J Neurosurg (Spine 1) 94:51–60, 2001

Biomechanics of Grade I degenerative lumbar spondylolisthesis. Part 2: Treatment with threaded interbody cages/dowels and pedicle screws

SEDAT ÇAGLI, M.D., NEIL R. CRAWFORD, PH.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 determine the biomechanical effectiveness of threaded interbody cages or dowels compared with that achieved using pedicle screw instrumentation in resisting Grade I lumbar spine degenerative spondylolisthesis.

Methods. Thirty-three levels obtained from seven cadaveric lumbar spines were instrumented with cages or dowels, pedicle screw/rod instrumentation, or both. Entire specimens were loaded with nonconstraining torques. Each level was loaded with anteroposterior shear forces while an optical system was used to measure the specimen's motion at individual levels.

Pedicle screw/rods outperformed interbody cages and dowels in treating spondylolisthesis. Cages or dowels alone provided only moderate biomechanical stability, and their effectiveness depended heavily on the integrity of the ligaments and remaining annulus, whereas the success of pedicle screw fixation relied predominantly on the integrity of the bone for solid fixation. Little biomechanical difference was demonstrated between cages and dowels; both devices were susceptible to loosening with cyclic fatigue.

Conclusions. Biomechanically, cages or dowels alone were suboptimal for treating lumbar spondylolisthesis, especially compared with pedicle screw/rods. Threaded cages or dowels used together with pedicle screws/rods created the most stable construct.

KEY WORDS • degenerative spondylolisthesis • anterior lumbar interbody fusion • threaded interbody cages • biomechanics

SURGICAL treatment of Grade I lumbar spondylolisthesis (25% slippage) involves reducing the subluxation and inserting fixation devices and bone grafts to stabilize the spine and promote fusion. Two surgical options are threaded interbody cylinders[9] and pedicle screw/rod fixation.[3,12] Threaded interbody cylinders are available as metallic cages or as machined cadaveric cortical bone dowels (Fig. 1). The geometrical configurations of threaded titanium cages and threaded cortical bone dowels are similar, although the metallic cages have sharper threads. To our knowledge, the devices have not been compared biomechanically. Biomechanically, pedicle screw/rod fixation enhances the immediate mechanical strength of a threaded interbody cage system,[4,5] but the effects of pedicle screw fixation have not been compared with those of interbody threaded cylinder fixation for treating spondylolisthesis. In this study we therefore compared the relative biomechanical stability offered by threaded bone dowels, threaded titanium cages, and pedicle screw/rod fixation for the treatment of Grade I lumbar spondylolisthesis by using an in vitro model of spondylolisthesis.[1]

Abbreviations used in this paper: AP = anteroposterior; BMD = bone marrow density; EZ = elastic zone; NZ = neutral zone; ROM = range of motion; VB = vertebral body.

Materials and Methods

Lumbar Specimens

Thirty-three motion segments in seven human cadaveric lumbar spine specimens were studied (Table 1). The mean age of patients in whom the specimens were obtained after death was 52 years (range 23–66 years). Specimens were prepared for testing and handled as described in our companion article.[1]

Biomechanical Testing

Biomechanical flexibility was tested as described.[1] Briefly, potted specimens were loaded nondestructively with pure moments (maximum 5 Nm) to cause uniform bending or twisting of entire specimens, or they were loaded with AP shear forces (maximum 50 N) applied individually at each level to induce spondylolisthesis and retrolisthesis. Motion was measured optoelectronically.

Specimens were tested in the following conditions (Steps 1–6): 1) intact; 2) after induction of Grade I spondylolisthesis at all levels; 3) with pedicle screw/rod hardware; 4) with cages or dowels

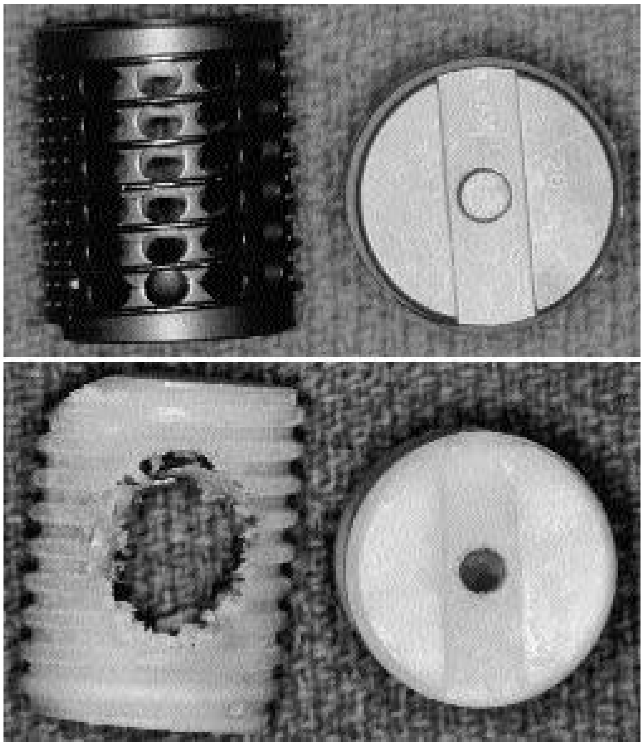

FIG. 1. Photographs of threaded cylindrical interbody devices. *Upper:* Titanium Novus LC cages are hollow cylinders with circumferential holes to allow bone ingrowth. *Lower:* Cortical bone dowels are solid cylinders with a single, large transverse hole drilled to allow bone ingrowth.

alone; 5) with pedicle screw/rod hardware combined with cages or dowels; and 6) with cages or dowels alone after 6000 cycles of fatigue (0.5–2 Hz, 0–5 Nm). Because pedicle screw/rod hardware was used across single vertebral levels only, the order of testing in Steps 2 and 3 or in Steps 4 and 5 depended on the positions of the rods. For example, a specimen might first be tested, after destabilization, with rods spanning L1–2, L3–4, and L5–sacrum. Hence, from a single test, data obtained for these fixed levels would correspond to Step 3, whereas data recorded for L2–3 and L4–5 would correspond to Step 2. The inverse would be true when rods were repositioned in a subsequent test to span L2–3 and L4–5. Fatigue loading (Step 6) consisted of 2000 cycles each of left axial rotation, right axial rotation, and extension applied at approximately 1 Hz between 0 Nm and 5 Nm. These loading modes were expected to be the most strenuous and the most likely to expel the cage or dowel.

After all testing was completed, all hardware was removed, and the BMD of the VBs was determined by obtaining dual-energy x-ray absorptiometry scans (Lunar DPX-L; Lunar Corp., Madison, WI) of laterally oriented specimens embedded in rice and water.

