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1 **Biomechanical analysis of segmental lumbar lordosis and risk of cage**
2 **subsidence with different cage heights and alternative placements in**
3 **transforaminal lumbar interbody fusion**

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2 **subsidence with different cage heights and alternative placements in**
3 **transforaminal lumbar interbody fusion**

4
5 **Abstract**

6 Cage subsidence in transforaminal lumbar interbody fusion (TLIF) is one of the
7 concerns. The objective was to numerically assess the resulting segmental lumbar
8 lordosis (SLL) and stresses at the bone-cage interface as functions of cage height
9 (8- vs. 10-mm) and cage placement (oblique asymmetric, vs. anterior symmetric)
10 for normal and. osteoporotic bone quality. A L4-L5 detailed finite element model
11 of TLIF.was subjected to the functional loadings of 10 Nm in the physiological
12 planes after the application of a 400 N follower-load. The SLL was increased by
13 0.9° (11%) and 1.0° (13%), respectively in oblique asymmetric and anterior
14 symmetric cage placement with 8-mm height; they were 1.4° (18%) and 1.7° (21%)
15 for the 10-mm cage. The maximum stresses at the cage-bone interface, in normal
16 bone model, were increased up to 16% and 41% with the 10-mm cage and
17 asymmetric oblique placement, respectively, and they increased up to 16% and
18 43% in osteoporotic bone model. The greater cage resulted to a higher simulated
19 SLL. Oblique asymmetric placement and the use of a greater cage may increase
20 the risk of cage subsidence. Due to the lower mechanical strength of osteoporotic
21 bone, the risk of cage subsidence should be higher.

22
23 Keywords: finite element analysis, TLIF (Transforaminal lumbar interbody
24 fusion), cage subsidence, interbody cage, biomechanics, spine

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1 **Introduction**

2 Transforaminal lumbar interbody fusion (TLIF) is a surgical procedure to restore the
3 intervertebral body height, the lumbar lordosis (LL), and spinal stability. This involves
4 the removal of the nucleus pulposus (NP) and a portion of the annulus fibrosus (AF),
5 followed by decompression of the segment and the placement of an interbody cage
6 through a unilateral approach. This is aimed at achieving an anterior interbody fusion in
7 addition to the posterior one by a solid segmental fixation (Agrawal and Resnick 2012;
8 Gum et al. 2016).

9 One of the mechanical complications of the TLIF surgical intervention is cage
10 subsidence, a situation where a cage enters into the vertebral body and consequently
11 results in the loss of intervertebral body height and segmental lumbar lordosis (SLL)
12 created intraoperatively by the instrumentation. Cage subsidence rates associated with
13 different cage designs and surgical techniques were reportedly from 10% to 38% (Le et
14 al. 2012; Malham et al. 2015; Lee et al. 2017). A cadaveric experimental study reported
15 that the subsidence stiffness and subsidence force with TLIF were significantly lower
16 ($p < 0.01$) than those for anterior (ALIF) and lateral (LLIF) lumbar interbody fusion
17 (Palepu et al. 2019). Several risk factors of cage subsidence in TLIF have been
18 identified, such as cage geometry (shapes and sizes) (Cho et al. 2008; Le et al. 2012;
19 Agarwal et al. 2013; Faizan et al. 2014; Kim JT et al. 2015; Deng et al. 2016; Kim CW
20 et al. 2016), single cage vs. paired cages (Xu et al. 2013), use of unilateral posterior
21 fixation vs. bilateral one (Chen et al. 2012; Ambati et al. 2015), and trabecular bone
22 volume fraction (Palepu et al. 2019).

23 Biomechanical analysis using finite element models showed that a larger cage
24 footprint (e.g., 490 vs. 280 mm²) allowed to bear about 300% more functional load and
25 reduced the maximum stresses by about 50%, resulting in a lower risk of cage

1 subsidence (Faizan et al. 2014). Biconvex shapes were shown to allow better cage
2 fitting, but with loads more concentrated in the medial region of the endplates with
3 relatively lower mechanical strength than peripheral cortical bone, thus higher risk of
4 cage subsidence (Cho et al. 2008). Using paired- vs. single-cage configurations resulted
5 in 55.2% lower stress at the bone-cage interface (49.77 vs. 77.23 MPa) and
6 subsequently lower risk of cage subsidence (Xu et al. 2013).

