investigating the kinematics of some pathologies, which involves fitting three dimensional patient specific 3D CT scans to dynamic fluoroscopic images of the hip during functional activities. This is the first study in which single plane fluoroscopy has been used for kinematic measurements of natural hip bones. The main focus of the study is on the out-of-plane translation and rotation movements which are difficult to measure precisely using a single plane approach. From our experimental results we found that the precision of our proposed approach compares favourably with that of the most recent dual plane fluoroscopy approach.**

I. Introduction

At present the most accurate and clinical standard approach for 3D modelling of joint component kinematics is Roentgen Stereo analysis (RSA) [1]. Measuring the kinematics of joints whether normal, disease-affected or following a surgical procedure is an integral part of developing effective musculoskeletal treatment. In the RSA approach, tantalum beads are implanted in the bones during the THA surgery and then X-rays are projected through the joint in two imaging planes to generate a 3D spatial model. Dynamic RSA is now capable of capturing these X-ray images at rates of 250 frames per second which allows the kinematics of the joints to be modelled.

In the RSA approach, the tantalum beads are implanted under general anaesthesia which is the main limitation of this technique. Therefore preoperative kinematics or controlled studies using healthy joints are very rare.

Researchers have been trying to develop realistic ways for kinematic analysis and image guided interventions by taking advantage of recent progress in image processing. Few studies have reported the measurement of in vivo and in vitro kinematics of the hip. Some researchers have used bi-planar X-rays or dual plane fluoroscopy and some have used single plane fluoroscopy for measuring THA component kinematics. Lin et al.[4] and Tsai et al.[5] presented algorithms based on dual plane fluoroscopy for measuring the kinematics of natural bones in the hip. Penney et al. [2] and Sorin et al. [3] proposed methods using single plane fluoroscopy for post-operative measurements of THA implant component position. Whereas, in our proposed algorithm, we use a single plane fluoroscopy approach for measuring the kinematics of natural bones in the hip which has not previously been reported.

Lin et al. [4] and Tsai et al. [5] proposed similar algorithms in which the main focus was to validate a non-invasive dual fluoroscopic imaging system (DFIS) for measurement of hip kinematics. The objective of Tsai et al. [5] was to validate a DFIS for determination of the THA kinematics using both in-vitro and in-vivo approaches. They also presented a comparison between their method with RSA. Though the initial results of these methods were encouraging our main focus is on single-plane fluoroscopy.

Penney et al. [2] presented a method using 2D-3D registration to align both the prosthesis and the preoperative computed tomography (CT) volume of the hip bones to an X-ray image. Results show that their method is significantly more accurate than the plain-film method for calculating cup anteversion. The measurement technique described in their work has been designed for the purpose of validating a computer assisted orthopaedic surgery (CAOS) system. However, measurements taken directly from postoperative X-ray images require accurate patient positioning. Consequently it is difficult to obtain reliable results.

Sorin et al. [3] proposed a CT/X-ray matching algorithm named Xalign. The goal of this algorithm was to allow 3D anatomic pelvic and acetabula measurements on 2D anterior-posterior X-rays. The postoperative cup abduction, version and pelvic flexion angles were determined in three different ways: using CT images directly, applying the Xalign method, and finally by performing conventional (abduction only) measurements on AP pelvic X-rays. The cup orientation measured on CT images was taken as the ground truth. The Xalign measurement errors were defined as the difference between the CT cup values and those obtained by applying the matching method. The main source of errors for the algorithm was the matching of the synthetic X-ray with the

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real radiograph.

Improved methods for measuring in vivo three dimensional hip kinematics are needed to increase our understanding of pathologies as diverse as femoroacetabular impingement and total hip arthroplasty edge loading. In the present study, a 2D-3D registration algorithm has been developed which uses a new similarity measure and a fast gradient-based algorithm. The main focus of this paper is to compare the kinematic measurements produced by this new algorithm with those produced by RSA.

The remainder of the paper is organized as follows: Section II describes the 2D-3D registration algorithm and section III describes our experimental method. In section IV, we describe experimental results and finally section V contains our conclusions.

