A novel image registration application for the *in vivo* quantification of joint kinematics

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Abstract. The transformation of rigid objects in 3 dimensional space can be defined by a 3x4 affine matrix, and these transformations can be used to quantify the kinematics of joints such as the knee. If good quality MRI data are available, image registration methods can be used to calculate the transformation matrix automatically, and such techniques for performing rigid transformations are described in the literature [1]. However, when acquiring pseudo-dynamic MRI data for kinematic studies, the fast acquisition time, typically 1-2 seconds per slice, results in low resolution widely spaced image slices which are impossible to register to each other due to the sparse nature of the acquired data. The aim of this work was to develop a registration technique that can produce quantitative kinematic data from the low spatial resolution scans available. The technique is based on registering a full high spatial resolution segmented MRI volume to the sparse slices of the dynamic data. The acquisition of both static and dynamic data sets could be carried out in approximately 30 minutes. It is shown that the registration of the static volume to sparse dynamic data is possible, and that reliable kinematic information can be extracted. This method is expected to be more reliable that other techniques such as surface marker tracking as it considers the objects of interest explicitly and does not require any inference of bone locations from surface data.

Introduction

The development of subject specific kinematic analyses is one that is becoming increasingly important as biomedical engineering seeks to move from generalised joint modelling, and clinicians endeavour to utilise bespoke treatment plans and prostheses. This research is part of a pan European project called SIMBIO (www.simbio.de), which is concerned with the development of a general biomedical simulation environment in which rapid construction and solution of patient specific finite element models can be carried out. As part of the model validation it is necessary to compare the actual kinematics of the test subjects to those produced by the finite element simulations. Methods for subject-specific kinematics have been published in the literature, but generally take the form of motion capture methods. These techniques involve placing reflective surface markers onto the skin surface and tracking them in 3D using multiple camera acquisition systems. The problem with using such systems is that they do not measure the variables of interest (the bone positions) directly, because the skin surface moves, sliding to a certain extent over the underlying structures. For large-scale investigations such as gait analysis the errors produced are acceptable, but when quantitative kinematics at millimetre resolution is required, the small surface marker errors can become significant. A solution to this problem would be to use dynamic anatomical scans in which the bones can be identified and tracked explicitly throughout the motion of interest. The main problem with such imaging techniques is the time required to collect good quality volumetric data. The imaging modality of choice for this type of investigation is magnetic resonance imaging (MRI). This modality has the advantage of being safe, and can image soft tissues such as cartilage, ligaments and blood vessels. Unfortunately, it takes approximately 10 minutes to collect a good quality MRI volume using the standard T1 or T2* acquisition protocols. This limitation makes it impractical to investigate dynamic motion using high resolution MRI data. As a compromise it is possible to collect a small subset of the full volume of interest quickly, in the order of a few seconds, which can provide qualitative information on the motion in what is termed a pseudo-dynamic movement. Similar investigations into kinematic quantification have involved the acquisition of velocity images which when integrated can produce measures of joint displacement [2]. Here we describe the use of image registration to extract quantitative data on the kinematics of the knee joint from these pseudo-dynamic MRI scans which tracks the joint displacements explicitly.

Method

Data acquisition. All imaging was performed on a whole-body system based around a cylindrical superconducting magnet operating at 1.5T (Eclipse, Philips Medical Systems). Two protocols were used for data acquisition, firstly the high quality static volume which was collected while the subject's knee remained stationary, followed by the dynamic sequence which was acquired while the leg was under load in a custom-designed rig. The static volumes were collected using a T2*-weighted gradient-echo sequence (TR=47ms, TE=15ms, Flip angle 30°) with image dimensions of 512 x 512 x 60, producing an in-plane resolution of 0.35mm and effective slice thickness of 2mm. Collection of the image volume takes approximately 20 minutes, due to the high resolution that is required for other parts of the project. It is expected that 256 x 256 x 60 voxels would be more than adequate for the purposes of a kinematic analysis, which would reduce acquisition time by half.