7 A thicker cage is generally more effective for SLL restoration, but requires more
8 intervertebral distraction for its placement, which increases the risk of cage subsidence
9 due to the higher compressive forces at the bone-cage interface (8.8 N with 6-mm cage
10 vs. 21.5 N with 8-mm cage in a biomechanical experiments using cadaveric lumbar
11 spines) (Truumees et al. 2008; Le et al. 2012). Clinical studies showed that a kidney-
12 shape cage placed 16% more anteriorly vs. a medial placement of a bullet-shape cage,
13 resulted in an SSL increase of 2.11° (Kim JT et al. 2015) and reduced the risk of cage
14 subsidence by shifting the bone-cage contact more to the peripheral region of the
15 endplates with superior mechanical strength. The results of a controlled cadaveric test
16 reported that using shorter cage (with length of 22- vs. 27-mm) in TLIF can potentially
17 restore the segmental lordosis up to 8.7° (Robertson et al. 2018). Wedged cages (vs. flat
18 cages) are reported to allow better lordosis restoration; increasing the wedge angle from
19 4° to 15° , and increased the resulting SLL from 2.6° to 6.5° (Hong et al. 2017).

20 Clinical studies, experiments using cadaveric spines, and numerical analyses
21 have been done on the use of interbody cages of different shapes, configurations, and
22 heights. However, the effects of essential cage parameters are not yet fully understood;
23 therefore, systematic biomechanical investigations remain to be performed to acquire
24 comprehensive biomechanical facts to help realize and reduce the risk of cage
25 subsidence. The objective of this study was to numerically assess the biomechanics of

1 TLIF in terms of the resulting SLL and stresses at the bone-cage interface as functions
2 of the cage height and its placement strategy with two tested bone qualities. This
3 objective aims at comparing two common cage placement strategy in surgical procedure
4 while surgeon may choose between two cages with different heights.

5

6 **Material and methods**

7 *Finite element model of the L4-L5 segment*

8 A detailed finite element model (FEM) of L4-L5 functional spinal unit was created
9 based on a previously developed and validated FEM of the spine (El-Rich et al. 2008;
10 El-Rich et al. 2009) (Figure 1a). The FEM was adapted and refined to simulate the
11 biomechanics of the TLIF, including intervertebral space preparation, cage insertion,
12 and posterior fixation (Agrawal and Resnick 2012; Gum et al. 2016). The geometric
13 model of the spine was reconstructed using medical images acquired through a CT-scan
14 (0.6 mm slice thickness) of a 50th percentile healthy man (El-Rich et al. 2008; El-Rich
15 et al. 2009). The model consisted of the vertebral body (cancellous and cortical bones),
16 posterior arches, intervertebral disc, facet joints, and seven ligaments, i.e. the anterior
17 longitudinal ligament (ALL), posterior longitudinal ligament (PLL), ligamentum flavum
18 (LF), capsular ligaments (CL), intertransverse ligament (ITL), interspinous ligament
19 (ISL), and supraspinous ligament (SSL) (Figure 1a).

20 Each vertebra was meshed with 4-node solid elements, representing the
21 trabecular bone enveloped by a layer of cortical bone with changing thickness in five
22 regions: endplates and anterior walls of the vertebral body (0.4 mm), upper pedicle (2
23 mm), lower pedicle (1.87 mm), posterior processes (1 mm), and insertion area of
24 pedicle screws (0.8 mm) (Silva et al. 1994; Hirano et al. 1997; Bianco et al. 2017)

1 (Figure 1b). The AF was modelled with five concentric layers of 8-node solid elements
2 between the two vertebrae, reinforced by spring elements to simulate the collagen fibres
3 oriented at $\pm 35^\circ$. The NP was meshed with 8-node elements. All ligaments were meshed
4 with 4-node shell elements, except the CL, which was represented by 3-node shell
5 elements. To balance the computation cost and analysis accuracy for this study, we
6 performed a mesh convergence study to determine adequate element sizes (Figure 1a
7 and Figure 1b).

