Basic Science

Bilateral pedicle screw fixation provides superior biomechanical stability in transforaminal lumbar interbody fusion: a finite element study

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Abstract

Background context

Transforaminal lumbar interbody fusion (TLIF) is increasingly popular for the surgical treatment of degenerative lumbar disease. The optimal construct for segmental stability remains unknown.

Purpose

To compare the stability of fusion constructs using standard (C) and crescent-shaped (CC) polyetheretherketone TLIF cages with unilateral (UPS) or bilateral (BPS) posterior instrumentation.

Study design

Five TLIF fusion constructs were compared using finite element (FE) analysis.

Methods

A previously validated L3–L5 FE model was modified to simulate decompression and fusion at L4–L5. This model was used to analyze the biomechanics of various unilateral and bilateral TLIF constructs. The inferior surface of the L5 vertebra remained immobilized throughout the load simulation, and a bending moment of 10 Nm was applied on the L3 vertebra to recreate flexion, extension, lateral bending, and axial rotation. Various biomechanical

parameters were evaluated for intact and implanted models in all loading planes.

Results

All reconstructive conditions displayed decreased motion at L4–L5. Bilateral posterior fixation conferred greater stability when compared with unilateral fixation in left lateral bending. More than 50% of intact motion remained in the left lateral bending with unilateral posterior fixation compared with less than 10% when bilateral pedicle screw fixation was used. Posterior implant stresses for unilateral fixation were six times greater in flexion and up to four times greater in left lateral bending compared with bilateral fixation. No effects on segmental stability or posterior implant stresses were found. An obliquely-placed, single standard cage generated the lowest cage-end plate stress.

Conclusions

Transforaminal lumbar interbody fusion augmentation with bilateral posterior fixation increases fusion construct stability and decreases posterior instrumentation stress. The shape or number of interbody implants does not appear to impact the segmental stability when bilateral pedicle screws are used. Increased posterior instrumentation stresses were observed in all loading modes with unilateral pedicle screw/rod fixation, which may theoretically accelerate implant loosening or increase the risk of construct failure.

Keywords: Finite element method; Spine biomechanics; Transforaminal lumbar interbody fusion; Interbody cage; Pedicle screw fixation; Lumbar spine

Introduction

Transforaminal lumbar interbody fusion (TLIF), initially described by Harms and Jerszenszky [1] in the early 1990s, continues to gain popularity as a surgical option to treat degenerative spinal disorders and achieve circumferential arthrodesis from a posterior-only approach. Transforaminal lumbar interbody fusion is commonly supplemented with bilateral pedicle screw instrumentation and is a widely accepted method for maintaining initial segmental stability [2–6]. There have been numerous clinical and in vitro biomechanical studies evaluating the efficacy of the TLIF procedure [7–12], and the addition of bilateral pedicle screws has been found to provide rigid fixation and confer both biomechanical and clinical advantages for TLIF constructs [13].

Advantages of the transforaminal approach include reducing the risk of excessive neural tissue retraction and epidural fibrosis when compared with a wider posterior lumbar interbody fusion (PLIF) exposure, as well as avoiding the potential complications associated with anterior lumbar interbody fusion, such as damage to the great vessels during mobilization or disruption of the presacral sympathetic plexus that may lead to retrograde ejaculation in male patients. Transforaminal lumbar interbody fusion also offers a theoretically lower risk of adjacent segment instability because the posterior laminar arch and posterior longitudinal ligament complex are preserved [14]. Preservation of the posterior tension band may also prevent the retropulsion of the interbody device and bone graft into the spinal canal [14].

Compared with the traditional TLIF procedure, which uses a midline incision and bilateral pedicle screw fixation, advances in instrumentation and tubular retractor systems have allowed the development of minimally invasive techniques that now commonly use a paramedian approach and radiographic guidance to achieve less tissue disruption. Given that discectomy and decompression are typically performed from only one side in these procedures, some authors have advocated further minimizing the tissue disruption with the use of unilateral pedicle screw instrumentation. The use of unilateral instrumentation not only decreases soft-tissue disruption, but also decreases operative time and results in lower implant costs. However, only a small number of clinical reports have evaluated this method. These studies report acceptable stability, reduced blood loss, and decreased operative and hospital time, with no difference in fusion rates or complications [11,13,15–18]. Nonetheless, controversy remains, as some investigators believe unilateral pedicle screw fixation provides inadequate stability that may lead to higher rates of instrumentation failure and pseudoarthrosis [19–22].

