Biomechanical Evaluation of Cortical Bone and Traditional Trajectories for Lumbar Single Level Segment Fixation

Yuan Dai
Huashan Hospital Fudan University
Jianyuan Jiang
Huashan Hospital Fudan University
Minghao Shao
Huashan Hospital Fudan University
Feizhou Lu (zytgdily@126.com)
Huashan Hospital Fudan University https://orcid.org/0000-0002-3512-2046

Research Article

Keywords: Cortical bone trajectory, Finite element model, Lumbar spine fixation, Biomechanics

Posted Date: February 10th, 2021

DOI: https://doi.org/10.21203/rs.3.rs-177841/v1

Abstract

Background: The aim of this study was to evaluate the biomechanical stability of a lumbar internal fixation system with 3 different fixation techniques by the establishment of a three-dimensional finite element (FE) model of lumbar single level fixation.

Methods: A three-dimensional osseoligamentous nonlinear FE model of osteoporosis lumbar 4-5 (male, aged 57 years, height 170 cm, weight 70 kg, bone mineral density T value -2.8SD) was built to detect the biomechanical stability of an internal fixation system with the following 3 screw trajectories: traditional pedicle screw trajectory fixation (TT), cortical bone trajectory (CBT) screw and hybrid pedicle screw fixation (CBT + TT). The location and area of maximal equivalent stress and the angular displacement of this lumbar model with different screw trajectories were measured in anterior bending, posterior extension, lateral bending and during rotation.

Results: The angular displacement of this lumbar model with the 3 different screw trajectories was similar, all of which could restrict the angular displacement of lumbar vertebrae. The maximal equivalent stress was located at the border of the CBT screw and rod, with the hybrid screw fixation technique in axial rotation.

Conclusion: The use of CBT and TT screws in lumbar internal fixation had similar stability. The CBT screw could be an alternative solution to lumbar short-segmented fixation, but temporary immobilization would be required to avoid the failure of CBT screw fixation due to the increasing stress in position of extensive lateral bending and rotation.

Background

A pedicle screw fixation system with excellent biomechanical stability is beneficial to the reconstruction of stability of spinal sequence and its fusion. It is the key to the curative effect of surgery for lumbar fusion [1, 2]. However, cases of internal fixation failure are not uncommon, due to loosening of the pedicle screw, screw exit or breakage and other complications. This failure occurs more commonly in elderly patients [3, 4]. The current study suggests that elderly patients with osteoporosis or degenerative lumbar instability need more powerful internal fixation and holding forces to maintain the stability of postoperative spinal stabilization [4].

In 2009, Santoni et al. [5] reported a new method – cortical bone trajectory (CBT). They tried to increase the screw in the nail holding force by changing the screw trajectory. Compared to the traditional trajectory (TT), axis of the pedicle was applied as a channel, so that the inclined screw could make better use of the pedicle cortical bone part, to ensure that the screw and cortical bone are embedded in each other. A subsequent biomechanical study confirmed its pullout strength was better than a TT screw [5].

At present, a CBT screw has been applied to lumbar internal fixation systems by some surgeons, and they found that the fusion rate in patients with CBT screw internal fixation was similar to those with traditional

screws; The operative time and intraoperative blood loss were significantly reduced [6, 7]. However, the CBT screw has not been widely applied in the clinic, which may be mainly attributed to the fewer biomechanical studies on CBT screw internal fixation, with the clinical use of a CBT internal fixation system still lacking a sufficient theoretical basis.

The FE method is a standard engineering technique in general used in the design of airplanes, machines and bridges. In the analyze of bone joints such as spines, which have complicated shapes, load and boundary conditions, FE method can be a useful tool [8]. Using special software, it allows the modeling of complex structures by splitting the structure into numerous, simple FEs, each of which is easy to characterize and model mathematically. The lumbar FE model includes the vertebrae, intervertebral discs, endplates, posterior bony elements and all seven ligaments (the anterior and posterior longitudinal ligaments, ligamentum flavum, facet capsules, and the intertransverse, interspinous and supraspinous ligaments) had been set up during last decades and facilitated the study of lumbar biomechanical characteristics [9–12].

Therefore, we conducted this study to investigate the biomechanical characteristics of TT and CBT screw internal fixation by the establishment of a three-dimensional FE model of lumbar spine short segmental fixation.

Methods

The establishment of the FE model

All study procedures and experiment protocols were performed in accordance to standard guidelines approved by the Ethics Committee of Experimental Research, Huashan Hoapital, Fudan University (Shanghai, China). An osteoporosis volunteer (male, 57 years old, height 170 cm, weight 70 kg, bone mineral density T value − 2.8SD) was used to create a three-dimensional osseoligamentous nonlinear FE model (Fig. 1). No Congenital bone disease and lumbar disease were found in volunteer. The geometry of the bony structures was generated from computed tomography images (slice thickness 0.625 mm) of L4-L5 segments. After that, a 3D reconstruction of the model based on computed tomography images was performed using Mimics 10.0. Then we used Geomagic 12.0 software to translate the reconstruction to a Nurbs surface model according to its anatomical structure. At last, the FE model was conducted in Hypermesh 12.0 and the output imported into Abaqus 6.12 for nonlinear FE analysis. Written informed consent was obtained from this volunteer and ethical approval was given by the medical ethics committee of Fudan university.