TABLE 1

Summary of clinical data obtained in cadaveric specimens

Specimen No.	Vertebral Levels	Age (yrs) at Death, Sex	Cause of Death
1	T12–sacrum	59, M	heart disease
2	T12–sacrum	55, F	subarachnoid hemorrhage
3	T12–sacrum	55, M	heart disease
4	T12–L4	66, M	cirrhosis
5	T12–sacrum	52, M	brain tumor
6	T12–sacrum	54, M	cerebrovascular accident
7	T12–sacrum	23, M	motor vehicle accident

TABLE 2

*Spondylolisthesis and BMD at 33 vertebral levels**

Specimen & Level	VB Width (mm)	Shear Instability (mm)	Slippage (%)	BMD (g/cm²)
Specimen 1				
L1–2	44	3.3	7.4	0.69
L2–3	44	5.0	11.3	0.77
L3–4	43	5.0	11.7	0.86
L4–5	44	4.1	9.3	0.99
L5–sacrum	40	10.2	25.4	1.07
Specimen 2				
L1–2	39	2.4	6.0	0.45
L2–3	39	2.6	6.6	0.49
L3–4	39	3.0	7.7	0.62
L4–5	39	4.5	11.4	0.69
L5–sacrum	34	2.1	6.1	0.66
Specimen 3				
L1–2	41	2.5	6.1	0.53
L2–3	40	2.6	6.4	0.52
L3–4	43	4.4	10.2	0.55
L4–5	43	4.9	11.3	0.66
L5–sacrum	45	3.4	7.5	0.71
Specimen 4				
L1–2	38	1.2	3.2	0.30
L2–3	39	2.9	7.5	0.34
L3–4	35	2.0	5.7	0.42
Specimen 5				
L1–2	39	3.3	8.4	0.51
L2–3	40	3.2	8.1	0.57
L3–4	39	3.7	9.6	0.61
L4–5	41	3.6	8.9	0.70
L5–sacrum	40	3.4	8.6	0.76
Specimen 6				
L1–2	40	5.8	14.5	0.89
L2–3	40	4.9	12.3	0.90
L3–4	38	6.2	16.2	1.00
L4–5	37	5.2	14.1	1.17
L5–sacrum	38	5.0	13.2	1.29
Specimen 7				
L1–2	35	2.7	7.7	0.95
L2–3	35	2.5	7.1	0.96
L3–4	35	3.3	9.3	0.98
L4–5	33	2.8	8.5	1.00
L5–sacrum	34	2.3	6.6	1.01

*The overall means $\pm$ standard deviations were 39.2 $\pm$ 3, 3.8 $\pm$ 1.7, 9.5 $\pm$ 4.1, and 0.75 $\pm$ 0.25 for VB width, shear instability, slippage, and BMD, respectively.

Instrumentation and Its Application

The Texas Scottish Rite Hospital pedicle screw and rod system (Sofamor Danek, Memphis, TN) was used for posterior fixation and was placed according to standard techniques.[2] Stainless-steel 6.35-mm-diameter rods for connecting pedicle screws were cut to the appropriate length for single-level fixation (4–5 cm) and attached to the pedicle screw ends by using the swiveling connectors. Immediately before testing, a torque wrench was used to verify that the torque applied to the connectors was at the manufacturer's recommended level of 12 Nm.

Self-tapping, titanium, threaded, cylindrical interbody cages (Novus LC cages; Sofamor Danek) and nonself-tapping, cylindrical, threaded, cortical allograft bone dowels (University of Florida Tissue Bank, Inc., Alachua, FL) were used for anterior interbody fixation (Fig. 1). Diameters of 16 mm, 18 mm, and 20 mm were available in lengths from 23 to 29 mm.

Appropriate cage diameters and lengths were selected by studying radiographs and directly measuring the anatomical dimensions. A cage or dowel diameter was chosen to penetrate the vertebral endplate but not to compromise the remaining lateral annulus fibrosus.

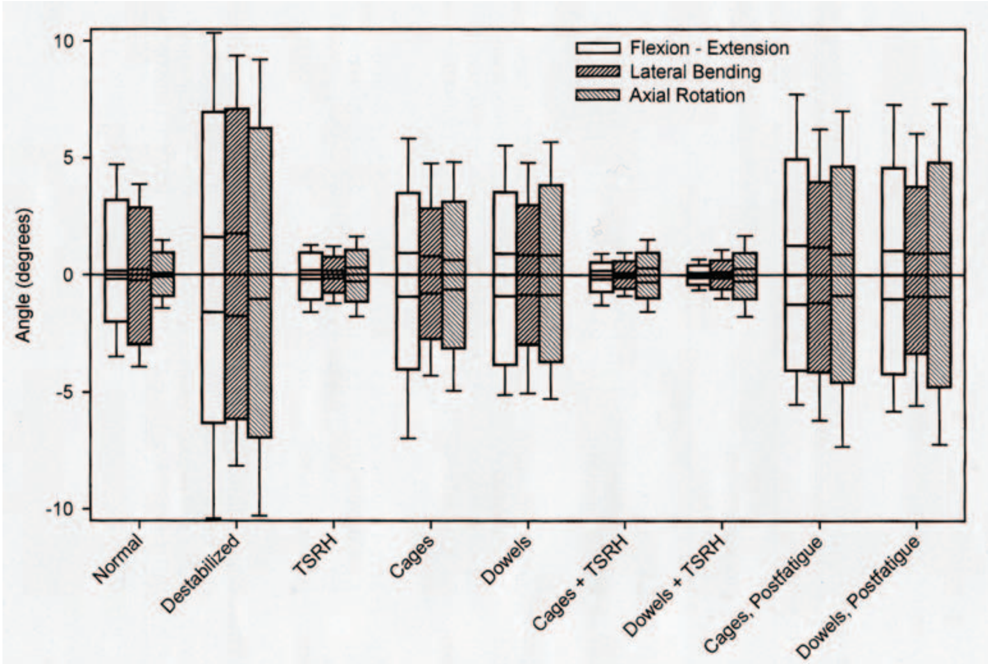

FIG. 2. Graph showing the mean angular motion in each condition studied. Flexion, left lateral bending, and left axial rotation are represented as positive motions, and extension, right lateral bending, and right axial rotation are shown as negative. *Horizontal lines* on each bar partition the NZ from the EZ. *Error bars* show standard deviation of the ROM. TSRH = Texas Scottish Rite Hospital pedicle screw system. Used with permission from Williams & Wilkins.

Cage length was chosen to span 70 to 80% of the VB's AP dimension. Sometimes the largest disc-space distractor for the chosen cage diameter was inadequate to distract the disc space to a normal height and to apply sufficient tension to the ligaments and lateral annulus. In such cases, the next sized larger cage diameter was used. A satisfactory cage size fit was found in all cases. Overall, 19% of the fixated levels required 16-mm-diameter inserts, 39% required 18-mm devices, and 42% required 20-mm devices.