8 Non-linear material properties were implemented to model the mechanical
9 behaviour of the spinal elements in physiological loading conditions. The cortical and
10 trabecular bones were modelled as homogenous isotropic materials governed by the
11 elastoplastic Johnson-Cook constitutive law (Wagnac et al. 2012). The NP and AP were
12 modelled as Mooney-Rivlin hyperelastic material while collagen fibres were
13 incorporated as one-dimensional spring elements acting in tension only. The non-linear
14 behaviour of the spinal ligaments was modelled with a generalized Maxwell-Kelvin-
15 Voigt constitutive law, and the failure criteria was incorporated based on the maximum
16 tensile strain level (Wagnac et al. 2012). The material properties of the elements were
17 initially defined using numerical results from the literature (Tables 1, 2, and 3). Material
18 properties of osteoporotic bone were modelled by reducing Young's modulus of cortical
19 (33%) and trabecular bone (66%) (Polikeit et al. 2003; Salvatore et al. 2018). To model
20 the zygapophyseal facet joints, a general purpose contact was used with an initial gap of
21 0.5 mm (Faizan et al. 2014) and Coulomb friction coefficient of 0.2 (Li et al. 2015)
22 between the two facets of the articulation. Tied contacts were modelled between the
23 ligaments and cortical bone at their attachment sites. The mechanical properties of the
24 aforementioned modelling elements were adjusted and calibrated such that the load-
25 displacement results of functional loading simulations corresponded to the results from

1 experiments on cadaveric lumbar spines (Heuer et al. 2007). To validate the FEM, the
2 ROMs of the intact model under simulated pure bending of 8 Nm in flexion-extension,
3 lateral bending, and axial rotation were compared with the reported ROMs of similar
4 experimental cadaveric tests (Dahl et al. 2013; Jaramillo et al. 2016). To support the
5 decisions in the context of use (COU) which was the risk of cage subsidence in TLIF,
6 the credibility of developed FEM was established. For this purpose, the assumption and
7 model inputs were tested in their ranges to assure that the results are still applicable to
8 the COU with the quantified uncertainty of the predictions.

10 *Simulation of TLIF procedure*

11 The surgical procedure of TLIF was modelled by partial discectomy through the
12 unilateral approach was modelled by removing the elements corresponding to the
13 posterior-left portion of the AF and NP. A facetectomy was simulated by removing the
14 elements corresponding to the zygapophyseal joint to virtually make a window for the
15 cage insertion (Figure 2Error! Reference source not found.a). Four pedicle screws (40
16 mm long, 6.5 mm diameter; CD HORIZON® LEGACY™, Medtronic, Memphis TN)
17 were modelled as rigid bodies, and their external surfaces were meshed with triangular
18 shell elements. They were aligned with their corresponding vertebra based on a typical
19 lumbar pedicle screw insertion technique (Agarwal et al. 2013; Bianco et al. 2017). To
20 identify the proper element size, a mesh convergence study initially was conducted by
21 testing various element sizes (from 0.5 to 1.5 mm) at the endplate-cage interface of the
22 oblique asymmetric placement of 10-mm cage, up until the variation of the maximum
23 Von-Mises stress at the bone-cage interface was lower than 5%. Boolean operations
24 between the screw and the vertebral models were performed to remove the cortical layer

1 and trabecular core model elements (Bianco et al. 2017). A point-to-surface contact
2 with a Columb friction of 0.2 was modelled to represent the bone-screw interface.

