

A Biomechanical Comparison of Facet Screw Fixation and Pedicle Screw Fixation

Effects of Short-Term and Long-Term Repetitive Cycling

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Study Design. A biomechanical study was conducted to assess the stabilization performance of transfacet pedicle screw fixation.

Objective. To compare the biomechanical effects of short-term and long-term cyclic loading on lumbar motion segments instrumented with either a pedicle screw or a transfacet pedicle screw construct.

Summary of Background Data. Facet screw fixation is an alternative to pedicle screw fixation that permits the use of a minimally invasive strategy. It is not known whether facet screw fixation can provide stability equivalent to pedicle screw fixation during cyclical loading. Therefore, transfacet pedicle screw fixation and standard pedicle screw fixation techniques were compared biomechanically.

Methods. Lumbar motion segments were tested under short-term and long-term cyclic loading conditions. For the short-term phase, specimens were tested intact for six cycles (to 400 N or 4 Nm) in compression, flexion, extension, lateral bending, and torsion. The specimens then were instrumented with bilateral semicircular interbody spacers and pedicle screw instrumentation or transfacet pedicle screws, and the testing sequence was repeated. For the long-term phase, 12 specimens were instrumented in a similar manner and loaded to 6 Nm of flexion bending for 180,000 cycles.

Results. For the short-term phase, both fixation systems had significantly greater stiffness and reduced range of motion, as compared with the intact state. No differences were observed between the fixation systems except in flexion, wherein transfacet pedicle screw specimens were significantly stiffer than traditional pedicle screw specimens. For the long-term phase, the stiffness and range of motion did not significantly increase or decrease over repetitive cycling of the instrumented specimens. Furthermore, no significant difference between the fixation systems was observed.

Conclusions. The stability provided by both transfacet pedicle screw fixation and traditional pedicle screw fixation was not compromised after repetitive cycling. In this model, transfacet pedicle screw fixation appears equivalent biomechanically to traditional pedicle screw fixation.

[Key words: cyclic fatigue, pedicle screw fixation, spinal stabilization, transfacet pedicle screw fixation, translaminar facet screw fixation] **Spine 2003;28:1226–1234**

The use of dorsal lumbar rigid or semirigid (constrained) fixation for stabilization of the spine, with the explicit intent of increasing the chance of acquiring a solid arthrodesis, is becoming increasingly popular. The added advantage of providing a degree of stiffness that immediately diminishes mechanical back pain is, in most circumstances, of theoretical incidental, but significant, clinical value. Pedicle screw fixation has been the gold standard for such stabilization since the late 1980s. Its longevity as a surgical technique makes a vividly clear statement regarding its efficacy. However, its safety, facility of insertion, and morbidity have been questioned since the early years of its use. The chance of screw malposition, with the potential risk of neurologic and vascular injury and suboptimal fixation, has been clearly and consistently reported in the literature.^{1–12} In addition, the often excessive lateral paraspinal muscle retraction required to place pedicle screws adequately has been shown to injure these muscles functionally and increase the volume of devitalized tissue, which can lead to an increased incidence of infection and other related complications.^{1–4,7,9–12}

The use of a less invasive and equally efficacious method of fixation would clearly be of interest to surgeons. Facet screw fixation has been used by a select few surgeons over the past several decades.^{13–26} Two facet screw fixation techniques have been described, although the terminology referring to these techniques has not been clearly defined. Boucher¹⁴ described the “true transfacet” technique in 1959, and Magerl²² described the “translaminar transfacet” technique in 1984. The translaminar transfacet technique was thought by most surgeons to be of greater biomechanical efficacy, and therefore it achieved the greatest popularity. This is despite the fact that it is more technically demanding and arguably more dangerous than the true transfacet technique. The danger of the latter results from the requisite long passage of crossing screws through the lamina before they traverse the facet joint.

Both facet screw fixation techniques have been considered inferior biomechanically to pedicle screw fixation strategies. In addition, they have been thought prone to failure at the screw–bone interface during repetitive

From the Cleveland Clinic Foundation, Spine Research Laboratory, Cleveland, Ohio.

