Interbody cage stabilisation in the lumbar spine

BIOMECHANICAL EVALUATION OF CAGE DESIGN, POSTERIOR INSTRUMENTATION AND BONE DENSITY

T. Lund, T. R. Oxland, B. Jost, P. Crompton, S. Grassmann, C. Etter, L.-P. Nolte

From the University of Bern, Switzerland

We performed a biomechanical study on human cadaver spines to determine the effect of three different interbody cage designs, with and without posterior instrumentation, on the three-dimensional flexibility of the spine. Six lumbar functional spinal units for each cage type were subjected to multidirectional flexibility testing in four different configurations: intact, with interbody cages from a posterior approach, with additional posterior instrumentation, and with cross-bracing. The tests involved the application of flexion and extension, bilateral axial rotation and bilateral lateral bending pure moments. The relative movements between the vertebrae were recorded by an optoelectronic camera system.

We found no significant difference in the stabilising potential of the three cage designs. The cages used alone significantly decreased the intervertebral movement in flexion and lateral bending, but no stabilisation was achieved in either extension or axial rotation. For all types of cage, the greatest stabilisation in flexion and extension and lateral bending was achieved by the addition of posterior transpedicular instrumentation. The addition of cross-bracing to the posterior instrumentation had a stabilising effect on axial rotation. The bone density of the adjacent vertebral bodies was a significant factor for stabilisation in flexion and extension and in lateral bending.

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T. Lund, MD, Resident Surgeon
T. R. Oxland, PhD, Research Group Head
B. Jost, MD, Resident Surgeon
P. Crompton, MS, Graduate Student
S. Grassmann, MS, Research Engineer
L.-P. Nolte, PhD, Division Head
M. E. Müller Institute for Biomechanics, University of Bern, Murtenstrasse 35, PO Box 30, 3010 Bern, Switzerland.

C. Etter, MD, Surgeon
Wirbelsäulenchirurgie, Schmerzklinik, Hirschgässlein 11-15, 4051 Basel, Switzerland.

Correspondence should be sent to Dr T. R. Oxland at the Department of Orthopaedics, Faculty of Medicine, University of British Columbia, Room 3114, 910 West 10th Avenue, Vancouver, British Columbia, Canada V5Z 4E3.

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The technique of posterior lumbar intervertebral fusion (PLIF) was introduced independently by Jaslow¹ and by Cloward^{2,3} in the mid-1940s. The theoretical basis is that mechanical stability is provided by the intervertebral fusion, the original disc height is restored and the intervertebral foramina are distracted. Later authors, including Lee, Vessa and Lee,⁴ noted that excision of the nucleus pulposus eliminated many of the possible biomechanical causes of chronic pain. There has been, however, a lack of enthusiasm for PLIF, mainly because of the technical difficulty of the operation and the complications, especially the risk of neural damage and graft retropulsion into the spinal canal.⁵ Moreover, clinical studies have shown that the postoperative increase in disc height tends to return to the preoperative level, with or without additional posterior fixation, and regardless of the type of bone graft.⁶⁻⁹ Whether this occurs because of graft subsidence into the adjacent vertebral body or graft collapse is unknown.

In the last few years, several interbody cages of different designs have been developed for use through an anterior or posterior approach.¹⁰⁻¹³ The aim was to provide mechanical support to the segment being fused with biocompatible implant material, and to allow the use of autogenous bone to promote fusion. Theoretically, these new implants give more lasting restoration of disc height and better stabilisation as a result of a more repeatable surgical technique and more consistent initial behaviour.

In 1993 Brantigan and Steffee¹⁴ first described the use of interbody implants and reported successful fusion in all 28 patients after the use of a carbon-fibre-reinforced polymer implant with pedicle screw fixation. In a more recent study by Tullberg et al¹⁵ a fusion rate of 86% was achieved using the same implant with posterior fixation; all but one of their 51 patients maintained the immediate postoperative disc height for a year. Clinical results for interbody implants without posterior fixation have been reported with fusion rates of between 83% and 100%.^{12,16,17}

