failure may be precipitated by a stifled state caused by the compressed fragments.1,2 This may in turn contribute to the results described in this report, and may be associated with clinically relevant implications.

Conclusions The data presented in this study suggest that peritendineous pedicle screw fixation provides a degree of biomechanical performance that is superior to that of pedicle screw fixation systems. Even after long-term repetitive cycling, the bone-metal interface maintained its mechanical integrity for both types of fixation systems studied (transpedicular pedicle screw and pedicle screw fixation) when used with an interbody spacer. Facet engagement, using a lag screw, appears to augment stability. The stability measured by the transfacet pedicle screw fixation did not degrade over 180,000 cycles. In this model, transfacet pedicle screw fixation appears equivalent biomechanically to traditional pedicle screw fixation.

Key Points

Transfacet pedicle screw fixation was shown to have effectiveness in stabilizing the lumbar motion segments and limiting the range of motion similar to that of a transfacet screw under short-term cyclic loading conditions when interbody spacers were used.

- In short-term cyclic loading conditions, both transfacet pedicle screw fixation and pedicle screw fixation provided similar stability. Stability did not decrease for other groups after 180,000 loading cycles in the model used.

- The stability of motion segments did not decrease with repetitive spinal loading up to 180,000 cycles.

Transfacet pedicle screw fixation using a lag screw decreased the range of motion during the period required for a successful fusion to occur.

References

lateral fusion with transalaran facet screw fixation in the lumbar spine. Although transalaran facet screw fixation differs slightly from the transfacet pedicle screw fixation performed in the current study, multiple studies have demonstrated that both insertion techniques lead to adequate stabilization for a highly successful dorsolateral spinal fusion, while minimizing facet capsule disruption.\(^{13-17,19,2,23,25-28}\) Biomechanically, similar studies have found no significant performance differences among transalaran facet fixation, transfacet pedicle fixation, and pedicle screw fixation systems for dorsolateral fusion.\(^{31,32}\) High fusion rates that range from 90% to 98% have been observed with transfacet pedicle and transalaran facet screw fixation. This is comparable with that of pedicle screw fixation.\(^{1,11,17,19,21,23}23\) The associated complication rates with respect to transfacet pedicle screw breakage were small compared with that for pedicle screw breakage. Overall, transfacet pedicle screw fixation results in less dorsal destruction of the bony elements and has the biomechanical advantage of stabilizing the dorsal column and the possibility of percutaneous placement. A lag screw for the transfacet pedicle screw was used in the technique presented in this report. Lag screws have been used conventionally for the compression of bone components. This type of fixation takes advantage of the design, and is used to compress the facet joint, thus "engaging" the facets. Biomechanical assessment of the aforementioned "engagement strategy" demonstrates that similar mechanical fixation is provided by transfacet pedicle screw fixation and traditional pedicle screw fixation. These data are consistent with the data of others who have biomechanically compared nondestructive testing of transalaran facet screw fixation with that of pedicle screw fixation.\(^{31}39\) It is emphasized that facet engagement via the aforementioned lag effect may have precluded the spacers, thus conferring an element of stiffness. This indeed may have affected the short-term study results. Fatigue testing, as accomplished with the long-term phase of this study, should negate this effect, if present. It is therefore not likely that any significant confounding effect via facet engagement exists regarding the overall interpretation of the results from this study.

**Short-Term and Long-Term Stability**

The short-term and long-term cycling of the specimens instrumented with transfacet pedicle screws showed no decrease in mechanical stability, as compared with that of the pedicle screw construct. In fact, transfacet pedicle screws performed significantly better than the pedicle screw system in fixation during the short-term phase. The favorable influence of facet screws in flexion has been observed in other studies.\(^{34,40}\) Long-term biomechanical structural integrity of the system was not observed at the bone-implant interface, and stiffness was not compromised over 180,000 cycles of repetitive loading.

The cyclic loading strategy used in this study simulated approximately 6 weeks of daily spinal loading. This exceeds the period of initial bony incorporation across a fusion site.\(^{25}\) The extended loading period represents an appropriate laboratory scenario for the modeling of the spine during the initiation of the fusion process.\(^{1,17-19}\) The use of a lag screw design for the facet screw with its ability to "engage" the facets aggressively and maintain most of the facet capsule provides a biomechanical advantage over traditional pedicle screw fixation. Such facet engagement may increase stability and reduce micromotion across the FSU.

