Biomechanical Analysis of Various Footprints of Transforaminal Lumbar Interbody Fusion Devices

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Abstract

Study Design: A biomechanical Finite Element modeling study of human lumbar spine.

Objective: To evaluate the effects of transforaminal interbody device’s footprint on lumbar spine biomechanics to further examine the potential subtle biomechanical differences not captured in previous studies.

Summary of Background Data: In recent years, the evolution of interbody fusion devices has provided the surgeons with a multitude of options. An articulating TLIF device is developed in order to overcome the surgical challenges associated with insertion of a large footprint interbody device through a small incision.

Methods: Finite element model of L3-S1 lumbar segment was modified to simulate replacement of various TLIF constructs with different cage designs including an articulating vertebral interbody TLIF device (AVID) and a generic TLIF device placed in different configurations. The instrumented models were subjected to 400N follower load along with 10 Nm bending moment at different physiological planes. The kinematics, loads and stresses were compared among various models.

Results: Simulated cage designs provided similar kinematical stability within the treated segments. However, the articulating and the double TLIF implants allowed for better load sharing through the anterior column. These implants resulted in lower endplate and pedicle screw stresses and more homogenous stress distribution concentrated across the peripheral region of the endplate.

Conclusions:
An articulating, large footprint, peripherally placed TLIF device affords substantial biomechanical advantages. This device may be able to reduce the incidence of subsidence due to its ability to reduce and distribute the endplate stresses in the stronger peripheral region. It may also reduce the posterior hardware failure incidence owing to its ability to reduce the screw stresses as compared to traditional TLIF. Although double TLIF was demonstrated similar biomechanical advantages as AVID, complications associated with double TLIF (i.e. larger surgical incision, longer surgical procedure, placement and alignment challenges) support AVID as a better optimized alternative.

Key words: TLIF cage, Lumbar spine, Finite element modeling, Biomechanics, Load sharing.
Introduction:

Lumbar interbody fusion has become state-of-the-art procedure to treat segmental instability and associated low back pain. The main aims of this procedure are to restore functionality and relieve pain by correcting mechanical deformity, and stabilizing the motion segment. The biomechanical goals of lumbar interbody fusion are to restore alignment, restore intervertebral and foraminal height, and promote fusion by providing immediate stability (ability to limit segmental motions under physiologic load). Interbody cages play an important role in the biomechanical outcome of the fusion process through preservation of the disc height as the bone graft incorporation occurs. Interbody cages can aid in achieving these goals, especially bone graft resorption/disc height loss. Dennis et al. demonstrated that stand-alone constructs for anterior lumbar interbody fusion utilizing bone graft alone are associated with high rates of subsidence as high as 100%. Further it has been shown that the loss of disc height as a result of subsidence will adversely affect the foraminal size and any mechanical deformity correction.

Various types of interbody fusion devices are available and commonly used nowadays. These devices are inserted posteriorly, anteriorly or laterally and are often supplemented with posterior instrumentation (pedicle screw and rod systems) to enhance the segmental stability and facilitate the fusion process within the disc space. A multitude of surgical procedures (i.e. TLIF, PLIF, ALIF) have been developed in order to place an interbody implant in the disc space to stabilize the segment. Transforaminal lumbar interbody fusion (TLIF) is one of the common surgical procedures in order to place the cage through an unilateral approach.
In a normal person, about 80% of compressive load passes through the vertebral bodies, and the remaining 20% passes through the posterior elements.\textsuperscript{2, 36} Considering this load distribution pattern, a considerable amount of load will be shared by the interbody device following implantation. Based on Wolff’s law, this exerted compression on the implanted interbody device filled with bone graft will result in enhanced bone fusion. A large footprint interbody device will act to enhance segmental stability compared to the one with smaller footprint. Also, it may be theoretically hypothesized that a larger footprint interbody device will reduce the stress risers in the bone-implant interface and lead to more homogenous stress distribution across the vertebral endplate region.

