The *in vivo* three-dimensional motion of the human lumbar spine during gait

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Abstract

Lumbar spine pathology accounts for billions of dollars in societal costs each year. Although the symptomatology of these conditions is relatively well understood, the mechanical changes in the spine are not. Previous direct measurements of lumbar spine mechanics have mostly been performed on cadavers. The methods for *in vivo* studies have included imaging, electrogoniometry, and motion capture. Few studies have directly measured *in vivo* lumbar spine kinematics with in-dwelling bone pins.

This study tracked the *in vivo* three-dimensional motion of the entire lumbar spine (L5 to S1) in 10 healthy, young-adult subjects. Two 1.55 mm (0.062 in.) diameter Kirshner wires were inserted into each vertebra’s spinous process under anesthesia. Motion capture cameras were used to track a triad of passive markers attached to the wires. Offsets between anatomical landmarks and tracking markers were established with a CT scan for each individual vertebra. Subjects were asked to perform various exercises including walking and voluntary range of motion.

Subjects were able to complete all of the exercises. All subjects reported being adequately informed of all of the procedures and there were no neurological or orthopaedic complications. The range of the average inter-segmental range of motion was 4.26° – 4.38° in the sagittal plane, 2.61° – 4.00° in the coronal plane, and 4.11° – 5.24° in the transverse plane.

Using a direct (pin-based) *in vivo* measurement method, the motion of the human lumbar spine during gait was found to be triaxial. This appears to be the first three-dimensional motion analysis of the entire lumbar spine using indwelling pins. The results were similar to previously published data derived from a variety of experimental methods.

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1. Introduction

Adult degenerative spine disease is a significant problem, accounting for an estimated societal cost of 25–100 billion dollars in medical expenses, disability, and lost work time [1]. As the spine degenerates, it is assumed that the mechanical properties change, and while the symptomatology of these conditions is relatively well understood, the mechanical changes in the spine are not.

Most of our understanding of the motion of the lumbar spine is based upon cadaveric models or *in vivo* models using clinical goniometric measurements, radiographic imaging or surface optical markers [2–28]. Cadaveric models allow invasive, rigid fixation of bone pins and optical markers, but eliminate the effect of normal muscle contraction and limit the ability to measure motion during activities of daily living [28]. Videofluoroscopic imaging has been used to measure flexion and extension, as well as lateral bending during common
activities, but this technique is limited in its ability to detect translational and rotational movements about the lumbar spine \[6,13,14,17,29–31\]. Electrogoniometers have also been used to describe three-dimensional motion of the spine \[5,19,21,24\]. Several studies have reported on the intra- and interobserver variability of these techniques with reassuring results. However, most devices, such as the commonly used OSI CA 6000, are limited in their ability to discern segmental motion, and only measure global lumbar movements. Surface optical markers have been utilized in many series, but movements in the transverse plane have not been possible to measure due to the inherent error of motion at the bone – subcutaneous tissue/skin interface \[10\].

Motion capture technology has been used to directly measure vertebral motion \textit{in vivo}. Dickey \textit{et al.} used active motion capture LEDs attached to pedicle screws (L4 or L5 and S1) in nine subjects \[3\]. Steffen \textit{et al.} mounted electromagnetic tracking devices on Kirshner wires inserted into the spinous processes (L3 and L4) of 16 subjects \[22\]. These studies provided unprecedented insight into inter-segmental vertebral motion and validation for future pin/screw based studies.

The purpose of the present study was to build on previous work to develop a method for visualizing and quantifying the three-dimensional, \textit{in vivo} motion of the entire human lumbar spine.

2. Materials and methods

2.1. Preliminary procedure

After attaining approval from the University of Minnesota Institutional Review Board (HSC #0601M80446), 10 healthy, adult, asymptomatic volunteers were recruited to evaluate the normal kinematics of their lumbar spines. Subjects with a previous history of back pain or back surgery, body mass index \(>25\), susceptibility to infection, allergy to local anesthetics, coagulopathy or spinal deformity were excluded. If subjects met the criteria for inclusion, the nature of the project and the risks associated with the study were thoroughly discussed. After obtaining informed consent, screening anterior–posterior and lateral radiographs were obtained to confirm the presence of five lumbar vertebrae and the absence of significant spinal pathology (e.g. previous fracture, scoliosis, spondylolisthesis or spondylolysis). In addition, flexion/extension lateral radiographs were taken to measure the global lumbar range-of-motion (ROM) from L1 to S1 according to the method described by Cobb. Each subject was given antibiotic prophylaxis (500 mg Cephalexin) prior to the procedure to avert any potential infection due to wire insertion. A short physical examination was undertaken prior to the wire placement to determine the anthropometric measurements required for the standard gait model.