Previous biomechanical studies have reported that unilateral fixation after TLIF provides less rotational stability compared with bilateral screw fixation [4,21,22]. Yucesoy et al. [22] performed a biomechanical evaluation that found that unilateral pedicle screw fixation provided inadequate stability with significantly more range of motion (ROM) compared with bilateral fixation in a two-level construct. However, the optimal amount of stability required to promote a spinal fusion remains unknown. In fact, some researchers [23–25] have demonstrated that excessively rigid fixation may produce undesired adverse effects, including device-related osteoporosis because of stress shielding and the absorption of grafted bone [26,27].

In addition to determining the best method for posterior stabilization in TLIF procedures, the optimal method for interbody reconstruction, with regard to the shape and number of interbody cages, also remains unknown. Interbody reconstruction during TLIF is typically performed using a synthetic or metal cage and contributes load sharing capability [28] and stability to the fusion construct. Previous biomechanical studies have shown increased construct stiffness with combined anterior and posterior fixation compared with posterior fixation alone [29,30], with reported fusion rates between 90% and 100% in these reports [31]. In a biomechanical study, Polly et al. [30] assessed the effects of interbody cage placement in a simulated single-level spinal fusion and found that construct stiffness increased dramatically, ranging from 6-fold to 18-fold for different cage positions with posterior fixation compared with implants without posterior fixation.

Because of the load-sharing structural function of the interbody cages, overall construct stability may be affected by the cage shape, material properties, cage orientation, and the number of cages [32]. Numerous in vitro and finite element (FE) studies have evaluated the influence of these factors on the overall construct performance [33–37]. However, these studies did not consider the cage-vertebra interface stresses for the reported TLIF configurations. Different cage

configurations may alter the load transfer mechanics, and thereby affect stress transfer at the cage-vertebra interface. With recent studies showing regional variations in end plate strength [38], there has been an increased interest in applying this information to guide interbody reconstruction methods. In light of the existing information on TLIF approaches, determining an optimal anteroposterior construct is critical to achieving adequate stabilization, reducing instrumentation failure, and potentially optimizing the rates of arthrodesis. The objective of the current investigation was to compare the acute stability of different TLIF reconstructions using an FE model. Using a previously validated L3–L5 FE model modified to simulate decompression and fusion at L4–L5, this study compared the biomechanics of the following TLIF constructs: unilateral cage+unilateral pedicle screw fixation (UC+UPS); unilateral cage+unilateral pedicle screw fixation (CC+UPS); unilateral cage+bilateral pedicle screw fixation (UC+BPS); crescent cage+bilateral pedicle screw fixation (CC+BPS); and bilateral cage+bilateral pedicle screw fixation (BC+BPS). Flexion, extension, lateral bending, and axial rotation and posterior implant and end plate stresses were modeled for each construct.

Methods

A nonlinear, three-dimensional (3D), ligamentous FE model of the L3–L5 segments was used in this study (Fig. 1). The geometry of the bony structures was procured from axial computed tomography scans of a cadaveric spine specimen. Each vertebra was modeled as consisting of a cancellous inner core surrounded by a 1 to 1.5 mm cortical shell using hexahedral (C3D8) elements. A 0.5 mm bony end plate was simulated on either end of each vertebra. The intervertebral disc was modeled as consisting of annulus ground substance surrounding an incompressible nucleus pulposus. The fluid-like behavior of the disc was simulated using 3D hybrid hexahedral elements (C3D8H) with hyperelastic Mooney-Rivlin formulation. Annular fibers were embedded on the ground substance, with these fibers acting at 30° and 150° from the horizontal in eight layers using rebar definition. Fiber cross-sectional area and Young modulus were assigned to decrease from 550 MPa in the outermost layer to 357 MPa in the innermost layer. The articulating facet joints were modeled with surface-to-surface contact elements. A cartilaginous layer was simulated between the facet surfaces using an exponential pressure overclosure relationship. The major ligaments of the lumbar region were incorporated in this model as tension-only 3D truss elements (T3D2) via hypoelastic material definition. The material properties of the various spinal components were derived from the literature as specified in Table.