Human anatomy and TLIF surgical techniques pose great challenges when it comes to inserting a large footprint interbody device through a small incision. In order to overcome this issue, new articulating TLIF devices have been developed. These devices use the same incision size as the traditional TLIF implants; however, they articulate inside the disc space allowing for the placement of a relatively large implant in the disc space. Articulating vertebral interbody device (AVID) is one such recently developed implant which has limited clinical history as well. The AVID device is inserted into the disc space in a way similar to that of insertion of any regular TLIF device. However, the AVID device has articulating capabilities which make it possible to insert such a large footprint device using the TLIF approach. The sequential articulation used for inserting the device is depicted in Figure 1. Using the special insertion instrument, surgeon can insert and articulate the device in the disc space to achieve the desired orientation.

In a retrospective study, McClellan et al. followed 51 patients who underwent TLIF procedure using a large footprint AVID device.\textsuperscript{26} The patient’s average preoperative lumbar back pain
using the visual analogue scale was compared to their average postoperative lumbar pain at 4 weeks and 3 months. The authors concluded, “We feel that the larger footprint and peripheral placement of the cage played a role in our ability to achieve excellent pain control by the first postoperative visit”.

Additionally, previous biomechanical studies have also advocated the use of large footprint interbody devices to achieve superior stability. It was proposed that these devices would avoid the incidence of subsidence as well. Tsitsopoulous et al. have also demonstrated that anteriorly placed TLIF implants provide better construct stability.

In order to gain deeper insight, the current study uses a 3-dimensional (3D), experimentally validated finite element (FE) model of L3-S1 spine segment to study the biomechanical effects of various footprint TLIF interbody cages. We hypothesized that a large footprint, peripherally placed interbody device would provide superior stability to the segment and would lead to lower stresses in the endplate region, potentially minimizing the incidence of subsidence. Moreover, it will also share higher load through the disc space resulting in lower loads passing through the posterior instrumentations, potentially minimizing the likelihood of posterior instrumentation failure.

**Materials and Methods**

An experimentally validated non-linear ligamentous FE model of the L3-S1 spine was used. This model has been widely validated and has been used in a number of biomechanical studies to investigate diverse spine related complications (Figure 2). This FE model contains all main physiological features of the intact spine which includes vertebral bodies, all seven ligament groups, intervertebral
disc components and posterior facet joints and the capsules. A concise description of the model specifications and characteristics is outlined in the following section.

**FE Model Specifications**

The FE model was developed based on imaging data of a normal healthy adult free from any signs of abnormalities, disorders and previous surgeries. The high resolution transverse computational tomography (CT) images (1.5 mm slice thickness) were then imported into the ImageJ software package (National Institutes of Health, Bethesda, MD, USA) in order to develop the eight node hexahedral mesh. The mesh was then imported into Abaqus FE package (Dassault Systems, Simulia Corp., Providence, RI, USA) to define individual sections representing physiological components of the spine. The final developed FE model consists of all main physiological features of the intact spine including vertebral bodies (with cortical and cancellous regions), ligaments, intervertebral disc components and posterior facet joints and the capsules. While the boney structures and intervertebral discs were modeled using 3D elements, the ligamentous structure was simulated using uni-axial truss elements with prescribed cross sectional area. The intervertebral disc was modeled as two individual components representing Nucleus Pulposus and Annulus Fibrosus. The Nucleus Pulposus was simulated using 3D hexagonal elements with incompressible fluid behavior. The Annulus component was defined as a composite structure including a ground substance embedded with concentric layers of fibers. The fibers were oriented at 30° alternating angles to the horizontal with increased thickness and stiffness in radial direction. Further, the fiber properties were set to just take tensile loads. The Gap elements were used to define the facet joints at each level. These elements would provide compression resistance, with a non-linear stiffness, between nodes along a single direction as the facet joint closes. Proper material properties adopted from literature (Table 1) along with appropriate interactions and constrains were
assigned to each component in order to closely mimic the physiologic conditions. Subsequently, the
developed model was extensively validated against cadaveric measures of kinematics and intra-discal
pressure under a wide range of loading conditions.\textsuperscript{11, 14, 21} In a recent study we have utilized the current
model along with cadaveric experiment to compare the effect of interbody fixation with lateral cage on
segmental biomechanics.\textsuperscript{22} The kinematic data of intact and instrumented case derived from FE model
was well in agreement with the experiments in all loadings (Table 2).