2.2. Operating room

In the operating room, an experienced orthopedic surgeon and full support staff were present during the procedure. Subjects lay prone and were supported by a curved frame in order to maximize lumbar kyphosis and increase the spacing between the spinous processes. By doing so, skin traction during the testing procedure was minimized. After anesthetizing the skin and periosteum with a dilute solution of lidocaine:epinephrine, the surgeon used fluoroscopy to locate the spinous processes of the L1–S1 vertebrae. Subjects were monitored closely for signs of discomfort. Guided by fluoroscopy, two 1.6 mm (0.062 in.) Kirshner wires were inserted approximately 1 cm into each spinous process (L1–S1) (Fig. 1A). One Jurgan pin ball was fixed to each pair. A motion capture marker triad was attached to each Jurgan pin ball (Fig. 1B).

![Fig. 1. The protocol for the current study included: (A) insertion of Kirshner wires in the operating room, (B) attachment of marker triads, (C) full lumbar CT scan, (D) digitization and segmentation of the CT scan, and (E) synchronization of the CT data with the motion capture data.](image)
2.3. Triads

The triads of motion capture markers were custom built from 1.5 mm diameter carbon fiber rods. Carbon fiber was chosen to minimize the inertial properties of the triad, as it is rigid yet light. The rods were fixed together with epoxy to form “t” shapes approximately 4 cm wide and 5.5–7 cm long. One end of the “t” used a ring terminal connector that allowed attachment to the Jurgan pin ball. Motion capture markers 10 mm in diameter were affixed to the other three ends. Each Jurgan pin ball was modified to allow an extra screw with which to mount the marker triad. The two wires inserted into each vertebra were clamped together to minimize the possibility of rotational artifact about the wire axis. The rigid marker triads were arranged in such a way to maximize marker visibility and minimize triad overlap (Fig. 1B).

2.4. Imaging

Once all of the Kirshner wires were inserted and triads mounted, each subject underwent an axial CT scan of the lumbo-sacral spine (0.5 mm resolution) (Fig. 1C). Transfer from the CT scanner to the motion analysis lab (two floors, several hallways and doorways, plus an elevator ride) was undertaken using a gurney to ensure that the triads were not inadvertently moved relative to the vertebrae during transport.

2.5. Motion capture

The three-dimensional positions of the markers were tracked in the motion lab using a 12 camera Vicon MX system (Vicon, Oxford, UK). Each subject went through a standard gait data collection protocol including static and functional model calibration. The functional model calibration consisted of range of motion trials for both hips and both knees that were used to calculate the appropriate joint parameters [32]. Additional virtual markers were defined at the pelvis and trunk. The subjects were then asked to complete five walking trials. As a subject specific reference, maximum voluntary range of motion in the sagittal, coronal and transverse planes were collected. Three trials were also collected for each of several other functional tasks including jogging, sit-to-stand, and lifting.

2.6. Post-procedure

Once the motion capture was completed, the wires were removed and subjects were observed for 30 min. The entire procedure, from wire insertion to wire extraction, lasted less than 2.5 h. Subjects were contacted both 2–3 days and 6–9 months after the procedure to monitor for any complications. In addition, subjects were asked to complete a follow-up questionnaire to appraise participant perception of the procedure.

2.7. Data processing

All of the trajectories in the motion capture trials were filtered using a low-pass filter with a 5 Hz cut-off frequency. The markers on each of the vertebral triads were also constrained to maintain a constant relative geometrical relationship using the procustes method [33]. This method fits the three trajectories of a triad to an average reference position frame by frame, minimizing the amount each marker must be adjusted.

