In vitro biomechanical analysis of three anterior thoracolumbar implants

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Object. The goal of this study was to evaluate the comparative efficacy of three commonly used anterior thoracolumbar implants: the anterior thoracolumbar locking plate (ATLP), the smooth-rod Kaneda (SRK), and the Z-plate.

Methods. In vitro testing was performed using the T9–L3 segments of human cadaver spines. An L-1 corpectomy was performed, and stabilization was achieved using one of three anterior devices: the ATLP in nine spines, the SRK in 10, and the Z-plate in 10. Specimens were load tested with 1.5-, 3-, 4.5-, and 6-Nm in flexion and extension, right and left lateral bending, and right and left axial rotation. Angular motion was monitored using two video cameras that tracked light-emitting diodes attached to the vertebral bodies. Testing was performed in the intact state in spines stabilized with one of the three aforementioned devices after the devices had been fatigued to 5000 cycles at ± 3 Nm and after bilateral facetectomy.

There was no difference in the stability of the intact spines with use of the three devices. There were no differences between the SRK- and Z-plate–instrumented spines in any state. In extension testing, the mean angular rotation (± standard deviation) of spines instrumented with the SRK (4.7 ± 3.2˚) and Z-plate devices (3.3 ± 2.3˚) was more rigid than that observed in the ATLP-stabilized spines (9 ± 4.8˚). In flexion testing after induction of fatigue, however, only the SRK (4.2 ± 3.2˚) was stiffer than the ATLP (8.9 ± 4.9˚). Also, in extension postfatigue, only the SRK (2.4 ± 3.4˚) provided more rigid fixation than the ATLP (6.4 ± 2.9˚). All three devices were equally unstable after bilateral facetectomy. The SRK and Z-plate anterior thoracolumbar implants were both more rigid than the ATLP, and of the former two the SRK was stiffer.

Conclusions. The authors’ results suggest that in cases in which profile and ease of application are not of paramount importance, the SRK has an advantage over the other two tested implants in achieving rigid fixation immediately postoperatively.

Key Words • spine • spine fusion • bone screw • bone plate • internal fixation • biomechanics

For a variety of reasons, anterior implants are being used with greater frequency in the management of spinal cord and cauda equina compression syndromes. Anterior approaches allow greater access to the ventral aspect of the spinal canal without sacrificing the spinous processes, laminae, facets, or intervening ligaments. By implanting a graft in place of the fractured VB and applying a lateral implant, the anterior column is reconstructed. To optimize fusion, bone grafts should be maintained under compression, and this is better achieved via an anterior than a posterior approach. Also, the anterior approach generally requires fixation extending to only one level above and one level below the fracture, whereas posterior instrumentation often extends two to three levels above and below the fracture. The posterior approach requires some sacrifice of the posterior elements, that is, the laminae and facets, if decompression after a burst fracture is to be achieved. The aforementioned advantages of the anterior approach are also reflected in a lower rate of complications, failure, and kyphosis compared with outcomes achieved using the posterior approach.

A large number of titanium anterior devices have been approved for clinical use by the United States Food and Drug Administration. Some, such as the SRK (AcroMed, Raynham, MA), consist of rods and four constrained bicortical screws that allow compression and distraction. Others, like the Z-plate (Sofamor-Danek, Memphis, TN) consist of a slotted plate that provides compression by means of two constrained bicortical bolts. The ATLP (Synthes, Paoli, PA), on the other hand, affords limited compression by means of two temporary bone screws and fixation with four constrained screws.

Because of the different biomechanical features of the...
three aforementioned devices, a biomechanical testing protocol was instituted. To simulate implantation in patients, our study was performed in human cadaver spines subjected to a total L-1 corpectomy, the attached anterior and posterior longitudinal ligaments excised, and an anterior strut graft placed.

**Materials and Methods**

Human spines (T9–L3) frozen at −20°C and stored in plastic double bags were thawed and their musculature dissected, leaving the ligaments and discs intact. The spines were refrozen at −20°C until testing. Radiographic studies were obtained to exclude spines with preexisting fractures or bone disease. Bone mineral densities at T11, T12, L1, and L2 were quantified in the anteroposterior orientation by using dual-energy x-ray absorptiometry.