Fig. 1 Finite element model of an L3–L5 spine segment.

Table Material properties of spinal components

Element set	Element type	Young modulus (MPa)	Poisson ratio
Vertebral cancellous bone	C3D8	100	0.2
Vertebral cortical bone	C3D8	12,000	0.3
Posterior bone	C3D8	3,500	0.25

End plates	C3D8	1,000	0.3
Nucleus pulposus	C3D8H	Hyperelastic (Mooney-Rivlin) 1 ($C_1=0.12$, $C_2=0.03$)	0.499
Annulus (ground)	C3D8	Hyperelastic (Mooney-Rivlin) 1.75 ($C_1=0.2333$, $C_2=0.0583$)	0.45
Annulus fibers	Rebar	357–550	0.3
Facet joint	Contact elements	Exponential pressure overclosure relationship (softened contact)	
Ligaments	Tension-only truss elements		
Anterior longitudinal	T3D2	12.8–15.0	0.3
Posterior longitudinal	T3D2	10.0–20.0	0.3
Capsular	T3D2	8.0–20.7	0.3
Intertransverse	T3D2	10.0–58.7	0.3
Interspinous	T3D2	2.8–5.0	0.3
Supraspinous	T3D2	2.8–7.0	0.3
Ligamentum flavum	T3D2	10.0–15.5	0.3

FE model validation

The intact L3–L5 FE model was validated against the results of previously published studies [39,40] and experimental results for prediction accuracy [41] (the validation manuscript was submitted for publication [42]). Nonlinear behavior of the FE model was verified over the entire moment-rotation curve under the conditions of flexion, extension, lateral bending, and axial rotation [42].

Modeling of implants

Two different shapes of commonly used interbody cages were studied: standard and crescent, with model implant dimensions selected based on the measurements of the cadaveric L4–L5 disc. The fusion constructs simulated in the study were modeled using Abaqus version 6.10 (Dassault Systemes Simulia Corp., Providence, RI, USA). The solid models were then imported into HyperMesh version 10.0 (Altair Engineering, Inc., Troy, MI, USA) and meshed in 3D continuum elements. The FE L3–L5 model was modified to simulate the TLIF surgical procedure at the L4–L5 level via a transforaminal approach by the application of facetectomy (unilateral or bilateral), partial annulotomy, and discectomy. The following TLIF reconstructive options were then applied to the decompressed segment for stabilization: unilateral cage+unilateral pedicle screw fixation (UC+UPS); crescent cage+unilateral pedicle screw fixation (CC+UPS); unilateral cage+bilateral pedicle screw fixation (UC+BPS); crescent cage+bilateral pedicle screw fixation (CC+BPS); and bilateral cage+bilateral pedicle screw fixation (BC+BPS).

Figs. 2 and 3 show the cage and posterior construct configurations adopted in the present study. The simulated standard cage (9 mm height) is box-shaped, with flat superior and inferior surfaces. Two different cage placement options, unilateral and bilateral, were included in this analysis. Cages were inserted into the disc space through the annulus window via an oblique approach for unilateral cage simulation and symmetric to the midline of the annulus for bilateral cage placement. The simulated crescent cage (9 mm height) is banana-shaped, with flat superior and inferior surfaces, the crescent cage was positioned anteriorly and centered on the middle sagittal plane in the disc space. All cage representations were tested in conjunction with either unilateral or bilateral posterior instrumentation. The posterior instrumentation consisted of transpedicular screws (55 mm long, 6 mm diameter) and longitudinal rods (45 mm long, 6 mm diameter) spanning between adjacent screws. For unilateral screw simulations, the left side of the model was assigned to undergo instrumentation. Titanium ($E=110$ GPa) and polyetheretherketone ($E=3.6$ GPa) material properties were defined for the posterior instrumentation and interbody cages [43].

Fig. 2 Transforaminal lumbar interbody fusion cage positions: (Left) unilateral standard cage, (Middle) bilateral standard cages, and (Right) crescent-shaped cage.

Fig. 3 Posterior instrumentation: (Left) unilateral pedicle screw system and (Right) bilateral pedicle screw system.

