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Precision Analysis of a Multi-slice Ultrasound Sensor for Non-invasive 3D Kinematic Analysis of Knee Joints

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Abstract—**Currently the standard clinical practice for measuring the motion of bones in a knee joint with sufficient precision involves implanting tantalum beads into the bones to act as fiducial markers prior to imaging using X-ray equipment. This procedure is invasive in nature and exposure to ionizing radiation imposes a cancer risk and the patient's movements are confined to a narrow field of view. In this paper, an ultrasound based system for non-invasive kinematic evaluation of knee joints is proposed. The results of an initial analysis show that this system can provide the precision required for non-invasive motion analysis while the patient performs normal physical activities.**

I. Introduction

Kinematic analysis allows the motion of individual bones in the knee joint to be estimated. These measurements can be used for the proper understanding of normal and abnormal joint trajectories, designing artificial knee components, evaluating different strategies for ligament reconstruction, identifying pain and wear inducing motion and developing therapeutic techniques to eliminate this motion in its early stage. During flexion, the two main bones in a knee joint, the tibia and femur, do not bend like a door hinge. Instead they rotate about several specific axes. Hence it is necessary to conduct the kinematic analysis in 3D space to capture the complete relative motion of bones in the knee joint.

Conventional motion analysis systems utilize optoelectronic or video based systems to track markers attached to the skin. These systems are non-invasive, easy-to-operate and widely used in motion analysis, computer graphics and animations. However, studies [1-4] have shown that they do not provide the precision necessary for many clinical applications due to the large relative movement of skin and soft tissues with respect to the underlying bones during dynamic activities. The clinical standard for 3D modelling of joint kinematics, implant performance and implant bearing wear has been *Roentgen Stereo Analysis* (RSA) [5]. This approach involves implanting tantalum beads or markers in the bones and then taking X-rays projected through the joint in two imaging planes to model 3D kinematics. However, it is essentially an invasive procedure. The beads are implanted during knee replacement surgery and therefore RSA is mainly used for post-operative assessments. RSA also does not allow 3D motion analysis of the knee joint while the patient performs normal everyday activities owing to the limited field of view of the X-ray equipment.

As stated previously, non-invasive skin mounted optical motion tracking is prone to errors due to the relative motion of the skin marker to the underlying bone. This relative motion is usually referred to as "skin wobble". The tracking system can accurately measure the 3D position of the sensor attached to the skin but this does not necessarily account for the position of the underlying bones in the joint. We propose to use an "intelligent" skin mounted sensor which contains ultrasound (US) transducers that can record images of the internal muscle tissue and bone surface while the patient is performing a particular activity. Figure 1(a) shows an illustration of this concept in 2 dimensions with the red arrow indicating the relative position of the skin-mounted marker to the underlying bone in the knee (the femur in this case).The position of the bone relative to the skin-mounted marker will be determined by registering the bone's surface in an initial US frame with the bone's surface in subsequent US frames.

Once the position of the skin-mounted sensor and the position of the bone relative to the sensor is known, these two distance vectors can be added together to find the true 3D position of the bone. As shown in Figure 1(b) for the 2D case, the movement of the bone relative to the sensor can be used to compensate for the movement of the sensor due to "skin wobble". Once the skin wobble has been removed, a much more accurate measurement of the true 3D position of the bones can be obtained. This measurement is shown as the green arrow in Figure 1(b). The merits of our proposed system over previous techniques are that it is non-invasive, and can be used while the patient is performing normal everyday activities.

In an earlier version of our proposed system we showed that it was possible to estimate five out of the six possible 3D rigid body movements between a sensor and a model femur [6]. The sensor used in our previous approach produced two orthogonal ultrasound slices. In this paper we will show how the use of a third ultrasound slice allows all six possible 3D movements between the sensor and the underlying bone to be estimated with high precision.