The spines were potted rostrally and caudally. Dry-wall screws were placed at L3 to prevent disruption of the L2–3 disc space. The spine was then placed in a mold while the horizontal position of L1 was maintained. A potting mixture of two-thirds body filler (Bondo; Mar-Hyde Corp., Atlanta, GA) and one-third fiberglass resin was then poured around the L3 VB, with firm purchase assured by placement of the screws. A loading frame was attached to T9 and similarly potted. The potted spine was thawed overnight and then attached to a solid, immovable base plate before testing.

Three infrared LEDs were attached to each neural arch at T11, T12, L1, and L2 (transverse processes, laminae, and spinous processes) as well as the base plate. Motion of the spine was recorded by applying the stereophotogrammetry method, which involves spatial localization of a point in space based on a set of photographs taken from different camera positions. The three-dimensional motion of the LEDs is tracked by two photosensitive cameras (Selspot II system; Innovision Systems, Inc., Warren, MI) that sense the intensity of infrared light from each LED. This system is accurate to within less than 0.5 mm in translation and to within less than 10 mrad in rotation. Using the base plate as a fixed reference point, the three-dimensional position of each LED was quantified according to an X,Y,Z cartesian axis system. Processing these positional data generated the angular rotations of the spine in response to the applied loads.

Using a system of weights and pulleys, quasi-static loads were applied to each spine sequentially in opposite directions, creating pure moments of 0, 1.5, 3, 4.5, and 6 Nm. To prevent the viscoelastic effect, the spines were ranged maximally in all orientations prior to load application. These loads were applied in six directions in the following order: flexion, extension, right and left lateral bending, and right and left axial rotation, as described by Goel and colleagues. Specimens were allowed to “creep” for 30 seconds after load application and before data collection. The spines were load tested using the following four steps. 1) The intact spines were load tested prior to any manipulation. 2) Following L1 corpectomy, grafts were placed in the spines with a 1.25-in-(32-mm)-diameter dowel made of oak; stabilization was achieved with the ATLP device in nine spines (Fig. 1 left), the SRK in 10 (Fig. 1 center), and the Z-plate in 10 (Fig. 1 right). Wooden dowels were used for convenience and to eliminate variations in graft material. In addition, these dowels have been shown to equal or exceed cortical bone in stiffness and have been used for this purpose in similar investigations. All implants were applied on the left side in a manner similar to that used in the operating room and in accordance with the manufacturer’s specifications. In the case of the ATLP, compression force was exerted upon the graft by means of two temporary 4 × 35-mm bone screws engaging the vertebrae and by tapered holes in the plates. The plate was affixed to the spine by using four unicortical constrained bone screws measuring 7.5 mm in diameter and 45 mm in length. In the SRK, 6.25-mm-diameter bone screws were applied; a length of 45 to 55 mm was selected to achieve bicortical purchase. These bone screws are constrained to two 6.35-mm rods by means of set screws. In turn the rods were spanned by two couplers to prevent translation and axial rotation. In the case of the

![Fig. 1. Left: Photograph showing the ATLP with four constrained unicortical screws applied between T-12 and L-2. Center: Photograph of the SRK device showing four bicortical screws applied across T-12 and L-2. Right: Photograph showing the Z-plate applied across T-12 and L-2 with two constrained bicortical bolts and two nonconstrained bicortical screws.](image-url)
Z-plate, bicortical constrained 7-mm-diameter bolts were implanted posteriorly, and nonconstrained 6.5-mm-diameter screws 45 to 55 mm in length were implanted anteriorly for bicortical purchase. 3) The spine was removed from the testing cage and affixed to a materials-testing system (Material Testing Systems, Minneapolis, MN). The specimen was kept in a plastic hood attached to a humidifier providing 100% humidity. The stabilized spine was fatigued for 5000 cycles of flexion–extension at a frequency of 0.5 Hz, using a load of ±3 Nm as measured by the load cell. The specimen was then returned to the testing cage and retested as described above. 4) Finally, load testing was repeated after bilateral facetectomy.

Each specimen was regularly irrigated with 0.9% saline to prevent dehydration during load testing. The angular rotation of T-12 relative to L-2 was measured in degrees in each of the six aforementioned directions. Thus an increase in angular rotation of T-12 relative to L-2 (T12–L2) meant a decrease in stability or stiffness. Analysis of angular rotation as a measure of stiffness was performed in accordance with published studies from this laboratory12–16 as well as others.1,3,23 In previous studies1,3,12,23 the authors have shown that large moments (6–6.4 Nm) are necessary to elicit significant differences in angular rotation between devices. Thus in this study we played a greater degree of stiffness in flexion (3.9°) and played a greater degree of stability on flexion was noted after bilateral facetectomy, with an increase in rotation from 3.9° to 5.2° and 2.1° for the A TLP and SRK spines, respectively (p = 0.0252).