Contact definitions

A finite sliding algorithm with a coefficient of friction of 0.2 was defined between the cage and end plate to allow for any small relative displacements between the two contacting surfaces. Rigid fixation was simulated using a “Tie” constraint at the following interfaces: pedicle screw and pedicle/vertebral body and pedicle screw and rod. The pedicle screws were placed such that they engaged about two-thirds of the vertebral body.

Loading and boundary conditions

A motion protocol was defined for all reconstructive options and the intact lumbar spine condition. The inferior surface of the L5 vertebra was immobilized throughout the load simulation. The nodes on the uppermost surface of the L3 vertebra were coupled to a reference node for load application. A bending moment of 10 Nm was applied to this reference node on the superior surface of the L3 vertebra to represent movements of flexion/extension, lateral bending, and axial rotation. Abaqus 6.10 (Dassault Systemes Simulia Corp., Providence, RI, USA) was used to perform numerical analyses. The segmental ROM in the intact lumbar condition was used as the baseline. The ROMs across the index (L4–L5) and superior adjacent levels (L3–L4), as well as peak Von Mises stresses in the posterior instrumentation and L4 inferior end plate, were computed and compared for each model condition.

Results

Range of motion

Flexion/extension

The FE analysis of L3–L5 indicated no ROM differences in flexion or extension for the superior adjacent level between any of the simulated fusion constructs and the intact condition (Fig. 4, Top), but there was a significant reduction in ROM at the level of fusion (L4–L5) for all reconstructive configurations when compared with the intact model (Fig. 4, Bottom). The shape and number of interbody spacers did not appreciably affect the ROM achieved with bilateral fixation in flexion or extension. Variation in stability was noted with unilateral posterior fixation when compared with bilateral fixation.

Fig. 4 (Top) Range of motion values at L3–L4 in intact and implanted models, (Bottom) range of motion values at L4–L5 in intact and implanted models. UC, unilateral cage; UPS, unilateral pedicle screws; CC, crescent cage; BPS, bilateral pedicle screw; BC, bilateral cage; flex, flexion; ext, extension.

Lateral bending

Lateral bending revealed no ROM differences in lateral bending at the superior adjacent level between any of the simulated fusion constructs and the intact condition (Fig. 4, Top), but there were differences noted between unilateral and bilateral posterior fusion constructs at the fusion level (Fig. 4, Bottom). In left lateral bending, more than 50% of intact ROM remained after unilateral posterior fixation, compared with less than 10% of intact ROM after bilateral posterior fixation. In right lateral bending, ROMs achieved with unilateral and bilateral posterior fixations compared with intact ROM were less than 25% and 10%, respectively. In both right and left lateral bending, the shape and number of cages did not significantly influence the motion achieved with bilateral fixation.

Axial rotation

Similar to flexion/extension and lateral bending, axial rotation ROM was drastically reduced for all reconstruction techniques compared with the intact condition (Fig. 4). At the level of fusion, axial rotation ROM was reduced by greater than 63% of intact ROM for unilateral fixation and 90% of intact ROM for bilateral fixation. There were no differences in the ROM observed within either the unilateral or bilateral fixation groups between the different types of cages tested. There were no ROM differences in axial rotation at the superior adjacent level between the intact condition and any of the simulated fusion constructs.

Posterior instrumentation stress analysis

Fig. 5 summarizes peak Von Mises stresses in the posterior instrumentation for the study constructs under various loading modalities at 10 Nm. In flexion, the maximum stresses calculated for the caudal screws ranged from 2.7 to 6.3 times greater for unilateral screws compared with bilateral screws. In extension, the maximum stresses for unilateral screws ranged between 1.2 and 1.7 times greater than those for bilateral screws. Similarly, the peak stresses in the unilateral screw constructs were 2.0 to 4.1 times greater than those for bilateral screws in the left lateral bending and 1.4 to 2.4 times greater than those for bilateral screws in right lateral bending. In axial rotation, there was little difference in peak stresses between the unilateral or bilateral posterior

configurations. The calculated stresses for the caudal pedicle screw/rod interface were generally greater for constructs using unilateral fixation compared with those using bilateral fixation. Among the bilateral fixation constructs, BC+BPS demonstrated the greatest pedicle screw stress, followed by CC+BPS and UC+BPS.