Two parallel cages or threaded dowels were inserted in each level from L1–2 to L5–sacrum, alternating between cages or dowels from level to level. All cages or dowels were inserted with their long axes oriented anteroposteriorly by using a double-barreled guide tube. This method facilitates the symmetrical placement of cages or dowels. Before a cage or dowel was inserted, a small window was cut with a scalpel in the anterior longitudinal ligament through which a discectomy would normally be performed. Discectomy, however, was performed as part of the destabilization procedure,[1] and only some residual medial annulus needed to be removed with a curette and pituitary rongeurs. The anterior window extended laterally just far enough to accommodate the cages or dowels.

At this point in the procedure, the destabilized specimens devoid of surrounding musculature were extremely mobile, which made reaming the pilot holes in the vertebral endplates difficult. Therefore, after the disc space distractors and guide tube were inserted, pedicle screw/rod hardware was temporarily and loosely engaged at the level at which cages or dowels were to be implanted. The pedicle screw/rod hardware prevented movement of the vertebrae relative to each other, allowing precise reaming. Although this procedure is not used during surgery, it was considered to better approximate the stability provided by surrounding tissues in patients than the uneven reaming that would have resulted from unrestricted movement. Because the pedicle screw hardware was locked down after the distractors were inserted, the final orientation of a cage or dowel should have been unaffected by this procedure.

Holes were tapped before bone dowels were inserted but not before cages, which were self-tapping, were inserted. The cages or dowels were screwed into place until the ventral end was recessed approximately 1 mm below the anterior margin of the VB.

Data Analysis

From the raw data, the ROM, NZ, EZ, and flexibility coefficient were calculated.[1] The reduced data were analyzed using paired Student's t-tests to determine whether each progressive step in destabilization or fixation procedures significantly increased or decreased motion. One-tailed t-tests were used when the directionality of the motion change could be predicted (for example, it could be predicted that motion should decrease as hardware was attached to the unstable spine). Otherwise, two-tailed t-tests were used (that is, to compare data with those obtained in the intact condition). Nonpaired t-tests were used to compare destabilized or instrumented conditions in independent groups of specimens. In all cases, probability values less than 0.05 were considered significant. Increases in the NZ, EZ, ROM, and flexibility values indicated increased instability in the fixated spine. In addition to these comparisons, BMD in relation to NZ and ROM was correlated using the Pearson product moment correlations to determine the extent to which the fixation quality of the different devices was affected by bone quality before and after fatigue.

Results

Anteroposterior Slippage

After destabilization, the mean amount of AP shear instability or spondylo-/retrolisthesis was 3.8 mm, ranging from 1.2 to 10.2 mm (3.2–25.4%) of the AP width of the VB (Table 2).

Bone Mineral Density

On dual-energy x-ray absorptiometry scans, the BMD values of the specimens varied from 0.3 to 1.29 g/cm² (mean 0.75 g/cm²; Table 2). In specimens fitted with cages or dowels, lower BMD values correlated with larger ROM values during flexion–extension ($p = 0.041$), lateral bend-

TABLE 3

*Differences in angular motion compared with normal motion**

Angular Motion Parameter & Direction	Destabilized Spine	Pedicle Screw–Rod	Cage	Dowel	Cage + Pedicle Screw–Rods	Dowel + Pedicle Screw–Rod	Postfatigue Cage	Postfatigue Dowel
	Type of Fixation Device (p Value)							
NZ								
flexion–extension	2.8 ± 2.8 (0.0000)†	0.0 ± 0.4 (0.8572)	1.5 ± 1.2 (0.0004)†	1.5 ± 1.1 (0.0007)†	0.0 ± 0.4 (0.7344)	−0.1 ± 0.2 (0.1627)	2.3 ± 1.8 (0.0010)†	1.7 ± 1.0 (0.0002)†
lat bending	3.1 ± 4.1 (0.0001)†	−0.1 ± 0.4 (0.0935)	1.1 ± 1.3 (0.0056)†	1.2 ± 1.6 (0.0197)†	−0.3 ± 0.2 (0.0008)†	−0.2 ± 0.2 (0.0075)†	1.9 ± 2.4 (0.0201)†	1.4 ± 1.6 (0.0170)†
axial rotation	1.9 ± 4.5 (0.0196)†	0.4 ± 0.6 (0.0002)†	1.1 ± 0.8 (0.0001)†	1.5 ± 0.8 (0.0000)†	0.5 ± 0.4 (0.0006)†	0.4 ± 0.4 (0.0154)†	1.6 ± 1.2 (0.0008)†	1.7 ± 1.1 (0.0004)†
EZ								
flexion	2.3 ± 3.4 (0.0004)†	−2.2 ± 1.4 (0.0000)†	−0.2 ± 2.2 (0.7371)	−0.2 ± 1.9 (0.7021)	−2.4 ± 1.1 (0.0000)†	−2.5 ± 1.3 (0.0000)†	0.9 ± 2.7 (0.2696)	0.6 ± 2.7 (0.4809)
extension	2.9 ± 4.1 (0.0003)†	−0.9 ± 1.4 (0.0004)†	1.2 ± 1.3 (0.0027)†	1.6 ± 1.0 (0.0002)†	−1.3 ± 1.8 (0.0168)†	−1.1 ± 0.5 (0.0000)†	1.3 ± 1.2 (0.0030)†	1.7 ± 1.1 (0.0004)†
lat bending	2.2 ± 1.1 (0.0000)†	−2.1 ± 1.0 (0.0000)†	−0.9 ± 1.3 (0.0269)†	−0.6 ± 1.4 (0.1952)	−2.4 ± 0.8 (0.0000)†	−2.2 ± 1.0 (0.0000)†	−0.2 ± 1.7 (0.7498)	−0.1 ± 1.8 (0.7956)
axial rotation	4.7 ± 1.6 (0.0000)†	−0.1 ± 0.5 (0.5371)	1.8 ± 1.4 (0.0003)†	2.0 ± 1.2 (0.0001)†	−0.1 ± 0.4 (0.6096)	−0.2 ± 0.4 (0.0674)	3.1 ± 2.0 (0.0002)†	2.9 ± 1.7 (0.0002)†
ROM								
flexion	3.7 ± 3.3 (0.0000)†	−2.2 ± 1.5 (0.0000)†	0.6 ± 2.4 (0.3837)	0.5 ± 2.2 (0.4291)	−2.4 ± 1.2 (0.0000)†	−2.6 ± 1.4 (0.0000)†	2.1 ± 3.1 (0.0406)†	1.5 ± 2.7 (0.1070)
extension	4.3 ± 3.7 (0.0000)†	−0.9 ± 1.5 (0.0011)†	2.0 ± 1.5 (0.0003)†	2.3 ± 1.0 (0.0000)†	−1.3 ± 1.9 (0.0244)†	−1.1 ± 0.6 (0.0001)†	2.5 ± 1.2 (0.0000)†	2.6 ± 1.2 (0.0000)†
lat bending	3.7 ± 1.9 (0.0000)†	−2.1 ± 1.1 (0.0000)†	−0.3 ± 1.9 (0.5184)	0.0 ± 2.0 (0.9348)	−2.5 ± 0.9 (0.0000)†	−2.3 ± 1.0 (0.0000)†	0.8 ± 2.3 (0.2655)	0.5 ± 2.2 (0.4327)
axial rotation	5.7 ± 3.1 (0.0000)†	0.2 ± 0.7 (0.1555)	2.3 ± 1.7 (0.0002)†	2.7 ± 1.4 (0.0000)†	0.2 ± 0.5 (0.1667)	−0.1 ± 0.5 (0.6788)	3.9 ± 2.5 (0.0002)†	3.8 ± 2.1 (0.0001)†
Flexibility								
flexion	−0.20 ± 0.17 (0.0000)†	−0.31 ± 0.15 (0.0000)†	−0.14 ± 0.11 (0.0003)†	−0.14 ± 0.13 (0.0034)†	−0.36 ± 0.13 (0.0000)†	−0.38 ± 0.14 (0.0000)†	−0.15 ± 0.21 (0.0383)†	−0.19 ± 0.16 (0.0022)†
extension	−0.03 ± 0.12 (0.1039)	−0.09 ± 0.15 (0.0016)†	0.18 ± 0.20 (0.0038)†	0.27 ± 0.24 (0.0029)†	−0.14 ± 0.17 (0.0083)†	−0.15 ± 0.08 (0.0000)†	0.14 ± 0.17 (0.0138)†	0.14 ± 0.21 (0.0536)
lat bending	−0.03 ± 0.11 (0.0685)	−0.23 ± 0.12 (0.0000)†	−0.07 ± 0.11 (0.0305)†	−0.02 ± 0.14 (0.5978)	−0.27 ± 0.12 (0.0000)†	−0.25 ± 0.09 (0.0000)†	−0.04 ± 0.11 (0.2765)	−0.04 ± 0.12 (0.3106)
axial rotation	0.32 ± 0.11 (0.0000)†	0.04 ± 0.08 (0.0103)†	0.24 ± 0.12 (0.0000)†	0.25 ± 0.14 (0.0001)†	0.03 ± 0.08 (0.1455)	0.01 ± 0.09 (0.6725)	0.25 ± 0.11 (0.0000)†	0.22 ± 0.11 (0.0001)†