Supported in part by a grant from NuVasive, San Diego, California. Acknowledgment date: February 21, 2002. First revision date: May 31, 2002. Second revision date: July 23, 2002. Acceptance date: November 14, 2002.

The devices and drugs are approved by the FDA or by a corresponding national agency for this indication.

Corporate and industry funds were received to support this work. No benefits in any form have been or will be received from a commercial party related directly or indirectly to the subject of this article.

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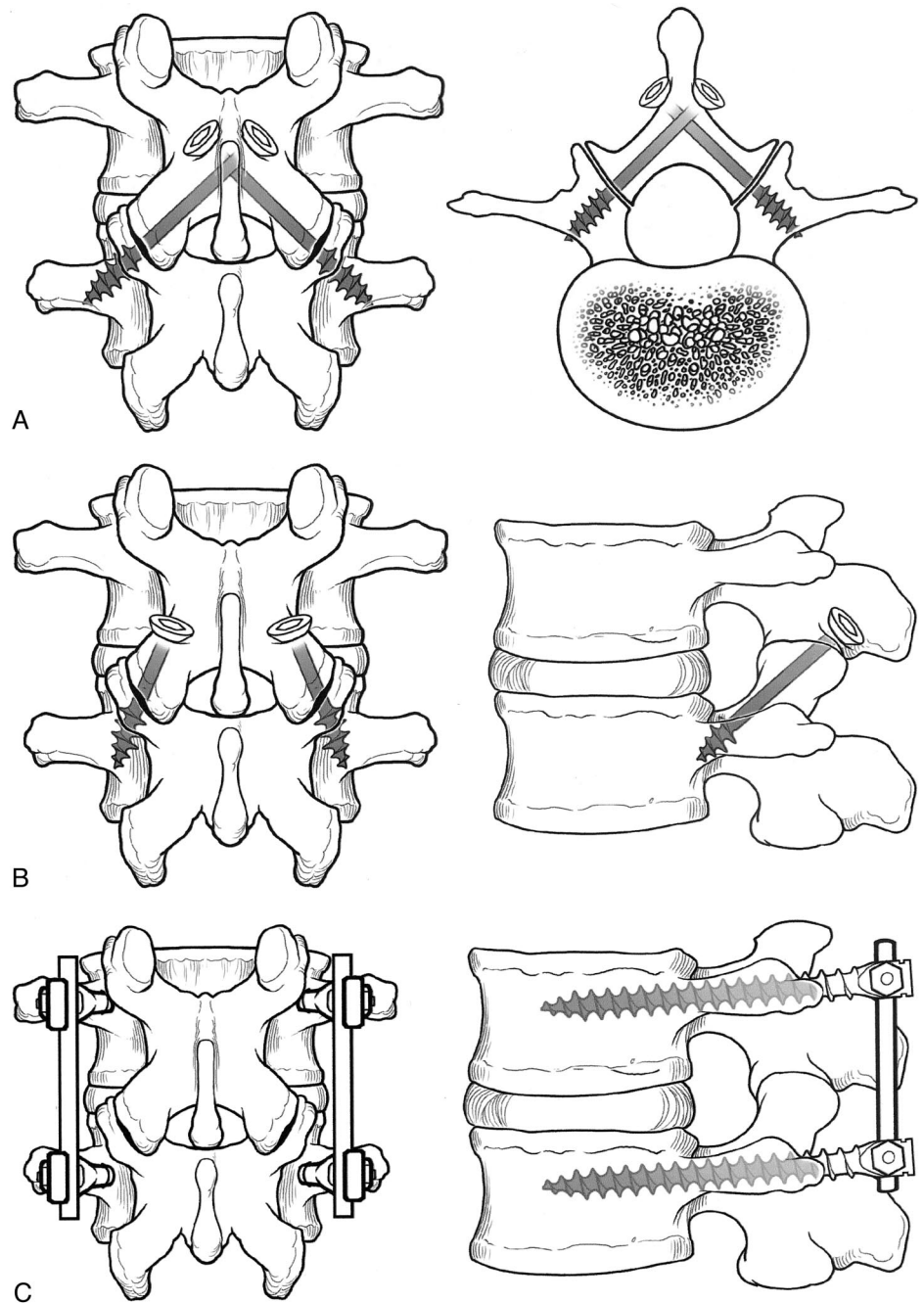


Figure 1. Three fixation techniques. **A**, Translaminar facet screw fixation (contralateral facet fixation from the site of insertion). **B**, Transfacet pedicle screw fixation (ipsilateral facet fixation from the site of insertion). **C**, Traditional pedicle screw fixation.

cycling, yet insufficient documentation exists to corroborate this assumption. These and related factors have diminished the enthusiasm of many surgeons for the use of facet screw fixation strategies.

