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A fast, accurate, and reliable reconstruction method of the lumbar spine vertebrae using positional MRI

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Abstract

In vivo measurement of lumbar spine configuration is useful for constructing quantitative biomechanical models. Positional magnetic resonance imaging (MRI) accommodates a larger range of movement in most joints than conventional MRI and does not require a supine position. However, this is achieved at the expense of image resolution and contrast. As a result, quantitative research using positional MRI have required long reconstruction times and are sensitive to incorrectly identifying the vertebral boundary due to low contrast between bone and surrounding tissue in the images. We present a semi-automated method used to obtain digitized reconstructions of lumbar vertebrae in any posture of interest. This method combines a high-resolution reference scan with a low-resolution postural scan to provide a detailed and accurate representation of the vertebrae in the posture of interest. Compared to a criterion standard, translational reconstruction error ranged from 0.7 mm to 1.6 mm and rotational reconstruction error ranged from 0.3 degrees to 2.6 degrees. Intraclass correlation coefficients indicated high interrater reliability for measurements within the imaging plane (ICC = 0.97–0.99). Computational efficiency indicates that this method may be used to compile data sets large enough to account for population variance, and potentially expand the use of positional MRI as a quantitative biomechanics research tool.

Keywords

kinematics; spine modeling; *in vivo*; iterative closest point

1. Introduction

Conventional magnetic resonance imaging (MRI) provides high-resolution images of soft-tissue and bone geometries, and is collected in a noninvasive manner; however, several positioning restrictions may limit the applicability of conventional MRI in orthopedic modeling. Conventional MRI requires the person to lay supine inside a horizontal bore. The supine position reorients gravity and alters the configuration of multi-joint structures such as the spine compared to a natural orientation. For example, the supine position reduces the convex curvature of the lumbar spine, known as lordosis, compared to a neutral standing position²³. To approximate the normal effects of gravitational loading on the spine, some investigators have applied axial compressive loads to the participant while lying

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supine^{11,12,22}. In addition, the horizontal bore limits the range of motion of most joints. In the lumbar spine, investigations of vertebral joint mechanics are limited to axial rotations because functional sagittal plane flexion-extension or lateral bending positions are not feasible^{7,9}.

Positional MRI, a subcategory of Open MRI, is an adaptation of conventional MRI technology that allows image acquisition during standing and seated postures within a capture volume that is less restrictive than conventional MRI. Because a person can stand or sit in an open capture volume, pathologies can be viewed by physicians and researchers under natural loading (both gravitational and muscle loading), and joint biomechanics can be assessed across the functional range of motion. Soft tissue pathologies such as lumbar disc herniation are sensitive to loading conditions and torso position, which supports using positional MRI as a diagnostic tool²⁶. Positional MRI was recently used to characterize lumbar spine geometry during natural gravitational loading. Cargill *et al.*⁵ investigated changes in intervertebral position of the lumbar spine during seated flexion and extension and Meakin *et al.*¹⁵ quantified changes in lumbar curvature in response to a range of axial loads while standing.

Collecting images in natural weight-bearing postures with short acquisition times provides a unique opportunity to investigate *in vivo* intervertebral translation and rotation throughout a range of torso positions. These data may be used to enhance musculoskeletal models¹⁴, diagnose pathologies that are sensitive to loading and posture^{1,10}, and develop motion preserving orthopedic implants². When coupled with efficient processing, the short acquisition times will allow researchers to power imaging and computational investigations to account for intersubject variability.

Although positional MRI can accommodate larger joint range of motion than conventional MRI, this is achieved with reduced resolution. Positional MRI uses mid-field magnet strength (0.5-0.7 Tesla) and the large field of view typically increases the filling factor, which can result in lower image resolution compared to conventional MRI collected with high-field magnets. In addition, the probability of incurring motion artifact in an image is increased when collecting images in non-supine postures. To reduce the likelihood of participant motion and limit participant fatigue, short image acquisition times are often used with positional MRI. Protocols that use long acquisition times to obtain the highest possible image resolution from a mid-field magnet are impractical in some positions of interest such as partial lumbar flexion.