Processing of the CT images included individual renderings of each vertebra and motion capture markers. The center of the rendered volume for each marker was used as its reference location. The three markers of each triad were also rendered together to create one solid volume per triad. Triangular meshes were created from each of the renderings and saved in standard Wavefront OBJ file format (Fig. 1D) and imported into the motion capture software (Polygion v2.4, Vicon, Oxford, UK) (Fig. 1E). Identification of anatomical landmarks was undertaken within the CT visualization software environment (Analyze 7.0, Analyze Direct, Lenexa, KS, USA).

The CT-based anatomical landmarks and motion capture marker reference points were transformed into the corresponding dynamic marker-based technical coordinate systems (CS) using the following equation:

\[
v^b_i = \beta T_B^M T_M^T S^T S_0^T v_S^i, \tag{1}
\]

where

\[
S_0^T S = \begin{pmatrix}
0 & 1 & 0 & 0 \\
1 & 0 & 0 & 0 \\
0 & 0 & 1 & 0 \\
0 & 0 & 0 & 1
\end{pmatrix} \tag{2}
\]

is the transformation from the left handed CS of the CT scanner, to a right handed scanner prime CS. The transformation from the scanner prime CS to the motion capture marker based CS is

\[
M_T S = \begin{pmatrix}
M R_S^T & M^T S_0 \\
0_{1x3} & 1
\end{pmatrix} \tag{3}
\]

where

\[
M R_S^T = \begin{pmatrix}
\hat{e}^M_1 \\
\hat{e}^M_2 \\
\hat{e}^M_3
\end{pmatrix} \tag{4}
\]

is the local CS derived from the centers of the rendered motion capture markers in the scanner prime CS and

\[
M^T S_0 = M R_S (O_S - O_M) \tag{5}
\]

is the translation from the origin of marker based CS to the origin of the scanner prime CS, \(O_S\), defined as (0,0,0). The origin of the marker based CS, \(O_M\), is at the midpoint between the most lateral marker on the triad and the most superior marker on the triad.

The transformation from the marker based technical CS to the bone based CS \((^B T_M)\) is simply a translation from the origin of the marker based CS to the origin of the bone based CS, which is the midpoint of the two points of symmetry on the vertebra. The rotation from the CS aligned with the marker technical CS to the anatomical CS is

\[
^B T_M = \begin{pmatrix}
^B R_B & 0_{3x1} \\
0_{1x3} & 1
\end{pmatrix} \tag{6}
\]
where

\[ B_R = \begin{pmatrix} \hat{e}_{b1} \\ \hat{e}_{b2} \\ \hat{e}_{b3} \end{pmatrix} \]  

(7)

is the rotation from technical to anatomical CS.

The vertices of the vertebral meshes were also transformed using the same Eqs. (1)–(7). The entire surface of each vertebra was then expressed in a motion capture marker-based coordinate system, allowing the meshes to be displayed along with the motion capture data, and providing a dynamic visualization of the relative vertebral motion.

2.8. Anatomical coordinate systems

The local anatomical vertebral coordinate systems are defined similarly to those of Stokes [34]. The y axis points to the subject’s left and is parallel to two symmetric landmarks on the superior surface of the pedicles. The z axis points up and is perpendicular to the superior end-plate. The x axis points forward and is perpendicular to the y and z axes. The origin is at the midpoint of the two pedicle landmarks (Fig. 2).

2.9. Kinematics

Once the anatomical coordinate systems were determined, the relative vertebral kinematics were calculated. Cardan angles (rotation sequence: y, x, z) were calculated for each vertebral pair from S1 to L1. The inferior vertebra in the pair was considered fixed, resulting in flexion/extension calculated about the y axis of the inferior vertebra, lateral bending calculated about a rotated x axis, and axial rotation calculated about the z axis of the superior vertebra.

2.10. Statistics

The range of motion was calculated in all three anatomical planes for each exercise. A trial was included in the analysis if valid data existed at the extrema of one complete cycle of the exercise. Subject averages for all valid trials for a specific exercise were averaged together at each inter-vertebral level. This resulted in, at most, 60 ROM values for each subject (5 inter-segmental levels × 4 exercises × 3 planes). The mean and 95% confidence interval was then calculated across subjects. T-tests were run to determine if the group means were significantly non-zero. For each combination of exercise and plane, an ANOVA was run to determine if inter-segmental level was a significant factor between groups.