II. 2D-3D Registration Algorithm

In this section, our proposed registration algorithm for the application of registering 3D CT data to 2D single plane fluoroscopy frames will be described. Image registration is the process of spatially aligning two or more images of the same scene taken at different times, from different viewpoints and/or by different sensors [6]. In 2D-3D registration there are two core components: a similarity measure and an optimization technique.

The similarity measure gives a numerical value, which indicates how well the two sets of extracted information are aligned. In this work we utilize a new similarity measure called the sum of conditional variances (SCV) as it has been shown to produce more accurate results than the more commonly used Mutual Information (MI) measure [1], [6], [8], [9].

In our context, 3D rigid body motion of an object is described by 6 motion parameters. These 6 parameters control translation in the x (anterior-posterior), y (proximal-distal) and z (medial-lateral) directions (denoted by T_x, T_y and T_z respectively) and rotation about the x, y and z axes (denoted by R_x, R_y and R_z respectively).

In fluoroscopic imaging, X-rays are emitted from a point source, pass through the object and strike the image intensifier. The image intensifier then produces an image based on the attenuation of the detected X-rays. This property of fluoroscopy imaging is essential when measuring out-of-plane translation in 2D-3D registration. A digitally reconstructed radiograph (DRR) is generated by simulating X-rays passing through the CT volume and calculating the cumulative attenuation of the ray from each voxel it passes through.

In the present study, during every iteration of the registration algorithm, a 3D rigid-body geometric transform is applied to the THA component to produce a change in the 3D position of the bone. A perspective transform is also applied to duplicate the conical spread of the X-rays during the capture of fluoroscopy frames. The 3D volume is then reduced to a 2D digitally reconstructed radiograph (DRR) by summing the voxel values of the transformed component in the z direction.

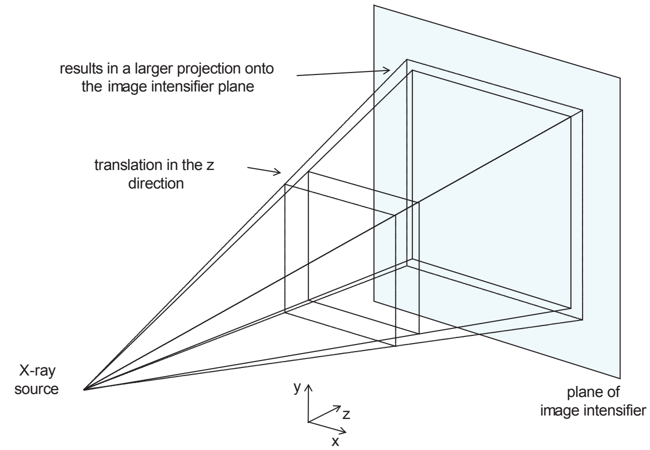

Fig. 1. As an object moves closer to the X-ray source it will produce a larger image at the image intensifier of the fluoroscopy unit.

Before the registration algorithm is performed, the images must be pre-processed to enhance the registration performance. After generating the DRR from the segmented CT we use a Laplacian-of-Gaussian (LoG) filter. The registration is performed in 3 stages and the width of the LoG filter is reduced in each stage to provide progressively sharper edges in the images.

To describe the calculations required for the SCV measure we denote pixel values of the filtered digitally reconstructed radiograph (DRR) by I_i and pixel values of the filtered fluoroscopy frame are denoted by R_i. Here, the subscript i indicates the pixel value at co-ordinates (x_i', y_i') and (x_i, y_i) in I and R respectively for $i = 1, 2, \cdots, N$ pixels in the images. To construct the joint histogram, the images I_i and R_i were quantized from the original 256 possible values to 64 possible values. Hence, the joint histogram has 64×64 bins corresponding to the 64 possible quantized values in I and R.