To take full advantage of positional MRI, an accurate, reliable, and efficient reconstruction method is needed to obtain intervertebral measurements from mid-field MR images. Past investigations have relied on manual segmentation and registration to reconstruct the vertebrae⁵. Automated segmentation methods that rely on image contrast between different tissue types are commonly used with computed tomography (CT) and high-field MRI scans. However, these are not applicable to positional MRI because of insufficient image contrast between bone and the surrounding soft tissues. Therefore, manual segmentation must be used for positional MR images. Manual segmentation is time consuming and is susceptible to boundary misidentification. Cargill *et al.*⁵ performed multiple manual registrations for each vertebra and averaged the results across raters to reduce these errors. Although effective, performing multiple segmentations and reconstructions drastically increases processing time and precludes using large data sets to power generalizable investigations.

In this investigation, we present a semi-automated reconstruction method (manual segmentation and automated registration) specifically designed for positional MR images of the lumbar spine. Section 2.1 presents the reconstruction method, which expands the

research capability of positional MRI in three ways. First, it provides quantifiable measurements of *in vivo* lumbar vertebral configuration with known accuracy. Second, it introduces automated registration to reduce data processing times. Third, it uses manual segmentation techniques that are insensitive to rater error. Section 2.2 presents an evaluation of the reconstruction method using a criterion measurement standard. An array of manual segmentation methods are tested for accuracy and reliability with respect to the position and orientation measurements using an ovine lumbar spine specimen. We also evaluate the processing efficiency (speed) of each of the manual segmentation methods. Section 2.3 applies the reconstruction method to *in vivo* human lumbar spine data collected throughout a range of flexion-extension postures and evaluates interrater reliability.

2. Methods

2.1. Vertebral Reconstruction Method

The vertebral reconstruction method uses an iterative closest point (ICP) algorithm to combine two different types of T1-weighted scans collected from a FONAR 0.6-Tesla Upright MRI using a planar coil. All scans were acquired using a Fast Spin Echo pulse sequence with 30x30x8 cm³ field of view and 480x480 resolution. The first type of MRI scan, the *reference scan*, was performed in a seated position. The reference scan protocol was designed to maximize image resolution, but requires a longer acquisition time than would be sustainable while performing standing postures. Posture (vertebral configuration) is irrelevant during the reference scan, so a position that is easy to maintain such as seated or supine should be selected. The second type of MRI scan, the *postural scan*, was collected in the posture of interest with a short acquisition time. The postural scan protocol makes use of short acquisition times to minimize motion artifact and participant fatigue. However, the postural scan protocol provides less detail than the reference scan due to a larger slice thickness, increased slice interval, and fewer total slices through the anatomy (Table 1).

A reconstruction of the postural configuration with high-resolution surface geometry was achieved by registering each segmented vertebra from the reference scan with the corresponding segmented vertebra from the postural scan. Bone surface geometries were segmented from all scans to create three-dimensional (3D) point clouds using ScanIP (Simpleware, Exeter, UK). Each vertebra from the reference scan was registered to the corresponding postural vertebra using an ICP algorithm (Figure 1) that aligns the reference scan vertebra with the equivalent postural vertebra such that the distance between vertebral surface points is minimized⁴. Prior to performing the ICP algorithm, a right-handed body-fixed coordinate system was assigned to each vertebra and sacrum from the reference scan and to the sacrum from the postural scan. These coordinate systems were assigned according to the International Society of Biomechanics (ISB) recommendation²⁴. This was accomplished using the 3D point clouds with SolidWorks (Dassault Systèmes, Vélizy, France) through a four-step process (Figure 2).

Step 1: Planar rectangles were defined tangent to the caudal and cranial endplates. The rectangular edges were positioned tangent to the anterior, posterior, and lateral edges of each endplate.

Step 2: The origin was defined at the midpoint of a line-segment connecting the center points of each rectangle. This line segment also defined the y-axis of the coordinate system.

Step 3: The x-axis was defined by a line that projects anteriorly from the coordinate system origin and intersects a line segment that joins the mid-points of the anterior edges of the caudal and cranial rectangles.

Step 4: The z -axis was defined as the cross product of the x -axis and y -axis.

The sacral coordinate system was defined using the same coordinate axis convention. However, only one rectangular plane located on the superior sacral endplate was defined. The coordinate system origin was positioned at the rectangle center point (Figure 1).