Fig. 5 Maximum Von Mises stresses (MPa) in the posterior instrumentation for various transforaminal lumbar interbody fusion constructs: UC+UPS, unilateral cage and unilateral pedicle screws; CC+UPS, crescent cage and unilateral pedicle screw fixation; UC+BPS, unilateral cage and bilateral pedicle screw fixation; CC+BPS, crescent cage and bilateral pedicle screw fixation; BC+BPS, bilateral cage and bilateral pedicle screw fixation.

L4 inferior end plate stress analysis

Fig. 6 demonstrates the peak Von Mises stresses for the L4 inferior end plate obtained for each construct at 10 Nm. For both unilateral and bilateral posterior fixations in all loading modes, the peak end plate stresses calculated for the crescent-shaped cage were greater than those for the obliquely-placed standard cage. In unilateral fixation, the maximum end plate stresses calculated for the CC+UR group exceeded those of the UC+UR group by 1.7 times in flexion, 2.0 times in extension, 1.3 times in left lateral bending, 3.4 times in right lateral bending, 2.0 times in left rotation, and 2.2 times in right rotation. In bilateral fixation, the maximum end plate stresses calculated for the CC+BR group exceeded the UC+BR group by 1.3 times in flexion, 2.5 times in extension, 1.4 times in left lateral bending, 1.7 times in right lateral bending, 1.7 times in left rotation, and 2.1 times in right rotation. Although standard cages generated lower peak stresses, paired cages with bilateral fixation (BC+BR) produced stresses similar to those calculated for the crescent-shaped cage groups. Fig. 7 includes stress contour plots for the L4 inferior end plate for each cage construct, with bilateral fixation in flexion, extension, lateral bending, and axial rotation at 10 Nm. Higher stresses were noted at the L4 inferior end plate under the conditions of flexion, rotation, and lateral bending with crescent-shaped cages compared with the standard cages. For all cage shapes, higher end plate stresses were noted at the area of contact between the end plate and cage surface.

Fig. 6 Maximum Von Mises stresses (MPa) in the L4 inferior end plate for various transforaminal lumbar interbody fusion constructs: UC+UPS, unilateral cage and unilateral pedicle screws; CC+UPS, crescent cage and unilateral pedicle screw fixation; UC+BPS, unilateral cage and bilateral pedicle

screw fixation; CC+BPS, crescent cage and bilateral pedicle screw fixation; BC+BPS, bilateral cage and bilateral pedicle screw fixation.

Fig. 7 Stress (MPa) plots of single standard, paired standard, and crescent-shaped cages with BR fixation under loading conditions at 10 Nm. UC, unilateral cage; CC, crescent cage; BC, bilateral cage; BR, bilateral rod fixation.

Discussion

Transforaminal lumbar interbody fusion is an increasingly popular technique for the treatment of degenerative conditions of the lumbar spine. This procedure is traditionally supplemented with bilateral posterior stabilization using pedicle screws and rods. Although unilateral pedicle screws and rod augmentation has been proposed as an alternative posterior supplement for TLIF procedures, there is limited literature evaluating the biomechanical stability of such constructs compared with bilateral stabilization [20–22,44,45]. There is also continued debate regarding whether there are differences in long-term clinical and radiographic outcomes when comparing unilateral versus bilateral posterior instrumentation after TLIF. Several previous clinical studies have evaluated TLIF with unilateral pedicle screw fixation and reported excellent fusion rates (97%–100%) with no neurologic complications [11,15]. However, these studies were limited by being retrospective and including only small sample sizes with short-term follow-up. More recently, Xue et al. [13] performed a randomized, prospective analysis comparing unilateral versus bilateral posterior instrumentation for single-level TLIF in 80 patients and found no statistically significant differences between bilateral versus unilateral fixation for fusion rates (95.4% bilateral vs. 91.9% unilateral), clinical outcomes, rates of screw failure, or complications. The authors did find significant differences in favor of unilateral fixation in operative time, blood loss, implant costs, and in-hospital time [13]. It is worth noting, however, that this study was likely underpowered to evaluate the differences in rates of screw failure or complications, and there were significant differences in the surgical technique between the two groups. In this study, unilateral fixation was performed using a minimally invasive paramedian approach, whereas bilateral fixation was performed using a traditional midline open approach. In addition to this study, two other randomized, prospective clinical studies have demonstrated no clinical benefit or difference in radiographic outcomes between bilateral and unilateral instrumentation in

lumbar posterolateral fusion [18,46]. Despite these good clinical outcomes reported for unilateral pedicle screw fixation in both TLIF and posterolateral fusion procedures, the scientific community still lacks a well-powered, randomized study to evaluate unilateral versus bilateral posterior fixation in TLIF.

