ARTICLE IN PRESS

Journal of Biomechanics ∎ (∎∎∎∎) ∎∎∎–∎∎∎

Contents lists available at ScienceDirect

Journal of Biomechanics

journal homepage: www.elsevier.com/locate/jbiomech
www.JBiomech.com

A non-ionizing technique for three-dimensional measurement of the lumbar spine

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ARTICLE INFO

Article history:
Accepted 27 October 2016

Keywords:
Ultrasound
Magnetic resonance imaging
Registration
Lumbar spine
Segmental kinematics

ABSTRACT

Comprehensive assessments of scoliotic deformity and spinal instability require repetitive three-dimensional (3D) measurements of motion segments at different functional postures. However, accurate 3D measurement of the spine is a challenging task. In this paper, we present a novel, non-invasive, non-ionizing technique to quantify 3D poses of lumbar motion segments in terms of clinically meaningful anatomical coordinates. The technique used ultra-short echo time (UTE) magnetic resonance (MR) images to construct subject-specific geometrical models of individual vertebrae and registered them with 3D ultrasound dataset acquired during pose measurements. A hierarchical registration approach was used to minimize the detrimental effects of speckle noise and artifacts within soft tissues on registration accuracy. The technique was validated using a human dry bone specimen as well as a fresh porcine cadaver. Registration errors were determined by comparing with a gold standard fiducial-based registration. Results showed that the technique is accurate and reliable with bias in sub-degree and sub-millimeter level (except for the flexion–extension of the porcine cadaver experiment, which was $-1.74°$), and average precision of 1.11° in rotation and 0.86 mm in position for the human dry bone experiment, and 1.26° and 1.23 mm for the porcine cadaver experiment. Given its non-ionizing nature, the UTE MR–ultrasound registration technique is particularly useful for repeated measurements and longitudinal follow-up. With further refinement and validation, it could be a powerful tool for 3D spinal assessment.

1. Introduction

Two-dimensional radiograph has been widely used for functional, developmental, diagnostic, and treatment-effect evaluations of the spine. For instance, current clinical practice relies on measuring the Cobbs angle from coronal plane radiograph to quantify scoliotic deformity (Cobb, 1948). Although this method provides valuable information about the sideways deformity, it cannot detect any out-of-plane deformities and therefore does not provide a complete picture of the actual deformity. Likewise, even though spinal instability (a presumed mechanical cause of back pain and an indication for spinal fusion) has been typically quantified by sagittal plane radiographs captured at the end-range of motion of a flexion–extension maneuver (Fujiwara et al., 2000), it has been suggested that axial rotations (Haughton et al., 2002) and coupled motions (Ochia et al., 2006) are more relevant measures of spinal instability. These clinical examples highlight the pressing need of developing three-dimensional (3D) measurement tools for the spine.

Over the past few decades, several 3D spinal measurement methods including skin marker tracking (Gracovetsky et al., 1995), implanted marker tracking (Dickey et al., 2002), bi-plane radiography (Pearcy, 1985), dual fluoroscopy (Li et al., 2009), computed tomography (CT) (Ochia et al., 2007), magnetic resonance imaging (MRI) (Ishii et al., 2004), etc. have been proposed. However, these methods are either prone to substantial errors, highly invasive, subject to repetitive ionizing radiation, or require the subjects to be tested at non-functional and unloaded states. We recently developed and validated an intensity-based hierarchical CT to ultrasound registration technique to non-invasively quantify the 3D poses of lumbar vertebrae and motion segments from 3D freehand ultrasound and one-time computed tomography under functionally loaded conditions (Koo and Kwok, 2016). The basic premise of this technique is to apply geometric transformations to CT-reconstructed vertebral models so that they align with the 3D freehand ultrasound dataset of the lumbar spine acquired during pose measurements. Although the use of 3D freehand ultrasound

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http://dx.doi.org/10.1016/j.jbiomech.2016.10.048
0021-9290/