Each vertebra from the reference scan was registered to the postural scan using an ICP algorithm based on k -d tree point matching. The ICP algorithm identified the closest matching point pairs between the reference scan and postural point clouds and determines the rigid transformation that minimizes the root mean squared distance between all matched points. The algorithm was limited to 150 iterations, and a 5% worst-point match rejection was included to improve convergence time and minimize influence from spurious data that could be introduced through MR imaging noise or segmentation error¹⁹.

Two preprocessing alignment steps were used to increase the likelihood of locating a global minimum with the ICP algorithm.

Step 1: The reference scan vertebrae and sacrum were collectively translated and rotated so that the coordinate system from the reference scan sacrum was directly aligned with the coordinate system from the postural scan sacrum. This realignment of the reference scan establishes a common coordinate system for both scans and corrects for any rotational misalignments established during scan setup.

Step 2: Each reference scan vertebral point cloud was translated so that the coordinate system origin was positioned at the spatial mean of the corresponding postural scan point cloud. ICP registration was performed after the two alignment steps were completed.

Alignment steps and ICP registration were implemented using Matlab (MathWorks, Natick, Massachusetts). The ICP algorithm converged after approximately 40 iterations (3-4 seconds) for all vertebrae and segmentation methods.

Following reconstruction, position and orientation of each vertebra were measured in the sacral coordinate system. Positions were represented as rectangular coordinates (x , y , z) and rotational orientations were represented by Cardan angles (α , β , γ). The first Cardan angle (α) is a rotation about the original z -axis (i.e. a rotation in the sagittal plane). The second (β) and third (γ) Cardan angles are rotations about the successively rotated x - and y -axes. Cardan angles were chosen to describe the vertebral rotations because the primary rotations occur about axes that closely imitate the anatomic joint movements²⁰.

2.2. Selection of Segmentation Method using In Vitro Specimen

Manual segmentation introduces human error into the reconstructions and is the most time-consuming step in the reconstruction method. Therefore, we measured accuracy, reliability, and processing efficiency of the vertebral reconstruction method across eight manual segmentation methods using a cadaveric ovine spine specimen. The specimen included four vertebrae (T12, T13, L1, and L2) with intact intervertebral discs and partially intact musculature. To establish a criterion standard for measurement, the specimen was scanned using CT with 0.625mm thick axial slices (0.625mm gap, 1.25mm interval) from a GE HiSpeed QX/i scanner. The specimen was also scanned with a FONAR 0.6-Tesla Upright MRI using the reference and postural protocols (Table 1) and reconstructed using the procedure described in the previous section. The ovine specimen was vacuum sealed in heavy-gauge plastic and kept near freezing throughout the imaging procedures to maintain fixed intervertebral positions during all scans for comparison.

The eight manual segmentation methods differed according to the continuity of the boundary assigned to segmented structures (vertebral body, spinous process, or transverse process) from each MR slice (Figure 3). The continuity conditions ranged from selecting the “corner points” of these structures plus one additional point on the boundary between corner points to selecting a continuous boundary. For each of the seven point selection techniques, points were placed along the boundary with consistent spacing between points. The continuous boundary was created using a closed spline curve on each vertebral boundary from the MR images. Raters were instructed to use as many points as necessary to create the spline, which accommodates the irregular shapes of the vertebral structures. In ScanIP, the point-based segmentation methods produced point clouds consisting of only the points selected from each image slice. The continuous boundary condition in ScanIP generated an interpolated surface, which was exported as a point cloud.

Because the ovine specimen maintained a fixed vertebral configuration throughout all scans, a postural configuration was simulated by translating and rotating each vertebral point cloud from the reference scan prior to performing reconstruction. The simulated geometry (+80 mm along the *x*-axis, -50 mm along the *y*-axis, and 45 degrees about the *z*-axis) was equivalent to the typical configuration between L1 and the sacrum in a human performing extreme lumbar flexion. Reconstruction was performed on the T13, L1, and L2 vertebrae. Position and orientation of the T13 and L1 vertebrae were expressed in the L2 coordinate system because the ovine specimen did not include a sacrum. The postural scans were segmented using all eight of the segmentation methods performed by three different raters (CS, HX, NF), and provided 27 total sets of postural point clouds. The reference scans were segmented using only the continuous boundary method.