The L4-L5 vertebra were simplified as isotropic and elastic materials and described by 2 parameters, namely the elastic modulus and Poisson’s ratio. The material property was assigned in the FEA module of software Mimics 10.0 according to the grey value. The number of materials was set to 15. The Poisson ratio of the L4-L5 vertebra was 0.3. The relationship between density and grey value was determined using the following empirical equation:

$$P = 1.112 * HU + 47$$

The relationship between density and Young's modulus was determined using the following empirical equation:

$$E = 1.92 * \rho - 170$$

The intervertebral disc consisted of nucleus pulposus, annulus fibrosus and cartilaginous endplates. The nucleus pulposus comprised 43% of the total disc volume and was positioned slightly posterior to the disc center [9]. The annulus consisted of annulus fibers and annulus ground substance. The orientations of the annulus fibers were ± 23° at the anterior side and increased to ± 58° at the posterior side. The ratio of fiber volume to the volume of surrounding annulus ground substance was 5% at the inner layers and increased to 23% at the outer layers [10, 13]. The annulus ground substance and nucleus were modeled with tetrahedral solid elements (C3D8), and the annulus fibers were modeled with tension-only truss elements (T3D2). The surfaces of the facet joints were simulated using a cartilaginous layer (C3D6). The contact between the facet joints was simulated with surface-to-surface hexahedral solid elements without friction. All 7 ligaments (anterior and posterior longitudinal ligaments, inter- and supraspinous ligaments, facet capsular ligaments, ligamentum flavum, and intertransverse ligaments) were modeled as tension-only truss elements with nonlinear elastic properties. The materials used in the model were assumed to be homogeneous and isotropic. The material properties of the various spinal components were derived from the literature as specified in Table 1 [11, 12].

Table 1
Material properties of the various spinal components

Element set	Element type	Young modulus	Poisson ratio	Cross-sectional area
		(MPa)	(μ)	(mm2)
End plates	C3D6	12,000	0.3	
Nucleus pulposus	C3D8	1	0.49	
Annulus (ground)	C3D8	4.2	0.45	
Annulus fibers	T3D2	357.5–550	0.3	0.00601–0.00884
Facet joint	C3D6	3,500		
Titanium	C3D8	145,000	0.3	
Anterior longitudinal ligaments	T3D2	20	0.4	74
Posterior longitudinal ligaments	T3D2	20	0.4	14.4
Supraspinous ligaments	T3D2	10	0.3	30
Intersupraspinous ligaments	T3D2	10	0.3	40
Ligamentum flavum	T3D2	10	0.3	40
Intertransverse ligaments	T3D2	10	0.3	1.8
Facet capsular ligaments	T3D2	10	0.3	34

C3D6 is the 6-node isoparametric triangular shell element, C3D8 is the 8-node isoparametric hexahedral elements, and T3D2 is the 2-node isoparametric truss element in the Abaqus Software

FE models of implants

The size of the TT pedicle screw was 45 mm × 6.5 mm. The size of the CBT pedicle screw was 35 mm × 5.5 mm. We placed TT screws into the vertebral body along an anatomical axis of the pedicle and parallel to the vertebral endplate. According to Matsukawa's research, the origin of CBT was situated at the lateral aspect of the pars interarticularis projecting in the 5 O'clock orientation in the left pedicle and the 7 O'clock orientation in the right pedicle. As a result,CBT screws were placed 10°laterally in the axial plane and 25°cranially in the sagittal plane [14].

Three types of fixation technique using TT and CBT were studied: traditional trajectory pedicle screw fixation (TT) (Fig. 2a), cortical bone trajectory for lumbar pedicle screw fixation (CBT) (Fig. 2b), and hybrid pedicle screw fixation (cortical bone trajectory screws in upper vertebrae and traditional trajectory screws in lower vertebrae) (Fig. 2c). The compressive constructs were modeled by Abaqus version 6.12. The FE L4–L5 model underwent bilateral facetectomy and partial laminectomy to replicate the improved

transforaminal lumbar interbody fusion surgery when titanium alloy screws and connecting rods were used. The friction coefficient among the screws, connecting rods, and bone was set as infinite. A tie constraint at the interface of the pedicle screw and rod was used to simulate the rigid fixation.