* All data are presented as the means ± standard deviations, and all data, except those for flexibility, are recorded in degrees/Newtonmeter.
† Statistically significant.

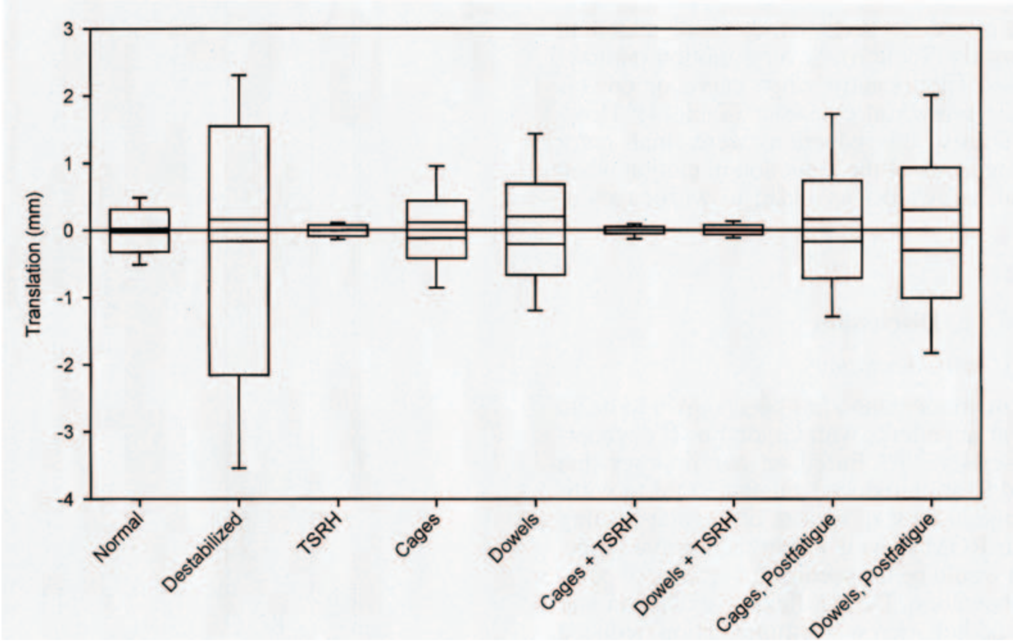

FIG. 3. Graph showing the mean shear ROM both anteriorly (positive) and posteriorly (negative) in each condition studied. *Horizontal lines* on each bar partition the NZ from the EZ. *Error bars* show standard deviation of the ROM. Used with permission from Williams & Wilkins.

ing ($p = 0.0002$), and axial rotation ($p = 0.008$) and with larger NZ values during lateral bending ($p = 0.004$). In specimens in which the pedicle screw system was placed, lower BMD values correlated significantly with higher ROM values (lateral bending, $p < 0.0001$; axial rotation, $p = 0.0005$) and with higher NZ measurements (lateral bending, $p = 0.002$; axial rotation, $p = 0.005$).

Instrumented Spine Compared With Intact Spine

Immediately after the surgical procedure, we deemed several levels instrumented with interbody cages or dowels to be nonrepresentative of a clinically adequate construct. The procedural failures were caused by visibly laterally torn ligaments and annulus due to overdistraction, asymmetrical reaming, or poorly controlled resection during the destabilization procedure. These problems mostly occurred in the initial specimens in which fixation was performed. The questionable data obtained in these levels were excluded from analysis, reducing the number of levels for comparison from 33 to 26 (cage levels, 14; dowel levels, 12).

The different methods of fixations did not always reduce angular motion to within normal ranges, especially during axial rotation (Fig. 2 and Table 3). During axial rotation, pedicle screw/rod hardware reduced the angular EZ to within the normal range but did not reduce the angular NZ. Pedicle screw/rod hardware, either alone or in combination with cages or dowels, achieved significantly better results than cages or dowels alone in reducing the angular and shear parameters to within normal limits. Although the use of cages or dowels alone reduced some angular parameters to within normal limits (Table 3), most angular parameters remained larger than normal. Cages or dowels alone reduced no shear parameter to within normal limits (Fig. 3 and Table 4).

Fatigue Testing

Fatigue of motion segments instrumented with cages or dowels alone significantly increased several angular and shear motion parameters in multiple directions of loading (Figs. 2 and 3; Table 5). With fatigue, the increases in shear parameters were more pronounced and more often significant than those in angular motion parameters (Table 5). During lateral bending, BMD values correlated with increases in angular NZ measurements ($p = 0.013$). There was no correlation with other modes of loading or with increases in ROM values ($p > 0.107$).

