Biomechanics of Grade I degenerative lumbar spondylolisthesis. Part 1: In vitro model

NEIL R. CRAWFORD, PH.D., SEDAT ÇAGLI, M.D., VOLKER K. H. SONNTAG, M.D., AND CURTIS A. DICKMAN, M.D.

Spinal Biomechanics Research Laboratory, Division of Neurological Surgery, Barrow Neurological Institute, St. Joseph’s Hospital and Medical Center, Phoenix, Arizona

Object. The authors sought to create and to evaluate an in vitro model of Grade I degenerative (closed-arch) spondylolisthesis.

Methods. The model of spondylolisthesis was created by two primary procedures: 1) resection of the disc; and 2) stripping of anterior and posterior longitudinal ligaments away from the vertebral bodies (VBs). In 13 vertebral levels obtained from three cadaveric lumbar spines, the tissues were resected sequentially in alternating order to determine the relative contribution of each resection to spinal instability. The entire specimens were loaded with nonconstraining torques and then individual levels were loaded with anteroposterior shear forces. The motion values were measured optoelectronically for each specimen at individual levels.

Conclusions. The integrity of the disc was more important than attachment of the ligaments to the VB, but the resection of both structures was necessary to achieve substantial destabilization. The structures of the spine are highly resilient, and destabilization is difficult to achieve without performing extensive resection. Using the techniques described in this paper to alter normal spines, a level of spinal instability (Grade I; 25% slippage) that may represent spondylolisthesis can be modeled in vitro.

KEY WORDS • degenerative spondylolisthesis • lumbar spine • biomechanics • in vitro modeling • cadaver

In several in vitro biomechanical studies of the lumbar spine, authors have assessed the stability provided by certain fixation devices when attached to normal spines; however, the fixation devices were not inserted in spines representative of a patient with the degree of spinal instability that requires surgery. Therefore, it is debatable whether the positive or negative results reported in such studies accurately represent clinical situations. Furthermore, a particular fixation device might be less effective for the treatment of one type of instability than for another. A thorough investigation of a fixation device would quantify its effectiveness in treating several types of instability.

In laboratory studies, it would be ideal to undertake stabilization procedures in cadaveric specimens obtained from patients who actually required surgery for spinal instability. Because obtaining such specimens is infeasible, a reasonable solution is to alter normal cadaveric spines to create a model representative of instability. In several studies of the lumbar spine the investigators have modeled acute traumatic or iatrogenic instability by impacting or resecting appropriate structures in cadaveric spines.

A common form of spinal instability that has not been previously modeled in vitro is Grade I degenerative (closed-arch) spondylolisthesis, which is defined as anterior slippage up to 25% of the AP width of the VB. Because this condition typically develops over months or years, it is more difficult to model than that of acute instability. The goal of this study was to develop a model of degenerative spondylolisthesis by surgically modifying normal cadaveric tissues. The model was then used to study the stabilizing properties of pedicle screws and threaded interbody devices.

Several factors can contribute to the onset of degenerative spondylolisthesis: disc degeneration, structural abnormalities (for example, anomalous facet orientation), and an increase in the stresses to the joint. To model degenerative spondylolisthesis, we first assumed that complete facet joint incompetence was necessary to allow anterior slippage. We also assumed that some amount of degeneration of the disc and the ligaments surrounding the disc space was required to produce Grade I slippage. To avoid excessive resection of the disc in favor of the ligaments (or vice versa), we evaluated the relative importance of these two tissues.

Abbreviations used in this paper: AP = anteroposterior; EZ = elastic zone; NZ = neutral zone; ROM = range of motion; VB = vertebral body.
Materials and Methods

Lumbar Specimens

Thirteen motion segments obtained in three human cadaveric lumbar spines were studied (Table 1). Specimens were obtained fresh frozen, thawed in a bath of normal saline at 25°C, and carefully cleaned of muscle tissue without damaging ligaments, discs, or joint capsules.

The sacrum (or L-4 in Specimen 2; Table 1) was prepared by inserting household wood screws at various locations and passing 1.5-mm-diameter surgical guide wires through the structure. The screws and wires were potted in a block of polymethylmethacrylate, which was clamped in an angle vise attached to the base of the testing apparatus. The vise was adjusted such that the anterior margins of the upper lumbar VBs formed a vertical line. The body of T-12 was potted using screws and polymethylmethacrylate in a metal fixture for the application of torque loads. Two wire loops (1.5-mm-diameter stainless-steel surgical guide wires) were passed through each VB so that AP shear loads could be applied.