The dynamic sequences are acquired under load in a custom-designed rig, intended to mimic the action of pushing a clutch pedal. The rig has optical instrumentation (Measurand Inc., Canada) to monitor the primary flexion of the knee joint in order to produce evenly-spaced flexion angles. The sequence used is a T2* weighted (TR=29ms, TE=13.4ms, Flip angle=20°) high speed, gradient echo sequence, with an in-plane resolution of 1.64mm, slice thickness of 3mm and has 6 slices across the knee joint which are positioned manually to span the joint. (The lower in-plane resolution of the dynamic slices is due to the much larger field of view necessary to capture the joint flexion). The acquisition time for the dynamic data is approximately 1s per slice, with 6 slices per flexion angle and 6 flexion angles throughout the full range of joint motion 0-40 degrees. Figure 1 shows the knee rig, and associated instrumentation used during dynamic data capture.

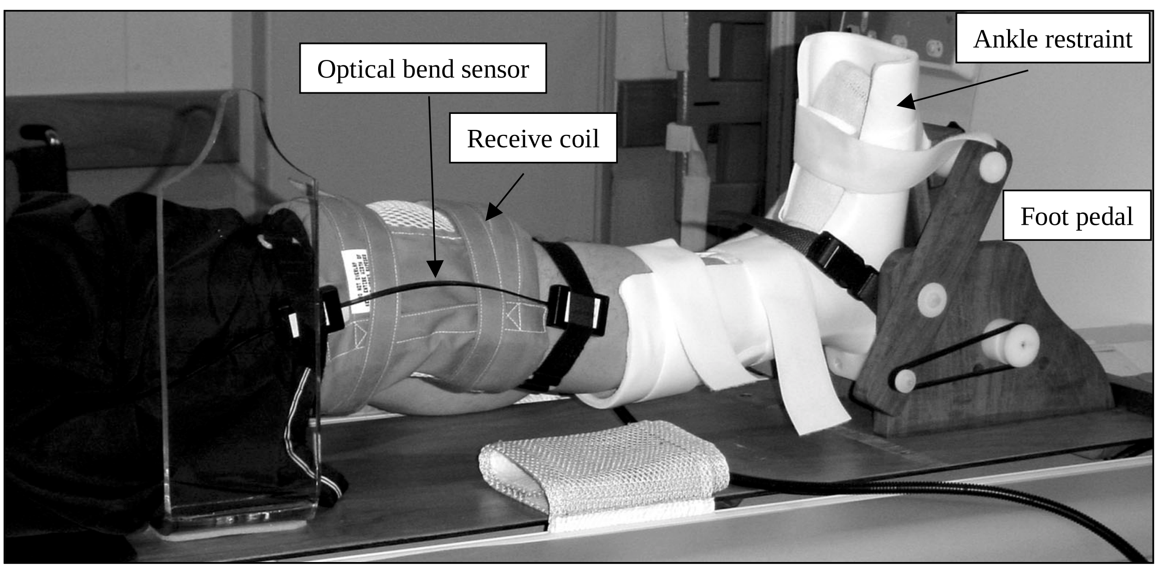

Figure 1: Photograph of the knee rig, and instrumentation used in the MRI scanner

Registration process. The registration process is a rigid affine transformation algorithm. The algorithm is a modification of the motion correction algorithm developed by [1] which adjusts the affine matrix at each iteration to remove all scaling effects via an SVD. The requirement to compute the (rigid) motion of the individual bones in the knee means that the bones must be segmented from the image data. At present, the segmentation is performed by hand, but tools are under development as part of the SIMBIO project that should automate this process. Each complete volume bone segment in the static image is then registered to the sparsely sampled bone segment from the dynamic data. We have developed the following registration equation.

$$f - m = \frac{1}{2}\left[\Delta u(r)\frac{\partial f}{\partial u(r)} - \Delta u(r)^{-1}\frac{\partial m}{\partial u(r)}\right]$$

Where f is the fixed or target image, m is the moved or starting image, $\Delta u(r)$ is the mapping function which maps m to f and $\Delta u^{-1}(r)$ the inverse function which maps f to m. Making the assumption that $\Delta u(r) \approx -\Delta u^{-1}(r)$ and (gradient of f ~ gradient of m) it is possible to reduce this equation to either of the following:

$$f - m = \Delta u(r)\frac{\partial f}{\partial u(r)} \quad \text{or} \quad f - m = -\Delta u(r)\frac{\partial m}{\partial u(r)}$$

In the present work the mapping function is parameterised (following the approach of Lee) by three rotation and three translation parameters. Lee's method drives the registration parameters directly from the registration equation.