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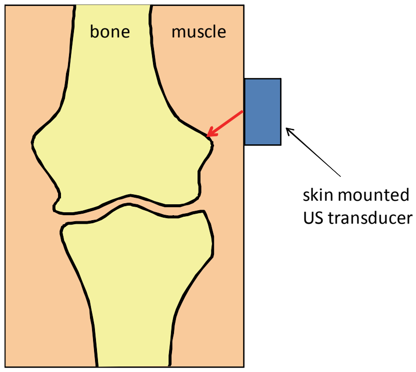
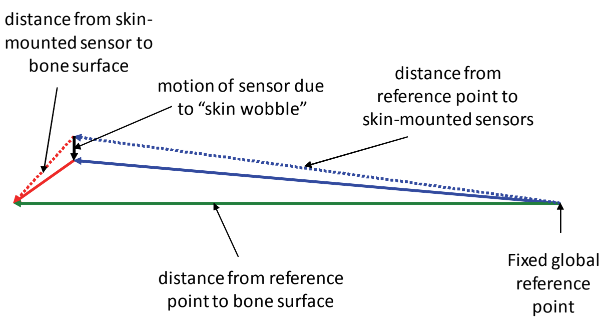


(a)



(b)

Figure 1. (a) Illustration of the intelligent sensor concept, (b) The bone-to-sensor distance can be used to compensate for the skin wobble of the sensor.

II. Measuring Skin Wobble Using US-US Image Registration

To find the 3D position of the femur and the tibia relative to the sensors, three image "slices" arranged in a novel 'H' configuration were captured using US transducers positioned on each bone. In our previous work [6] we showed that two orthogonal slices can be used to estimate five out of six motion parameters (three in-plane translations and two in-plane rotations). In the current system the remaining motion parameter (the out of plane rotation) can be extracted using a third image slice. The approximate projection of these slices on the bone surface is shown in red in Figure 2(a) for the femur and tibia. Consider the slices AA', BB' and CC', the corresponding slices from a B-mode US scan are shown in Figure 2(b). Slices AA' and CC' are parallel to each other while slice BB' is orthogonal to both thus forming a shape similar to the letter 'H'. For US image acquisition three US arrays are mounted firmly so there is no relative motion between the three image slices. By registering these slices simultaneously to the slices from an initial US scan, the current position of the bone relative to the sensor can be measured.

In our registration algorithm we use the optimization procedure proposed by Lucas and Kanade [7] with a new similarity measure called the sum of conditional variance (SCV) [8] to efficiently determine the affine transform parameters required to align the bone surface in the two images. Originally SCV was proposed for multimodal medical image registration [8] but recently it has been used for mono-modal registration technique where non-linear illumination change was an issue [9]. In [9] the authors showed that its performance was superior when compared to the mutual information (MI) and cross-cumulative residual entropy (CCRE) similarity measures .The benefits of using SCV can be summarized as follows: it is invariant to view dependent non-linear intensity variation, it provides a large convergence radius and is also computationally inexpensive.

The image registration algorithm can be explained as follows: Assume the current US image of the bone is $I(x_i, y_i)$ to be registered with the initial image $R(x'_i, y'_i)$. The coordinates (x'_i, y'_i) and (x_i, y_i) denote the locations of the pixels in I and R respectively for $i = 1 \ldots N$ where N is the number of pixels in the image. In our approach, the relationship between the locations of corresponding pixels in I and R is modelled by an affine transform.