Despite the growing clinical interest in intervertebral cages, few studies have been performed to evaluate their biomechanical behaviour. DeBowes et al¹⁸ observed bone ingrowth into and around a stainless-steel implant in horses; other investigators have made similar observations on other animal models.¹⁹⁻²¹ Butts, Kuslich and Bechtold²² addressed the immediate flexibility of using one or two

cylindrical implants in calf and pig spines and found that two implants were more stable than one. Wilder et al.²³ evaluated a cylindrical, threaded device (BAK; Spine-Tech Inc, Minneapolis, Minnesota) in a baboon model and found good stabilisation. In a multifaceted study in a calf spine model, Brodke et al.²⁴ found that the BAK system with pedicle fixation provided stability similar to that of pedicle fixation with interbody bone blocks. Tencer, Hampton and Eddy²⁵ evaluated a cylindrical cage (Ray TFC; Surgical Dynamics Inc, Concord, California) in different positions and orientations in the interbody space in calf and human spines. They found that the interbody cages changed only the intervertebral joint laxity (i.e., neutral zone) and not the joint stiffness. Only Tencer et al.²⁵ used human cadaver specimens for biomechanical evaluation. It is possible that there are significant differences between the animal models and man, because many animal models have immature endplates²⁶ and interbody implants rely on the endplate for fixation.

The immediate three-dimensional changes in flexibility due to cage insertion and the compression strength of the construct with a cage are important. The stabilisation provided by fixation is an indication of the relative movement occurring at the bone-implant interface and is therefore related to the potential for bony ingrowth and subsequent fusion. The compression strength of the construct provides an indication of the ability of the system to maintain a distracted disc height. This has been addressed by Jost et al.²⁷ Our aim in this study was to investigate the stabilisation provided by interbody cages in a human cadaver model.

Materials and Methods

Specimen preparation. We used 18 human cadaver lumbar functional spinal units (FSU; 9 L2-L3, 9 L4-L5). The age and gender of 12 of them were known. Their ages ranged from 25 to 67 years and there were three female and nine male specimens. They had an evenly distributed range of degeneration from mild to severe. All non-ligamentous soft tissue was removed before storage at -20°C . Care was taken to keep the specimens moist throughout the preparation and testing phases.

Before testing, the bone mineral density (BMD) of the upper and lower vertebrae of every specimen was determined using dual-energy X-ray absorptiometry (DEXA; QDR 1000, Hologic Corporation, Waltham, Massachusetts) from both lateral and anteroposterior (AP) directions, giving four measurements of bone quality expressed in g/cm^2 .

For all measurements, the entire L2-L5 specimens were placed in a plastic container and surrounded by granular semolina to simulate the soft tissues around the spine. After measurement the vertebrae were mounted in polymethylmethacrylate (PMMA) so that the disc space was positioned horizontally.

Fig. 1

The apparatus for multidirectional flexibility testing. Marker carriers with LEDs were attached to the specimen. Pure moments (here a lateral bending moment), were applied to the upper vertebra and its motion with respect to the lower vertebra, was measured. A compressive preload of 200 N was used throughout the testing.

Flexibility measurement. A specially designed apparatus allowed the precise application of known loads to the spine and the measurement of intervertebral movements in an unconstrained manner (Fig. 1). A 200 N compressive preload was applied throughout and positioned so that no observable rotation was produced on loading in a neutral posture. Pure moments of flexion and extension, bilateral axial rotation and bilateral lateral bending were applied individually to the upper vertebra in steps of 2.5 Nm to a maximum of 10 Nm. At each load step, the specimen was allowed to creep for 30 seconds before measurement.

The movement of the upper with respect to the lower vertebra was measured using an optoelectronic camera system (Optotrak 3020; Northern Digital, Waterloo, Canada). The vertebrae and the intervertebral cages were equipped with marker carriers, each of which had four non-collinearly-arranged infrared light-emitting diodes (LEDs). The optoelectronic camera monitored the spatial position of the markers and could locate the position of each LED marker with an accuracy of 0.15 mm at a distance of 2 m. Customised software was used to calculate the rotations of the upper with respect to the lower vertebra in terms of Euler angles.

For each applied moment and for all testing configurations the movement in the direction of the moment was

Fig. 2a

Fig. 2b

Fig. 2c

Fig. 2d

Photograph and lateral radiographs of the different cage designs. Figure 2a – A top view of the Stratec, Ray and Brantigan cages from left to right, showing the part of the implant which would interface with the adjacent vertebrae. For all designs, two parallel interbody cages were inserted. Figure 2b – The Stratec cage, a porous titanium implant designed to fit to the endplate contours. Figure 2c – The Brantigan cage, a rectangular, porous carbon-fibre implant. Figure 2d – The Ray cage, a cylindrical, threaded, porous titanium implant.

analysed and the total movement under the maximum moment, the range of motion (ROM), was calculated. The intervertebral joint laxity (i.e., neutral zone) was calculated²⁸ but was consistently of small magnitude, less than 2°, and therefore these data are not presented.