**Study Limitations**

Transfacet pedicle screw fixation is an alternative form of spinal stabilization that appears to provide stability similar to that achieved with pedicle screw fixation. This study assessed the biomechanical stability conferred by a lag transfacet screw. The advantage of the lag effect may be important as evidenced by the results presented in this report. This is the first study in the literature to document the performance of a lag screw for facet fixation and stability. The facet joints were not obliterated or disrupted in any fashion for either fixation system. This permitted the attainment of a statistically fair comparison. However, it may be emphasized that during the application of the transalaran facet screw fixation, facet joint disruption commonly occurs. It is emphasized that the ideal degree of stability and reduction of micromotion across a bone graft is unknown and does not necessarily predict the ability of a fixation system to obtain a solid arthrodesis. Limited profile, less bony invasion, and the potential for percutaneous placement are benefits of transfacet pedicle screw placement, making it potentially more appealing than traditional pedicle screw fixation. However, the long-term cycling effects observed in this study were tested in a flexion mode on an instrumented motion segment. Therefore, the long-term biomechanical stability in other loading directions (extension and lateral bending) should be addressed.

The current study was limited by specimen availability and the decomposition rate of cadaveric specimens, making it impossible to test each specimen intact initially and then over long-term cycling (180,000 cycles). The number of cycles evaluated for the long-term phase simulated 6 weeks of daily spinal loading during postoperative healing. The fatiguing effect of repetitive cycling on bone and the extended amount of time required for the long-term phase (50 hours per specimen) prevented analysis of intact specimens.

Finally, the fatiguing (cyclical loading) of bone and soft tissue leads to spinal integrity degradation in a cadaveric specimen, in which physiologic bone remodeling does not exist. This degradation is a result of small microfractures in the bone, desiccation of the disc and ligaments, and eventual tearing of soft tissues. This eventually results in spinal segment failure. This end-stage
than specimens instrumented with pedicle screw systems. No significant differences were found between the intact and instrumented specimens in compression or torsion (P > 0.05).

For flexion, extension, and lateral bending, both fixation systems (transfacet pedicle screws and traditional pedicle screw techniques) led to significantly reduced range of motion. For flexion, there were no significant differences between transfacet pedicle screws and the pedicle screw system for any of the testing modes. Table 3 shows the reductions in range of motion for each fixation system.

Long-Term Phase

The flexural stiffness at various intervals was calculated for each specimen. The stiffness variations over 180,000 loading cycles for pedicle screw specimens with interbody spacers and for transfacet pedicle screw specimens with interbody spacers are shown in Figures 4 and 5. There were no significant changes in stiffness over time for either type of fixation system (P > 0.05).

A two-tailed unpaired t test was used to compare the stiffness of pedicle screw specimens with that of transfacet pedicle screw specimens. The mean stiffness across 180,000 cycles was analyzed for each specimen. There were no significant differences in the mean stiffness of traditional pedicle and that of transfacet pedicle screw specimens with bilateral interbody polymer spacers (P > 0.05).

Discussion

Traditional Pedicle Screw Fixation

Traditional pedicle screw fixation is widely used for internal stabilization of the lumbar spine. Fusion procedures usually are augmented with instrumentation to minimize the motion across an interbody bone graft, thus hopefully resulting in an increased fusion rate. However, rigid fixation can have detrimental effects if the fixation does not allow adequate stresses and micro-motion to be transmitted to the bone graft. Excessive rigidity of the spinal implant can cause stress shielding. This inhibits the bone graft’s ability to experience stresses incurred with daily spinal loading, thus resulting in resorption of the graft.12,23,24 Conversion of excessive motion across a bone graft can contribute to a pseudarthrosis and early instrumentation failure. The optimum degree of rigidity required for successful spinal fusion is unknown.

Transfacet Pedicle Screw Fixation Versus Translaminar Facet Screw Fixation

Both types of facet screw fixation have been investigated as a less-invasive alternative for dorsal spinal integrity augmentation across a fusion site.4,13,22,23 Reich et al.13 successfully demonstrated the clinical success of dorsal

Table 4. Long-Term Phase Range of Motion Data

<table>
<thead>
<tr>
<th>Test Mode</th>
<th>Range of Motion</th>
</tr>
</thead>
<tbody>
<tr>
<td>Pelvic screw fixation with interbody grafts</td>
<td>1.9 ± 0.7</td>
</tr>
<tr>
<td>Transfacet screw fixation with interbody grafts</td>
<td>3.4 ± 0.6</td>
</tr>
</tbody>
</table>

The mean ± standard deviation range of motion (in degrees) for flexion/extension for the transfacet screw and pedicle screw specimens are given.