The goal of the optimized registration process is to minimize the sum of conditional variance (SCV) and in so doing estimate the motion parameters that model the relationship between the coordinates of corresponding pixels. The SCV, S can be written as a function of the vector of motion parameters $\mathbf{m} = [m_1, m_2, \ldots m_6]^T$ (where $[.]^T$ denotes the transform operation) and computed using I and R as follows:

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

where $\hat{R}_i$ is the conditional expectation given by

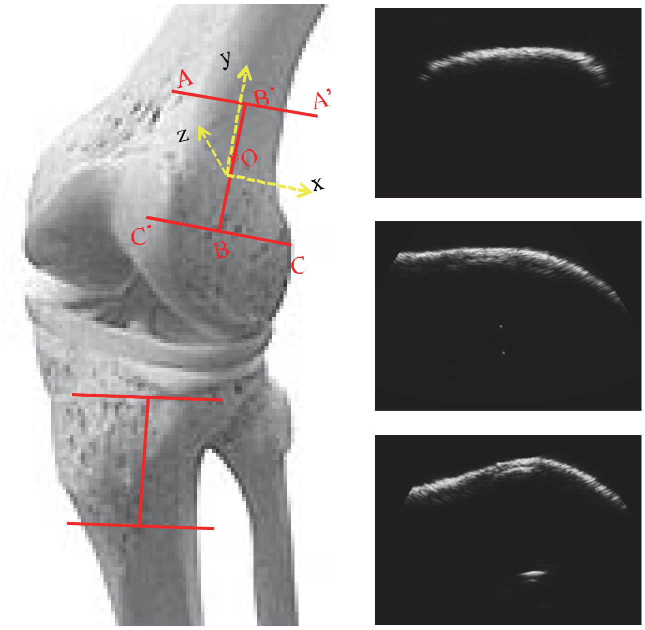
$$\hat{R}_i = E\left(I_i \,\middle|\, R_i \in \Delta_k\right) \tag{2}$$

which is calculated using the joint probability distribution of I and R. $E(\ )$ denotes the expectation operator and Δ_k is the histogram bin which includes R_i and $\hat{R}_i$.

The optimization procedure is required to find the value of the motion parameters which minimizes S. The minimization process uses an estimate of the values of S in a small neighborhood around the current value of $\mathbf{m}$ using a second order Taylor series approximation as follows:

$$S(\mathbf{m}+\mathbf{p}) = S(\mathbf{m}) + \mathbf{p}^T \nabla S(\mathbf{m}) + \tfrac{1}{2}\mathbf{p}^T \nabla^2 S(\mathbf{m})\mathbf{p} \tag{3}$$

where $\mathbf{p}$ is a vector that is added to the current vector of motion parameters to change the location of $\mathbf{m}$. In fact it is the value of the vector $\mathbf{p}$ that minimizes $S(\mathbf{m}+\mathbf{p})$ that is calculated at each iteration of the optimization procedure using the gradient-descent approach described in [7]. The parameter update vector $\mathbf{p}$ is added to the motion parameters at each iteration and a new version of I is calculated with pixel locations defined by the new motion parameters. In this



(a) (b)

Figure 2. (a) Approximate position of US transducer arrays, (b) B- mode US scan of a model bone at slice AA' (top), BB' (middle) and CC' (bottom).