Experimental protocol. The 18 specimens were first tested in the intact condition and then after the insertion of one of three different interbody implant designs. Six specimens were tested for each implant type. The implants were:

- 1) a porous titanium cage designed to fit the endplate contours (Stratec; STRATEC Medical AG, Oberdorf, Switzerland);
- 2) a rectangular carbon-fibre cage (Brantigan; Acromed Corporation, Cleveland, Ohio); and
- 3) a cylindrical threaded titanium cage (Ray TFC; Surgical Dynamics Inc, Concord, California).

Before insertion, the cages were filled with autogenous bone. All specimens had two intervertebral implants inserted from a posterior direction according to the manufacturer's guidelines for surgical technique which included the removal of the medial portion of the articular facets. All disc spaces were distracted to achieve a tight annulus. Photographs of the different cages and lateral radiographs

are shown in Figure 2. All specimens with intervertebral cages then had titanium transpedicular fixation (Universal Spine System; STRATEC Medical AG, Oberdorf, Switzerland). A compression load of 200 N was applied to the specimen before the bolts were tightened. For the fourth test a cross-link was added to the pedicle fixation.

Statistical methods. Since there were only six specimens per group, we could not be certain that the data were normally distributed and therefore we used non-parametric methods. To determine the effect of cage design, the cage movements were normalised with respect to the intact FSU movement and a Kruskal-Wallis analysis of variance (ANOVA) was performed across the three cage types. The critical level of significance was 0.05.

To determine the sequential effects of cage insertion, posterior instrumentation and cross-bracing the different interbody implant designs were grouped and a Friedman repeated-measures ANOVA was performed. The grouping of the different cage designs was allowable only if there were no differences between the designs. If the Friedman ANOVA results indicated differences between the groups, paired comparisons were made between specific fixations using a Wilcoxon signed-rank test with Bonferroni correc-

tion for multiple comparisons. Since three paired comparisons were deemed relevant (intact to cage, cage to posterior instrumentation and cage, and posterior instrumentation and cage to cross-bracing on posterior instrumentation and cage), the critical significance levels were 0.01667, 0.03333 and 0.05 for the largest to smallest differences.²⁹

To determine the effect of bone density on cage performance, Pearson correlation coefficients were determined between the normalised cage movements and the DEXA BMD measurements. A *p* value of less than 0.05 was considered significant.

Results

The median ratios of cage to intact movement with quartiles and ranges for the different cages are plotted in Figure 3 for all loading directions. The median ranges of motion with quartiles in flexion, extension, bilateral axial rotation and bilateral lateral bending for the different test conditions are shown in Figures 4 to 7, respectively.

Flexion and extension. In flexion, the median cage move-

ments were 44% to 72% of intact movement, a statistically significant difference ($p = 0.004$). The differences between the cages, however, were not statistically significant ($p = 0.53$, Fig. 3). In extension, the median cage movements were 80% to 107% of intact movement which was not a significant difference ($p = 0.55$). The differences between the cage designs were not statistically significant ($p = 0.31$, Fig. 3).

The ranges of motion in flexion and extension are illustrated in Figures 4 and 5. The median movements with posterior instrumentation were 11% and 18% of intact movement which were significantly different from the conditions after cage insertion (flexion $p = 0.0003$; extension $p = 0.0002$). The cross-bracing had no significant effect in flexion and extension from the posterior instrumentation condition (flexion $p = 0.55$; extension $p = 0.57$).

Axial rotation. The median cage to intact movement ratios and the ranges of motion for bilateral axial rotation are shown in Figures 3 and 6. Overall, the cage insertions resulted in significantly increased intervertebral movement ($p = 0.03$). The median movement with the Ray cage was similar to the intact movement while the median Brantigan

Fig. 3

Median ratios of cage to intact movement for all loading directions. Quartiles and ranges are shown as boxes and whiskers. For each cage type, six specimens were tested. There were no significant differences between the cage designs for any loading direction: flexion $p = 0.53$, extension $p = 0.31$, axial rotation $p = 0.12$, lateral bending $p = 0.10$.

Fig. 4

Data for median range of motion (with quartiles) for flexion loading. The different testing configurations were contrasted for all cages grouped together (18 specimens in each group). The cages provided significant stabilisation from the intact ($p = 0.004$). The cage w/posterior instrumentation (cage w/PI) was different from the cage condition ($p = 0.0003$) while the addition of cross-bracing (cage w/PI & CB) was not significant ($p = 0.55$).

Fig. 5

Data for median range of motion (with quartiles) for extension loading. The different testing configurations were contrasted for all cages grouped together (18 specimens in each group). The cages did not provide significant stabilisation from intact ($p = 0.55$). The cage w/posterior instrumentation (cage w/PI) was different from the cage condition ($p = 0.0002$) while the addition of cross-bracing (cage w/PI & CB) was not significant ($p = 0.57$).