Table 1. Mean rotations (± standard deviation) across T12–L2 in the intact, stabilized, and fatigued states, and after bilateral facetectomy.

<table>
<thead>
<tr>
<th>Range of Motion (°)</th>
<th>Flexion</th>
<th>Extension</th>
<th>Rt Lateral Bending</th>
<th>Lt Lateral Bending</th>
<th>Lt Axial Rotation</th>
<th>Rt Axial Rotation</th>
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<td>Z-plate</td>
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<tr>
<td>intact</td>
<td>4.4 ± 1.6</td>
<td>4.1 ± 2.5</td>
<td>5.1 ± 1.8</td>
<td>5.4 ± 1.8</td>
<td>2.6 ± 1.9</td>
<td>2.4 ± 1.4</td>
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<tr>
<td>stab</td>
<td>4.3 ± 2.6</td>
<td>3.3 ± 2.3</td>
<td>1.0 ± 0.9*</td>
<td>3.6 ± 2.2*</td>
<td>2.7 ± 1.5</td>
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<td>fatigued</td>
<td>4.9 ± 3.3</td>
<td>4.4 ± 2.7</td>
<td>1.3 ± 1.7*</td>
<td>4.4 ± 2.5*</td>
<td>3.3 ± 2.2</td>
<td>4.0 ± 2.1</td>
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<tr>
<td>bilat facet</td>
<td>7.0 ± 4.3</td>
<td>8.2 ± 6.4</td>
<td>2.4 ± 2.3*</td>
<td>6.4 ± 4.3*</td>
<td>3.7 ± 2.0</td>
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<td>5.3 ± 4.2</td>
<td>6.2 ± 4.2</td>
<td>5.6 ± 2.4</td>
<td>5.6 ± 2.0</td>
<td>2.3 ± 1.6</td>
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<tr>
<td>stab</td>
<td>5.6 ± 5.6</td>
<td>4.7 ± 3.2</td>
<td>1.7 ± 1.5</td>
<td>2.3 ± 2.1</td>
<td>2.6 ± 1.7</td>
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<td>fatigued</td>
<td>4.2 ± 3.2</td>
<td>2.4 ± 3.4</td>
<td>1.1 ± 1.5†</td>
<td>1.2 ± 3.4</td>
<td>2.6 ± 1.3</td>
<td>2.3 ± 1.3</td>
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<tr>
<td>bilat facet</td>
<td>7.1 ± 6.8</td>
<td>6.3 ± 7.2</td>
<td>3.9 ± 5.3</td>
<td>5.5 ± 8.3</td>
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<tr>
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<td>5.0 ± 1.3</td>
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<td>1.7 ± 0.7</td>
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<td>3.9 ± 2.1*†</td>
<td>9.0 ± 4.8*</td>
<td>2.8 ± 2.4</td>
<td>3.9 ± 2.6</td>
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<tr>
<td>fatigued</td>
<td>8.9 ± 4.9</td>
<td>6.4 ± 2.9</td>
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<td>8.0 ± 3.4</td>
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<td>5.3 ± 3.3</td>
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* Significantly different from each other within a device in opposite motion directions (p ≤ 0.05). † Significantly different from each other within a device in the same motion direction (p ≤ 0.05). ‡ Significantly different from the intact value (p ≤ 0.05). Data on intact spines are included for the sake of comparison. Data correspond to 6-Nm bending moment. Abbreviations: bilat facet = bilateral facetectomy; stab = stabilized.

Results

The donors were equally distributed between sexes, and ages ranged between 53 and 89 years for the entire group, with a mean (± SD) of 69 ± 8 years for the ATLP-, 75 ± 10 years for the SRK-, and 68 ± 9 years for the Z-plate–instrumented spines. There was no significant difference among groups in terms of age distribution (p = 0.190). The average BMD values were 0.7 ± 0.1 g/cm² for the ATLP spines, 0.64 ± 0.09 g/cm² for the SRK spines, and 0.80 ± 0.21 g/cm² for the Z-plate spines. There were no significant differences in the BMD values of T-12 (p = 0.0531), L-1 (p = 0.1704), or L-2 (p = 0.1043) among the three devices. The BMD of T-11 measured 0.63 ± 0.08, 0.84 ± 0.23, and 0.68 ± 0.08 g/cm² for the SRK, Z-plate, and ATLP, respectively (p = 0.0252).