\textit{FE Analysis}

\textit{Surgical Procedure and Implantation}

The developed FE model of the intact lumbar segment was modified to simulate the surgical procedure
for placement of TLIF interbody cage along with supplemental posterior instrumentation. For this
purpose, the 3D CAD models of AVID, an articulating TLIF inter body cage (Custom Spine Inc.,
Parsippany, NJ, USA), and a traditional TLIF interbody cage device along with the models of a pedicle
screw-rod fixation construct were acquired (Figure 3). Each component was meshed and imported into
the FE package. The polyether-ether-ketone (PEEK) material property (E=3.4 GPa, $\mu=0.4$) was assigned to
all cage models while the material properties of a Titanium alloy (E=115 GPa, $\mu=0.34$) was assigned to
the components of the posterior instrumentation (screw-rod system).

Further, the following surgical scenarios were simulated at L3-L4 segment:

\textbf{AVID TLIF}: A novel articulating TLIF implant which was peripherally placed in the disc space.

\textbf{Double TLIF}: Two traditional TLIF implants placed in the disc space.

\textbf{Symmetric Single TLIF}: A Single traditional TLIF implant placed symmetrically in the disc space.
Asymmetric Single TLIF (Asym. TLIF): A single traditional TLIF implant placed asymmetrically in the disc space.

The simulated surgical procedure for TLIF cases included the removal of entire nucleus, part of annulus at the posterior-left side and excision of the left facets. A specific methodology was employed in order to simulate realistic surgical procedure for the placement of each cage into the spine segment. In this method, the cage was fitted into the intervertebral area through distraction of the segment, as performed during surgery to tightly fit the implant into the disc space. For this purpose, the cage was aligned to the inferior endplate (L4), then the segment was distracted until the superior endplate (L3) came to a height slightly above the superior surface of the cage. Further, a rough friction (with infinite coefficient of friction, representing rigid bounding) was defined between the cage and the endplate and the segment was set free allowing for the upper vertebra to sit on the cage (Figure 4). This procedure guaranteed that a compressive load was applied to the cage after placement into the intersegmental area, as occurs in the actual surgical procedure. All the cages were filled with cancellous bone to simulate the embedded bone graft within the implanted cage. Following the interbody cage placement, the posterior instrumentation was performed by adding the screw-rod fixation construct posterior to the treated segment into the pre-defined pedicle holes (Figure 5). Figure 5 shows a posterior lateral view of the simulated surgery plus a top view of the inferior endplate (L4) of the index level implanted with various interbody cages.

**Boundary Conditions**
Subsequent to implantation the physiological loading conditions were applied to the intact and all implanted models. For this purpose the most inferior surface of the S1 vertebra was fixed in all degrees of freedom and the segment was subjected to a 400N compressive follower pre-load. Further, additional 10 Nm bending moments were applied to simulate physiological flexion (Flex), extension (Ext), left and right lateral bending (LB & RB) and left and right axial rotation (LR & RR) as demonstrated in Figure 6.

Finally, The range of motion, normal load and the peak stresses at the vertebral endplates as well as the peak stresses in the screws were computed and compared among the simulated cases.