Fig. 6

Data for median range of motion (with quartiles) for axial rotation loading. The different testing configurations were contrasted for all cages grouped together (18 specimens in each group). The cages did not provide significant stabilisation from intact but instead, produced significantly greater movement than intact ($p = 0.03$). The cage w/posterior instrumentation (cage w/PI) was different from the cage condition ($p = 0.0002$). The addition of cross-bracing (cage w/PI & CB) was also significant in this loading direction ($p = 0.0005$).

Fig. 7

Data for median range of motion (with quartiles) for lateral bending loading. The different testing configurations were contrasted for all cages grouped together (18 specimens in each group). The cages provided significant stabilisation from intact ($p = 0.004$). The cage w/posterior instrumentation (cage w/PI) was different from the cage condition ($p = 0.0002$) while the addition of cross-bracing (cage w/PI & CB) was not significant ($p = 0.20$).

cage increased axial rotation compared with the intact state by 19% and the Stratec cage by 49% (Fig. 3). The difference between the cages was not significant ($p = 0.12$). Posterior instrumentation reduced axial rotation to below intact levels and this change was significant ($p = 0.0002$, Fig. 6). The effect of cross-bracing was significant for all

cage types and it reduced movement consistently to a maximum of 0.5° ($p = 0.0005$).

Lateral bending. The median cage to intact movement ratios and the ranges of motion for bilateral lateral bending are shown in Figures 3 and 7. Cage insertion produced a reduction in median movement to between 58% and 80%

of the intact movement ($p = 0.004$, Fig. 3). The differences between the cage designs were not significant ($p = 0.10$). The effect of posterior instrumentation was to reduce movement to 9% of intact movement which was statistically different from the cage movement ($p = 0.0002$, Fig. 7). Cross-bracing had no effect on lateral bending movement ($p = 0.20$).

Effect of bone density. To determine the effect of vertebral bone density on the cage stability, all cage types were grouped together. This was deemed feasible since no significant differences had been observed between the different cage designs. Four BMD values were available for each tested specimen, i.e., for the upper and lower vertebrae of the FSU from both lateral and AP directions. The AP values

Fig. 8

Scatterplots of the ratio of cage to intact movement versus the upper vertebral lateral BMD in flexion and extension (a), axial rotation (b) and lateral bending (c). Significant correlations were observed in flexion and extension ($R^2 = 0.26$, $p = 0.032$) and lateral bending ($R^2 = 0.44$, $p = 0.003$).

were higher than the lateral values, presumably due to the influence of the posterior elements. Since our focus was the vertebral body, we used the lateral BMD values. The upper and lower vertebral densities were highly correlated and for this analysis, the upper vertebral density was used.

Scatterplots between the normalised cage movements (relative to intact) and the upper vertebra lateral BMD are shown in Figure 8 for flexion and extension, axial rotation, and lateral bending. In flexion and extension and lateral bending, there was a significant relationship between the movement and BMD (flexion and extension $R^2 = 0.26$, $p = 0.032$; lateral bending $R^2 = 0.44$, $p = 0.003$). Thus, the stabilising effect of cage insertion was greatest as bone density increased. In axial rotation, there was no significant relationship between movement and BMD (axial rotation $R^2 = 0.02$, $p = 0.60$).

Discussion

Effect of cage design. No significant differences between the cage designs were observed. A marginally significant difference was seen in axial rotation ($p = 0.12$) in which the Ray cage allowed the same motion as intact but the Stratec and Brantigan cages allowed greater movement than intact. This trend may be due to the Ray cage penetrating the endplate and providing an interference fit to resist the local shear loading associated with an axial torque. The other two implants rest on the endplate and therefore rely on friction to resist shear.

Reductions in movement after cage insertion were significant in flexion and lateral bending. This is consistent with the findings of most previous investigations²²⁻²⁴ and particularly of Tencer et al²⁵ who found significant reductions in laxity only in flexion and lateral bending.

The lack of reduction in movement in extension has not been consistently described probably because most investigators report movement in the sagittal plane as flexion and extension together.^{22,24} Wilder et al²³ found stabilisation in extension, but this may be because they used a baboon model which has been shown to have higher extension stiffness than the human spine.³⁰ In their tests on human spines, Tencer et al²⁵ did not find any stabilisation in extension, which is similar to our results.