Finally, it has been assumed by most surgeons that less than optimally rigid fixation strategies (*e.g.*, the perception regarding facet screw fixation) are not adequate for the acquisition of an arthrodesis unless an interbody buttressing device, such as a strut or cage, is placed.^{17,27-31}

To assess the aforementioned issues, a biomechanical laboratory study is presented that directly addresses the relative efficacy of the facet screw and pedicle screw fixation techniques, the need for an interbody buttress, and the effect of cycling on the screw–bone interface integrity

for both pedicle screw and facet screw fixation techniques. For the purpose of this study, the following terms are used to describe fixation techniques: translaminar facet screw fixation (contralateral facet fixation from the site of insertion; Figure 1A), transfacet pedicle screw fixation (ipsilateral facet fixation from the site of insertion; Figure 1B), and traditional pedicle screw fixation (Figure 1C). The aim of this biomechanical study was to compare the effects of nondestructive cyclical loading on cadaveric lumbar motion segments buttressed by interbody plastic spacers and instrumented using transfacet pedicle screw fixation with the effects of the gold standard, traditional pedicle screw fixation. Translaminar facet screw fixation was not assessed in the current study.

Table 1. Cadaveric Specimen Histories for the Short-Term and Long-Term Phases

Specimen	Levels	Age (yrs)	Sex	Cause of Death	BMD (g/cm ²)
Short-term specimens					
37632	L1–L2	66	M	Hypertensive, cerebrovascular disease	0.94
	L3–L4				1.00
39197	L1–L2	39	M	Respiratory failure caused by C1–C2 dislocation	0.89
	L3–L4				1.01
37676	L1–L2	48	F	Congestive heart failure	0.81
	L3–L4				0.89
39167	L1–L2	55	F	Respiratory failure with pneumonia	0.89
	L3–L4				0.95
39292	L1–L2	62	M	Cardiac arrest	0.95
38446	L1–L2	47	M	Craniocerebral trauma contact	1.07
37786	L1–L2	55	F	Cardiopulmonary arrest	0.80
	L3–L4				0.95
39151	L1–L2	52	M	Hypoxic cardiopulmonary arrest	1.29
38355	L1–L2	48	F	Adult respiratory distress syndrome	0.75
	L3–L4				0.82
37858	L1–L2	59	M	Acute hemopericardium	0.91
	L3–L4				1.00
37677	L1–L2	52	M	Functional impairment of prosthetic aortic valve	0.96
Long-term specimens					
37688	L1–L2	49	F	Myocardial infarction	0.90
	L3–L4				1.00
38352	L1–L2	44	F	Cardiac arrest	1.20
	L3–L4				1.31
41211	L1–L2	51	M	Metastatic carcinoma of the colon	0.84
	L3–L4				0.84
41213	L1–L2	56	M	Primary pruritorial mesothelima	1.10
	L3–L4				1.14
41235	L1–L2	44	F	Cardiac arrest	0.81
	L3–L4				0.92
41215	L1–L2	48	M	Liver cirrhosis	0.86
	L3–L4				1.02
41231	L1–L2	52	M	Ventricular fibrillation	0.74
	L3–L4				0.79
41203	L1–L2	60	M	Cardiac arrest	1.18
	L3–L4				1.24

BMD = bone mineral density.