Please cite this article as: Koo, T.K., Kwok, W.E., A non-ionizing technique for three-dimensional measurement of the lumbar spine. Journal of Biomechanics (2016), http://dx.doi.org/10.1016/j.jbiomech.2016.10.048

completely eliminates ionizing radiation during repeated pose measurements (in contrast to the bi-plane X-ray method (Pearcy, 1985) with reported radiation dose of 24 mSV for a single lumbar pose measurement), like other model-based tracking techniques (e.g. Anderst et al., 2011; Haque et al., 2013), it still relies on an one-time CT to construct geometrical models of individual vertebrae. Given that the estimated effective dose of a lumbar spine CT scan is about 5.6 mSv, with an estimated cancer risk of 1 in 3200 (Richards et al., 2010), it is highly desirable to replace the one-time CT by MRI to make the hierarchical registration approach a truly non-ionizing 3D measurement of the spine.

Major challenges of using MRI for bone imaging are that cortical bone usually has no observable signal in MRI due to its short sub-millisecond T2 relaxation time and, tendons and ligaments also appear to be dark in MRI due to their short T2 of under 10 ms (Robson and Bydder, 2006; Chang et al., 2015). As a result, cortical bone may not be separable from its neighboring connective tissues, leading to difficulties in image segmentation of the vertebrae. In recent years, ultra-short echo-time (UTE) sequences with echo time down to tens of micro-seconds have been developed (Chang et al., 2015). They enable MRI signal with very short T2 relaxation to be detected before it decays to undetectable levels as in conventional MR sequences, and hence, they may be used to distinguish cortical bone of the vertebrae from surrounding connective tissues. We recently conducted a pilot study that utilized the UTE technique for spine MR imaging. Our in vivo and in vitro pilot data suggested that UTE MRI is able to improve the visualization and segmentation of the vertebrae (Kwok and Koo, 2015). Based on these promising results, the objective of this study was to explore the feasibility of using UTE MRI within the context of the intensity-based hierarchical registration algorithm by evaluating its bias and precision, and comparing with the CT–ultrasound registration.

2. Materials and methods

2.1. Freehand 3D ultrasound

A freehand 3D ultrasound system (Koo et al., 2014) was used to acquire 3D ultrasound dataset of the lumbar spine. The system consists of an ultrasound scanner (Ultramark 400c; ATL Ultrasound Inc., Bothell, WA) with a 3.5–5 MHz curvilinear transducer to image the lumbar spine, an optoelectronic measurement system (Northern Digital Inc., Waterloo, Canada) to track the transducer, and a personal computer with a frame grabber and a data acquisition card installed to capture ultrasound images and synchronize with the tracking data. The ultrasound transducer was calibrated using a point-based method (Muratore and Galloway, 2001). After calibration, we further located 30 points within the field of view of the calibrated transducer and digitized them using a digitizing probe (Northern Digital Inc., Waterloo, Canada). The accuracy (i.e. average distance between the two corresponding point sets) was determined to be 1.48 mm. An image coordinate system was defined so that its x and y axes aligned with the horizontal and vertical directions of the acquired ultrasound images respectively and its origin was at the center of the most convex point on the transducer's scanning surface. During the ultrasound scan, local coordinates of each pixel from each ultrasound image was transformed to a common laboratory coordinate system using the calibration matrix and the tracking information, forming a 3D ultrasound dataset.

2.2. UTE MR imaging

UTE MR images were acquired by a Siemens TRIO 3 Tesla system (Siemens Healthcare, Erlangen, Germany) with a spine matrix coil. A Siemens work-in-progress 3D UTE sequence with isotropic radial sampling was used to image the lumbar spine using the following parameters: repetition time = 5.8 ms, echo time = 0.07 ms, flip angle = 6°, field-of-view = 32 cm, number of radial views = 100,000, number of data points per radial view = 320, receive bandwidth = 781 Hz/pixel, and imaging time = 9 min 40 s. The data was gridded onto a Cartesian matrix with size $320 \times 320 \times 320$ using Kaiser–Bessel algorithm (window width = 3 and $\beta = 4.2054$). Due to radial sampling, the data were oversampled in the inner part of k-space and undersampled in the outer part of k-space. Sampling density pre-compensation was performed using a rho filter that plateau toward the outer k-space to correct for undersampling. The reconstructed image resolution was $1.0 \times 1.0 \times 1.0\ \text{mm}^3$. The imaging parameters were selected based on the results of a pilot study that aimed to optimize the scanning parameters for improved visualization and segmentation of in vivo human vertebrae yet within an acceptable imaging time (Kwok and Koo, 2015).