Loading and boundary conditions

The L5 vertebra in the load simulation was restrained, while the superior surface of L4 was given a 500 N compressive preload to mimic upper body weight. The nodes on the upper surface of the L4 vertebra were used as a reference node for load application. The flexion/extension, lateral bending, and axial rotation were simulated by a bending moment generated through a 15 Nm force applied to the superior surface of the L4 vertebra. Data analyses were performed by Abaqus 6.12.

Observed indicators

The location and area of maximal equivalent stress and the angular displacement of this lumbar model with different screw trajectories were measured in anterior bending, posterior extension, right lateral bending and right rotation.

Results

The validation of FE model

In order to ensure the accuracy of the intact L4–L5 FE model, we compared the results with previous studies [11, 15, 16]. The entire moment-rotation curve on the conditions of flexion, extension, lateral bending and axial rotation demonstrated the nonlinear behavior of our FE model. Furthermore, the axial displacement curve of the vertebral body under different distribution pressure was compared with the previous FE studies in literature [17–19].

Angular displacement of FE models and stress of implants

The angles of the normal lumbar model in anterior bending, posterior extension, lateral bending and rotation were 1.17°, 1.17°, 1.15°and 1.24°, respectively. The axial displacement of lumbar 4–5 FE model was 0.29 mm, 0.54 mm, 0.78 mm, and 1.03 mm under a pressure of 500 N, 1,000 N, 1,500 N and 2,000N, which was in accordance with previous studies [20, 21].

Figure 3 shows the normal lumbar model and its angular displacement with the 3 types of fixation technique in anterior bending, posterior extension, right lateral bending and rotation. The angular displacement in the CBT lumbar model was slightly higher than those in the other 2 models; however, the 3 types of fixation technique could restrict obviously the angular displacement of lumbar vertebrae when compared with the angular displacement of normal lumbar spine.

The maximal equivalent stress with the 3 types of fixation technique was located at the border of the screw and rod, but the individual location had some slight change for these techniques (Fig. 4). The maximal equivalent stresses in anterior bending with TT, CBT and hybrid fixation techniques were located

at the border of lower vertebral screw and rod, which were 176 Mpa, 207.6 Mpa and 152.2 Mpa, respectively. Similarly, the maximal equivalent stresses in the posterior extension with TT, CBT and the hybrid fixation technique were also located at the border of the lower vertebral screw and rod, which were 151.3 Mpa, 217.4 Mpa and 232.2 Mpa, respectively. In the condition of right bending, the lower vertebral screw at the left side was bearing a higher load for the TT and CBT screws fixation (220 Mpa, 288 MPa). The maximal equivalent stress with the hybrid fixation technique was located at the CBT screw in the left upper vertebra (380.7 MPa). In the condition of axial rotation to the right, the stress concentration with the TT screw was located at the border of the upper vertebral screw and rod, with 320.1 MPa of maximum equivalent stress. For CBT screw internal fixation, the maximum equivalent stress was located at the border of the lower vertebral screw and rod (778.9 MPa). For hybrid screw internal fixation, the maximum equivalent stress was located at the border of CBT screw and nail rod (812 Mpa) (Fig. 5).

Discussion

At present, most tests have been performed through cadaver simulation of pedicle screws in screw detection, and the results have shown that the CBT screw has good biomechanical properties, but the results of the same index are not entirely consistent. For example, in the cadaver experiments of Santoni et al. [5] and Matsukawa et al. [14], the CBT screw had a stronger anti-pullout force and torque. But in the study of Baluch et al. [22], the anti-pullout strength of the two kinds of screws was not obviously different. In the experiment of Perez-Orribo et al. [23], CBT screw and TT screw lumbar specimens had the similar performances in the aspect of anti-axial pullout. The differences are closely related to the objects of the researches. Although lumbar specimens can accurately reflect the biomechanical and morphological characteristics of lumbar vertebrae, during CBT screw placement, it cannot ensure the consistency and accuracy of the angle of pedicle screws. The deviation of position and the standard requirements of nail will lead to biased results. But in the case of screw misplacement, specimens cannot be replaced, so the repeatability of experiments is poor. Unlike cadaver studies, three-dimensional FE software with good repeatability and accuracy can completely copy each structure and simulate the lumbar biomechanical characteristics, and at the same time can choose the screw entry point and screw direction in accordance with appropriate standards [24].

The main purpose of lumbar internal fixation is to restrict segmental motion, thus providing a good environment for fusion. Oshino [25] found that intervertebral stability after CBT fixation was similar to that of TT fixation by using a sheep cadaveric spine model. Through this experimental research, we found that the angular displacement of this lumbar model with the 3 different fixation techniques was a little different in anterior bending, posterior extension, lateral bending and rotation, but all could restrict the angular displacement of lumbar vertebrae significantly when compared with normal lumbar vertebrae, thereby indicating that the 3 types of internal fixation techniques could provide a relative stable fusion environment.