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Devices were compared with one another. There was no difference in the stability of the intact spines of the three devices (Table 1). There were no differences between the SRK- and Z-plate–instrumented spines in any state (Table 1). In extension, the spines stabilized with the SRK (4.7 ± 3.2°) and Z-plate (3.3 ± 2.3°) were more rigid than those stabilized with the ATLP (9 ± 4.8; p = 0.0047; Table 1, Fig. 2). In flexion postfatigue, only the SRK (4.2 ± 3.2°) provided greater stiffness than the ATLP (8.9 ± 4.9°; p = 0.0314; Table 1, Fig. 3). Also postfatigue, only the SRK (2.4 ± 3.4°) was more rigid than the ATLP in extension (6.4 ± 2.9°; p = 0.0336; Table 1, Fig. 3). All three devices were equally unstable after bilateral facetectomy (Table 1, Fig. 4). For technical reasons, one SRK spine failed during fatigue testing with screws pulling through the inferior endplate of T-12. Otherwise, the study was completed in all spines.

Types of manipulation were compared in each device. In the ATLP-instrumented spines, a significant decrease in stability on flexion was noted after bilateral facetectomy, with an increase in rotation from 3.9 ± 2.1° to 11.6 ± 6.6° in the stabilized spines following bilateral facetectomy (p = 0.029; Table 1, Fig. 4). The ATLP-stabilized spines displayed a greater degree of stiffness in flexion (3.9 ± 2.1°) than in extension (9 ± 4.8; p = 0.009). Postfatigue, SRK-stabilized spines remained more rigid in right lateral bending (1.1 ± 1.5°) than intact spines (5.6 ± 2.4°; p = 0.0209;
Table 1, Fig. 3). After fatiguing (1.3 ± 1.7˚) and bilateral facetectomy (2.4 ± 2.3˚) the Z-plate–stabilized spines were more rigid (1 ± 0.9˚) than the intact spines (5.1 ± 1.8; p = 0.0001) in right lateral bending. The Z-plate–stabilized spine was more rigid in right lateral bending than left lateral bending before (p = 0.003) and after fatigue (p = 0.004) and bilateral facetectomy (p = 0.023). In right axial rotation, Z-plate–stabilized spines became unstable after bilateral facetectomy (5 ± 2.1˚) compared with intact spines (2.4 ± 1.4˚) (p = 0.0644).

The power estimates for comparison between devices for the intact spines ranged from 0.072 to 0.275%. This was due to the small differences that we observed between the intact Z-plate– and SRK-instrumented spines in right axial rotation (0.4˚) and flexion (1.1˚), as well as between the intact SRK- and ATLP-stabilized spines in extension (1.1˚) (Table 1). It would take a difference of approximately 2˚ to achieve a 50% power. The observed differences in rotational units of 3˚ or more has a greater than 90% power in this experiment. Hence significant differences encountered in the results reported here represent sizable changes in rotation.

Discussion

Several anterior fixation devices are currently available for anterior thoracic and lumbar stabilization. Some consist of rods for longitudinal components affixed to the vertebral bodies, such as the Kaneda or Kostuik–Har-lington (Zimmer, Warsaw, IN) devices. Other devices feature a plate to bridge the gap between bone screws; these include the Syracuse I plate (AO Foundation, Bern, Switzerland), the ATLP, the Z-plate, and the University plate (AcroMed Corp, Cleveland, OH). With the exception of the I plate, these instruments are dynamic, allowing for some compression on the graft before final fixation.

As difficult and inconvenient as it may be, comparative testing of devices should be performed in the manner in which they are designed to be used.16,34 All anterior spinal procedures, except for the treatment of scoliosis, entail anterior grafting with bone, methyl methacrylate, titanium, or other materials. This anterior graft is an integral component of the construct created using anterior devices whether rods or plates. Testing similar to ours has been conducted using the Kaneda device, the Z-plate, the University plate, and the TSRH (Sofamor-Danek) in calf spines.1 Load testing was conducted in the lumbar spine after an L3–4 discectomy and grafting. All devices exerted a significant stabilizing effect that exceeded that of the intact spine in flexion, extension, and lateral bending. All devices restored the stability of the spine in axial rotation, but only the SRK device exceeded the intact specimen in torsion. Our results showed that in terms of stiffness only the Z-plate–stabilized spine exceeded that of the intact spine in right lateral bending. Otherwise, our three devices were able to restore stability to a level comparable to but not exceeding the intact spine (Table 1, Fig. 2). The difference between our results and those of An, et al.,1 is due
to differences in testing models. Their model involved calf spines destabilized by discectomy alone, whereas ours entailed a total L-1 corpectomy. Thus it was easier for their implants to achieve significantly greater stiffness than the intact spine. Our results are comparable to theirs, however, in that in both models the SRK device conferred the most rigidity (Fig. 2).