Materials and Methods

Specimen Preparation. This study was conducted in two phases: a short-term phase that involved 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 (L1–L5) were obtained. The bone mineral density (BMD) for each spine was determined *via* dual-energy x-ray absorptiometry (DEXA; Hologic QDR 4500A, Waltham, MA). Specimens with a *t* score less than -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 ± 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 0.99 ± 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 dissected into L1–L2 and L3–L4 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 embedded in customized gripping fixtures using a polyester resin (Bondo/Mar-Hyde, Atlanta, GA). Wood screws were placed in the upper and lower vertebra of each motion segment in a multiplanar fashion to secure the motion segment in the embedding material. The upper and lower vertebrae were

embedded into the polyester resin to their midbodies. 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 transfacet pedicle screw fixation construct (NuVasive, San Diego, CA) by a spine fellowship-trained spine surgeon. The facet joints were not ablated or disrupted 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 transfacet pedicle screw instrumentation). All 12 FSUs had bilaterally placed interbody spacers.

Interbody Spacer Technique. The interbody spacers used were plastic replicas of femoral ring segments (NuVasive). Appropriately sized lateral anulotomies, ranging from 10 to 15 mm in height, were created for bilateral graft insertion, 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 distractor 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) then was inserted from the opposing side using a distracting tool in a transverse fashion until positioned properly. The spacer then was rotated 90° so that the saw-toothed edges were orthogonal to the axis of loading and resting against the endplates to resist translation and expulsion of the graft during loading. The distractor was removed, and the second spacer was inserted in a similar manner.

Transfacet Pedicle Screw Insertion. Initially, a small notch was made in the middle of the articular facet's cortical surface for the reception of 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 caudal edge of the lamina at an angle of 45°, and passed through the two facets of the 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 facet complex 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 a titanium alloy (Ti6Al4V). The screws were placed medially from the inferior facet and directed laterally toward the superior facet (Figure 1B).¹⁴ The insertional torques were recorded for the last three or four screw turns using a torque wrench calibrated to ±3% (Sturtevant Richmond, Franklin, IL).

Pedicle Screw Insertion. The Texas Scottish Rite Hospital pedicle screw system (TSRH; 6.5 × 40 mm; Medtronic Sofamor Danek) was used for dorsal fixation of the motion segments. An awl initially created the screw holes, and pedicle screws were inserted to 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 Magerl.²² Care was taken to ensure that the screw was inserted parallel to the axis of the pedicle and the pedicle cortex, and breaching of the ventral cortex was avoided. Pretapping and drilling of the pedicle screw hole was not performed to increase screw purchase within the bone. Titanium rods (diameter, 6.3 mm) were locked into place without the use of cross-fixation (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 Alliance RT/10, MTS Corporation, Eden Prairie, MN). All 18 of the FSUs were initially tested intact. After intact testing, each FSU underwent a controlled discectomy with spacer placement, accompanied by the application of dorsal instrumentation according to the procedure previously described. Nine motion segments (L1–L2 and L3–L4) were instrumented randomly with traditional pedicle screw fixation, and the remaining nine were instrumented with transfacet pedicle screw fixation systems.

Each FSU was nondestructively tested in five sequential modes: compression, flexion, extension, left lateral bending, and torsion. For the first four modes, each specimen was mounted into custom testing fixtures that allowed free rotation in the sagittal plane (*i.e.*, both the upper and lower fixtures were free to rotate). The rotation was recorded by rotational potentiometers. Initially, the center of rotation for each FSU

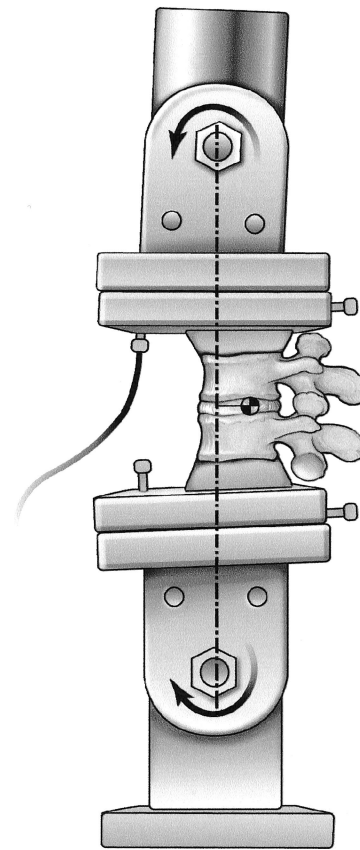


Figure 2. Experimental test setup on the material testing machine for the short-term phase. The lumbar motion segment was housed in gripping fixtures anchored to the testing system. Both the upper and lower fixtures were permitted to rotate about the sagittal plane, and rotational potentiometers recorded angular variations.