3. Results

Ten subjects completed the study (four females, six males). A follow-up questionnaire was completed by nine of the ten subjects. These showed that all responding subjects felt adequately informed about the procedure. There was also unanimous agreement among responders that the study met their expectations. Seven of the nine subjects stated that they would repeat the procedure. When asked how long it took to feel “normal”, five of the subjects reported feeling normal again within 2 weeks, with four taking up to 2 months, and one subject reporting that his “back remained sore during stretching of lumbar region for several months” and that he had “decreased range of motion for over a year.”

Fig. 2. Example of the rendering of an L1 vertebra with motion capture markers. The anatomical coordinate system is shown. The markers, as measured by the motion capture system, are in red with the overlaid marker mesh from the CT scan shown in blue. Although the meshed markers are not perfect spheres, the alignment with the motion capture data appears to be good.

Fig. 3. Averages and 95% confidence intervals for the various ranges of motion. The numbers above the error bars indicate how many subjects had valid data that was included in the computation. The only significant differences were seen in the coronal plane during lateral bending between the L2/L3 ROM and the L5/S1 ROM as well as the L3/L4 ROM and the L5/S1 ROM (curly brackets).
When asked for how long they experienced pain, five reported 1 week or less, three reported between a week and a month, and one reported 60 days.

The sagittal plane ROM from S1 to L1 during the maximum flexion–extension exercise was \(70.0^\circ \pm 13.8^\circ\). This was similar to the ROM found using lateral flexion/extension radiographs (\(66.7^\circ \pm 10.1^\circ\)).

The relative ROM for each inter-segmental level (level) was selected as the primary result for this study (Fig. 3 and Table 1). For some subjects, no valid data were available at some of the levels for some of the exercises. This information was left out of the analysis. The percentage of excluded data at each level was: L5/S1 – 13.2%, L4/L5 – 15.8%, L3/L4 – 15.8%, L2/L3 – 13.2%, and L1/L2 – 5.3%.

For walking, there was no significant difference between the levels for any of the anatomical planes. For the maximum voluntary ranges, the only significant difference was found in the lateral bend exercise between L2/L3 and L5/S1 and between L3/L4 and L5/S1 (\(p < .05\)). All of the ranges of motion for all exercises, at each level, in each plane, were significantly different from zero (\(p < .01\)).

4. Discussion

One of the main goals of this study was to determine if using bone pins and passive marker motion capture technology was a safe and practical way to analyze in vivo three-dimensional motion of the lumbar spine. Safety was assessed by monitoring patient discomfort and complication rate. Subjects reported mild discomfort during pin placement, and descriptions of the pain after the procedure ranged from “tender” to “extremely sore/stiff.” When asked how long they experienced pain after the study, all of the nine subjects who completed the follow up survey reported less than 2 months. Five subjects reported feeling pain for a week or less. All of the nine responders felt the study met their expectations and that they were adequately informed of the procedures. Also, there were no wound complications or nerve injuries.

The lumbar vertebrae were successfully tracked for the majority of subjects, levels, and motions. The data appeared qualitatively correct upon inspection, i.e. the vertebrae were in the correct order and orientation and followed the motion of the subject. The marker triads, as rendered from the CT scan, were overlaid on the marker trajectories with the correct position and alignment. Subsequently, there was close apposition of facet joints with limited interpenetration/distraction.

Quantitatively, walking was determined to be a fundamentally triaxial exercise. Ranges of motion significantly different from zero were calculated for all three anatomical planes. However, these motions were small, especially when compared to the available range of motion. The triaxial nature of lumbar motion is consistent with the known net motion between the proximal trunk and the pelvis [35]. The complex multi-planar lumbar motion is what allows the lower extremities to remain in a largely planar pattern. The lumbar spine must accommodate the axial rotation of the pelvis as the leg is extended to prepare for initial contact. The obliquity of the pelvis must also be absorbed through the lumbar spine. In the Gillette Childrens Specialty Healthcare Center for Gait and Motion Analysis, the control data for pelvic obliquity ranges from 10\(^\circ\) to 10\(^\circ\) during free speed gait [36] Without compensation through the spine this would push the sternum laterally by a distance that is 17% (sin 10\(^\circ\)) of the height of the trunk. This, in turn, would lateralize the center of mass and subsequently the ground reaction force, and alter the joint reaction forces and net moments throughout the body. Loss of this compensa-

