PRECISION ANALYSIS OF AN INTELLIGENT SKIN MARKER FOR NON-INVASIVE KINEMATIC ANALYSIS OF KNEE JOINTS

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ABSTRACT

Kinematic analysis of normal and artificial knee joints is an important diagnostic tool for the orthopaedic research community. Abnormal motion in a knee joint gives rise to pain and eventually causes chronic injury. Currently the standard way to measure the motion of bones in a knee joint with high precision is by implanting tantalum beads in the bones prior to imaging using X-ray equipment. This technique is invasive and the required exposure to ionizing radiation imposes a significant cancer risk. Moreover, during the procedure the patient cannot perform normal everyday activities due to their confinement to the limited field of view of the X-ray equipment. In this paper, we propose a non-invasive US-US registration based framework which can successfully capture motion of bones in a knee joint with the required precision using dynamic images produced by dual ultrasound transducer arrays. Our results show that this approach is capable of producing the precision required for 3D kinematic analysis of knee joint without the requirement for any invasive procedures and can be used while the patient performs normal activities.

Index Terms— joint kinematics, image registration; artificial knee, fluoroscopy, ultrasound

1. INTRODUCTION

Visualization of the motion of individual bones in a knee joint not only facilitates proper intervention procedures but also reveals significant details to orthopaedic surgeons for the planning of knee replacement and reconstruction surgery. This type of study has several important applications including: providing valuable information during knee replacement surgery, enabling the comparison between the motion of normal and abnormal knees for designing artificial knee components, investigating how abnormal motion influences the resulting early wear of the components in an artificial knee joint, evaluating different types of techniques for ligament reconstruction, identifying pain and wear inducing motion and developing therapeutic strategies to prevent this motion in its early stages.

Analysis of knee joint kinematics should be conducted in 3D due to the natural way that bones in the knee move relative to each other. As the knee flexes, 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 in a complete normal bend. Hence it is necessary to evaluate the kinematic analysis in 3D space to capture the complete relative motion of bones in the knee joint.

Traditional motion analysis methods 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. As a result, pre-operative or controlled studies are very rare. The beads are implanted during knee replacement surgery and therefore RSA is mainly used for post-operative assessments.

Hence RSA suffers from the following disadvantages: a) in practice it is usually only applied to artificial knee joints due to the requirement of marker implantation for achieving better accuracy. b) It requires the patient to be exposed to ionizing radiation during the capture of the X-ray images. This exposure has an associated elevated risk of the patient contracting cancer. c) It does not allow 3D motion analysis of the knee joint while the patient performs normal everyday activities due to the limited field of view of the X-ray equipment.

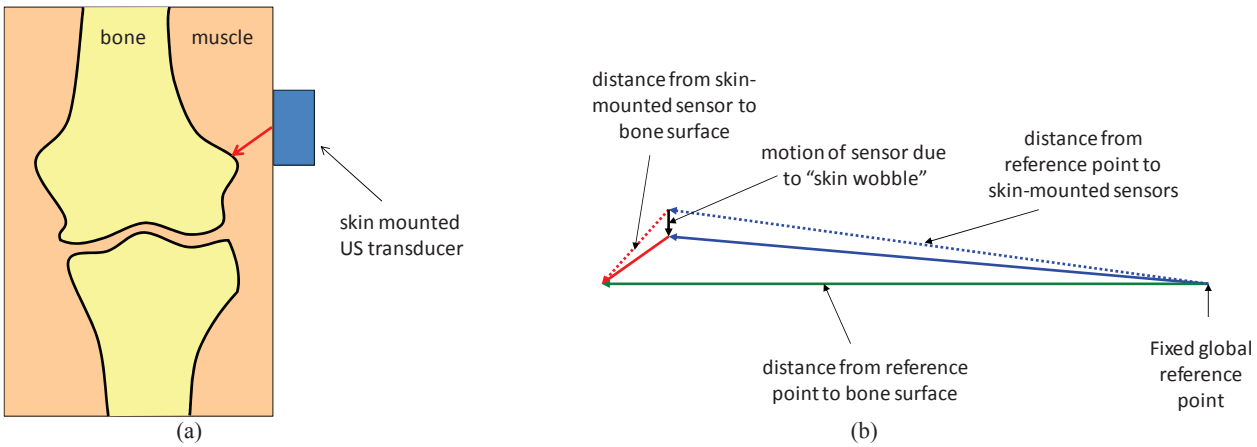


Figure 1. (a) Illustration of the intelligent marker concept, (b) The bone-to-marker distance can be used to compensate for the skin wobble of the marker.