Materials and Methods

Specimen Preparation

This study was conducted in two phases: a short-term phase involving six cycles per testing mode and a long-term phase that involved 180,000 cycles of flexion-compression loading. For this study, 15 fresh cadaveric lumbar spines (L3-L5) were obtained. The bone mineral density (BMD) for each spine was determined via dual-energy x-ray absorptionmetry (DEXA; Hologic, QDR 4500A, Waltham, MA). Specimens with a bone density < −2.5 were deemed osteoporotic and inadequate for the purposes of this study. The nine cadaveric lumbar spines used for the short-term phase had a mean age of 52 ± 2.9 years and a mean BMD of 0.94 ± 0.12 g/cm². The six cadaveric spines used for the long-term phase had a mean age of 51 ± 6 years and a mean BMD of 0.59 ± 0.18 g/cm². The specimen criteria for both phases of testing are listed in Table 1.

The surrounding musculature was removed from each spine, leaving all ligamentous structures intact. The spines were disected into L1–L2 and L3–4 segments, or functional spinal units (FSUs), yielding 18 FSUs for short-term biomechanical testing and 12 FSUs for long-term biomechanical testing. Each FSU was instrumented in custom-made gripping frames using a polyurethane resin (Bondo/Max-Style, Atlona, GA). Wood screws were placed in the upper and lower vertebrae of each motion segment in a multiaxial fashion to secure the motion segment in the embedding material. The upper and lower vertebrae were embedded into the polyester resin to their midpoints. The disc and facet joints were free of embedding material and accessible for the application of instrumentation.

Specimen Instrumentation. Each FSU was instrumented in a random manner with either a pedicle screw fixation construct (Texas Scottish Rite Hospital, Medtronic Sofamor Danek, Memphis, TN) or a translaminar facet screw fixation construct (Navisys, San Diego, CA) by a spine fellowship-trained spine surgeon. The facet joints were not ablated or discrepant for either instrumentation procedure. For the short-term phase, all 18 FSUs incorporated bilateral plastic semicircular interbody spacers. For the long-term phase, 12 FSUs were instrumented with dorsal instrumentation (6 with traditional pedicle screw fixation and 6 with translaminar facet screw instrumentation). All 12 FSUs had bilaterally placed interbody spacers.

Interbody Spacer Technique. The interbody spacers used were plastic replicas of hemispherical rings (Navisys). Appropriately sized lateral anularomies, ranging from 10 to 15 mm in height, were created for bilateral graft sites, and the nucleus pulposus was removed bilaterally. Sequentially sized Cobb curettes and interspace shapers were used to prepare the disc space for an interbody graft. The parallel 12-mm spacer was introduced into the disc space from one side to distract the
height of the disc space during graft insertion. The interbody spacer (height, 12 mm) was inserted from the opposing side of the disc space. The soft tissue was trimmed so that the disc space was restored. The graft was inserted, and the disc space was restored. The spacer was inserted in a similar manner.

Transfacet Pedicle Screw Insertion. Initially, a small rongeur was made in the middle of the pedicle’s cortical surface for the purposes of drilling a drill point. This was necessary to prevent wandering of the drill point during rotation. The drill was directed downward and outward parallel to the cooled edge of the lamina at an angle of 45°, and passed through the facetal articulation, creating a tunnel for the insertion of the screw. The inferior and superior facets were drilled with a drill bit 3.5 mm in diameter aimed toward the pedicle. The pedicle core was tapped using a cortical tap 4.5 mm in diameter, and two 4.5 × 40 mm facet screws (NuVasive Inc., San Diego, CA) were inserted bilaterally.

The facet screws were lag screws (partially threaded) composed of titanium alloy (Ti6Al4V). The screws were placed medially from the inferior facet and directed laterally toward the superior facet (Figure 1B). The insertion torques were measured for the last three or four screw turns using a torque wrench calibrated to ±1.3% (Stereotact Franklin, IL).