3D Doctor software (Able software Corp., Lexington, MA) was used to manually segment individual vertebrae from the UTE MR images and construct the vertebral models. Posterior surfaces that supposed to be visualized by the B-mode ultrasound were then extracted from each vertebral model using a modified forward ray tracking method (Koo and Kwok, 2016).

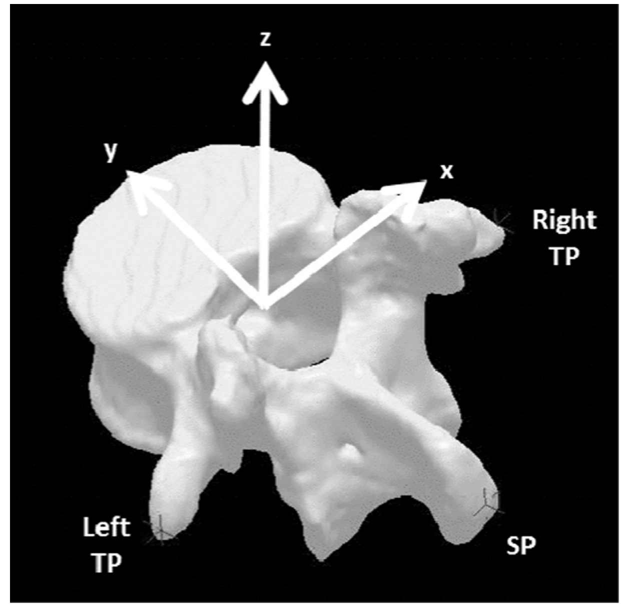

Fig. 1. An anatomical coordinate system of a vertebra**.** The x, y, and z axes are defined as the calculated principal axes that point to the right, anterior, and superior directions respectively.

2.3. Local coordinate system on vertebra

In order to quantify vertebral and intervertebral kinematics in a way that makes sense to clinicians, an anatomical coordinate system was defined on each vertebra using the information from all voxels of the vertebral model. Origin was defined as the centroid of the vertebral model. A set of three orthogonal axes representing the major axes of mass distribution within each vertebra were calculated using the principle axis transformation technique (Tsao et al., 1998). X, Y, and Z axes of each vertebra were defined as the calculated axes that pointed to the right, anterior, and superior directions respectively (Fig. 1).

2.4. Hierarchical UTE MR to ultrasound registration

The overview of the hierarchical UTE MR–ultrasound registration technique is illustrated in Fig. 2. For each vertebra, the process began with a landmark-based registration that searched for a reasonable initial guess of 3 rotational $[\alpha_1, \beta_1, \gamma_1]$ and 3 positional $[x_1, y_1, z_1]$ parameters that roughly transformed each UTE MR-reconstructed vertebra to the laboratory space of the actual assessment. This was accomplished by registering 3 corresponding bony landmarks between the ultrasound and UTE MRI dataset. We then used an intensity-based registration approach to guide the search of a set of rotational and positional parameters $[\alpha_2, \beta_2, \gamma_2, x_2, y_2, z_2]$ closest to the ground truth. This was achieved by applying a backward ray tracing technique (Yan et al., 2011) to extract the ultrasound pixels in the vicinity of posterior vertebral surface and registered them with the posterior vertebral surface extracted from the UTE MR-reconstructed vertebral model. This surface extraction process was designed to remove the speckle noises and other artifacts within soft tissues (Fig. 3), facilitating the registration to converge to a realistic solution space. Lastly, we further refined the transformation parameters by constraining the search space within $\pm 8°/8$ mm of $[\alpha_2, \beta_2, \gamma_2, x_2, y_2, z_2]$ and re-registered the extracted posterior surfaces of the UTE MRI-based model with the original ultrasound dataset to get the final solution $[\alpha_f, \beta_f, \gamma_f, x_f, y_f, z_f]$.

2.5. Validation

2.5.1. Pose measurements

We opted to use the same ultrasound datasets that were acquired from a human dry bone specimen and a porcine cadaveric specimen to facilitate a direct