was established by applying a 400-N compressive load to the upper jig. The specimen was realigned, and the load was reapplied until no angular motion was detected by the potentiometers, indicating a centered specimen. Once the center of rotation was determined, it was clearly marked on the fixtures for each specimen tested. The specimens then were tested in pure compression with the application of an axial load at an actuator rate of 0.25 cm/min. The maximum load limit was set 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 Testworks 4 and a DATAQ data acquisition system (DATAQ Instruments, Akron, OH) on a Compaq Deskpro EN Series PC.

For flexion and extension testing, a compressive load was placed 1 cm ventrally or dorsally (for flexion or extension) to 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 at approximately the center of rotation. For left lateral bending, an axial compressive load was applied 1 cm left of the center of rotation at a rate of 0.25 cm/min. In flexion, extension, and left lateral bending, each specimen was preconditioned for three cycles. Data then were sampled for an additional three cycles following the testing scheme used in compression.

For torsion testing, a servohydraulically driven biaxial In-

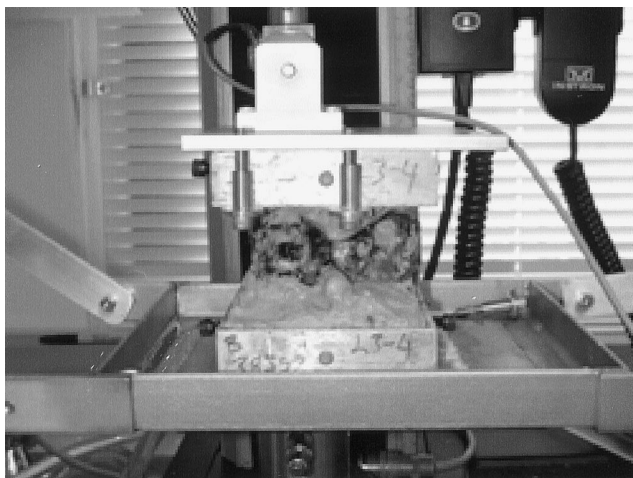


Figure 3. Experimental test setup on the material testing machine for the long-term phase. The lumbar motion segment was rigidly affixed to the inferior platform of the testing system. The upper fixture was permitted to rotate about the sagittal plane, and a rotational potentiometer recorded angular variations.

stron materials testing system was used (Instron 8874; Instron, Canton, MA). A 25-N axial compressive load was applied to each specimen to simulate the weight of the upper torso. Under torque control, a ± 5 -Nm force was applied to each specimen bidirectionally. Each specimen ran for a total of seven cycles to allow three complete preconditioning cycles and four sampled cycles. Continuous data were sampled at a rate of 20 Hz using Fastrack Advanced Fatigue Software (Instron).

Long-Term Phase. Long-term cyclical loading of the instrumented spines was necessary to examine the mechanical behavior at the bone-metal interface for more realistic periods of bone healing. Therefore, to simulate approximately 6 weeks of normal daily spinal loading that occurs *in vivo*, 180,000 cycles were chosen as an optimal period for *in vitro* testing.³² Clinically, incorporation and integration of interbody bone graft usually is initiated within the first 2 weeks after surgery. Hence, a 6-week time frame, as replicated in this study, represents a study time frame that clearly provides unique and clinically meaningful information. Theoretically, if a fusion has not begun to incorporate within this time frame, excessive stress and strain will be placed on the spinal instrumentation and bone at the implant-bone interface, thereby increasing the risk of early construct failure. The goal of the long-term cyclical portion of this study, therefore, was to evaluate the ability of the study implants to limit motion effectively during this critical time frame for healing.

All long-term cyclical testing was conducted on a servohydraulically driven uniaxial materials testing apparatus (Instron DynaMight 8841; Instron). Each motion segment was secured to the Instron machine before testing (Figure 3). The superior jig, housing each specimen, was rotated during loading, and the angles were recorded using a rotational potentiometer. The inferior jig was clamped to the test platen and not allowed to rotate. The specimen's center of rotation was established by applying a 500-N maximum compressive load to the upper jig in a fashion similar to that described for the short-term phase. The specimen was realigned, and the load was reapplied until no angular motion was detected by the potentiometer, establishing the center of rotation.