While exposed during preparation and testing, specimens were wrapped in gauze that was frequently sprayed with 0.9% saline to prevent dehydration. To limit exposure, specimens were refrozen at the end of each day of testing. This treatment is probably the best method of mitigating changes to the specimen’s mechanical properties when long exposure is required.1,2,21

Modeling of Spondylolisthesis

To create instability resembling Grade I spondylolisthesis, the initial strategy was to resect tissue in small increments until AP motion in response to manually applied shear loads equaled 25% of the width of the VB at each motion segment. In two pilot specimens, the facets were resected first using standard Kerrison rongeurs, but the tendency for AP shear changed little. Next, the nucleus pulposus and annulus fibrosis were removed using pituitary rongeurs and curettes (Fig. 1 upper left). Consistently, a small wall (approxi-

---

TABLE 1

Summary of clinical data obtained in cadaveric specimens

<table>
<thead>
<tr>
<th>Specimen No.</th>
<th>Vertebral Levels</th>
<th>Age (yrs) at Death, Sex</th>
<th>Cause of Death</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>T12–sacrum</td>
<td>55, M</td>
<td>heart disease</td>
</tr>
<tr>
<td>2</td>
<td>T12–L4</td>
<td>66, M</td>
<td>cirrhosis</td>
</tr>
<tr>
<td>3</td>
<td>T12–sacrum</td>
<td>52, M</td>
<td>brain tumor</td>
</tr>
</tbody>
</table>

---

Fig. 1. Drawings depicting the technique for modeling spondylolisthesis. In all specimens, the facets were first removed to destroy the major buttress against AP motion. **Upper Left:** Disc material was removed through four holes in the perimeter of the disc. **Upper Right:** The anterior and posterior longitudinal ligaments were separated from the bone to allow gapping instead of ligamentous tensioning as the VB moved anteroposteriorly. **Lower Left and Right:** After the ligaments were detached and the disc and facets were resected, spondylolisthesis occurred.
mately 2 mm wide) of the lateral annulus fibrosus was left in place bilaterally because this tissue is essential in creating a tension-band effect when interbody cages or dowels are inserted, as will be demonstrated in Part 2 of this study. Surprisingly little AP motion occurred following disc removal. Finally, the anterior and posterior longitudinal ligaments were separated from the VB by using a periosteal elevator (Fig. 1 upper right). Only after all three steps were completed was a condition resembling low-grade spondylolisthesis obtained (Fig. 1 lower left and right). We had not predicted that the residual annulus and ligaments surrounding the disc would contribute substantially to preventing AP shear. In the process of gradually resecting the discs in pilot specimens, the spondylolisthesis grade was visually estimated while shear forces were manually applied to the specimen. Typically, it was easier to create rotational instability than shear instability. The overall extent of resection required to induce obvious AP shear instability exceeded our expectations; in most cases, we were unable to reach 25% slippage.

To compare the relative contributions of the ligaments and disc in maintaining spinal stability in response to shear forces, the order of tissue resection was changed in the next three specimens, which comprise the dataset presented here (13 levels total). The ligaments were separated from alternating levels and no disc material was removed (Fig. 1 upper right; seven levels), or disc material was removed without separating the ligaments (Fig. 1 upper left; six levels). Specimens were tested in each partially destabilized condition and then in a fully destabilized condition. In each case the standardized measure was the extent of tissues resected or separated rather than the percentage of slippage achieved. The extent of tissue resection was approximately the same as that in the pilot specimens.

**Biomechanical Testing**

A noncontacting stereophotogrammetric optical system (Optotrak 3020; Northern Digital, Waterloo, ON, Canada) was used to monitor the three-dimensional motion of the specimen. Three 1.5-mm-diameter stainless-steel end-threaded surgical guide wires were inserted in a noncollinear arrangement in each VB and in the sacrum and cut to 4 to 9 cm. Infrared-emitting diodes were rigidly glued to the free ends of the guide wires (Fig. 2). A digitizing probe (accessory to Optotrak) was used to establish the spatial relationship between markers on the vertebrae and the local vertebral coordinate systems. Custom-designed software was used to convert marker movement to vertebral angles about each of the anatomical axes; this method mathematically models each motion segment as overlapping cylinders. The software also allowed specified locations on the vertebrae to be tracked during testing.