Accuracy, reliability, and efficiency were quantified for reconstructions performed using each of the eight manual segmentation methods. Accuracy was quantified by the measurement error, or differences in vertebral translation and orientation measured on the CT data (criterion standard) and the MRI reconstruction. Measurements for both vertebrae (T13 and L1) were pooled for analysis. Interrater reliability was quantified based on the standard deviation of the measurement error from all reconstructions. Processing efficiency was quantified by the time taken to complete the segmentation.

2.3. Reliability of Reconstructions from In Vivo Human Participants

Because a criterion standard was not available when scanning and measuring *in vivo* human participants, error cannot be assessed. However, sensitivity of the reconstruction measurements to manual segmentation performed by different raters can be evaluated. Positional MR images were collected from three participants (2 males, 1 female, 31.3±9.5 years old) with no history of low back pain or injury. Each participant provided informed consent in accordance with the University of Denver Institutional Review Board. Each participant performed a sitting reference scan and five standing postural scans: neutral standing, maximal lumbar flexion, partial lumbar flexion, maximal lumbar extension, and partial lumbar extension. All scans were collected with a FONAR 0.6-Tesla Upright scanner according to the imaging protocols in Table 1.

Each MR image was manually segmented by three raters (CS, HX, NF). All images (postural scans and reference scans) were segmented using the continuous boundary method. Each researcher segmented 75 vertebrae (3 participants, 5 positions), which provided 225 vertebrae for analysis.

Interrater reliability of reconstructed vertebral position and orientation was assessed by using the intraclass correlation coefficient (ICC) model (2,1)¹⁸. The dependent variables were rectangular positions (*x*, *y*, *z*) and Cardan angles (*α*, *β*, *γ*) of each vertebra with respect

to the sacral coordinate system. To calculate ICC model results, mixed-model analysis of variance (ANOVA) with fixed effect of rater and random effect of vertebra were calculated. Mean square error, between subjects mean square, and between raters mean square were taken from the ANOVA tables to calculate ICCs.

3. Results

Reconstruction error was calculated as the difference between MRI and CT measurements from reconstructions of the ovine spine in simulated flexion. Measurement errors are reported as the average of the T13 and L1 reconstruction errors produced by all three raters (Figure 4). Translational reconstruction error ranged from 0.68 mm to 1.6 mm and rotational reconstruction error ranged from 0.28 degrees to 2.6 degrees across the eight segmentation methods. The smallest error in the z -direction was 0.98 mm, achieved with the 4-intermediate point segmentation method. The lowest x -position error was 0.68 mm, achieved with 8-intermediate point segmentation. The lowest y -position error was 0.91 mm, achieved with continuous boundary segmentation. The lowest α -angle reconstruction error was 0.28 degrees, achieved with 3-intermediate point segmentation. The lowest β -angle error was 1.19 degrees, achieved with 4-intermediate point segmentation. The lowest γ -angle error was 0.14 degrees, achieved with 6-intermediate point segmentation.

The fastest segmentation time was recorded for the two intermediate-points technique, which required 5.8 ± 3.0 minutes and yielded point clouds comprised of 109-127 points (Figure 5). The longest average segmentation time was recorded for the 10 intermediate-points technique, requiring 15.0 ± 5.0 minutes on average and yielded point clouds comprised of 359-419 points. Continuous boundary segmentation required 12.1 ± 3.4 minutes and provided point clouds that ranged from 12000 to 13000 points. Segmentation using the 6-, 8-, and 10-intermediate point methods requires longer segmentation time than continuous boundary segmentation. ICCs indicated high interrater reliability for sagittal plane positions, x (ICC= 0.99) and y (0.99), but poor reliability for the z (0.26) position. ICCs indicated good interrater reliability for the sagittal plane rotation, α (0.97), but poor reliability for second and third rotations, β (0.18) and γ (0.26).

4. Discussion

The goal of this investigation was to evaluate a novel vertebral reconstruction method used to measure *in vivo* intervertebral configuration with positional MRI. This method of reconstruction addresses inherent limitations in image resolution that prevent automated segmentation methods. The vertebral reconstruction method combined two distinct positional MR scan types within a systematic and semi-automated reconstruction and demonstrated promising results. Reconstruction accuracy based on a criterion standard was within 1mm for sagittal plane positions (x , y) and 1.5mm for lateral (z) positions. Accuracy of sagittal plane rotation (α) was within 1 degree. Segmentation times were approximately 12 minutes per vertebra and registration time was less than 10 seconds per vertebra. Reliability of *in vivo* reconstructions was high for sagittal plane positions and rotations but low for out of plane (lateral) positions and rotations.