The joint histogram for images I and R will be denoted by $H(u, v)$ where u and v have values of 0,1,2,$\cdots$, $L-1$ and L is the number of quantized values in I and R. The values of the joint histogram represent the frequency that pixels at the same spatial location in quantized images I and R have the values u and v respectively. This joint histogram can be considered to consist of multiple conditional histograms corresponding to each of the possible quantized values of R_i. It is possible to estimate the conditional expectation (conditional mean) of the distributions represented by these histograms using the following formula:

$$E(I_i|R_i = v) = \frac{\sum_{u=0}^{L-1} uH(u, v)}{\sum_{u=0}^{L-1} H(u, v)} \tag{1}$$

Now, given these conditional mean values, an estimate of R_i can be found which has pixel values that correspond linearly to the pixel values in I_i. This estimate is found by replacing the pixel values in R with the conditional mean for that value of R and is given by:

Fig. 2. 400 mm×400 mm perspex calibration box.

Fig. 3. Cadaveric femur and hemi-pelvis with Tantalum marker beads.

$$\hat{R}_i = \begin{cases} E(I_i|R_i = v) & \text{when } R_i = v \\ 0 & \text{otherwise} \end{cases} \tag{2}$$

for v = 0,1,2,$\cdots$, $L-1$.

The new image $\hat{R}$ now has pixel values which correspond linearly to those in I. However, the spatial mis-alignment between I and $\hat{R}$ will be the same as that between I and R. So the transformation required to register I and R will be the same as that required to register I and $\hat{R}$. However, since I and $\hat{R}$ are linearly related, the sum-of-the-squared difference between I and $\hat{R}$ can be minimized. This new similarity measure is given by:

$$S = \sum_{i=1}^{N} \left(I_i - \hat{R}_i \right)^2 \tag{3}$$

Since this measure is the sum of the squared difference between values of I_i and their conditional mean, this equation actually represents a sum of the conditional variances (SCV) over all values of R_i [7].

The Standard Gauss-Newton Gradient based optimization technique is used in this study as it has higher accuracy than other techniques.

In the fluoroscopy unit the X-ray beams diverge from a common point as they pass through the knee and arrive at the image intensifier (as shown in Fig. 1). When an object moves closer to the X-ray source, it will produce a larger image at the image intensifier. Hence the out-of-plane translation in the z direction is estimated by a change in scale of the object in the fluoroscopy frame. However, this change in scale in the z direction is difficult to measure precisely as a large translation in the z direction will result in only a very small scale change in the fluoroscopy frame.

To determine the scale change due to the out-of-plane translation (In this case in the z direction), a 400 mm×400 mm perspex calibration box was used (shown in Fig. 2). Tantalum beads were implanted in the front and back planes of the box in a regular grid pattern. The perspective transform parameter W_z can be calculated as follows:

$$W_z = \frac{\frac{a}{b} - 1}{d} \tag{4}$$

Here d is the distance between the front and back plane of the calibration box, a is the distance between two beads in the front plane which is closer to the X-ray source as projected in the fluoroscopy frame and b is the projected distance between two parallel beads in the back plane of the grid box. The beads are spaced equally in both the planes but their projection will differ due to the conical spread. For our fluoroscopy unit, we have measured the perspective transform parameter to be W_z = 0.009. That is, a movement of 1 pixel in the z direction will correspond to a scale change of 0.009.

A. Kinematic Analysis

To evaluate the performance of the proposed algorithm, a cadaveric femur and hemi-pelvis were utilized. Tantalum marker beads were implanted into the cadaveric femur and hemi-pelvis and then a CT scan was captured to generate a 3D model(shown in Fig. 3). A custom apparatus was used, in which the hip was flexed from approximately 0^o to 60^o in five discrete movements, first with the hip abducted approximately 20 degrees throughout the hip flexion range, then repeated with the hip adducted approximately 20 degrees.

Fluoroscopy images were taken in each position and RSA was also used to measure the positions of the femur and hemi-pelvis. The fluoroscope axis was positioned approximately parallel to the acetabular rim. The orthogonal coordinate system chosen for comparison between RSA and fluoroscopic images was defined using the plane of the image intensifier of the fluoroscopy machine. This plane was defined as the $x-y$ plane and the z axis was defined as the axis normal to this plane. The kinematic measurements obtained from registering the digitally reconstructed radiograph (DRR), produced from the CT volume, with the fluoroscopic images were validated against the RSA measurements.