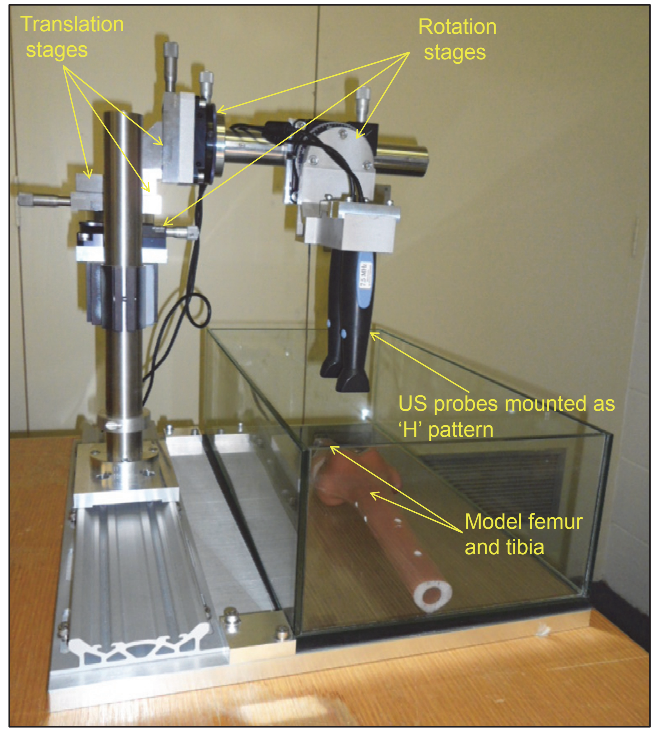


Figure 3. Experimental setup for both femur and tibia movement measurements.

way the image *I* is progressively transformed until it matches the image *R*. The process continues until some threshold is reached, e.g. a maximum number of iterations or a minimum change in *S*.

III. EXPERIMENTAL PROCEDURE

To validate the framework for determining the movement of the sensor relative to the bones in the knee we used the specially designed experimental apparatus shown in Fig. 3. Three Interson USB ultrasound probes were attached to the calibration apparatus. The experimental apparatus has three rotation and three translation stages to allow the probes to be accurately positioned. The rotation stages have an angular precision of 1/60th of a degree and the translation stages have a precision of 10 microns.

An artificial model of a femur and tibia were placed in a water-filled container as shown in Fig. 3. The water in the tank acted as a coupling medium and simulated the behaviour of muscle tissue. The frequency of the ultrasound signals used to capture the images was 24 MHz. From calibration experiments we determined that the resolution of the images captured using these probes was 0.132 mm/pixel.

The probes were then translated by ±4 mm in steps of 0.5mm in the x, y and z directions away from the initial positions and by ±3 degrees in steps of 1 degree around the *x*, *y* and *z* axes. At each position, the US probes were used to capture 2D B-mode scans of the surface of the model bone. These images were then registered to US images captured at the initial position of the probe. The amount of translation or rotation required to register the images was taken as an estimate of the movement of the probe relative to the surface of the bone. In-plane translations (along the x, y and z axes) and two in plane rotations (around the x and y axes) were determined from the two orthogonal US scans captured at the positions of AA' and BB'. The last out-of-plane rotation (about the z axis) was determined using parallel image slices at AA' and CC'. The estimates of the in-plane translations and rotations came directly from the output of the image registration. The out-of-plane rotation around the z axis was estimated using the equal and opposite translations in the parallel scans at AA' and CC'. From the known radii of rotation, OB' and OB, the angular displacement of the two parallel probes was calculated and averaged to produce the estimate of the rotation around the z axis.

IV. RESULTS

The error between the displacements measured by the registration of the US images and the true displacements of the probes measured using the stages on the experimental apparatus are shown in Fig. 4. (a),(b) for translation and Fig. 4. (c),(d) for rotation. The mean and standard deviation of these errors are shown in Table I. *Tx, Ty* and *Tz* denote translations in the x, y and z directions respectively and *Rx, Ry* and *Rz* denote rotations about the x, y and z axis respectively. These results show that the shape of the bone's surface at the positions which were scanned by the US probe have enough detail to allow an accurate measurement of the relative position of the US probe from one scan to another. They show that our proposed registration system is able to