Cages Compared With Dowels

Few angular or shear parameters differed visibly when cages were compared with dowels, and none was significantly different when analyzed using the nonpaired t-tests (Figs. 2 and 3). The most noteworthy difference was in the angular and shear NZ values, both of which were smaller in those with cages than in motion segments fixated with bone dowels (Figs. 2 and 3). In all three directions of angular loading, the angular NZ value increased more with fatigue testing in levels fitted with cages than in those with dowels, reducing the difference in angular NZ measurement between the two groups (Table 5). However, the larger increase in NZ values in the cage-fitted specimens was not significant ($p = 0.18–0.29$). Before and after fatigue, the shear NZ value was slightly larger in the bone dowel–fitted specimens than those instrumented with the cage (not significant; minimum $p = 0.28$).

Effects of Additional Hardware

Data were also analyzed in terms of the effects of one hardware system being enhanced by the addition of the other. All angular and shear motion parameters decreased significantly when pedicle screws–rods were added to cages or bone dowels (Table 6). Several motion parameters also decreased significantly when cages or dowels were added to the screw/rod construct (Table 7). However, the magnitudes of the reductions were small compared with the magnitude of the reduction in motion when pedicle screw/rod hardware was added to with cages or dowels.

Discussion

Pedicle Screw/Rod Instrumentation

Pedicle screw instrumentation has been shown to be an effective treatment in patients with Grade I or II degenerative spondylolisthesis.[7,8,12,15] Based on our findings that pedicle screw/rod fixation reduced angular ROM to within the normal range in most directions of angular loading and reduced shear ROM to well within the normal range, successful fusion would be the predicted outcome of pedicle screw instrumentation. During flexion–extension and lateral bending, pedicle screw instrumentation reduced angular motion parameters to within normal limits more effectively than during axial rotation (Table 3). This finding partially reflects the typically smaller amount of axial rotation associated with the lumbar spine compared with flexion–extension or lateral bending. In axial rotation, the intact spine rotated by less than half the amount that was demonstrated during lateral bending or flexion–extension (Fig. 2). Thus, after destabilization, reducing axial rotation to within the normal range means reducing it by a considerably larger amount than that required for the other two directions of motion.

Threaded Interbody Instrumentation

Threaded interbody devices (metal cages or bone dowels) effectively restored the height of the disc space. Consequently, tension was again placed on the remaining intact ligaments and annulus, restoring some biomechanical stability. After placement of the interbody system, the main region where the ligaments were not intact was anteriorly, where the cages entered the disc space. The missing ligament (the anterior longitudinal ligament) provides important resistance against extension in all regions of the spine.[14] After some specimens were subjected to fatigue testing, mouthing adjacent to the cage– or dowel–vertebra interface occurred during extension. Accordingly, during extension and axial rotation, cages were least effective in bringing motion to within near the normal range (Table 3).

Our results compare well with those reported by Rathonyi, et al.,[11] who found that anterior threaded interbody cages stabilize the lumbar spine during flexion and lateral bending but not during extension or axial rotation. In contrast to our findings, Nibu, et al.,[10] found that during axial rotation anterior cages significantly reduced the ROM when compared with normal limits. However, they did not injure the spine before inserting cages. This finding implies that the presence of the facets, the integrity of the peripheral annulus and ligaments or both are important to how effectively anterior cages or dowels achieve biomechanical stability during axial rotation.

Application of cages or dowels did not reduce the shear

TABLE 4

*Differences in shear parameters compared with normal**

Shear Force Parameter & Direction	Type of Fixation Device (p Value): Destabilized Spine	Pedicle Screw/Rod	Cage	Dowel	Cage + Pedicle Screw/Rod	Dowel + Pedicle Screw/Rod	Postfatigue: Cage	Postfatigue: Dowel
NZ (mm)								
AP	0.25 ± 0.34 (0.0002)†	−0.05 ± 0.07 (0.0002)†	0.20 ± 0.33 (0.0552)	0.33 ± 0.47 (0.0349)†	−0.03 ± 0.04 (0.0299)†	−0.07 ± 0.06 (0.0024)†	0.29 ± 0.32 (0.0083)†	0.52 ± 0.72 (0.0372)†
EZ (mm)								
anterior	1.11 ± 0.68 (0.0000)†	−0.21 ± 0.17 (0.0000)†	0.06 ± 0.38 (0.5856)	0.17 ± 0.44 (0.2087)	−0.22 ± 0.09 (0.0000)†	−0.24 ± 0.16 (0.0002)†	0.30 ± 0.87 (0.2606)	0.31 ± 0.74 (0.1917)
posterior	1.65 ± 1.28 (0.0000)†	−0.20 ± 0.17 (0.0000)†	0.06 ± 0.31 (0.4996)	0.14 ± 0.25 (0.0751)	−0.20 ± 0.10 (0.0000)†	−0.26 ± 0.15 (0.0001)†	0.30 ± 0.45 (0.0406)†	0.39 ± 0.47 (0.0212)†
ROM (mm)								
AP	2.80 ± 1.43 (0.0000)†	−0.59 ± 0.36 (0.0000)†	0.11 ± 1.07 (0.7252)	0.44 ± 1.25 (0.2509)	−0.64 ± 0.35 (0.0000)†	−0.77 ± 0.35 (0.0000)†	0.79 ± 1.45 (0.0866)	1.00 ± 1.87 (0.1072)
flexibility (mm/kN)								
anterior	9.3 ± 6.0 (0.0000)†	−4.2 ± 3.9 (0.0000)†	0.6 ± 5.6 (0.7002)	2.6 ± 6.9 (0.2091)	−4.4 ± 2.2 (0.0000)†	−5.1 ± 4.0 (0.0010)†	3.0 ± 14.8 (0.4995)	7.1 ± 13.7 (0.1188)
posterior	20.2 ± 13.5 (0.0000)†	−4.4 ± 3.8 (0.0000)†	1.1 ± 5.6 (0.4904)	3.2 ± 5.1 (0.0532)	−4.2 ± 2.3 (0.0000)†	−5.6 ± 3.7 (0.0003)†	4.8 ± 5.9 (0.0170)†	7.1 ± 6.5 (0.0047)†

* All values are presented as the means ± standard deviations.
† Statistically significant.