Nonconstraining, nondestructive pure moments (torques) were applied to each specimen through a system of cables and pulleys in conjunction with a standard servohydraulic test system (MTS, Minneapolis, MN), as has been previously described (Fig. 2 left). Other researchers have successfully applied torque loads to T12–sacrum specimens. Torque loading is independent of the site...
of application; the same-magnitude torque is distributed evenly to each vertebral level in the same global axis in which it is applied, regardless of the condition of the levels below.\textsuperscript{13} Torques were applied about the appropriate anatomical axes of T-12 while holding the sacrum fixed to induce three different types of motion: flexion–extension, lateral bending, and axial rotation. The main drawback in the application of torques occurs when the anatomical alignment of the different levels is substantially different or becomes different during movement in the coupled (secondary) motion directions. In the lumbar spine, the LS–sacrum region is more lordotic than the upper motion segments. Hence, the torques required to induce T12–L1 lateral bending or axial rotation may not be applied in the best direction to achieve these motions in the LS–sacrum region. However, if the secondary alignment remains unchanged—as it generally was in this study regardless of the degree of instability—motion measurements obtained from different tests are directly comparable. Before data were recorded, the specimens were preconditioned at 5 Nm for 60 seconds three times. After preconditioning, the specimens were allowed 60 seconds for tissues to recoil before data collection was performed. During the period of data collection, loads were applied quasi-statically (each load held for 45 seconds) in 1-Nm increments to a maximum of 5 Nm. Data were recorded at 2 Hz.

After the nondestructive torque testing, nondestructive shear force was applied to investigate the AP translatory motion associated with spondylolisthesis. Shear forces were applied through the wire loops inserted through each VB (Fig. 2 upper and lower right). At each motion segment, positive and then negative shear forces were applied by attaching a single cable at each end to the wire loops on the superior and inferior vertebra of the motion segment. The hydraulic piston was raised to apply tension to the cables, similar to that in the pure moment technique. These forces caused spondylolisthesis and retroolisthesis (Fig. 2 upper and lower right). Shear forces were applied quasi-statically in 10-N increments to a maximum of 50 N. As in the torque-loading procedures, the specimens were preconditioned three times to maximum load and then allowed to rest for 60 seconds before we conducted data collection. To ensure that loads were applied in a uniform direction, excessive bending of the specimen was prevented using a “guy wire” attached to the wire loop in the vertebra below the level being studied. Before each test run, tension was applied to the guy wire by hand and locked in place.

Data Analysis

From the raw data, the ROM, NZ, EZ, and flexibility values were calculated. The ROM value indicates the maximum displacement (in degrees or millimeters) through which the specimen can move with the largest applied load; NZ, the portion (in degrees or millimeters) of the ROM that exhibits ligamentous laxity or laxity between the bones and hardware;\textsuperscript{14} and EZ, the portion of the ROM where ligaments, bones, or hardware substantially resist motion. The specimen angle (or the location of the specimen after releasing the third preconditioning cycle and waiting 60 seconds for tissues to recoil) was defined as the boundary between NZ and EZ.\textsuperscript{15} The flexibility value indicates the mechanical response once ligaments or hardware are engaged and must be measured within the EZ to obtain valid and reproducible results.\textsuperscript{15} Thus, angular flexibility coefficients were calculated from loads between 3 Nm and 5 Nm and corresponding angular deformations by using a least-squares fit in Microsoft Excel (version 5.0). Similarly, shear flexibility coefficients were calculated from loads between 20 N and 50 N and corresponding deformations in the AP direction. The units of flexibility are recorded as deformations (degrees/Newtonmeter or millimeter/kilo-Newton), which is an approximation of how much motion the joint or hardware allowed beyond the NZ for each Newton or Newtonmeter of load applied. Flexibility is the inverse of stiffness, which indicates the amount of resistance that the hardware or joint provides per degree or millimeter of motion.

Analysis of the reduced data was performed using paired one-tailed Student’s t-tests to determine whether each progressive step in destabilization significantly increased or decreased motion (p < 0.05). Increases in the NZ, EZ, ROM, or flexibility indicated increased instability within the region of motion that each parameter describes. Nonpaired two-tailed t-tests were used to compare specimens in which ligaments were first separated with those in which discectomy was first performed.