Results

The range of motion of the implanted level is plotted in Figure 7. As demonstrated, all simulated interbody devices were able to substantially reduce the segmental range of motion compared to the intact. Intact spine range of motion ranged from 2.2˚ to 4.9˚ in various modes of loading whereas simulated fusion fixations reduced the range of motion to less than 1˚. Only asymmetric TLIF allowed for more than 1˚ of motion (during right lateral bending), which was still much less than intact range of motion (4.5˚). The normal load exerted on the inferior endplate (L4) in the implanted models are presented in Figure 8. The maximum normal load value was observed under flexion, while the minimum value occurred under extension in all cases. The load range was 149-532N in AVID TLIF, 185-570N in Double TLIF, 30-396N in Single TLIF and 122-438N in Asymmetric TLIF. Moreover, the peak stresses on the inferior endplate under different loading conditions ranged from 48-110 MPa in AVID TLIF, 76-172 MPa in Double TLIF, 74-168 MPa in Single TLIF, and 102-238 MPa in Asymmetric TLIF model respectively.
as shown in Figure 9. In order to demonstrate the differences in stress distribution across the endplate under various cases, a typical stress contour plot is shown in Figure 10. This figure depicts the high intensity stress risers in the single TLIF cases whereas low stress riser regions were found in the Double TLIF and AVID cases. Moreover, the stresses in the AVID TLIF cases were distributed on the peripheral region of the endplate.

The peak stress across the pedicle screws in different simulated models are presented in Figure 11. The stresses were minimum in compression and maximum in lateral bending. The range of peak stresses was 41 to 91 MPa in AVID TLIF, 42 to 102 MPa in Double TLIF, 47 to 107 MPa in Single TLIF, and 54 to 133 MPa in Asymmetric TLIF model.

**Discussion**

In recent years, the evolution of interbody fusion devices has provided the surgeons with a multitude of options. TLIF approach has gained traction as the cage is inserted through an unilateral route avoiding significant complications associated with other surgical approaches such as dural sac retraction during PLIF approach. Biomechanical studies have shown that interbody fusion devices, especially when augmented with posterior screw/rod systems, may lead to improved segmental stabilization and eventually promote fusion.  

Cho et al. conducted a series of biomechanical experiments comparing straight and banana shaped cages to understand the effects of cage designs on the construct stability. They concluded that “The experimental results suggest that the geometry of cages, including shape (banana or straight), length, and surface profile (bi-convex or flat), does not affect construct stability when the cages are used in
conjunction with posterior fixation. With posterior fixation and surface serration, cage migration was minimal under cyclic loading for both biconvex and flat cages. Tsitsopoulos and associates investigated the effects of anatomically shaped interbody cages on potential superior stability. They found that “The parallel-shaped and anatomically shaped cages provided similar stability in a PLIF construct. The greater stability of the TLIF construct was likely due to a more anterior placement of the TLIF cage and preservation of the contralateral facet joint”. Although these studies discuss segmental stability (decreased range of motion), the issues associated with load sharing and stress distribution (i.e. subsidence, hardware failure) have been neglected. Due to critical role of these factors in the long term success of the fusion procedure, a thorough investigation of the effect of the interbody device design on the resultant load sharing and stress distribution patterns are necessary.

The current study utilizes FE method which has the ability to compute additional parameters such as loads and stresses that are challenging if not impossible to be obtain experimentally. This study aimed to evaluate the effects of interbody device’s footprint on the overall biomechanics including stability, loads and stresses in order to further examine the potential subtle biomechanical differences not captured in previous studies. We hypothesized that a large footprint, peripherally placed TLIF interbody device would provide improved biomechanics (as measured by anterior/posterior load sharing and endplate stresses) as compared to the traditional TLIF devices. Multiple TLIF interbody fusion procedure coupled with posterior instrumentation were simulated with the main focus on a novel articulating (AVID) interbody fusion device. This recently developed cage allows for peripheral placement of a substantially large footprint device (490 mm² footprint area) owing to its ability to articulate within the disc space. At the same time, TLIF techniques using traditional implants (280 mm² footprint area) were also simulated. For instance, traditional TLIF implants placed symmetrically (as recommended) and asymmetrically (as happens on many occasions when surgeon is not able to maneuver the implant in disc space) were simulated. Moreover, double TLIF (two TLIF implants inserted using the same incision to achieve greater
stability) procedures was also simulated in order to understand the effects of increasing the footprint (560 mm² footprint area).