Twelve FSUs were randomly instrumented: six with a pedicle screw construct and two bilateral plastic semicircular interbody spacers, and six with transfacet pedicle screws and bilateral interbody spacers. All the specimens were shifted dorsally 1.5 cm from the center of rotation to create a clinically and physiologically significant flexion and bending moment. Each FSU was loaded repetitively in flexion to 6 Nm for 180,000 cycles at a frequency of 1Hz. An automatic misting device housed each FSU and the accompanying gripping fixtures within a water-resistant chamber. Misting nozzles were programmed to moisten the specimen at 5-minute intervals during testing.

Statistical Analysis. For all testing modes, the stiffness of each FSU was calculated from a tangent line fit to the load-displacement data in the elastic region of the curve. Compressive stiffness was measured as the direct slope of the tangential line taken from the load-displacement curves. Flexion, extension, and lateral bending stiffnesses were calculated from the applied bending moment *versus* range of motion curves by quantifying the slope of the tangential 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 FSU 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 test phase, an ANOVA with a Newman-Keuls comparison was used to detect any statistical differences in stiffness and range of motion over 180,000 cycles within, between, and among specimens. In addition, a two-tailed, unpaired *t* test was performed to detect differences in mean stiffness between transfacet pedicle screw specimens and 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 intact mode ($P < 0.01$). Additionally, in flexion, transfacet pedicle screw specimens were significantly stiffer (30%)

Table 2. Short-Term Phase Stiffness Data

Test Mode	Intact	Transfacet Screw Fixation	Pedicle Screw Fixation
Compression (N/mm)	945.4 \pm 224.2	1,103.8 \pm 294.4	1,054.5 \pm 167.3
Flexion (Nm/°)	0.9 \pm 0.6	4.7 \pm 2.5	3.3 \pm 1.1
Extension (Nm/°)	1.4 \pm 0.6	4.5 \pm 1.7	3.6 \pm 0.9
Lateral bending (Nm/°)	0.9 \pm 0.5	2.1 \pm 1.2	2.8 \pm 1.3
CCW torsion (Nm/°)	2.4 \pm 1.2	2.9 \pm 1.4	3.7 \pm 1.9
CW torsion (Nm/°)	2.5 \pm 1.1	3.1 \pm 1.2	3.7 \pm 1.8

The mean \pm standard deviation stiffness for each testing mode for the intact, transfacet screw fixation, and pedicle screw fixation systems are given. CCW = counter-clockwise; CW = clockwise.

Table 3. Short-Term Phase Range of Motion Data

Test Mode	Intact	Transfacet Screw Fixation	Pedicle Screw Fixation
Flexion	5.5 ± 3.5	0.8 ± 0.4	1.0 ± 0.4
Extension	2.7 ± 1.6	0.7 ± 0.3	0.9 ± 0.2
Lateral bending	4.6 ± 2.9	1.8 ± 1.0	1.4 ± 0.8
CCW torsion	2.6 ± 1.5	2.0 ± 0.8	1.8 ± 0.9
CW torsion	2.4 ± 1.8	1.7 ± 0.6	1.6 ± 0.7

The mean ± standard deviation range of motion (in degrees) for each testing mode for the intact, transfacet screw fixation, and pedicle screw fixation systems are given.

CCW = counter-clockwise; CW = clockwise.

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 ($P < 0.01$). However, 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 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 plastic spacers ($P > 0.05$).

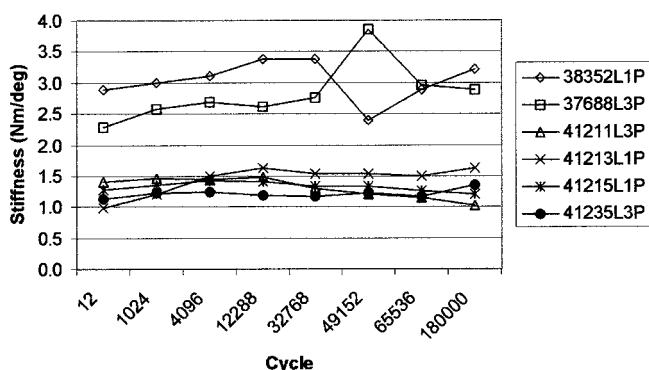


Figure 4. Stiffness variations of long-term specimens instrumented with pedicle screw systems and interbody grafts for 180,000 loading cycles (long-term phase).

