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Correction of step artefact associated with MRI scanning of long bones

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ABSTRACT

3D models of long bones are being utilised for a number of fields including orthopaedic implant design. Accurate reconstruction of 3D models is of utmost importance to design accurate implants to allow achieving a good alignment between two bone fragments. Thus for this purpose, CT scanners are employed to acquire accurate bone data exposing an individual to a high amount of ionising radiation. Magnetic resonance imaging (MRI) has been shown to be a potential alternative to computed tomography (CT) for scanning of volunteers for 3D reconstruction of long bones, essentially avoiding the high radiation dose from CT. In MRI imaging of long bones, the artefacts due to random movements of the skeletal system create challenges for researchers as they generate inaccuracies in the 3D models generated by using data sets containing such artefacts.

One of the defects that have been observed during an initial study is the lateral shift artefact occurring in the reconstructed 3D models. This artefact is believed to result from volunteers moving the leg during two successive scanning stages (the lower limb has to be scanned in at least five stages due to the limited scanning length of the scanner). As this artefact creates inaccuracies in the implants designed using these models, it needs to be corrected before the application of 3D models to implant design. Therefore, this study aimed to correct the lateral shift artefact using 3D modelling techniques.

The femora of five ovine hind limbs were scanned with a 3T MRI scanner using a 3D vibe based protocol. The scanning was conducted in two halves, while maintaining a good overlap between them. A lateral shift was generated by moving the limb several millimetres between two scanning stages. The 3D models were reconstructed using a multi threshold segmentation method. The correction of the artefact was achieved by aligning the two halves using the robust iterative closest point (ICP) algorithm, with the help of the overlapping region between the two. The models with the corrected artefact were compared with the reference model generated by CT scanning of the same sample.

The results indicate that the correction of the artefact was achieved with an average deviation of 0.32 ± 0.02 mm between the corrected model and the reference model. In comparison, the model obtained from a single MRI scan generated an average error of 0.25 ± 0.02 mm when compared with the reference model. An average deviation of 0.34 ± 0.04 mm was seen when the models generated after the table was moved were compared to the reference models; thus, the movement of the table is also a contributing factor to the motion artefacts.

1. Introduction

Magnetic resonance imaging (MRI) is theoretically designed to scan the soft tissues utilising the hydrogen nuclei as the source of signal. In recent studies, it has been shown to be a possible alternative to computed tomography (CT) for scanning of long bones [1,2]. This alternative provides researchers designing orthopaedic implants with an opportunity to acquire long bone image data from the young healthy human population, who represent nearly half of all trauma patients, without having to expose them to the ionising radiation of CT [3]. However, MRI still suffers from some limitations such as very long scanning times, motion artefact and poor contrast in certain anatomical regions. Of these limitations, the motion artefact is crucial as it reduces the accuracy of the 3D models reconstructed from such image data [2]. A lateral shift has been observed in the 3D models reconstructed from data sets; this is believed to occur as a result of random patient movements [4].

The design of an orthopaedic implant (e.g. joint replacement or fracture fixation plate) needs accurate 3D representations of the

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relevant bone geometry. This is important in order to have a correct anatomical alignment between two bone fragments or for a properly functioning joint. The current gold standard for acquisition of data for this purpose, CT, exposes a person to a high dose of ionising radiation. For an example: in a typical abdominal CT a patient exposes to about 10 mSv of radiation dose which is about 50 times that of a plane abdominal X-ray [16]. This radiation exposure limits CT to the scanning of cadaver bones which are, in most cases, more than 60 years old. Since most of the patients who have been implanted with a plate or intramedullary nail are from the younger population, the implants need to be designed to suit this age group. For this purpose, there is an urgent need for the acquisition of data from this younger population. MRI is a versatile alternative for this purpose as there are no radiation hazards involved in MRI scanning. The poor contrast of certain anatomical regions in the MRI scanning of long bones can be overcome to some extent by using a higher field MRI scanner, which benefit from higher signal to noise ratio [2,17]. However, artefacts due to random movements of a subject remain a problem which needs to be addressed in order to utilise the models for the intended application.