3 The interbody cage models were based on a generic cage (CAPSTONE®
4 interbody cage, Medtronic, Memphis TN). The length and width of the models were 26
5 mm and 10 mm, respectively. Two cage heights were tested, i.e. 8 and 10 mm. For each
6 model, we tested the oblique asymmetric and anterior symmetric placements, a total of
7 four interbody cage scenarios (Figure 3). The cages were meshed with 4-node
8 tetrahedral elements of 1.0 mm, and material properties of polyether-ether-ketone
9 (PEEK) were assigned ($E=3.4$ GPa and $\nu=0.4$ (Faizan et al. 2014)). The modelling of
10 the cage insertion was based on the documented surgical technique (Agarwal et al.
11 2013). First, the cage model was aligned to the superior endplate of L5, and a node-to-
12 surface contact with a minimum distance of 0.5 mm and Columb friction coefficient of
13 0.2 was applied to the interface. Then, a distractive force was applied between L4 and
14 L5 such that the intervertebral body space increased and there was no interference
15 between the cage and endplate geometries. Finally, the loads were released after node-
16 to-surface contact was modelled between the cage and the adjoining endplates (Figure
17 2b). The SSL was assessed before and after the simulation of the cage placement by
18 measuring the angle between the superior endplate of L4 and inferior endplate of L5
19 (Hong et al. 2017). After the simulation of the cage insertion, two titanium rod (4.5 mm)
20 models were aligned with screw head saddles and tied contacts were modelled between
21 them to simulate the posterior fixation (Figure 2c). The rods were meshed with 4-node
22 tetrahedral solid elements of 0.5 mm characteristic length, and the material properties of
23 Titanium alloy were adapted from literature ($E=115$ GPa and $\nu=0.34$ (Faizan et al.
24 2014)).

1 The resulting FEM was used to simulate physiological loading. The body weight
2 was modelled as a 400 N follower-load to the superior elements of L4 with the inferior
3 endplate of L5 fixed in space. A 10-Nm functional load was simulated in the three
4 anatomical planes, respectively, to simulate flexion (Fe), extension (Ex), right lateral
5 bending (RLB), left lateral bending (LLB), right axial rotation (RAR), and left axial
6 rotation (LAR). The ROM and maximum Von-Mises stresses were computed as a
7 measure of the risk of cage subsidence.

8 All the simulations were performed using the RADIOSS v14.0 finite element
9 package (Altair Engineering Inc., Troy, USA) in a quasi-static condition (Bianco et al.
10 2017).

11

12 **Results**

13 **The resulting ROMs of the non-instrumented FEM of the L4-L5 segment were 9.3°,**
14 **7.6°, and 4.1° in flexion-extension, lateral bending, and axial rotation, respectively.**

15 **These results were within the range of reported ROM of experimental cadaveric studies**
16 **in the literature (Dahl et al. 2013; Jaramillo et al. 2016) (Figure 4).** With the simulated
17 normal bone quality, the anterior symmetric and oblique asymmetric placement of the
18 cages increased the SLL by 0.9° and 1.0°, respectively, for the 8-mm height cage, and
19 by 1.4° and 1.7° for the 10-mm one. SLL restorations with simulated osteoporosis were
20 within 1.2% to those with the normal bone quality. With normal bone, the simulated
21 ROMs of the FSU after the TLIF procedure were lower than 1° in all loading directions,
22 while they ranged from 2° to 8° with the un-instrumented FSU (Figure 5a). With
23 simulated osteoporosis, the ROMs were slightly (about 8%) higher than those with
24 normal bone (Figure 5b). Oblique asymmetric vs. anterior symmetric placement

1 increased the ROM by 66% and 72% for the simulated normal and osteoporotic bone,
2 respectively. Insertion of the 8-mm cage vs. the 10-mm one increased the ROM of the
3 instrumented segment up to 43% and 48% in simulated normal and osteoporotic bone
4 models, respectively.

5 For the 8-mm cage with normal bone quality, the maximum stresses at the bone-
6 cage interface ranged from 82.1 to 98.4 MPa (anterior symmetric placement) and from
7 117.9 to 155.5 MPa (oblique asymmetric placement) (Figure 6a). For the 10-mm cage,
8 they were from 88.2 to 107.2 MPa (anterior symmetric) and between 134.4 and 176.4
9 MPa (oblique asymmetric) (Figure 6a). With osteoporosis, stresses at the bone-cage
10 interface were about 2.5% lower (Figure 6b). Oblique asymmetric as compared to the
11 anterior symmetric cage placement increased the maximum stresses by up to 41% and
12 43% for the simulated normal and osteoporotic bone, respectively. Insertion of the 10-
13 mm cage vs. the 8-mm one increased the maximum stresses by up to 16% in simulated
14 normal and osteoporotic bone models. The stress on the superior endplate of L5 is
15 displayed on Figure 7 for the 4 tested configurations under simulated flexion moment of
16 10 N m and 400 N follower-load.