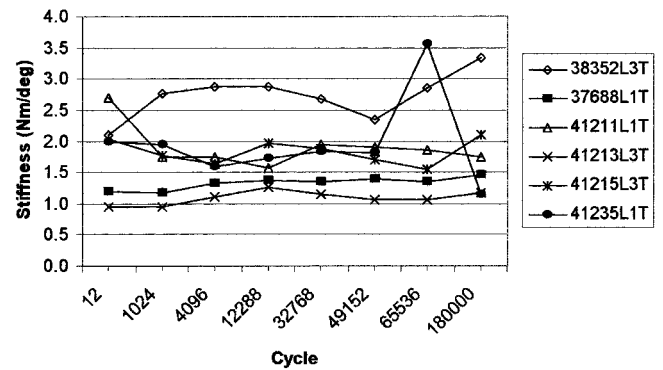


Figure 5. Stiffness variations of long-term specimens instrumented with transfacet screws and interbody grafts for 180,000 loading cycles (long-term phase).

The range of motion did not significantly change over time for any of the specimens, regardless of the type of fixation used. The range of motion was between 1° and 3° for all of the specimens (Table 4). No significant differences were noted between transfacet pedicle screw fixation and traditional pedicle screw fixation ($P > 0.05$).

Discussion

Traditional Pedicle Screw Fixation

Traditional pedicle screw fixation is used widely 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.³³⁻³⁶ Conversely, 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 less invasive alternatives for dorsal spinal integrity augmentation across a fusion site.^{14,21,22} Reich *et al*²⁵ successfully demonstrated the clinical success of dorso-

Table 4. Long-Term Phase Range of Motion Data

Test Mode	Range of Motion
Pedicle screw fixation with interbody grafts	1.9 ± 0.7
Transfacet screw fixation with interbody grafts	1.9 ± 0.6

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

lateral fusion with translaminar facet screw fixation in the lumbar spine. Although translaminar 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.^{13,15–17,19,22,23,25,31} Biomechanically, similar studies have found no significant performance differences among translaminar facet fixation, transfacet pedicle fixation, and pedicle screw fixation systems for dorsolateral fusion.^{31,37} High fusion rates that range from 90% to 98% have been observed with transfacet pedicle and translaminar facet screw fixation. This is comparable with that of pedicle screw fixation.^{1,16,17,19,21,25} 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 lag 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 translaminar facet screw fixation with that of pedicle screw fixation.^{37–39} It is emphasized that facet engagement *via* the aforementioned lag effect may have preloaded 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 flexion during the short-term phase. The favorable performance of facet screws in flexion has been observed in other studies.^{38,40} Long-term biomechanical decay of structural integrity was not observed at the bone–metal 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.³² The extended loading period represents an appropriate laboratory scenario for the modeling of the spine during the initiation of the fusion process.^{17,27–31}

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 must be emphasized that during the clinical application of pedicle 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

failure may be preceded by a stiffened state caused by the compressed fragments.^{41,42} 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 *in vitro* transfacet pedicle screw fixation provides a degree of biomechanical stability equivalent 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 (transfacet pedicle screw and pedicle screw fixation) when used with an interbody spacer. Facet engagement, using a lag screw, appears to augment stability. The stability conferred by 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 pedicle screw fixation under short-term cyclic loading conditions when interbody spacers were used.
- In flexion, transfacet pedicle screw fixation was stiffer than pedicle screw fixation after six loading cycles for the interbody load-sharing model used.
- Under long-term cyclic loading conditions, both transfacet pedicle screw fixation and pedicle screw fixation provided similar stability. Stability did not decrease for either group 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 design may provide adequate stability during the period required for a successful fusion to occur.

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