Our findings suggest that all the cage geometries provided sufficient biomechanical stability to the segment by substantially restricting the range of motion to less than 1°. Moreover, we found that only asymmetrically placed cage allowed for motion slightly larger than 1° (still much less than intact range of motion) due to its biased interbody placement which provided greater stability in left lateral bending than in right lateral bending. Facetectomy also contributed to the differences between left and right lateral bending/axial rotation motions. Our findings are congruent with those of Tsitsopoulos et al.³³ and Cho et al.⁷ with respect to the range of motion of the various models. It appears that regardless of cage design, all the posteriorly stabilized simulated cases resulted in similar biomechanical stability (decreased range of motion).

Stress distribution patterns across the L4 inferior endplate demonstrate a remarkable distinction between AVID and rest of the TLIF devices (figures 9 and 10). The peak stress generated with the AVID device was consistently lower than single TLIF cases. In few loading modes such as left rotation and flexion, stresses with single TLIF were as much as 3 to 4 times higher than stresses with AVID device. In addition, endplate stresses with asymmetric TLIF device were substantially higher than the rest of the cases which may be attributed to the asymmetric placement of the device resulting in disproportionate load distribution within the disc space.

The peripheral placement of AVID device plays an important role in distributing the stresses over a large area as well. This device is placed such that the majority of the contact with the endplate is over the peripheral region which has been shown to be relatively stronger compared to the central region.²⁴ Such a strategic placement provides a direct biomechanical advantage to the device.
Fukuta et al. conducted a retrospective analysis of factors related to kidney-type interbody spacer subsidence in TLIF.\textsuperscript{13} They studied the effect of cage/spacer positioning on the subsidence rate. Of the 82 spacers, 66 were located in the center of the disc space and 16 anteriorly. 18 subsidence cases were reported, mostly associated with central placement and old ages of the patients. These authors concluded that “kidney-type spacers should be located in the anterior portion of the disc space to prevent subsidence of the intervertebral body, especially in elderly population”. Similarly, Tan and colleagues studied various shapes and sizes of interbody devices.\textsuperscript{32} They reported: “larger devices should be used to increase contact area and subsequently increase failure load and therefore reduce device subsidence.”

In addition, Grant et al. studied the structural properties of the lumbosacral vertebral endplates.\textsuperscript{15} They warned against the potential pitfalls of the current state-of-the-art practices: “the center of the bone, where implants are currently placed, is the weakest part of the lumbar endplates and is not the strongest region of the sacral endplate”. Furthermore, Polikeit et al. also recommended peripheral placement of the interbody device for optimal biomechanical performance.\textsuperscript{29, 30} Also as mentioned above, Tsitsopoulos et al. also demonstrated the biomechanical advantages of anterior placement of the TLIF cage.\textsuperscript{33}

Therefore, a peripherally placed device such as AVID has clear biomechanical advantages compared to other TLIF devices which are more or less centrally placed. The endplate stress data from the current study corroborates the recommendation made by other investigators.\textsuperscript{13, 15, 29, 30, 32, 33} The data shows that peripheral placement of the large footprint device not only lowers stress riser points but also shifts the stress risers to the strong anterior/peripheral region of the endplate.

As expected, the load passing through the endplate are higher for the large footprint interbody devices including double TLIF and AVID. This is in turn due to the large contact area that these devices share.
with the vertebral endplates. Double TLIF, owing to largest footprint, allowed for maximum endplate loads which was closely followed by AVID device. Theoretically, the devices allowing for higher load sharing through anterior column would reduce the load sharing through the posterior hardware system (Figure 11).

Interbody devices are usually stabilized by supplemental screw/rod systems which provide adequate stability and a conducive environment for bone union to occur. A larger interbody device allows for greater load sharing through the disc space thereby reducing the loading on the posterior hardware. The results of the current study demonstrate that the larger footprint interbody devices reduce the peak stresses in the pedicle screws used in posterior instrumentation (figures 9 and 10). In most loading modes, screw stresses were in the same ballpark for both AVID and double TLIF cases. On the other hand, screw stresses were substantially higher (up to 48% increase compared to AVID case) for the asymmetric TLIF case due to its skewed placement in the disc space.