17 For the 8-mm cage with normal bone quality, the maximum stresses in the
18 posterior rods were between 128.9 and 230.3 MPa (anterior symmetric) and between
19 114.9 and 326.6 MPa (oblique asymmetric) (Figure 8a). For the 10-mm cage, they
20 ranged from 60.3 to 218.0 MPa (anterior symmetric) and from 69.6 to 262.5 MPa
21 (oblique asymmetric) (Figure 6a). With osteoporosis, stresses in the posterior rods
22 increased up to about 120% (Figure 8b). Oblique asymmetric vs. anterior symmetric
23 placement increased the maximum stresses by up to 55% and 48% for the simulated
24 normal and osteoporotic bone, respectively. In simulations with oblique asymmetric
25 placement, stresses in the rod on the opposite side of the cage were higher than the other

1 rod. Insertion of the 8-mm cage vs. the 10-mm one increased the maximum stresses up
2 to 59% and 54% in simulated normal and osteoporotic bone, respectively.

3

4 **Discussion**

5 A larger SLL restoration was observed in the simulations with a 10-mm vs. 8-mm cage.
6 This was expected from a geometric point of view because greater cage height means
7 greater anterior intervertebral distance, thus higher SLL. Consequently, stresses at the
8 bone-cage interface in simulations of 10-mm cage were always higher than the 8-mm
9 cage. Cages of greater height required more intervertebral distraction for its proper
10 placement, which initiated a higher compression force at the bone-cage interface
11 generated by the tightening of the soft tissues, which translated in higher structural
12 stiffness and lower ROM due to the non-linear mechanical behaviour of the
13 intervertebral ligaments. This could explain why the maximum stresses in the rods with
14 the 10-mm cage were lower than the 8-mm cage. The stresses generated by the
15 compression forces as a function of cage height agreed with the reported experiments
16 with cadaveric lumbar spines (Truumees et al. 2008; Ambati et al. 2015).

17 The simulated SLL restoration with the anterior symmetric cage placement was
18 very close to that with the oblique asymmetric placement, but the maximum stresses at
19 the bone-cage interface with anterior symmetric placement were consistently lower than
20 those with oblique asymmetric placement. This may be explained by the fact that the
21 resultant force at the bone-cage interface with anterior symmetric placement has a
22 longer lever arm with respect to the posterior fixation, providing a mechanical
23 advantage to balance the external loads. In this standpoint, with the use of similar
24 interbody cage footprints and in the presence of a smaller compression force at the
25 bone-cage interface, lower stress is expected with the anterior cage placement.

1 Compared with oblique asymmetrical placement, the anterior symmetrically placed
2 cage had more bone-cage contact area in the anterior part of the intervertebral body
3 space (Figure 4) where the endplates have superior mechanical strength (Tsitsopoulos et
4 al. 2012; Faizan et al. 2014). Increased stresses due to oblique asymmetric vs. anterior
5 symmetric (41%) seems to have a significant effect on the stress distribution at the
6 bone-cage interface. This difference is clinically important because the maximum stress
7 at the interface, in some cases, exceeds the failure stress of cortical bone (126 MPa)
8 (Hansen et al. 2008). On the other hand, using 10-mm cage does not significantly affect
9 the risk of subsidence since the value of the maximum stresses are still below the failure
10 stress of cortical bone. With the oblique asymmetric placement, reaction forces from the
11 rods had, therefore, shorter lever arms with respect to the cage centre – fulcrum point
12 between the upper and lower vertebral bodies, resulting in a higher stress in the rods to
13 balance the loads.

14 There was no difference in SLL restoration between normal and osteoporosis
15 bones. Although the maximum stresses at the bone-cage interface for the simulated
16 osteoporosis were identical to those of modelled normal bone, the risk of cage
17 subsidence should be higher because the osteoporotic bones also have between 20% to
18 40% lower mechanical strengths due to decreased bone mineral density (Dickenson et
19 al. 1981; Bono and Einhorn 2003). Also, clinical studies showed that the risk of cage
20 subsidence in osteoporosis spines was about 3 times higher than in spines with normal
21 bone (Formby et al. 2016; Oh et al. 2017). The simulated osteoporosis bones had lower
22 stiffness and provided less support of the functional loads as compared to the normal
23 bones, making the rods subjected to higher loads and stresses.