TABLE 5

*Average increases in angular motion and shear parameters with fatigue**

	Cages			Dowels		
Parameter & Direction	Mean Increase	Percent Increase	p Value	Mean Increase	Percent Increase	p Value
angular motion						
NZ (°)						
flexion–extension	1.0 ± 1.6	67 ± 89	0.0287†	0.3 ± 0.9	20 ± 48	0.1756
lat bending	0.9 ± 1.4	49 ± 65	0.0229†	0.4 ± 1.0	35 ± 63	0.1345
axial rotation	0.5 ± 0.5	35 ± 33	0.0042†	0.2 ± 0.5	8 ± 33	0.1033
EZ (°)						
flexion	1.0 ± 2.6	72 ± 120	0.1090	0.8 ± 1.1	23 ± 40	0.0163†
extension	0.1 ± 1.3	18 ± 44	0.3556	0.3 ± 0.8	11 ± 29	0.1254
lat bending	0.8 ± 1.0	57 ± 60	0.0092†	0.6 ± 0.5	29 ± 17	0.0009†
axial rotation	1.2 ± 0.8	47 ± 25	0.0001†	1.0 ± 0.7	33 ± 14	0.0003†
ROM (°)						
flexion	1.5 ± 2.2	59 ± 80	0.0221†	0.9 ± 1.2	23 ± 32	0.0118†
extension	0.6 ± 1.5	29 ± 43	0.0854	0.4 ± 0.7	11 ± 22	0.0463†
lat bending	1.3 ± 0.8	54 ± 40	0.0001†	0.8 ± 0.5	29 ± 17	0.0003†
axial rotation	1.4 ± 0.9	44 ± 22	0.0001†	1.1 ± 0.8	26 ± 14	0.0008†
flexibility (°/Nm)						
flexion	0.01 ± 0.14	1 ± 52	0.8675	−0.04 ± 0.10	−13 ± 34	0.1807
extension	−0.07 ± 0.19	−9 ± 34	0.2219	−0.13 ± 0.21	−20 ± 29	0.0716
lat bending	0.04 ± 0.06	16 ± 22	0.0567	0.00 ± 0.09	3 ± 28	0.9382
axial rotation	0.00 ± 0.05	4 ± 16	0.9115	−0.03 ± 0.08	−1 ± 21	0.3355
shear force						
NZ (mm)						
AP	0.32 ± 0.31	2100 ± 3193	0.0021†	0.58 ± 0.76	2150 ± 2593	0.0144†
EZ (mm)						
anterior	0.52 ± 0.86	1397 ± 2496	0.0299†	0.55 ± 0.71	875 ± 957	0.0133†
posterior	0.51 ± 0.44	883 ± 1289	0.0010†	0.66 ± 0.56	2775 ± 3671	0.0015†
ROM (mm)						
AP	1.46 ± 1.25	1407 ± 1232	0.0010†	1.79 ± 1.77	1405 ± 1280	0.0037†
flexibility (mm/kN)						
anterior	7.3 ± 14.8	657 ± 1329	0.0583	12.2 ± 13.7	766 ± 891	0.0072†
posterior	9.2 ± 5.5	507 ± 715	0.0001†	12.8 ± 8.9	738 ± 890	0.0004†

* All values, except p values, are presented as the means ± standard deviations.
† Statistically significant.

NZ to within the normal range. More surprisingly, they also did not reduce the shear NZ to within the destabilized range, especially after fatigue (Fig. 3 and Table 4). Hence, the loose region near the neutral position—at displacements before the ligaments or hardware began to resist motion—was unaffected by interbody cage or dowel fixation. Cage or bone dowel fixation mainly affected the amount of resistance that occurred once loading had begun. When ligament tension was restored, the EZ behaved more like the intact spine in cage- or dowel-fixed specimens than in destabilized specimens. (Table 4).

Interbody devices were extremely susceptible to the amount of fatigue applied, as indicated by significant alterations in several angular and linear parameters (Table 5). The applied fatigue was within normal physiological limits for the lumbar spine. Therefore, a few weeks after surgery, threaded interbody devices may not provide the same amount of biomechanical stability that they appear to provide at surgery. However, this conclusion is tentative because the study was performed in vitro: the ligaments may have lost strength during their extended exposure in the laboratory (see *Study Limitations*). The statistically significant correlation between BMD values and increase in NZ values with applications of fatigue indicates that some bone subsidence may also underlie the loosening of cages or dowels.

Cage Compared With Dowel Fixation

The geometrical characteristics of the bone dowels and threaded cages were almost identical. Consequently, little difference was expected biomechanically, and there were no statistically significant differences in the angular or linear NZ, EZ, ROM, or flexibility, either alone or in combination with pedicle screws and rods.

Another consideration in comparing bone dowels with threaded cages is surface texture. Because of the limited workability of cortical bone, the cutting edges of the threads cannot be made as sharp in bone dowels as in metal cages. Bone dowels, therefore, require tapping before insertion whereas metal cages do not. As a consequence, cages likely have a slightly tighter device–bone interface than dowels. This difference may account for the slight and insignificant prefatigue advantage that cages appeared to have over dowels in limiting shear NZ, EZ, and ROM values (Fig. 3).

Pedicle Screw/Rod Compared With Interbody Instrumentation

Pedicle screws and rods appeared to resist both angular

TABLE 6

*Average enhancement (reduction) in motion parameters demonstrated by adding pedicle screws/rods to cage or dowel fixation**

	Cages			Dowels		
Parameter & Direction	Mean Change	Percent Decrease	p Value	Mean Change	Percent Decrease	p Value
angular motion						
NZ (°)						
flexion–extension	−1.2 ± 0.5	−80 ± 14	0.0000	−1.6 ± 1.1	−88 ± 6	0.0003
lat bending	−1.3 ± 1.2	−85 ± 11	0.0020	−1.2 ± 1.3	−82 ± 9	0.0054
axial rotation	−0.7 ± 0.8	−48 ± 42	0.0063	−1.2 ± 0.8	−70 ± 23	0.0003
EZ (°)						
flexion	−2.4 ± 2.0	−85 ± 7	0.0010	−2.4 ± 1.5	−89 ± 6	0.0002
extension	−2.2 ± 1.5	−82 ± 13	0.0002	−2.6 ± 1.2	−89 ± 9	0.0000
lat bending	−1.6 ± 1.1	−71 ± 14	0.0003	−1.6 ± 1.0	−78 ± 7	0.0002
axial rotation	−1.9 ± 1.2	−71 ± 15	0.0001	−2.2 ± 1.3	−75 ± 11	0.0001
ROM (°)						
flexion	−3.0 ± 2.2	−85 ± 8	0.0003	−3.2 ± 1.9	−89 ± 5	0.0001
extension	−2.9 ± 1.6	−83 ± 10	0.0000	−3.4 ± 1.3	−89 ± 6	0.0000
lat bending	−2.2 ± 1.7	−76 ± 11	0.0004	−2.2 ± 1.6	−79 ± 7	0.0004
axial rotation	−2.2 ± 1.5	−68 ± 17	0.0002	−2.8 ± 1.5	−75 ± 13	0.0000
flexibility (°/Nm)						
flexion	−0.22 ± 0.09	−75 ± 17	0.0000	−0.25 ± 0.03	−80 ± 12	0.0000
extension	−0.34 ± 0.18	−75 ± 19	0.0000	−0.43 ± 0.28	−80 ± 21	0.0002
lat bending	−0.20 ± 0.10	−64 ± 17	0.0000	−0.23 ± 0.10	−72 ± 10	0.0000
axial rotation	−0.21 ± 0.09	−58 ± 19	0.0000	−0.25 ± 0.14	−62 ± 20	0.0001
shear force						
NZ (mm)						
AP	−0.22 ± 0.33	−78 ± 47	0.0154	−0.39 ± 0.51	−93 ± 19	0.0107
EZ (mm)						
anterior	−0.27 ± 0.37	−74 ± 32	0.0107	−0.41 ± 0.48	−75 ± 22	0.0063
posterior	−0.26 ± 0.33	−67 ± 34	0.0072	−0.41 ± 0.31	−76 ± 28	0.0004
ROM (mm)						
AP	−0.74 ± 0.89	−75 ± 20	0.0056	−1.21 ± 1.19	−82 ± 16	0.0024
flexibility (mm/kN)						
anterior	−5.0 ± 5.5	−83 ± 54	0.0031	−7.7 ± 7.8	−71 ± 22	0.0028
posterior	−5.3 ± 6.1	−77 ± 51	0.0045	−8.8 ± 6.4	−77 ± 30	0.0003