The screw stress data indicates that the design of the interbody devices may have significant influence on the posterior hardware. The integrity of the posterior hardware such as pedicle screws has implications on the older patients who have poor bone quality. These patients with poor bone quality are at higher risk of pedicle screw loosening due to compromised purchased of the screw in osteoporotic bone. Screw loosening may potentially lead to suboptimal clinical outcome. An interbody device which allows for offloading of the posterior fixation system may particularly be beneficial to the patients with osteoporotic or osteopenic bone. As a result, a device such as AVID may potentially improve the surgical outcome of TLIF procedure by reducing posterior hardware related complications.

The current FE study demonstrates the biomechanical advantages of a peripherally placed large footprint interbody TLIF device (AVID). At the time of preparation of this manuscript, there is only one published clinical study involving the AVID device. The study was presented by McClellan et al. who
reported short term follow up results of their patients. The authors found good clinical outcome in the patients who received AVID TLIF device. More clinical data is warranted to fully support the findings of our study.

Limitations

As with all the computational studies, there are inherent limitations associated with current study. These limitations included simplifications in the model geometry and simulation of mechanical characteristics of its components. In this study, the nucleus component was removed at the index level and the segment was subsequently stabilized. Absence of these structures was not expected to have a significant effect on the biomechanical predictions of the stabilized models. Lack of musculoskeletal structure may also affect the results. Nonetheless, a compressive follower pre-load was added to the models to simulate the muscle effects. Also, the model was validated against the data obtained from cadaveric experiments of the specimens from the elderly group donors in which the tissue characteristics might be different from those of an intact spine segment as simulated in the FE model. Nevertheless, the biomechanical data presented in this study should be viewed as a comparative analysis among different surgical cases and the intact model due to the inherent limitations of the model.
Conclusions

The current FE study attempted to delineate the biomechanical differences in various types of TLIF devices. We hypothesized that the large footprint devices would result in superior biomechanics. In conclusion, all the simulated cases were able to substantially stabilize the indexed segment (reduced range of motion), however only AVID intervertebral fusion device possessed substantial biomechanical advantages. This device resulted in higher anterior column load sharing and lower endplate stresses. Moreover, such a device also lowered stresses in the posterior pedicle screws. Therefore, a large peripherally placed interbody device may be able to reduce the incidence of subsidence and posterior hardware failure after TLIF procedure. Although double TLIF was demonstrated similar biomechanical advantages as AVID, complications associated with double TLIF (i.e. larger surgical incision, longer surgical procedure, placement and alignment challenges) support AVID as a better optimized alternative.
References:


Tables

**Table 1**: Assigned material properties to the components of the FE model.

**Table 2**: FE predicted versus experimentally measured segmental kinematics under different conditions.

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**Figure 1**: (A) Cross section of the disc space with the AVID device being inserted using the inserter instrument; (B) First device articulation after the device is inserted about halfway; (C) Second (final) device articulation after the device is pushed in further resulting in a C shaped implant in the disc space.

**Figure 2**: Finite Element model of the human L3-S1 spine.

**Figure 3**: Simulated models of TLIF interbody devices along with posterior instrumentation.

**Figure 4**: Simulated surgical procedure used to place fusion devices. (A) Segment distraction, (B) Cage placement, (C) Segment was set free allowing for the upper vertebra to sit on the cage, and (D) Posterior instrumentation.

**Figure 5**: FE spine with simulated TLIF surgical procedure with interbody cage plus posterior instrumentation (Top). Positioning of cage on the endplate for AVID, Double TLIF, Symmetric TLIF and Asymmetric TLIF surgical cases (Bottom).
**Figure 6:** Load application and simulated boundary conditions.

**Figure 7:** Treated level (L4-L5) Range of motion in various modes of motions including flexion-extension, lateral bending and axial rotation. The motions are presented in response to the compression plus bending moment (10 Nm) applied in anatomical planes.

**Figure 8:** Normal load exerted on the inferior endplate of superior vertebrae of the index segment (L4) for different surgical cases. The loads are presented for pure compression (400 N) and compression plus bending moment (10 Nm) applied in anatomical planes.