As mentioned previously, the main problem with skin mounted optical motion tracking is the relative motion of the skin marker to the underlying bone. This is usually referred to as "skin wobble". The tracking system can accurately measure the 3D position of the sensor but this does not necessarily correspond to the position of the bones in the joint. We propose to use an "intelligent" skin-mounted marker 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 marker and the position of the bone relative to the marker 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 marker can be used to compensate for the movement of the marker 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 this paper we will explain the proposed US-US registration approach and present the results of an experimental analysis of the proposed technique to determine the possible precision of such a device. These results show that our proposed method is able to provide measurement precision of less than 0.1 mm for translation and 0.1 degrees for rotation.

2. MEASURING SKIN WOBBLE USING US-US IMAGE REGISTRATION

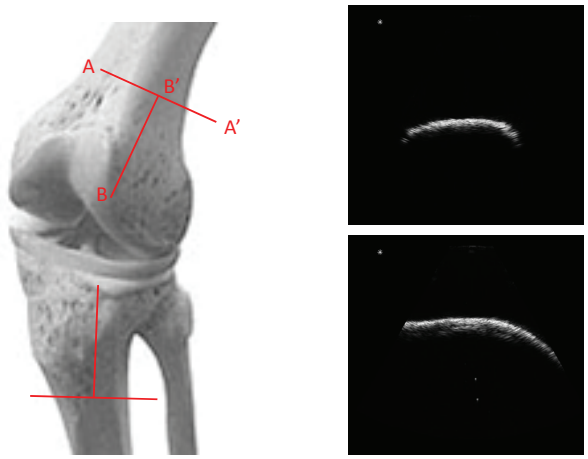
To find the 3D position of the femur and the tibia relative to the intelligent markers, two orthogonal image "slices" must be captured using US transducers mounted in markers positioned on each bone. The approximate position of these two slices is shown in red in Figure 2(a) for the femur and tibia. Consider the two slices AA' and BB', the corresponding slices from a B-mode US scan are shown in Figures 2 (b) and (c) respectively. The two arrays are aligned at 90 degrees to each other and fixed together so there is no relative motion between the two slices. By registering both 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 [6] to efficiently determine the affine transform parameters required to align the bone surface in the two images. This algorithm can be explained as follows: Assume the current US image of the bone $I(x_i', y_i')$ is 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 modeled by an affine transform defined by the following equations:

$$x' = m_1 x + m_2 y + m_3$$
$$y' = m_4 x + m_5 y + m_6 \quad (1)$$

where $m_1 \ldots m_6$ are the parameters that define the transform.

The goal of the optimized registration process is to minimize the sum of the squared difference (SSD) between the current and initial images and in so doing estimate the



(a) (b)

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

motion parameters that model the relationship between the coordinates of corresponding pixels. The SSD between I and R is given by

$$D(\mathbf{m}) = \sum_{i=1}^{N} \left(I(x_i', y_i') - R(x_i, y_i)\right)^2 \qquad (2)$$

In (2) the SSD is 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. The optimization procedure is required to find the value of the motion parameters which minimizes D. The minimization process uses an estimate of the values of D in a small neighborhood around the current value of $\mathbf{m}$ using a second order Taylor series approximation as follows:

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

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 $D(\mathbf{m}+\mathbf{p})$ that is calculated at each iteration of the optimization procedure using the gradient-descent approach described in [6].

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 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 D.

3. EXPERIMENTAL PROCEDURE

To validate the framework for determining the movement of the intelligent marker relative to the bones in the knee we used the specially designed experimental apparatus shown in Fig. 3. Two orthogonally aligned Interson USB ultrasound probes were attached to the calibration apparatus which 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.