* All values, except p values, are presented as the means ± standard deviations, and all values represent statistically significant enhancements (reductions).

motion and shear far better than the interbody devices, probably because pedicle screws can be inserted rigidly whereas the interbody devices performed only as well as the remaining ligaments allowed. Even the strongest and healthiest ligaments stretch substantially. Therefore, pedicle screws and rods would likely tend to outperform interbody devices in patients.

The finding that lower BMD values correlated with greater ROM and NZ values obtained in specimens fitted with either pedicle screws/rods or interbody cages or dowels indicates that both constructs are affected by bone quality. Pedicle screws inserted in poor quality–bone vertebrae would toggle more, allowing more movement than would be associated with high-quality bone. Interbody cages inserted between vertebrae with poor quality bone would subside more than those inserted between vertebrae with high-quality bone, limiting the effectiveness of the remaining ligaments in maintaining their tension band effects. Our data do not indicate which effect is greater.

Combined-Hardware Construct

Although pedicle screw/rod hardware outperformed the interbody devices, the latter still significantly enhanced biomechanical stability when they supplemented the former (Table 7). The clinical implication of this mechanical advantage is unknown. Because the magnitude of further reductions in motion parameters was relatively small when hardware was combined, the clinical enhancement to fusion rates from the corresponding improved stability may be inconsequential.

Other biomechanical factors also must be considered. Pedicle screw/rod hardware is attached posteriorly and must work through a large lever arm to resist loads that collapse the disc space (that is, flexion or compression). A large anterior column defect that is subjected to flexion or compression loading would most benefit from combining interbody devices with pedicle screw/rod hardware. A cage or dowel would act as a spacer and buttress against these loads, reducing the amount of stress that the pedicle screw/rod hardware would have to resist through a long lever arm. By relieving this stress on the pedicle screw/rod hardware, cages or dowels could lessen the chance that the former would fail from metal fatigue.

In cadaveric specimens, lumbar stiffness increases and ROM decreases when transfacet screws or translaminar facet screws are added to threaded interbody cages.[11,13]

TABLE 7

*Average enhancement (reduction) in motion parameters demonstrated by adding cages or dowels to pedicle screw/rod fixation**

	Cages			Dowels		
Parameter & Direction	Mean Change	Percent Decrease	p Value	Mean Change	Percent Decrease	p Value
angular motion						
NZ (°)						
flexion–extension	0.0 ± 0.3	1 ± 97	0.3168	−0.1 ± 0.2	−25 ± 45	0.0329†
lat bending	−0.1 ± 0.1	−29 ± 45	0.0064†	−0.1 ± 0.4	30 ± 95	0.1878
axial rotation	−0.1 ± 0.5	14 ± 59	0.2563	−0.1 ± 0.5	9 ± 72	0.2192†
EZ (°)						
flexion	−0.4 ± 0.3	−53 ± 33	0.0001†	−0.4 ± 0.3	−56 ± 33	0.0007†
extension	−0.5 ± 0.4	−48 ± 38	0.0008†	−0.4 ± 0.4	−51 ± 38	0.0014†
lat bending	−0.1 ± 0.2	−17 ± 30	0.0662	−0.1 ± 0.2	−26 ± 30	0.0124†
axial rotation	−0.1 ± 0.3	−23 ± 35	0.0404†	−0.1 ± 0.2	−18 ± 31	0.0315†
ROM (°)						
flexion	−0.4 ± 0.4	−45 ± 42	0.0015†	−0.5 ± 0.4	−53 ± 31	0.0007†
extension	−0.5 ± 0.5	−42 ± 43	0.0018†	−0.5 ± 0.4	−46 ± 39	0.0024†
lat bending	−0.1 ± 0.2	−22 ± 26	0.0254†	−0.2 ± 0.3	−25 ± 32	0.0402†
axial rotation	−0.2 ± 0.5	−20 ± 38	0.0869	−0.2 ± 0.3	−17 ± 34	0.0426†
flexibility (°/Nm)						
flexion	−0.06 ± 0.06	−43 ± 40	0.0024†	−0.08 ± 0.07	−49 ± 38	0.0020†
extension	−0.09 ± 0.09	−41 ± 43	0.0030†	−0.08 ± 0.08	−41 ± 52	0.0040†
lat bending	−0.01 ± 0.06	−10 ± 41	0.2333	−0.03 ± 0.04	−24 ± 27	0.0173†
axial rotation	−0.02 ± 0.06	−18 ± 34	0.1204	−0.02 ± 0.04	−15 ± 32	0.0817
shear force						
NZ (mm)						
AP	0.00 ± 0.03	−70 ± 205	0.3440	0.01 ± 0.04	−14 ± 561	0.2350
EZ (mm)						
anterior	−0.03 ± 0.05	−17 ± 116	0.0164†	0.00 ± 0.06	66 ± 294	0.4077
posterior	−0.05 ± 0.06	−59 ± 78	0.0039†	−0.04 ± 0.05	−39 ± 37	0.0069†
ROM (mm)						
AP	−0.06 ± 0.07	−28 ± 58	0.0040†	−0.03 ± 0.06	−15 ± 40	0.0781
flexibility (mm/kN)						
anterior	−0.59 ± 1.79	−24 ± 140	0.1298	−0.61 ± 2.92	−9 ± 107	0.2415
posterior	−0.81 ± 1.70	−43 ± 102	0.0551	−0.65 ± 1.45	−25 ± 83	0.0747

* All values, except p values, are presented as the means ± standard deviations.
† Statistically significant enhancement (reduction).