**Figure 9:** Peak von mises stress experienced by the inferior endplate of the superior vertebrae of the index segment (L4) for different surgical cases. The loads are presented for pure compression (400 N) and compression plus bending moment (10 Nm) applied in anatomical planes.

**Figure 10:** Von mises stress distribution on the inferior endplate of the superior vertebrae of the index segment (L4) for different surgical cases in flexion, compression and extension loadings.

**Figure 11:** The peak von mises stress observed by the screws of the posterior instrumentation system in different surgical cases at pure compression (400 N) and compression plus bending moment (10 Nm) loading scenarios.
Table 1: Assigned material properties to the components of the FE model.

<table>
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<tr>
<th>Component</th>
<th>Element Formulation</th>
<th>Modulus</th>
<th>Poisson's Ratio ((\nu))</th>
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<td><strong>Boney Structure</strong></td>
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<td></td>
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<tr>
<td>Vertebral Cortical Bone</td>
<td>Isotropic, Elastic, Hex elements</td>
<td>E=12,000 MPa</td>
<td>0.3</td>
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<td>Isotropic, Elastic, Hex elements</td>
<td>E=100 MPa</td>
<td>0.2</td>
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<tr>
<td>Posterior Cortical Bone</td>
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<td>E=12,000 MPa</td>
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<tr>
<td>Posterior Cancellous Bone</td>
<td>Isotropic, Elastic, Hex elements</td>
<td>E=100 MPa</td>
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<tr>
<td><strong>Intervertebral Disc</strong></td>
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<td>Annulus (Ground Substance)</td>
<td>Neo-Hookian, Hyperelastic, Hex elements</td>
<td>C10=0.348, D1=0.3</td>
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<td>Rebar elements</td>
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<td><strong>Ligaments</strong></td>
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<td>Anterior Longitudinal</td>
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<td>Posterior Longitudinal</td>
<td>Tension-only, Truss elements</td>
<td>10.0(&lt;11%), 20.0(&gt;11%)</td>
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<td>Ligamentum Flavum</td>
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<td>Intertransverse</td>
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<td>Capsular</td>
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<td><strong>Joint</strong></td>
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<tr>
<td>Apophyseal Joints</td>
<td>Non-linear Soft contact, GAPPUNI elements</td>
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Table 2: FE predicted versus experimentally measured segmental kinematics under different conditions.

<table>
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<tr>
<th>Simulated Cases</th>
<th>Range of Motion (degrees)</th>
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<td></td>
<td>Flex</td>
<td>Ext</td>
<td>LB</td>
<td>RB</td>
<td>LR</td>
<td>RR</td>
<td>Exp</td>
<td>FE</td>
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<tr>
<td>Intact</td>
<td>7.1±2.8</td>
<td>3.2</td>
<td>5.0±1.7</td>
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<td>2.6±1.8</td>
<td>2.4±1.7</td>
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<td>Cage Alone</td>
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<td>1.7±1.1</td>
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<td>1.7±1.0</td>
<td>0.6</td>
<td>0.9±0.5</td>
<td>0.2</td>
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<tr>
<td>Cage + Posterior Instrumentation</td>
<td>0.8±0.7</td>
<td>0.4</td>
<td>0.8±0.5</td>
<td>0.3</td>
<td>0.8±0.4</td>
<td>0.3</td>
<td>0.5±0.4</td>
<td>0.2</td>
</tr>
</tbody>
</table>
Fig 1

- **Insertion into the disc space** (A)
- **First articulation with advancement** (B)
- **Second articulation with advancement** (C)
Fig 3

Articulating TLIF Cage (AVID) (PEEK: E=3.4 GPa, U=0.4)

Traditional TLIF Cage (PEEK: E=3.4 GPa, U=0.4)

Posterior Instrumentation Components (Titanium: E=115 GPa, U=0.34)
Fig 9

L4 Inferior Endplate Stress (MPa)

- AVID TLIF
- Double TLIF
- Single TLIF
- Asym. TLIF

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