The motion artefacts occur when the protons of the tissue sample being scanned excited at one site are misregistered to another region of the image during the data acquisition [5]. This results when the position of the protons shifts relative to an applied gradient, causing a change in the gradient dependent frequency of the protons. This results in superimposed images of the moving structures along the phase direction, where the number and separation of the artefact images depends on the frequency of the motion [6]. The motion artefacts are of two types: the artefacts due to, periodic movements and the artefacts due to random movements [6]. The motion artefacts due to regular, periodic movements occurs (mainly) as a result of respiratory movements, flow of blood in vessels or peristalsis. These artefacts mainly affect the MRI scanning of the relevant anatomical regions. Although respiratory movements might have an impact on the scanning of long bones of the upper limb, they have minimal or no effect on the scanning of the long bones of the lower limbs (e.g. the femur and tibia). The artefacts resulting from random movements may be due to nerve excitation during the scanning, or the patient randomly moving the limb. Due to the limited linearity of the gradients and B_0 field of the MRI scanner, the lower limb of a subject has to be scanned in several stages (usually four to five) [18]. A preliminary investigation conducted by Schmutz et al. [4] using a MRI scanner has shown that the movement of the subject between these scanning stages produces a lateral displacement/shift in the final 3D model.

The artefacts that result from periodic movements can be minimised by using various scanning protocols that synchronise the movement with the data acquisition or by using post processing/filtering techniques [7–10]. The artefacts due to random movements, on the other hand, cannot be eliminated easily from acquired data in the absence of navigator echoes. However, navigator echoes have to be used at the expense of scanning time and the efficiency of the scanning. This can be achieved to some extent by immobilising the subject; however, immobilising a limb for ~60 min is not easily achievable. With regards to 3D model reconstruction, it can be achieved by 3D modelling techniques in which 3D models from the consecutive scanning stages can be aligned by using an iterative closest point (ICP) algorithm based technique [11].

The ICP algorithm is a widely used 3D-3D registration technique and has shown a high accuracy for translational as well as rotational alignments of 3D models. Lee et al. [1] conducted a preliminary registration test using the ICP algorithm in which a part of the bone model separated from the original model was matched perfectly to its original full model. In another study, the ICP algorithm was able to register a CT derived model to a real patient's model with an average error of $0.079 \pm 0.068°$ for rotation and 0.12 ± 0.09 mm on translations [12]. The ICP algorithm guarantees convergence to a local minimum from any given transformation of the data point set [11]. However, the obtained local minimum may not be the desired global minimum, as it depends on the initial registration. While the ICP algorithm has been used in numerous studies for aligning bone models, the effect of its initial position on the optimal global alignment has not yet been reported.

This study aimed to correct the lateral shift artefact that is associated with MRI scanning of long bones using the ICP algorithm for aligning the models. The dependency of the ICP algorithm on the initial position of the 3D surfaces to register them was also investigated.

2. Methods

MRI scans of five ovine femora (average length of a femur = 17.6 cm, average diameter of a femur at mid shaft region = 1.8 cm, average age = 7 years and average weight = 49 kg) obtained by scanning five intact sheep hind limbs were used for the study. A 3T MRI scanner with the following imaging protocol and the body matrix coil was used.

A Siemens (Erlangen, Germany) TIM TRIO 3T clinical MRI system was used to image the femora, which was positioned so the long axis of the bone was parallel with the bore of the magnet. The 3D vibe sequence was used with the following parameters: TR = 11 ms, TE = 1.83 ms, slice direction = transverse, field of view 240 (LR) × 120 (AP, phase direction), in plane image resolution 0.47 mm × 0.47 mm, slice thickness 1 mm, number of slices = 256, transverse images field of view = 240 (LR), number of averages = 1, and acquisition time = 9 min 37 s.

The femora was scanned using exactly the same parameters as described above with the only changes being the following combinations of slice stack position and bed position as described below. With the exception of the first step, the sample was scanned in two halves (proximal and distal) in all steps, while maintaining an approximately 7 cm overlap between the proximal and distal halves (Fig. 1).

1. The centre of the femur was positioned in the *z*-centre of the magnet and a complete scan was obtained using a single FOV (Fig. 1a).
2. With the centre of the femur still positioned at the *z*-centre of the magnet, the femora were then scanned in two halves using two *z*-offset FOVs without moving the table (Fig. 1b).
3. The femora were now scanned in two halves with the centre of each image stack aligned with the *z*-centre of the magnet, rather than the centre of the femora.
4. The scanning of the femora in two halves with the centre of each image stack aligned with the *z*-centre of the magnet was repeated, except after the distal half was scanned; the sample was shifted laterally to simulate the lateral shift caused by a volunteer moving their leg. Then, the proximal half was scanned (Fig. 1c).

3D models of bones were reconstructed from all MRI data sets using the multi-threshold segmentation method previously developed by the author [13]. This method combines a multilevel threshold approach with a method of selecting an appropriate threshold level for a particular anatomical region of a long bone. Two threshold levels calculated using the threshold selection method were used for the two anatomical regions: the distal/proximal region and the diaphyseal region. Most of the articular regions were segmented manually due to the presence of a number