To further examine this question, we conducted the FE analysis to evaluate the stability of the five TLIF constructs under physiologic conditions of loading, assessing motion at the index and an adjacent level, and estimating the stresses experienced by the posterior instrumentation and L4 inferior end plate. After simulated partial facetectomy and complete discectomy performed through a transforaminal approach, our study found that all reconstructive configurations provided a significant reduction in ROM compared with the intact lumbar spine. Bilateral pedicle screw and rod fixation reduced ROM at the index level up to 80% to 99% compared with intact conditions in all loading modes, consistent with the results from previous studies [15,47,48]. In contrast, unilateral pedicle screw and rod fixation reduced flexion 95% and extension in these loading modes from 43% when compared with intact conditions. Of particular note, ROM in lateral bending toward the side of instrumentation was reduced by only 50% by unilateral fixation. Moreover, the calculated stresses for unilateral posterior constructs were greater for every loading mode compared with bilateral posterior constructs, with maximal differences observed during flexion and left lateral bending. Our results suggest that unilateral pedicle screw and rod constructs provide less stability than bilateral constructs, particularly in lateral bending, with no differences in stability related to the shape or the use of one or two interbody cages. These results are consistent with a previous report by Cho et al. [34], who found that the shape, length, and surface profile of interbody cages did not affect construct stability. These results are concerning because inadequate stabilization, increased motion, and/or elevated implant stresses may theoretically accelerate screw loosening or construct failure under repetitive loading.

Based on our findings, the shape and use of one or two TLIF cages did not affect the stability as assessed by ROM of the instrumented level. However, our results demonstrated that the L4 inferior end plate stresses, which represent the cage-end plate interface stresses, vary with cage shape and number. Our results suggest that an obliquely-placed standard cage generates the least amount of end plate stress when combined with either unilateral or bilateral posterior fixation. This result is possibly because of the smaller cage surface area in contact with the end plate, in contrast to the larger area of a crescent-shaped cage. Furthermore, oblique positioning of the standard cage resulted in a further decrease in local stress concentration centrally and provided better structural stability with bilateral rod fixation in all loading modes (Fig. 6). These findings are consistent with those previously published by Tsuang et al. [49] using FE analysis to compare peak cage-end plate interface stresses for single, paired, and oblique PLIF cages, with and without posterior instrumentation. The authors of this study concluded that an obliquely-placed cage with bilateral posterior instrumentation induced less end plate stress than a one-side cage and provided adequate stability similar to paired cages. However, their study did not describe the end plate stresses for unilateral versus bilateral posterior instrumentation and differences in cage geometry were not considered. In another study, Chiang et al. [33] performed an FE analysis of paired and obliquely-placed single PLIF cages with bilateral posterior augmentation and concluded that the single PLIF cage approximated paired cages in its biomechanical stability and generated less stress when compared with paired implants.

Differences in segmental lordosis after nontapered cage implantation, depending on the shape and number of cages placed, were negligible. Thus, the high end plate stresses at the posterolateral aspect of the crescent cage and the posterior margin of the bilateral cage construct (Fig. 7), both of which increased during extension and rotation, likely represent areas of the cage-implant interface that undergo the most consistent contact throughout the segmental ROM. Intuitively, during extension, the posterior aspect of the cage will encounter the highest stress, whereas the posterolateral aspects of the crescent cage may remain the highest edge-loaded portions of the cage. Some surgeons advocate cage placement close to the anterior rim of the disc space, citing increased end plate strength at this location. The purpose of this anterior placement is to attempt to reduce subsidence, but anterior placement of the cage can be technically challenging and risks great vessel injury. Central cage placement is also not ideal because of reduced stability in this position and potential subsidence in this thinner end plate region. However, our results suggest that a crescent-shaped cage positioned on the anterior rim on the interspace stresses the end plate 1.3 to 3.4 times more than a unilateral cage positioned centrally at an oblique angle. For comparison, a biomechanical test performed by Lowes et al. [38] demonstrated that the anterior lateral and anterior central end plate regions of the lumbar vertebrae only require 1.2 and 1.26 times more force, respectively, for failure, compared with the central region of the vertebral body. Although the perimeter of the end plate is stronger than the center, our data suggest that the perimeter end plate stresses caused by the crescent-shaped cage may exceed that difference in strength. This finding calls into question the perceived advantages of an anterior rim cage position. In addition, the obliquely-positioned single cage demonstrated the least amount of pedicle screw stress. Together, these data demonstrating reduced stress on the end plate and the posterior instrumentation and equivalent fixation stability suggest that bilateral screws and rods with a unilateral, centrally-placed oblique cage appears to be the advantageous biomechanical construct.