To our knowledge, these data represent the first comparison of lumbar spine reconstructions from positional MRI to a criterion standard, and may help expand the use of this tool in clinical and research investigations. Baumgartner *et al.*³ used a Fonar 0.6-Tesla Upright scanner to investigate lordosis angle and disc height using single plane images of the lumbar spine during seated postures. This approach may be susceptible to measurement error if the image plane is not aligned with the midplane of each vertebra. Cargill *et al.*⁵ performed three-dimensional measurements of the lumbar vertebrae and demonstrated consistency of

the registration method, but do not include a criterion standard to evaluate measurement accuracy. The vertebral reconstruction we present expands on these investigations through a systematic registration and measurement process to create quantifiable accuracy and reliability. In addition, because the vertebral reconstruction method was applied to images collected while flexed and extended during standing, the estimates of accuracy and reliability are conservative if applied to postures easier to maintain, such as seated flexion.

The accuracy evaluation of the reconstructions from the mid-field positional MRI demonstrated promising results when compared with intervertebral measurements from conventional imaging. For example, Lim *et al.*¹³ reported 3D vertebral measurement accuracy of ± 1.0 mm and ± 1.0 degrees using CT scans (1.0 mm x 0.43 mm x 0.43 mm voxel size). More recently, Ochia *et al.*¹⁷ used a volumetric reconstruction method to evaluate lumbar axial rotation with CT scans, and reported measurement error less than 0.1 mm and 0.2 deg. We found only one comparable study that used positional MRI to perform quantitative intervertebral measurements. Cargill *et al.*⁵ used positional MRI for vertebral reconstruction, but no criterion standard was available to assess the measurement accuracy. In addition, a direct comparison was difficult because the measurements were presented as subject-specific intervertebral angles during various postures. Reconstruction accuracy can be influenced by specific bone geometry⁶; therefore, we compared these results only to intervertebral measurements.

The measurement accuracy using the vertebral reconstruction method (displacement error <1.5mm, sagittal plane rotational error <1deg) was comparable to values reported in previous clinical and biomechanical research applications. Clinicians use sagittal plane vertebral rotation angles measured from medical images to evaluate lumbar spine pathologies. Zhao *et al.*²⁵ collected sagittal radiographs from six *in vitro* lumbar spine specimens in flexion, extension, and neutral positions. Three experienced spine surgeons measured the sagittal rotation angle of each vertebrae and reported an average measurement error of $2.16 \pm 0.78^\circ$ compared to a criterion measurement. Subject-specific finite element models of lumbar spine have been developed to evaluate the relationship between tissue mechanics and kinematics. These models require accurate experimental measurements of vertebral displacement to establish boundary conditions and validate performance for *in vivo* modeling. Two such models^{8,16} use *in vitro* experimental measurements with more than 5° of rotational uncertainty for lumbar vertebrae during flexion and extension range of motion. Therefore, the methodology of acquisition and reconstruction may be used to address questions that require *in vivo* measurements of the lumbar spine with a proper gravity orientation.

The accuracy evaluation performed using ovine vertebrae provides a conservative accuracy estimate for measurements using human vertebrae, and the method should be capable of better accuracy in human imaging. Human and ovine vertebrae are geometrically similar, but ovine vertebral bodies have a smaller lateral dimension than human lumbar vertebrae²¹. This resulted in fewer MR slices intersecting each ovine vertebra and less geometric data available from the ovine specimens for performing the registration. Also, with fewer segmentation points, effects from inaccurate point placements were likely magnified. Note, however, that an *in vitro* specimen is not susceptible to motion artifact and the specimen can be consistently positioned relative to the coil and magnetic field isocenter.

Measurement accuracy based on MR images is affected by acquisition parameters such as magnetic-field homogeneity, tissue characteristics, location of the anatomy relative to magnetic field isocenter, and type of MRI coil. Therefore, images collected from the same scanner with the same protocol can vary in clarity. In our evaluation, we neglected any imaging error and assumed all measurement error was the result of the reconstruction

method, which is a conservative evaluation of the reconstruction method. A direct evaluation of the imaging error could be performed using an MR phantom with multiple articulating vertebral segments.