Results

The angular NZ, EZ, ROM, and flexibility values demonstrated in response to torque loading often changed significantly in the different states of destabilization. Compared with the normal condition of the spine, the change (usually an increase) observed in a given motion parameter after discectomy was significantly greater than that demonstrated in the same parameter after the ligaments were separated in eight of 15 parameters (Table 2). Corroborating these findings, we also found that the extent of instability additionally achieved when the specimens were fully destabilized was significantly less when the first destabilizing procedure was discectomy compared with ligament separation in eight of 15 parameters.

The translational NZ, EZ, ROM, and flexibility values determined in the AP direction in response to shear loading (Table 3) were similar to those demonstrated in the bending tests. Compared with the normal spine, the increases in shear translation were significantly smaller in four of six parameters when ligaments alone were separated first than when discectomy was performed first. After complete destabilization of the spine in two of six parameters the further increase in shear motion was significantly greater when ligaments alone were separated first compared with when discectomy was performed first. The total shear translation of 3.2 mm in the fully destabilized state (Table 3) did not approach Grade II spondylolisthesis, which would require a translation of 7 to 10 mm.

Table 2

<table>
<thead>
<tr>
<th>Parameter (*) &amp; Mode</th>
<th>Normal ROM (13 levels)</th>
<th>Ligaments Separated (6 levels)</th>
<th>Disc-ectomy (7 levels)</th>
<th>Full Destabilization (13 levels)</th>
</tr>
</thead>
<tbody>
<tr>
<td>NZ flex–ext</td>
<td>0.3 ± 0.2</td>
<td>1.0 ± 1.2†</td>
<td>2.6 ± 2.0†</td>
<td>2.3 ± 2.1†</td>
</tr>
<tr>
<td>NZ lat bend</td>
<td>0.3 ± 0.1</td>
<td>0.6 ± 0.4†</td>
<td>1.2 ± 0.4†</td>
<td>1.1 ± 0.4†</td>
</tr>
<tr>
<td>NZ axial rotat</td>
<td>0.1 ± 0.1</td>
<td>0.1 ± 0.1†</td>
<td>0.4 ± 0.1†</td>
<td>0.5 ± 0.2†</td>
</tr>
<tr>
<td>EZ flex</td>
<td>3.6 ± 1.6</td>
<td>3.1 ± 1.2‡</td>
<td>3.8 ± 1.3</td>
<td>5.6 ± 2.6‡</td>
</tr>
<tr>
<td>EZ ext</td>
<td>2.1 ± 1.6</td>
<td>3.2 ± 1.8†</td>
<td>2.4 ± 0.9†</td>
<td>2.4 ± 0.9</td>
</tr>
<tr>
<td>EZ lat bend</td>
<td>3.0 ± 0.9</td>
<td>3.1 ± 0.9‡</td>
<td>4.5 ± 1.1†</td>
<td>4.7 ± 1.1†</td>
</tr>
<tr>
<td>EZ axial rotat</td>
<td>1.1 ± 0.5</td>
<td>2.8 ± 1.0†</td>
<td>4.9 ± 1.1†</td>
<td>4.9 ± 1.0†</td>
</tr>
<tr>
<td>ROM flex</td>
<td>3.8 ± 1.8</td>
<td>4.1 ± 1.7†</td>
<td>6.4 ± 2.3†</td>
<td>7.8 ± 2.8†</td>
</tr>
<tr>
<td>ROM ext</td>
<td>2.3 ± 1.9</td>
<td>4.2 ± 2.9†</td>
<td>5.0 ± 2.3†</td>
<td>4.7 ± 2.7†</td>
</tr>
<tr>
<td>ROM lat bend</td>
<td>3.2 ± 0.9</td>
<td>3.7 ± 1.3†</td>
<td>5.7 ± 1.5†</td>
<td>5.8 ± 1.3†</td>
</tr>
<tr>
<td>ROM axial rotat</td>
<td>1.2 ± 0.5</td>
<td>2.9 ± 1.0†</td>
<td>5.3 ± 1.1†</td>
<td>5.4 ± 1.2†</td>
</tr>
<tr>
<td>flexibility (in N/Nm)</td>
<td>0.46 ± 0.18</td>
<td>0.24 ± 0.08†</td>
<td>0.19 ± 0.01†</td>
<td>0.26 ± 0.04†</td>
</tr>
<tr>
<td>flex</td>
<td>0.26 ± 0.13</td>
<td>0.38 ± 0.18†</td>
<td>0.27 ± 0.10</td>
<td>0.22 ± 0.06</td>
</tr>
<tr>
<td>ext</td>
<td>0.40 ± 0.11</td>
<td>0.31 ± 0.07†</td>
<td>0.37 ± 0.05</td>
<td>0.34 ± 0.09†</td>
</tr>
<tr>
<td>lat bend</td>
<td>0.17 ± 0.05</td>
<td>0.38 ± 0.10†</td>
<td>0.45 ± 0.09†</td>
<td>0.46 ± 0.10†</td>
</tr>
</tbody>
</table>

