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Abstract

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

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

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Introduction

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

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

Biomechanical analysis using finite element models showed that a larger cage footprint (e.g., 490 vs. 280 mm2) allowed to bear about 300% more functional load and reduced the maximum stresses by about 50%, resulting in a lower risk of cage subsidence (Faizan et al. 2014). Biconvex shapes were shown to allow better cage fitting, but with loads more concentrated in the medial region of the endplates with relatively lower mechanical strength than peripheral cortical bone, thus higher risk of cage subsidence (Cho et al. 2008). Using paired- vs. single-cage configurations resulted in 55.2% lower stress at the bone-cage interface (49.77 vs. 77.23 MPa) and subsequently lower risk of cage subsidence (Xu et al. 2013).

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

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

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

Material and methods

Finite element model of the L4-L5 segment

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

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

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

Non-linear material properties were implemented to model the mechanical behaviour of the spinal elements in physiological loading conditions. The cortical and trabecular bones were modelled as homogenous isotropic materials governed by the elastoplastic Johnson-Cook constitutive law (Wagnac et al. 2012). The NP and AP were modelled as Mooney-Rivlin hyperelastic material while collagen fibres were incorporated as one-dimensional spring elements acting in tension only. The non-linear behaviour of the spinal ligaments was modelled with a generalized Maxwell-Kelvin-Voigt constitutive law, and the failure criteria was incorporated based on the maximum tensile strain level (Wagnac et al. 2012). The material properties of the elements were initially defined using numerical results from the literature (Tables 1, 2, and 3). Material properties of osteoporotic bone were modelled by reducing Young's modulus of cortical (33%) and trabecular bone (66%) (Polikeit et al. 2003; Salvatore et al. 2018). To model the zygapophyseal facet joints, a general purpose contact was used with an initial gap of 0.5 mm (Faizan et al. 2014) and Coulomb friction coefficient of 0.2 (Li et al. 2015) between the two facets of the articulation. Tied contacts were modelled between the ligaments and cortical bone at their attachment sites. The mechanical properties of the aforementioned modelling elements were adjusted and calibrated such that the load-displacement results of functional loading simulations corresponded to the results from experiments on cadaveric lumbar spines (Heuer et al. 2007). To validate the FEM, the ROMs of the intact model under simulated pure bending of 8 Nm in flexion-extension, lateral bending, and axial rotation were compared with the reported ROMs of similar experimental cadaveric tests (Dahl et al. 2013; Jaramillo et al. 2016). To support the decisions in the context of use (COU) which was the risk of cage subsidence in TLIF, the credibility of developed FEM was stablished. For this purpose, the assumption and model inputs were tested in their ranges to assure that the results are still applicable to the COU with the quantified uncertainty of the predictions.

Simulation of TLIF procedure

The surgical procedure of TLIF was modelled by partial discectomy through the unilateral approach was modelled by removing the elements corresponding to the posterior-left portion of the AF and NP. A facetectomy was simulated by removing the elements corresponding to the zygapophyseal joint to virtually make a window for the cage insertion (Figure 2**Error! Reference source not found.**a). Four pedicle screws (40 mm long, 6.5 mm diameter; CD HORIZON® LEGACY™, Medtronic, Memphis TN) were modelled as rigid bodies, and their external surfaces were meshed with triangular shell elements. They were aligned with their corresponding vertebra based on a typical lumbar pedicle screw insertion technique (Agarwal et al. 2013; Bianco et al. 2017). To identify the proper element size, a mesh convergence study initially was conducted by testing various element sizes (from 0.5 to 1.5 mm) at the endplate-cage interface of the oblique asymmetric placement of 10-mm cage, up until the variation of the maximum Von-Mises stress at the bone-cage interface was lower than 5%. Boolean operations between the screw and the vertebral models were performed to remove the cortical layer and trabecular core model elements (Bianco et al. 2017). A point-to-surface contact with a Columb friction of 0.2 was modelled to represent the bone-screw interface.

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

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

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

Results

The resulting ROMs of the non-instrumented FEM of the L4-L5 segment were 9.3°, 7.6°, and 4.1° in flexion-extension, lateral bending, and axial rotation, respectively. These results were within the range of reported ROM of experimental cadaveric studies in the literature (Dahl et al. 2013; Jaramillo et al. 2016) (Figure 4). With the simulated normal bone quality, the anterior symmetric and oblique asymmetric placement of the cages increased the SLL by 0.9° and 1.0°, respectively, for the 8-mm height cage, and by 1.4° and 1.7° for the 10-mm one. SLL restorations with simulated osteoporosis were within 1.2% to those with the normal bone quality. With normal bone, the simulated ROMs of the FSU after the TLIF procedure were lower than 1° in all loading directions, while they ranged from 2° to 8°with the un-instrumented FSU (Figure 5a). With simulated osteoporosis, the ROMs were slightly (about 8%) higher than those with normal bone (Figure 5b). Oblique asymmetric vs. anterior symmetric placement increased the ROM by 66% and 72% for the simulated normal and osteoporotic bone, respectively. Insertion of the 8-mm cage vs. the 10-mm one increased the ROM of the instrumented segment up to 43% and 48% in simulated normal and osteoporotic bone models, respectively.

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

For the 8-mm cage with normal bone quality, the maximum stresses in the posterior rods were between 128.9 and 230.3 MPa (anterior symmetric) and between 114.9 and 326.6 MPa (oblique asymmetric) (Figure 8a). For the 10-mm cage, they ranged from 60.3 to 218.0 MPa (anterior symmetric) and from 69.6 to 262.5 MPa (oblique asymmetric) (Figure 6a). With osteoporosis, stresses in the posterior rods increased up to about 120% (Figure 8b). Oblique asymmetric vs. anterior symmetric placement increased the maximum stresses by up to 55% and 48% for the simulated normal and osteoporotic bone, respectively. In simulations with oblique asymmetric placement, stresses in the rod on the opposite side of the cage were higher than the other rod. Insertion of the 8-mm cage vs. the 10-mm one increased the maximum stresses up to 59% and 54% in simulated normal and osteoporotic bone, respectively.

Discussion

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

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

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

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

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

Conclusion

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

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List of figures

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

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

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

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

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

Figure 6 Maximum Von-Mises stress at the endplate-cage interface in different loading directions for normal (a) and osteoporotic (b) bone model (A08/A10: Oblique asymmetric placement of 8/10-mm cage; S08/S10: Anterior symmetric placement of 8/10-mm cage).

Figure 7 Stress (in MPa) on the superior endplate of L5 under simulated flexion moment of 10 N m and 400 N follower-load.

Figure 8 Maximum Von-Mises stress in the posterior rods in different loading directions for normal (a) and osteoporotic (b) bone models (A08/A10: Oblique asymmetric placement of 8/10-mm cage; S08/S10: Anterior symmetric placement of 8/10-mm cage)