The results of combining pedicle screw/rod hardware with interbody cages support these earlier findings. The addition of pedicle screws, however, significantly decreased flexibility and ROM in more modes of loading than were found in previous studies (Table 6). Furthermore, the magnitudes of the decreases were greater. Our findings indicate that pedicle screw fixation provides a greater stabilizing effect than transfacet or translaminar screw fixation. These findings are consistent with the expected superior stiffness obtained by pedicle screws, which are larger, engage more bone, and cross more columns of the spine than transfacet or translaminar screws.[6]

Study Limitations

An important consideration related to interbody devices is bone remodeling. Intuitively, to create a stable and strong mass, interbody devices would likely be more effective after some ingrowth or attachment of bone has occurred. The ligaments would then no longer be called on to provide biomechanical stability. This study addressed only the immediate and short-term postoperative condition. The effects of healing should be addressed in future comparisons of bone dowels and interbody cages through in vivo or clinical testing.

When patients stand, bone dowels or threaded cages are compressed from the weight of the upper body and from muscle contraction. We attempted to apply compressive preload forces on specimens with weights and pulleys, but it was too difficult to direct the compressive force axially so as to avoid disturbing the test load once the specimens had been bent substantially. Thus, the specimens were exposed only to the slight gravitational compression (9 N) exerted by the upper fixture of the apparatus. Patients, however, are obviously not always upright, and compression of interbody devices cannot be assumed to be constant. Large shear or bending forces can be generated when patients move from an upright to prone position and vice versa. The results of this study can be considered a worst-case scenario: the fundamental biomechanical stability provided by the devices without the benefit of compression. Compression would improve stability, although to an unknown degree.

By design, threaded interbody cages and bone dowels rely extensively on the integrity of the remaining lateral ligaments and annulus to keep them compressed between the vertebrae. In vivo, these tissues are nourished by the body. In vitro, they become necrotic and lose their ability to withstand loads. The specimens were kept at room temperature for several days during procedures for destabilization, instrumentation, and testing. Unlike interbody fixation devices, pedicle screw/rod hardware relies only on the integrity of the bone, not the ligaments. When exposed for long periods, bones do not become as necrotic as ligaments. This difference between the fixation systems may have given pedicle screw/rod hardware an advantage over the interbody devices that could have manifested in the results as smaller motions.

As is common, it was difficult to obtain a large number of specimens for this study. Furthermore, during the course of testing, several levels were lost to study because hardware had been attached poorly. The small number of specimens often limited statistical power (probability of avoiding a Type 2 or false-negative error) to less than 0.7. Although assertions of statistical differences were made with high confidence ($p < 0.05$), assertions of lack of differences, when the values appeared different, may actually reflect too few specimens to verify whether the observed trend was significant.

Conclusions

Pedicle screw/rod fixation stabilizes Grade I spondylolisthesis significantly more than anteriorly inserted threaded cages or dowels. The superior biomechanical stability of pedicle screw/rod fixation is evident during flexion, extension, lateral bending, axial rotation, and AP shear.

Because pedicle screw/rod hardware significantly enhances the spinal stability conferred by cages or dowels (and vice versa to a lesser extent), surgeons should consider using the two devices together. Combining the two forms of fixation may reduce the likelihood of hardware failure. The high biomechanical stability provided by the pedicle screw/rod hardware should increase the likelihood that fusion will occur across a cage or dowel. The results of this model suggest that cages or bone dowels should not be used as stand-alone devices to treat lumbar spondylolisthesis, although the limitations of the model should be kept in mind.

In specimens fitted with cages or dowels alone, fatigue caused significant destabilization. Thus, surgeons should probably limit the amount of postoperative lumbar movement (through external orthosis, prescribed bed rest, or both) in patients in whom cages or dowels have been placed without pedicle screws/rods until healing has occurred and a strong interface between cage/dowel and adjacent VBs has formed.

References

1. Crawford NR, Çagli S, Sonntag VKH, et al: Biomechanics of Grade I degenerative lumbar spondylolisthesis. Part 1: In vitro model. **J Neurosurg (Spine 1) 94:**45–50, 2000
2. Dickman CA, Fessler RG, MacMillan M, et al: Transpedicular screw–rod fixation of the lumbar spine: operative technique and outcome in 104 cases. **J Neurosurg 77:**860–870, 1992
3. Emery SE: In situ decompression and fusion techniques for spondylolisthesis. **Semin Spine Surg 11:**34–47, 1999
4. Jost B, Cripton PA, Lund T, et al: Compressive strength of interbody cages in the lumbar spine: the effect of cage shape, posterior instrumentation and bone density. **Eur Spine J 7:**132–141, 1998
5. Lund T, Oxland TR, Jost B, et al: Interbody cage stabilisation in the lumbar spine: biomechanical evaluation of cage design, posterior instrumentation and bone density. **J Bone Joint Surg (Br) 80:**351–359, 1998
6. Maiman DJ, Yoganandan N, Pintar F, et al: Biomechanics of the spine, in Youmans JR (ed): **Neurological Surgery.** Philadelphia: WB Saunders, 1996, pp 2221–2240
7. Mardjetko SM, Connolly PJ, Shott S: Degenerative lumbar spondylolisthesis. A meta-analysis of literature 1970–1993. **Spine 19:** S2256–S2265, 1994
8. Masferrer R, Gomez CH, Karahalios DG, et al: Efficacy of pedicle screw fixation in the treatment of spinal instability and failed back surgery: a 5-year review. **J Neurosurg 89:**371–377, 1998
9. Mizuno J, Nakagawa H: Threaded fusion cage for lumbar spondylolisthesis. **Neurol Med Chir 38:**155–160, 1998
10. Nibu K, Panjabi MM, Oxland T, et al: Multidirectional stabilizing potential of BAK interbody spinal fusion system for anterior surgery. **J Spinal Disord 10:**357–362, 1997
11. Rathonyi GC, Oxland TR, Gerich U, et al: The role of supplemental translaminar screws in anterior lumbar interbody fixation: a biomechanical study. **Eur Spine J 7:**400–407, 1998
12. Rechtine GR, Sutterlin CE, Wood GW, et al: The efficacy of pedicle screw/plate fixation on lumbar/lumbosacral autogenous bone graft fusion in adult patients with degenerative spondylolisthesis. **J Spinal Disord 9:**382–391, 1996
13. Volkman T, Horton WC, Hutton WC: Transfacet screws with lumbar interbody reconstruction: biomechanical study of motion segment stiffness. **J Spinal Disord 9:**425–432, 1996
14. White AA, III, Panjabi MM: **Clinical Biomechanics of the Spine.** Philadelphia: JB Lippincott, 1990
15. Yuan HA, Garfin SR, Dickman CA, et al: A historical cohort study of pedicle screw fixation in thoracic, lumbar, and sacral spinal fusions. **Spine 19:**S2279–S2296, 1994

Manuscript received September 9, 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., Neuroscience Publications, Barrow Neurological Institute, 350 West Thomas Road, Phoenix, Arizona 85013–4496. email: neuropub@chw.edu.