Continuous boundary segmentation provided the best combination of efficiency, reliability, and accuracy and was chosen as the segmentation method to evaluate *in vivo* reliability. Error variance was higher with the intermediate-point segmentation methods compared to continuous boundary segmentation, which indicated that continuous boundary segmentation has higher reliability when segmentation is performed by multiple raters. The continuous boundary segmentation was faster than 6-, 8-, and 10-intermediate point segmentation methods. This likely occurred because continuous boundary segmentation can be performed by selecting fewer than six intermediate points per side to create a continuous-boundary spline curve. Also, consistent point spacing is not required with continuous boundary segmentation, which simplifies the procedure compared to the other methods used. When processing large data sets using continuous boundary segmentation, these factors may help mitigate fatigue and increase productivity.

For the current study, MRI scan protocols were chosen to maximize reliability in the sagittal plane, but at the expense of the ICCs measured in nonsagittal-plane measurements. Short acquisition times are required during static postures to prevent participant fatigue. Therefore, MR images were collected with large slice intervals and were limited to single planar orientation (sagittal). ICCs are a ratio between true variance and measured variance (true + error), and low measured variance magnifies the effect of the error. The low measured variance in nonsagittal-plane measurements, which is expected during flexion-extension motions, will necessarily result in lower ICCs¹⁸. As a result, we are careful not to over interpret the meaning of the nonsagittal ICCs. We anticipate that ICCs of the nonsagittal measurements would be improved if sagittal and coronal plane images were collected within a single scan, but this would increase acquisition time. Future work will include validation of the reconstruction method for lateral bending and axial twisting postures and coronal imaging planes. Because lateral bending and axial twisting postures create coupled vertebral rotations, it is currently unclear how best to characterize kinematic rotations (Cardan sequence, helical axes, etc.), or how the accuracy and reliability will compare to analyses in the sagittal plane.

This reconstruction method provides the efficiency needed to construct large datasets and power investigations that rely on population variance. Cargill *et al.*⁵, which used manual segmentation and manual registration, indicated that data processing times were a limiting factor when performing vertebral reconstructions from positional MRI. Although manual segmentation is necessary with mid-field MRI, our evaluation of segmentation methods indicates that accurate results can be achieved using a relatively fast segmentation method. In addition, high interrater reliability indicates that data processing can be distributed to multiple trained personnel, and can reduce the overall processing time. The automated registration with ICP significantly reduces data processing times when compared with manual registration, and offers a systematic process to follow.

In conclusion, this vertebral reconstruction method provides a systematic approach with accuracy, reliability, and efficiency. A primary advantage of the method is its fast and accurate reconstruction, while using positional MRI to evaluate lumbar spine under proper orientation of gravity and a large range of motion without exposing the participants to ionizing radiation. The data processing steps are robust to user error and will provide efficient construction of large data sets. This approach advances current reconstruction methods, and will provide a useful tool to develop and validate biomechanical models of lumbar spine.

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FIGURE 1. Workflow used to reconstruct vertebrae from positional scans with quality of the reference scan using positional MRI.

FIGURE 2.
Illustration of the four-step process used to assign a body-fixed coordinate system to each reference scan vertebra.

FIGURE 3.

Segmentation of a mid-plane sagittal slice of an ovine L1 vertebra (left) and the digitized point clouds that are generated from the segmentation (right and far right). Position and orientation of the sagittal MR slices and gaps are clearly visible in the top view for the “point” segmentation methods.

FIGURE 4.

Ovine MRI reconstruction error for each plane and each segmentation method.

Reconstruction error is determined relative to a criterion standard, which was established by CT measurements. Error bars indicate one standard deviation.

FIGURE 5.

Time required to complete ovine segmentation using each segmentation method. Results indicate average time per ovine vertebra for two ovine vertebrae (T13 and L1) segmented across three raters. Error bars indicate one standard deviation.

Table 1

MRI scan protocol for vertebral reconstruction using a FONAR 0.6-Tesla Upright Scanner

	Acquisition Time (min:sec)	# of slices	Slice thickness (mm)	Slice interval (mm)	Repetition Time (ms)	Echo Time (ms)
Reference Scan	5:17	20	4	4	610	17
Postural Scan	3:02	10	5	8	350	17