In this study, as for most FE models and in vitro cadaveric studies, the results are limited by the absence of a consideration of the contributions of the musculature and follower load (upper body weight). In addition, the FE analysis presented did not evaluate the variations in bone mineral density, such as in osteoporotic bone, or in different sizes and positions of the interbody cages. As in similar studies, a lumbar FE analysis was used to better define the biomechanical outcomes of interest with less experimental variation before these results are verified with experimental data. Moreover, we did not specifically evaluate the stability of interbody cages alone, without posterior fixation, as this is an unlikely clinical scenario given the posterior destabilization after the facet resection required for access to the interbody space. Although certain in vitro biomechanical studies have shown promising results with TLIF cages alone [50], the stability afforded by cages when used as stand-alone devices may not be enough to achieve optimal clinical outcome, especially with dorsal structure violation [4,36] and was therefore, not a question of interest in this study.

Also, in this study, the "distraction-compression" principle used in the interbody cage implantation was not simulated. According to this belief, the annulus fibers contract after being placed under tension during disc distraction, producing compression between the cage and the end plates of the vertebral bodies and maintaining its position [51]. Further biomechanical and FE model analyses are necessary to evaluate the effects of pedicle screw instrumentation after TLIF for osteoporotic specimens and multilevel reconstructions. The findings of this study provide an insight into the biomechanical effects of current TLIF fusion constructs and support the need for additional in vitro analyses and a large, randomized, prospective trial comparing these constructs.

The most noteworthy finding from this study is that our results stand in general contradistinction to the published clinical literature regarding unilateral and bilateral posterior instrumentation in TLIF. As noted previously, some clinical literature comparing these two techniques have failed to find any major differences in outcomes, nor has any difference in fusion rates been noted [13,18,52], and the advent of minimally invasive TLIF techniques make unilateral posterior fixation an attractive tool [53]. However, as Hu et al. [52] note in their metaanalysis of these techniques, all of these studies rely on small sample size and vary in important characteristics—such as different functional outcome measures—that make any real, generalizable conclusions impossible. Other studies, however, have found that unilateral constructs may be subject to tensile coupled motion secondary to construct asymmetry, and therefore, may fail to provide enough stability after complete disc decompression [23,54]. Unilateral pedicle screw fixation also may only provide 50% the stiffness of bilateral fixation, especially in rotational motion [21]. Clinically, Aoki et al. [44] showed via their prospective, randomized controlled trial that unilateral fixation may result in less improvement in patients' symptoms, particularly back pain, leg pain, and lower extremity numbness. In fact, unilateral pedicle screw fixation may increase the risk of cage migration after TLIF [45], and bilateral screw fixation may demonstrate better long-term outcomes (up to 2-year follow-up) than unilateral fixation, despite improved perioperative results with less invasive unilateral screw fixation [55]. Our findings of decreased construct stiffness with unilateral fixation may support these clinical data that in some patients, bilateral pedicle screw fixation after TLIF is the most appropriate surgical option.

In conclusion, for single-level TLIF procedures, augmentation with bilateral posterior fixation offers greater immediate stability compared with unilateral posterior fixation. Moreover, the shape and number of interbody implants did not appear to greatly impact the segmental stability when bilateral pedicle screws were used. Based on our findings, an oblique, single TLIF cage augmented with bilateral posterior fixation provides maximal stability, while minimizing posterior instrumentation and end plate stresses. Further study is needed to determine if the biomechanical advantages of this construct lead to improved clinical outcomes, such as lower rates of cage subsidence, increased fusion rates, or reduced cost because of fewer hardware failures or complications.

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