If we choose the second of these equations it is only necessary to calculate the gradients of the moved image, which is a full volumetric data set, and therefore gradients in all three directions can be evaluated. If this simplification were not made, it would not be possible to register to the dynamic data set, as the single slice data does not have a gradient out of plane. Figure 2 shows an example of segmented measured data; the transparent femur volume is derived from segmentation of the high resolution static scan. The solid slice data is from the segmented dynamic scans, and the solution after the static volume is registered to the dynamic slices is shown.

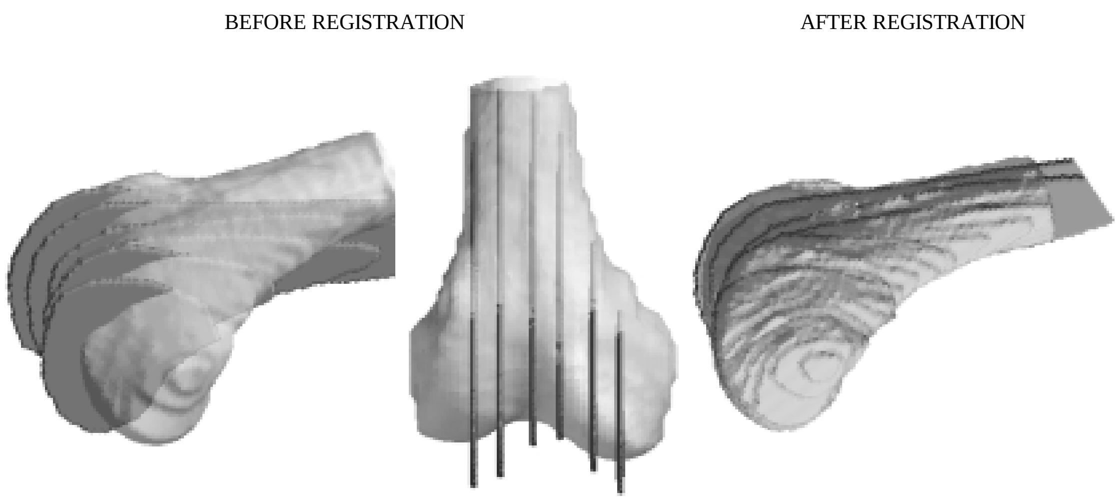

Figure 2: An example of the volume data, and sparse dynamic slices pre and post registration.

Results

Quantifying the errors in a study such as this is complicated by the fact that the only alternative method to non-invasively measure the *in vivo* rotations and translations of the bones is very complex, and the reported results are subject to a large standard deviation [2]. As a compromise, a simulation study has been carried out to estimate the magnitude of errors which might be expected from using this registration method. The process used was to take the femur volume from a static scan, shown in figure 2, apply a known rotation to it and then extract slices from the volume at the same locations as those which would have been acquired during a dynamic sequence. The original static volume data was then registered to the synthetic dynamic data and the rotations generated compared to the true rotations. As the primary variables of interest are the rotations of the femur relative to the tibia it was decided that these angles should be the measure of registration quality. An algorithm was written in Matlab to solve the non-linear problem of resolving the affine matrix components into the actual angles of rotation about the global co-ordinate system. This was an implementation of Broyden's method to find the roots of simultaneous non-linear equations. Figure 3 shows the results of these analyses. As can be seen the results are very good for rotations about the X and Z-axes, with less than 1 degree error at all points. The errors on the Y-axis are larger, and this is due to the fact that the shaft of the femur has some rotational symmetry about this axis and so does not ‘drive’ the registration as it does on the other two axes. However, the largest errors are found for small rotations around zero, and it is simple to add a rotation artificially to the starting position for this axis in order to ensure the result lies in the regions away from zero degrees.