### Table 1

<table>
<thead>
<tr>
<th>Plane</th>
<th>Motion</th>
<th>Walk</th>
<th>Flex/ext</th>
<th>Lat bend</th>
<th>Twist</th>
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<td>14.33</td>
<td>6.64</td>
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ory capability through injury or intervention may have significant effects on overall gait, as well as the motion and loading of adjacent vertebral segments. This and other questions are to be addressed in future studies involving subjects with spine pathology, and pre–post intervention analyses.

The sagittal plane ROM for the different levels during the maximum flexion exercise was similar to previously published results [6,13,22,28]. However, whereas this study found a maximum ROM at the L3/L4 level, Miyasaka et al. found a maximum at L4/L5 and Yamamoto et al. found a maximum at L5/S1. The difference may be due to the methods used in the studies. Miyasaka et al. conducted their study in vivo using functional radiographs. Yamamoto et al. used fresh frozen cadavers and dictated the motions based on applied forces.

The results for the lateral bending trials were similar to Dvorak et al. [6] and to Steffen et al. [22]. The axial rotation results were similar to Yamamoto et al. [28].

The inter-segmental ROM during walking has not been previously studied, therefore no direct comparison can be made to the current data.

### 4.1. Limitations

One of the issues contributing to the difficulty in calculating the full cycle kinematics was reconstruction noise and marker occlusion. This was most notable in situations when the two markers were in close proximity or possibly touching (marker merging). The design and placement of the marker triads will be improved to reduce the amount of overlap between the segments. Reducing the size of the motion capture volume can also improve marker merging and occlusion.

Pin bending is also of concern. The study by Steffen et al. measured the amount of strain in the pins of four of their subjects [22]. The pins in the current study were made of essentially the same material, and had the same cross-sectional shape (circular) as those in Steffen’s study. An estimate of the expected bending can thus be made by scaling Steffen’s results appropriately. Since the pin radius in the current study was 1.55 mm and the pin radius in the Steffen study was 2.5 mm, we can expect approximately 2.6 times the deformation. This would result in errors of less then 0.8°.

It is important not to overlook the discomfort of the subjects and risks to which they were exposed. Although all of the subjects reported that they were well informed, and most would undergo the procedure again, all of them experienced some level of pain. Pain and the other concomitant risks must be considered when deciding whether to perform a pin-based study, or to implement some less invasive procedure. Costs and logistics must also be evaluated since time in a fully staffed operating room, a CT scan, and several hours in the motion capture lab are necessary.

### 4.2. Directions for the future

Future plans include refining of the post-processing methods to optimize the reconstruction of the vertebral motions. One such refinement is to use the entire mesh of the marker triad to optimize the transformation from the CT data to the motion capture data. There is also data from the running, sitting, and lifting trials that have yet to be reported.

Evaluating clinical populations is also a goal. The effect of pathology on spinal mechanics in subjects with conditions such as spondylolisthesis or degenerative disk disease is not well understood. Neither is the effect of interventions such as spinal fusion or spine arthroplasty. The interaction between poor mechanics and pain or functional limitations is also of interest.

A detailed comparison of in vivo studies to cadaveric studies will be important to determine if the cadaveric studies actually represent true in vivo spinal motion.

This appears to be the first study in which the in vivo motion of the entire lumbar spine has been measured with bone pins. A complex protocol involving the surgical department, the radiology department, and the motion analysis lab was successfully implemented. The initial results are promising. The integration of pin-based motions and subject-specific bone meshes including anatomical landmarks, with a flexible three-dimensional workspace is novel in live human subjects. This allows for visualizations of the movement of the entire lumbar spine during functional tasks, as well as quantitative measurement of inter-segmental motions that have not previously been possible.

### Conflict of Interest

None of the authors had any financial or personal relationships with other people or organizations that could inappropriately influence this work.

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