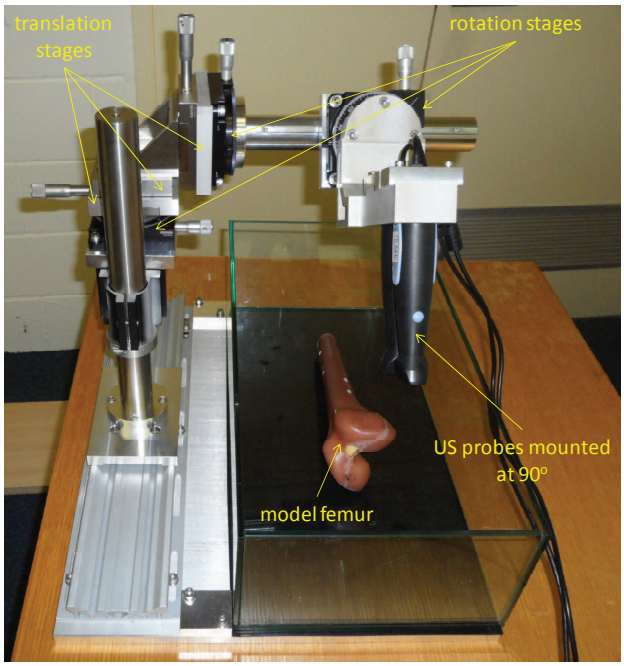


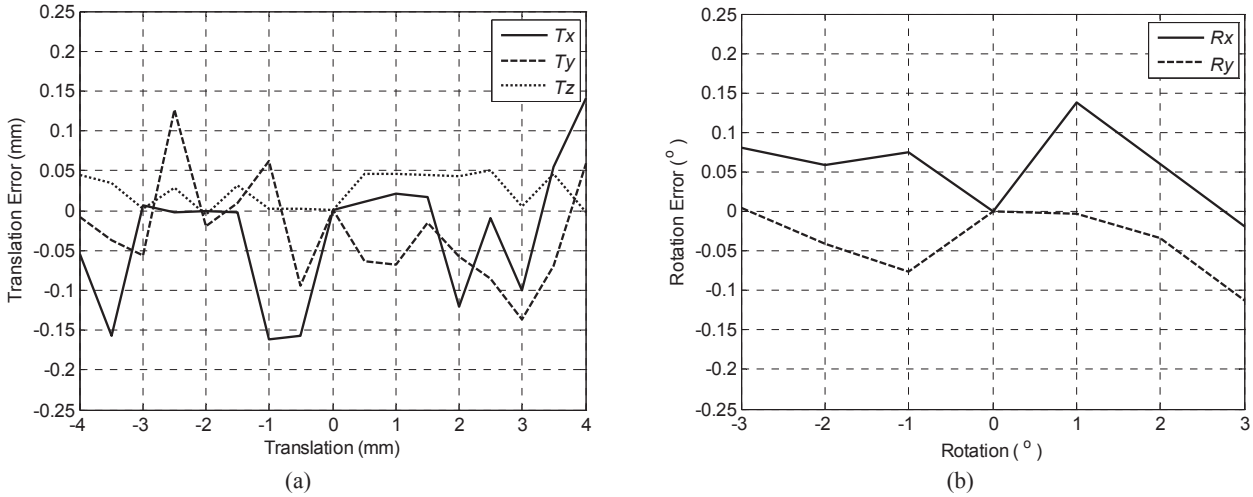
Figure 3. Experimental apparatus.

An artificial model of a femur was placed in a water-filled container as shown in Fig. 3. The water in the tank acted as a coupling medium and simulated the behavior of muscle tissue. The frequency of the ultrasound signals used to capture the images was 15 MHz. From calibration experiments we determined that the resolution of the images captured using these probe was 0.132 mm/pixel.

The probes were then translated by $\pm$ 4mm in steps of 0.5 mm in each direction away from the initial position and by $\pm$ 3 degrees in steps of 1 degree around the two axes normal to the planes of the US scans. At each position, the US probes were used to capture orthogonal 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.

4. RESULTS

The error between the displacement measured by the registration of the US images and the true displacements of the probes measured using the stages on the experimental



(a) (b)

Figure 4. Measurement error of the proposed US-US registration algorithm for (a) Translation and (b) Rotation.

apparatus are shown in Fig. 4. (a) for translation and Fig. 4. (b) for rotation. The mean and standard deviation of these errors is shown in Table I. *Tx*, *Ty* and *Tz* denote translation in the *x*, *y* and *z* directions respectively and *Rx* and *Ry* denote rotation about the *x* and *y* 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 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 favorably 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 MEASUREMENT ERRORS

	Translation (mm)			**Rotation (^{o})**	
	Tx	***Ty***	***Tz***	***Rx***	***Ry***
Mean	-0.030	-0.027	0.025	0.056	-0.038
SD	0.083	0.065	0.022	0.053	0.044

5. CONCLUSIONS

In this paper we have presented a novel non-invasive approach to measure motion of the bones in a knee using dual B-mode ultrasound probes and image registration. Our image registration method determines the position of bony landmarks relative to a B-mode ultrasound sensor arrays. The method takes advantage of the high spatial resolution of these ultrasound images to provide 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.

In future work for this project we aim to develop a system which can measure all three rotation angles. It is expected that this system will use image registration to register the 2D US scans with slices from a CT scan of the bone. We also intend to evaluate the system on knee cadavers to investigate the effect of the movement of soft tissue on the precision of the proposed system.

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