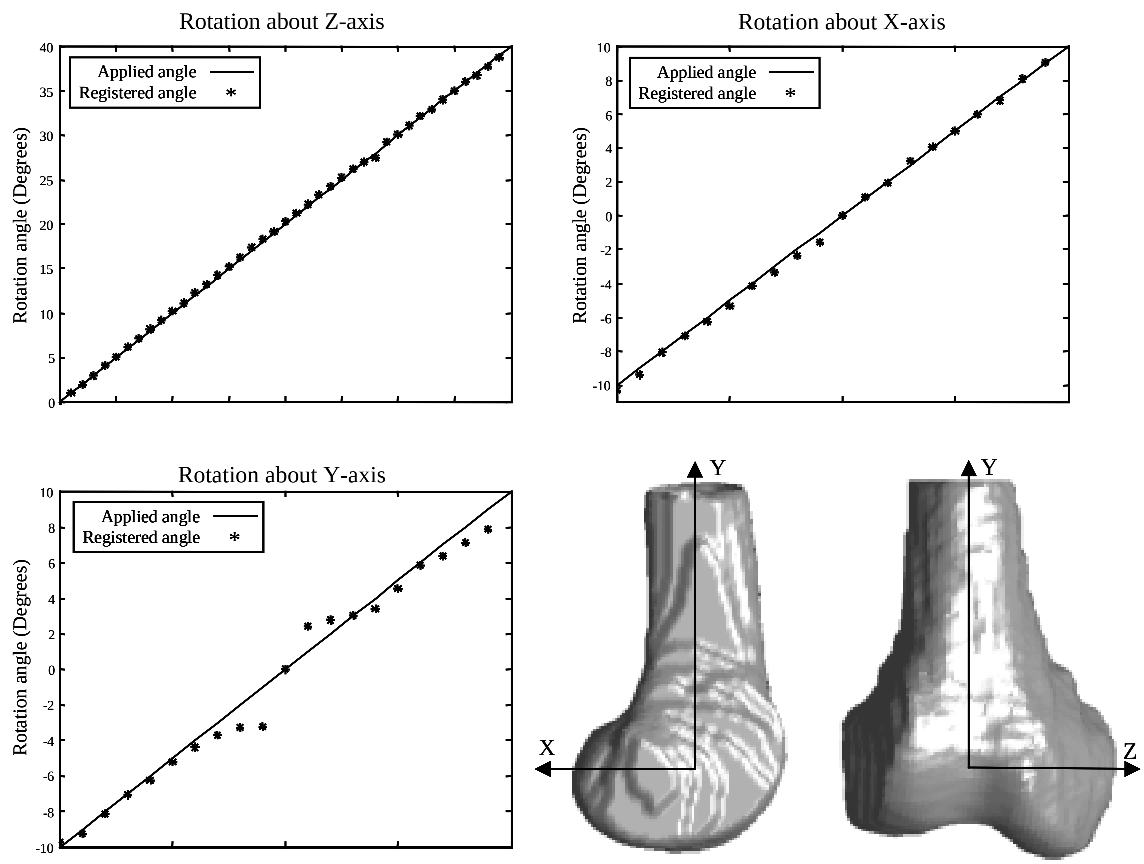

Figure 3: Results of registration process, against known angles of rotation about the 3 axes of rotation.

Conclusions

Presented here is a method that is capable of tracking the 3D motion of bones from pseudo-dynamic MRI data. It deals implicitly with out of plane motion of the joint, and can resolve all three rotational and translational aspects of the kinematic motion. Real data has been acquired and registered successfully, and a simulation study has shown that the method can produce accuracies of greater than one degree for rotations of the femur. However, the registration is only as good as the quality of the segmentation produced, and this is the subject for further research in the form of a phantom study. This method appears to be a robust way of accessing 3D kinematics in the knee, and this technique could be extended easily to other joints within the body.

References

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2. F.T. Sheehan & J.E. Drace "Quantitative MR measures of three-dimensional patellar kinematics as a research and diagnostic tool", *Medicine and Science in Sports and Exercise* 31(10): p. 1399-1405,1999.