Similar testing in calf spines following L-3 corpectomy has been performed using the SRK, TSRH, the contoured anterior spinal plate, and the Kostuk–Harrington devices. In flexion and axial loading, only the SRK and TSRH devices were able to restore spinal stability to a level equal to or exceeding that of the intact spine. In torsion, only the SRK was able to restore stiffness to the level of the intact spine. This superior performance by implants, previously described by us as well as by other investigators, is attributable to design features. Devices allowing maximum compression on the graft with screws constrained to the rods, such as the SRK and TSRH devices, provide greater rigidity.

Testing these devices in a corpectomy model made of artificial VBs (Sawbones model; Pacific Research Laboratories, Inc., Vashon, WA) or ultrahigh-molecular-weight polyethylene to create a worst-case scenario does not satisfy the recommended criteria for application of these devices. Although informative, such tests have often revealed failure of devices after fatigue testing in the absence of an interbody graft. One of the criticisms leveled against the use of cadaveric spines for biomechanical testing is that these spines vary in age and BMD. Our spines, although collected over a period of 1.5 years from different sources, were comparable, without significant differences among groups in age. Although a difference in BMD was encountered in T-11 (p = 0.0252), the intact spines in the three groups behaved similarly, without significant differences in rigidity as measured by angular rotation (Table 1, Figs. 2–4).

Testing devices for inherent rigidity is important for many reasons. Not only does an implant correct and maintain proper alignment, but its stiffness enhances fusion. Fusion rates in the spine have been shown to be enhanced after placement of instrumentation. Thus the more rigid implant is less likely to fail and is able to maintain spinal alignment and enhance fusion rates. The Z-plate–instrumented spines were more rigid than the intact spine in right lateral bending (Table 1). In right lateral bending, the SRK-stabilized spines were more rigid than the intact spines after fatigue (Table 1). Because of the rigidity of the SRK construct, no differences were encountered between right and left lateral bending in the stabilized, fatigued, and facetectomy-treated spines (Table 1). The Z-plate–stabilized spine, on the other hand, was significantly more rigid in right than left lateral bending after stabilization, fatigue testing, and facetectomy (Table 1). This difference in the Z-plate between right and left lateral bending can be attributed to its placement on the left side of the spine (Fig. 1 center). In right lateral bending, the graft restricts bending to the right and the Z-plate restricts distraction of the spine on the left. The ATLP was equally nonrestrictive in right or left lateral bending. In
extension, both the SRK- (4.7 ± 3.2˚) and Z-plate (3.3 ± 2.3˚)–implanted spines were more rigid than the ATLP-implanted spines (9.0 ± 4.8˚) (Table 1, Fig. 2). Because of the meager compression provided by the ATLP and the flexion-limiting effect of the graft, the ATLP-instrumented spines were significantly (p = 0.009) more rigid in flexion (3.9 ± 2.1˚) than extension (9.0 ± 4.8˚).

Although the SRK- and Z-plate–instrumented spine behaved similarly in flexion and extension, only the fatigued SRK spine (4.2 ± 3.2˚ and 2.4 ± 3.4˚, respectively) was more rigid than the fatigued ATLP spine (8.9 ± 4.9˚ and 6.4 ± 2.9˚, respectively) (Table 1, Fig. 3). This rigidity imparted to the spine by the SRK and Z-plate is attributable in part to the specified bicortical engagement of their bolts and screws as well as the ability of these devices to compress the graft between the vertebral end plates to the maximum degree. The ATLP provides only minimal compression by virtue of two 4 × 35–mm temporary screws.

The additional rigidity provided by the SRK device compared with the Z-plate is due to the fact that, in the SRK, the screws are constrained to the rods, whereas in the Z-plate only the bolts are constrained to the plates, not the anterior screws. Thus, for maximum rigidity in an anterior or spinal construct, the ideal device is one that provides graft compression by means of four bicortical constrained screws.

References