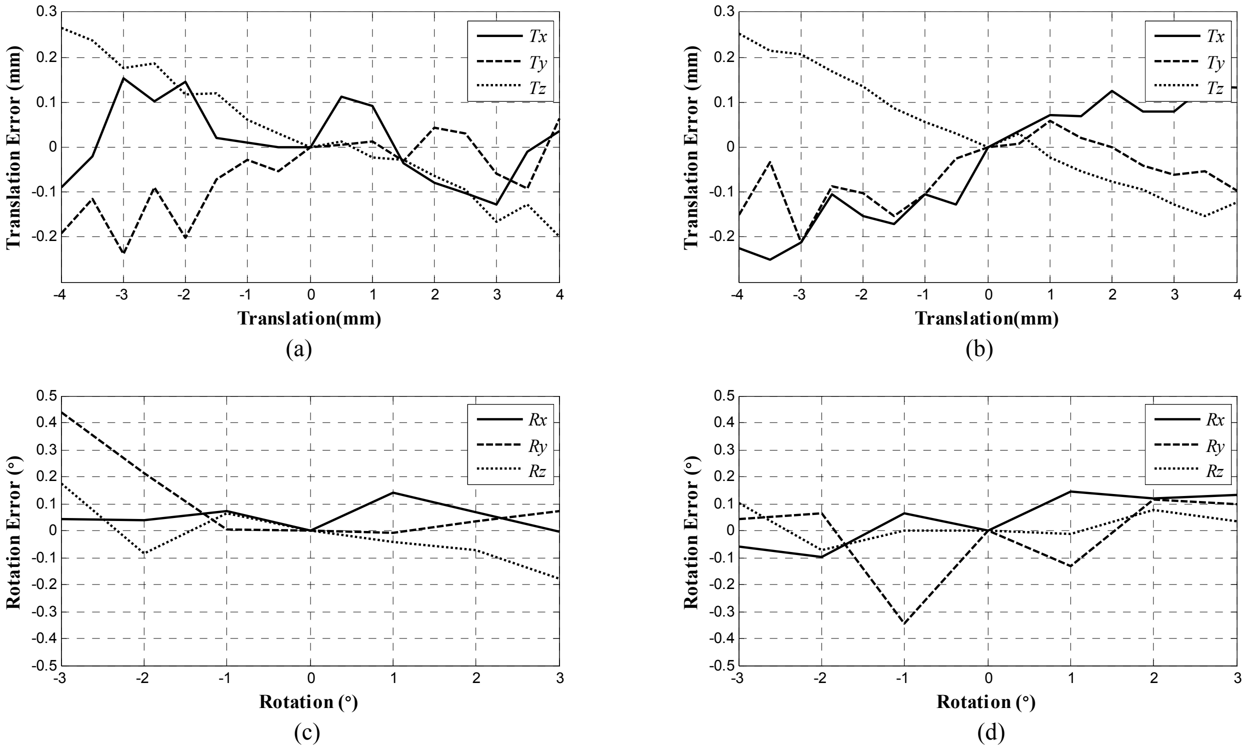


Figure 4. Measurement error analysis of the proposed system accounts for: (a), (c) femur translation and rotation, and (b), (d) tibia translation and rotation respectively.

estimate the motion of an ultrasound probe relative to the surface of a scanned bone with very high precision (standard deviation of error). The precision of the proposed system compares favourably with the current clinical standard of RSA which has a reported precision of 0.2 to 0.8 degrees for rotation and 0.1 to 0.5 mm for translation [5].

TABLE I. MEAN AND STANDARD DEVIATION (SD) OF ERROR

		Translation (mm)			Rotation (°)		
		Tx	*Ty*	*Tz*	*Rx*	*Ry*	*Rz*
Femur	**Mean**	0.009	-0.062	0.023	0.050	0.010	-0.022
	SD	0.083	0.085	0.134	0.046	0.152	0.105
Tibia	**Mean**	-0.028	-0.081	0.008	0.043	-0.022	0.018
	SD	0.133	0.068	0.125	0.090	0.151	0.054

CONCLUSION

In this paper we have presented a novel non-invasive approach to measure motion of the bones in a knee using multi-slice B-mode ultrasound and image registration. Our image registration method determines the position of bony landmarks relative to a B-mode ultrasound sensor array. The method takes advantage of the high spatial resolution of these ultrasound images to provide an average measurement precision of less than 0.1 mm and 0.1 degrees. The advantages of our proposed system over previous techniques are that it is non-invasive and can be used while the patient is performing normal everyday activities.

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