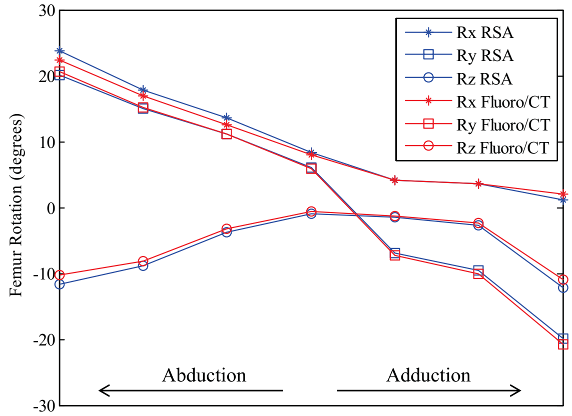

Fig. 4. Kinematic analysis for the cadaver femur (Rotation).

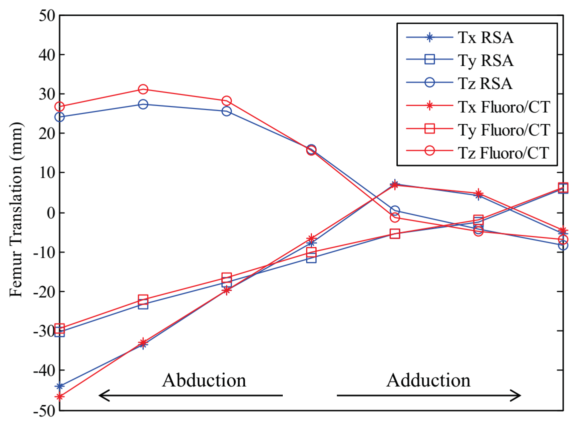

Fig. 5. Kinematic analysis for the cadaver femur (Translation).

III. Experimental Results

As mentioned previously, precise measurement for out-of-plane rotations (R_x and R_y) and translation (T_z) is difficult when using a single plane approach. Fig. 4 and Fig. 5 show the measured rotational and translational 3D rigid body transform parameters for the femur when using the proposed Fluoro/CT registration approach and RSA. The mean and standard deviation (SD) of the error between the RSA and Fluoro/CT measurements is given in Table 1. From Table 1 we can see that the mean and standard deviation (SD) of the error between RSA and Fluoro/CT is similar for the in-plane and out-of-plane rotation measurements. For translation, the in-plane precision is slightly better than for the out-of-plane measurement.

The overall errors measured in the current study were -0.70 ± 1.39 mm for all translations and -0.14 ± 1.68 degrees for rotation. As a comparison, the measured errors for the cadaveric study reported by Lin et al. [4], who used a similar validation procedure with RSA but used dual plane fluoroscopy, were 0.93 ± 1.13 mm for all translations and 0.59 ± 0.82 for all rotations. Hence, our results show that the overall precision of our proposed single plane fluoroscopy approach is similar to that obtained from the most recently proposed dual plane approach. However, our proposed approach has the benefits of a larger field of view for the patient to perform functional activities and a smaller dose of ionising radiation for the patient.

TABLE I

THE MEAN AND STANDARD DEVIATION (SD) OF THE ERRORS BETWEEN RSA AND FLUORO/CT (mm AND DEGREES)

	T_x	T_y	T_z	R_x	R_y	R_z
Mean	-0.10	-0.86	-1.14	0.93	0.30	-1.65
SD	1.24	0.49	1.99	1.79	0.97	1.02

IV. Conclusion

In this paper, we have described a new 2D-3D registration algorithm for the kinematic analysis of natural bones of the hip which uses single plane fluoroscopy. The precision provided by this new algorithm compares favourably with that of a previously proposed dual plane fluoroscopy approach. The major advantage of the new method is that it eliminates the need for implanting tantalum beads required for RSA or the use of a smaller field of view and a double dose of ionising radiation which is necessary in dual plane fluoroscopy based schemes.

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