24 Some simplifications and approximations were made in the modelling and
25 simulations in this study (i.e. the cortical and trabecular bones were modelled as

1 homogenous isotropic materials, the geometry and mechanical properties of the FSU
2 were based on a generic 50th model, and screw insertion was model as a geometric
3 Boolean operation between the screw and the vertebral models and with a contact
4 definition between the two). These modelling simplifications and approximations are
5 considered to have limited effects on the conclusions because this study focused on the
6 relative effects of the cage height, cage placement and bone quality which are common
7 in TLIF for most of the cases. The presented modelling technique might be adapted to
8 examine the biomechanics of multi-level TLIF, as well as the performance of any other
9 interbody cages in terms of risk of cage subsidence.

10

11 **Conclusion**

12 A detailed FEM was developed to simulate the biomechanics of the TLIF procedure.
13 The FEM allowed the assessment of the effects of the cage height, cage placement, and
14 bone quality on the SLL restoration and risk of the cage subsidence. It was found that
15 10- vs 8-mm cage height resulted in up to 0.7° higher SLL restoration and 16% higher
16 stresses at the bone-cage interface. Oblique asymmetric placement vs. anterior
17 symmetric placement had almost similar SLL restoration, but the stresses at the bone-
18 cage interface were up to 43% higher. Bone quality did not affect the achieved SLL; a
19 higher risk of cage subsidence is expected for the osteoporotic spines although the
20 maximum stresses at the bone-cage interface were 2.5% lower. The FEM presented in
21 this study was shown to be a relevant tool to assess the biomechanics of TLIF. It could
22 be further adapted to further assess the biomechanics of any interbody cage design, as
23 well as to evaluate reported clinical findings towards the improvement of the TLIF
24 procedure.

1

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6

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1 **List of figures**

2 **Figure 1** a) The un-instrumented FE model of the L4-L5 segment including the vertebrae,
3 seven spinal ligaments, and intervertebral disc; b) regional thickness of the cortical bone
4 and finer mesh of the trabecular bone around the screw imprint ALL: Anterior
5 Longitudinal ligament, PLL: Posterior Longitudinal Ligament, ITL: Intertransverse
6 Ligament, CL: Capsular Ligament, LF: Ligament Flavum, ISL: Interspinous Ligament,
7 SSL: Supraspinous Ligament, AF: Annulus Fibrosus, NP: Nucleus Pulposus.

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9 **Figure 2** Simulation of different surgical procedures of TLIF: (a) Partial discectomy and
10 facetectomy of L4-L5, (b) Cage placement by imposing distractive force and moment on
11 L4, while the inferior endplate of L5 was fixed in space, and (c) Implementation of the
12 posterior fixation followed by application of the follower-load and physiological
13 moments (flexion, extension, lateral bending, and torsion) on the superior endplate of L4
14 while the inferior endplate of L5 was fixed in space.

15 **Figure 3** Simulated placement scenarios of the cage: (a) Oblique asymmetric: (b)
16 Anterior symmetric.

17 **Figure 4** Simulated ROM of the FEM of L4-L5 segment was within the reported range
18 of similar experimental tests on human cadaveric spines (Dahl et al. 2013; Jaramillo et al.
19 2016).

20 **Figure 5 Range of motion (ROM)** of the instrumented spinal segment in different
21 loading directions for normal (a) and osteoporotic (b) bone models (A08/A10: Oblique
22 asymmetric placement of 8/10-mm cage; S08/S10: Anterior symmetric placement of
23 8/10-mm cage).

1 **Figure 6** Maximum Von-Mises stress at the endplate-cage interface in different loading
2 directions for normal (a) and osteoporotic (b) bone model (A08/A10: Oblique asymmetric
3 placement of 8/10-mm cage; S08/S10: Anterior symmetric placement of 8/10-mm cage).
4 **Figure 7** Stress (in MPa) on the superior endplate of L5 under simulated flexion
5 moment of 10 N m and 400 N follower-load.
6 **Figure 8** Maximum Von-Mises stress in the posterior rods in different loading directions
7 for normal (a) and osteoporotic (b) bone models (A08/A10: Oblique asymmetric
8 placement of 8/10-mm cage; S08/S10: Anterior symmetric placement of 8/10-mm cage)
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