Under an applied moment, an FSU must resist tensile and compression loads about a neutral axis, or centre of rotation, which is not fixed, but moves in a locus as the load changes. For simplification, we may assume that it is stationary and positioned in the intact condition approximately at the superior endplate of the lower vertebra, towards the posterior wall of the vertebral body.^{31,32} This implies that the tension is resisted anteriorly by the annulus fibrosus and that the compression is resisted posteriorly by the disc tissues and the posterior bony structures, particularly the facets. It has been established that the facets are loaded in compression under extension moments.³³ After cage insertion using the posterior approach, a medial facet-

ectomy and substantial FSU distraction have been produced. Under the same extension moment, the resistance provided by the anterior annulus in tension should be greater than that of the intact condition due to its increased stiffness after distraction.³⁴ Since the stability in extension is no better than the intact stability, the compression resistance provided by the posterior structures has probably been reduced. The effect of a bilateral medial facetectomy has also been studied previously³⁵ and found not to affect extension movement. It seems reasonable therefore that the distraction of the facet surfaces may be the primary reason for the lack of posterior resistance to the compression loads occurring during the application of an extension moment.

Effect of posterior instrumentation. A very consistent finding in our study was the dramatic stabilisation provided by posterior instrumentation together with interbody cages. This is similar to previous observations²⁴ when the most stable construct was a cage with posterior instrumentation. Tencer et al²⁵ observed a significant stiffening effect of pedicle screws after cage insertion only in extension but they did note that it may have been due to poor screw fixation in relatively degenerated specimens. In our study, the quality of the specimens appeared to be better since there was a complete range of degeneration from mild to severe.

The addition of cross-bracing to the posterior instrumentation had no effect in flexion, extension or lateral bending but had a small, but significant effect in axial rotation. It has been shown that cross-bracing has an effect only in axial rotation for highly unstable segments.³⁶ It seems therefore that in non-traumatic segments such as we studied that cross-bracing is not of great importance.

After pedicle screw and cage fixation the spinal movement is probably reduced as much as possible without actual fusion.

Effect of bone density. There was a significant relationship between bone density and the stability. As bone density increased, the stabilisation provided by the cages improved except in axial rotation when bone density had no significant effect. With an interbody cage, flexion and extension and lateral bending impose a substantial compressive load component to the surrounding vertebral bone. In cases of low bone density, the bone may fail locally resulting in somewhat higher flexibility. In axial rotation, however, the loading of the bone-implant interface is dependent on the degree of implant engagement into the vertebral body. Moreover, in axial rotation, after cage insertion and before fusion, the facet joints and the annulus fibrosus probably remain as the major stabilising structures.

Limitations. As in any study in vitro, it is important to acknowledge the limitations of the experimental protocol. The physiological loads in the lumbar spine are not completely known. We chose therefore to apply pure moments so that the loading applied was constant along the length of the specimen. Each specimen and test condition were subjected to identical loading. It is well known that muscles

stabilise the spine and thereby increase the compressive load in the spine. In general, this reduces intervertebral movements.^{28,37}

Immediate stabilisation. Our study addresses only the immediate, postoperative stability of the various implant configurations. The variation in stability, fusion rate or maintenance of disc height with time were not evaluated. There have been several reports dealing with these questions using various animal models.²⁰⁻²² Sandhu et al²⁰ reported that at a six-month follow-up, fusion in both groups was significantly stiffer than intact untreated spines. In the baboon model, Grobler et al¹⁹ observed similar trends. It is possible therefore that the immediate stability as measured in our study is a worst-case scenario.

Clinical relevance. The key question is the degree of immediate reduction in movement that is required for eventual interbody fusion. For a fusion to occur, bone growth must occur into and through the implant. It is known that if the movement at the bone-implant interface is excessive fibrous tissue develops. Basic studies for porous-coated implants have suggested that movements greater than 150 µm are deleterious to bone ingrowth into porous surfaces.³⁸

The relationship between local bone-implant micromotions and the intervertebral motions measured in this study is complex. It is dependent on many factors such as the deformation of the vertebrae and the location of the instantaneous centres of rotation. Further research is necessary to measure these interface motions.

Conclusions.

- 1) There were no significant differences in the three-dimensional stabilisation provided by the different cage designs.
- 2) All cages significantly stabilised the spine in flexion and lateral bending loading.
- 3) None of the cages stabilised the spine in extension or axial rotation.
- 4) All cage types included in this study provided the greatest stabilisation in flexion-extension and lateral bending when used together with posterior instrumentation.
- 5) In axial rotation, the addition of cross-bracing to the posterior instrumentation with cage condition produced a modest, significant increase in stabilisation.

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