* All values are presented as means ± standard deviation. Abbreviations: axial rot = left or right axial rotation; ext = extension; flex = flexion; lat bend = left or right lateral bending.
† Value represents a significant difference from normal condition.
‡ Value represents a significant difference from fully destabilized condition.
Lumbar biomechanics: part 1

### Discussion

By itself, discectomy was mostly responsible for creating the final state of angular destabilization; this was indicated by the fact that most angular parameters did not increase significantly from the time of discectomy to full destabilization (Table 2). Flexion EZ and ROM were exceptions: both showed large increases after the ligaments were separated. Likewise, few shear-related parameters increased significantly between the discectomy and full destabilization procedures, indicating that ligament separation was the less influential step (Table 3). However, the defining parameter for spondylolisthesis—the shear ROM—significantly increased as each phase of the destabilization process proceeded. Hence, although discectomy was the more important procedure, both destabilization procedures were necessary to create enough destabilization to model Grade I spondylolisthesis.

In contrast to other motion parameters, the angular flexibility coefficients often decreased rather than increased relative to normal after destabilization (Table 2). Thus, the motion that was allowed before stretching of the remaining ligaments and annulus fibrosus (NZ) increased substantially after destabilization, but these remaining tissues stretched more stiffly when loaded than the joints and tissues of the intact spine. This behavior may be attributed to the absence of the facet joints, whose capsules help distribute elastic stretching during flexion and lateral bending but whose bone articulations provide a stiff barrier against axial rotation and extension. Accordingly, after the facets were removed, flexibility increased during axial rotation and extension but decreased during flexion and lateral bending. It is to be expected that disc removal will increase the incidence of stiff bone-on-bone contact at the VBs during flexion, extension, or lateral bending, and discectomy may account for the final decrease in flexibility in all modes except axial rotation.

To our knowledge, no prior experimental model of lumbar spondylolisthesis has been reported. Although we attempted to attain anterior translation of 25% of the VB width through extensive surgical resection, it was difficult to induce anywhere near Grade II spondylolisthesis (25–50% slippage). This finding is a testament to the spine’s normal stability and resistance to spondylolisthesis. Because spondylolisthesis develops slowly, partially competent facets are present in patients suffering from degenerative spondylolisthesis. Although the defective facets may not prevent spondylolisthesis, they may still provide some resistance against bending motions. In our model, the facets were removed completely, creating a multidirectional instability that was probably more severe than that associated with the actual disease. In addition, the transection of disc material in our model does not accurately simulate the actual elongation of disc fibers associated with the disease. Our methods, however, were the only apparent solution to the given problem.

It was difficult to standardize exactly how much disc was removed during the destabilization of each specimen because the annular region could not be visualized during resection. This drawback may have increased the variability in destabilization among specimens, which was assessed in our companion article.

### Conclusions

The model of Grade I spondylolisthesis is a limited but probably adequate representation of actual clinical spondylolisthesis. The instability demonstrated in the model is likely more severe than that associated with the clinical condition but with less AP shearing. Such a model, however, may be the only option for studying this condition because obtaining large numbers of cadaveric specimens from persons who suffered from spondylolisthesis in life is not feasible.

### References

ment using threaded interbody cages/dowels and pedicle screws. J Neurosurg (Spine 1) 94:51–60, 2000

Manuscript received September 29, 1999. Accepted in final form August 9, 2000.
Current address for Dr. Çagli: University of Ege School of Medicine, Izmir, Turkey.
Address reprint requests to: Curtis A. Dickman, M.D., c/o Neuroscience Publications, Barrow Neurological Institute, 350 West Thomas Road, Phoenix, Arizona 85013–4496. email: neuropub@chw.edu.