Pedicle Screw Insertion. The Texas Scotts Rite Hospital pedicle screw system (TSR3, 6.3 × 40 mm; Medtronic Sofamor Danek) was used for dorsal fixation of the motion segments. As initially created the screw holes, and pedicle screws were inserted at a depth of approximately 80% (40 mm) of the distance from the surface of the lamina to the ventral cortex using the traditional method of Magnel.22 Care was taken to ensure that the screw was inserted parallel to the axis of the pedicle and the pedicle cortex, and branching of the ventral cortex was avoided. Kyping and drilling of the pedicle screw hole was not performed to increase screw purchase within the bone. Titanium screws (diameter, 6.3 mm) were locked into place without the use of cross-fitting (Figure 1C).

Biomechanical Testing.

Short-Term Phase. Specialized gripping fixtures were designed to align and secure each specimen to an electromechanically driven uniaxial materials testing apparatus (Figure 2; MTS Systems Corporation, Eden Prairie, MN). All 18 of the FSIs were initially tested intact. After intact testing, each FSI underwent a controlled disassembly with spacer placement, consistent with the application of dorsal instrumentation according to the procedure previously described. Nine motion segments (3.2-1.4°) were instrumented randomly with traditional pedicle screw fixation systems. Each FSI was nondestructively tested in five sequential moments: compression, flexion, extension, left lateral bending, and right lateral bending. Each specimen was loaded in tension with a 1100-lb. testing machine. Each specimen was randomly mounted onto custom testing fixtures that allowed free rotation in the sagittal plane (i.e., both the upper and lower fixtures were free to rotate) so that the rotation was recorded by rotational potentiometers. Initially, the center of rotation for each FSI was established by applying a 400-N compressive load to the upper and lower segments. The specimen was repositioned, and the load was reapplied until no angular motion was detected by potentiometers, indicating a centered specimen. Once the center of rotation was determined, it was easily marked on the facet for each specimen tested. The specimens were then secured in pure compression with the application of an axial load at an actuator rate of 0.25 cm/min. The maximum load level was held at 400 N. The gripping fixtures were not allowed to rotate in compression. Each specimen was preconditioned for three cycles. An additional three cycles were applied and sampled at 10 Hz using Testwave 4 and a DATAQ data acquisition system (DATAQ Instruments, Akron, OH) on a Gompac Desktop EN Series PC.

For flexion and extension testing, a compressive load was placed on the specimen to dorsally (for flexion or extension) or ventrally (for the center of rotation. An axial load was applied at a rate of 0.25 cm/min to produce a maximum bending moment of 4 Nm. During the center of rotation, the specimen was then fixed. In flexion, extension, and left lateral bending, each specimen was preconditioned for three cycles. Then, three cycles were sampled for an additional three cycles following the testing scheme used in compression. For torsion testing, a servohydraulically driven biaxial device (Instron Corp.) was used. The specimen was subjected to pure bending, and the load was recorded until no angular motion was detected by the potentiometers, establishing the center of rotation.

Statistical Analysis. For all testing modes, the stiffness of each FSI was calculated from a tangent line fit to the load-displacement data in the elastic region of the curve. The stiffness values were measured from the load-displacement data in the elastic region. The stiffness values were measured using the slope of the tangent line taken from the load-displacement curve. Flexion, extension, and lateral bending stiffnesses were calculated from the applied bending moment versus range of motion curves by quantifying the slope of the tangent line within the elastic zone in a fashion similar to that used for the compressive stiffness.

During all aspects of the study, range of motion was measured continuously from the rotational potentiometer data during spinal loading. Range of motion was defined as the total angular motion of the FSIs during loading. For the short-term phase, an analysis of variance (ANOVA) with a Newman-Keuls comparison was used to evaluate any significant differences in stiffness and range of motion between instrumentation systems. For the long-term phase, a two-way ANOVA with a Newman-Keuls comparison was used to detect any statistical differences in stiffness and range of motion over 180 cycles within and between systems. In addition, a t-test, unpaired t-test was performed to determine differences in mean stiffness between transfacet pedicle screw specimens and between the traditional pedicle screw specimens. All statistical tests were calculated using GraphPad Prism software, Version 3.02 (GraphPad Software, San Diego, CA). For all statistical tests, significance was defined by a P value less than 0.05.

Results

Short-Term Phase. The mean stiffness values for all test modes are shown in Table 2. In flexion, extension, and lateral bending, both fixation systems were significantly stiffer than the control mode (P < 0.001). Additionally, in flexion, transfacet pedicle screw specimens were significantly stiffer (30%) compared to the control group.

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