The maximum equivalent stress could estimate the risk of internal fixation fracture [4]. The results of the present studies suggested that the equivalent stress with a CBT screw was obviously larger than that with

a TT screw, which could result from: 1) the length of the CBT screw in the vertebral body being shorter; the CBT screw being shorter and only fixed at the lumbar posterior column, which could cause screw stress concentration when lumbar stress was transmitted to the screw, and this was different from the TT screw [14]; 2) The direction of the CBT screw did not match with the normal direction of lumbar activities. In axial rotation, the implanting angle of the contralateral CBT screw was in the opposite direction of lumbar spine motion, leading to increased stress at the border of the screw and rod. Therefore, the authors thought that temporary postoperative immobilization would be required to avoid CBT screw loosening or fracture due to increased stress.

Regarding the lumbar internal fixation method, surgeons should consider both the biomechanical characteristics of the internal fixation system and the intraoperative specific conditions. The entry point of the CBT screw was closer to the center line than that of the TT screw, thus less vertebral muscles detachment before screw implantation would be needed for a CBT screw, reducing the operative time and intraoperative bleeding. This would be beneficial when the operative field is hard to expose due to obesity and patients with poor surgery tolerance [6, 7]. With the advent of the CBT screw, some scholars used a combination of CBT and TT screws; the distance between the upper and lower vertebral screw nail was then shortened, which could shorten the incision in single segmental lumbar internal fixation surgery to 5 cm, thus further decreasing intraoperative blood loss [26]. Therefore, a CBT screw would be of certain advantages in the clinical application compared with a TT screw.

Our research has its limitations. In the clinic, screw tap importing and screw selection bias before nailing can affect screw stability. In this paper, in terms of FE analysis, Because of the direct screw placement in the preset position and the lack of clinical prenail operation, therefore, it is impossible to simulate errors caused by the preset. Although the study of the single segment lumbar spine proved the biomechanical of CBT screw in internal fixation of vertebral body. But there is still a need for long term follow-up, in order to understand the advantages and disadvantages of a CBT screw system in lumbar fusion fixation.

Conclusion

In this study, we establish 3 different FE models of single level lumbar fixation. According to our study, the use of CBT and TT screws in lumbar internal fixation could provide similar stability for single level lumbar fixation. The CBT screw could be an alternative solution to lumbar short-segmented fixation, but temporary immobilization would be required to avoid failure of CBT screw fixation due to the increasing stress in position of extensive lateral bending and rotation.

Abbreviations

FE: finite element; CBT: cortical bone trajectory; TT: trajectory fixation

Declarations

Ethics approval and consent to participate

All study procedures and experiment protocols were performed in accordance to standard guidelines approved by the Ethics Committee of Experimental Research, Huashan Hoapital, Fudan University (Shanghai, China).

Consent for publication

Signed informed consent was obtained from volunteer.

Availability of data and materials

The datasets used and/or analyzed during the current study are available from the corresponding author on reasonable request.

Competing interests

The authors indicated no potential conflicts of interest.

Funding

This work was supported by the National Natural Science Foundation of China (grant 81772385).

Authors' contributions

FZL and MHS conceived and designed the study. MHS and YD collected and assembled the data. JYJ, MHS, and YD analyzed and interpreted the data. MHS and YD performed the literature search. MHS and YD performed the writing of the manuscript. The authors read and approved the final manuscript.

Acknowledgements

None

References

1. McLain RF, Fry MF, Moseley TA, Sharkey NA. Lumbar pedicle screw salvage: pullout testing of three different pedicle screw designs. J Spinal Disord. 1995;8:62-8.
2. Chiba M, McLain RF, Yerby SA, Moseley TA, Smith TS, Benson DR. Short-segment pedicle instrumentation. Biomechanical analysis of supplemental hook fixation. Spine (Phila Pa 1976). 1996;21:288-94.
3. Dubousset J. Early failure of short-segment pedicle instrumentation for thoracolumbar fractures. A preliminary report. J Bone Joint Surg Am. 1995;77:648-9.
4. Davne SH, Myers DL. Complications of lumbar spinal fusion with transpedicular instrumentation. Spine (Phila Pa 1976). 1992;17 Supp l: S184-9.

5. Santoni BG, Hynes RA, McGilvray KC, et al. Cortical bone trajectory for lumbar pedicle screws. Spine J. 2009;9:366-73.
6. Glennie RA, Dea N, Kwon BK, Street JT. Early clinical results with cortically based pedicle screw trajectory for fusion of the degenerative lumbar spine. J Clin Neurosci. 2015;22:972-975.
7. Kasukawa Y, Miyakoshi N, Hongo M, Ishikawa Y, Kudo D, Shimada Y. Short-Term Results of Transforaminal Lumbar Interbody Fusion Using Pedicle Screw with Cortical Bone Trajectory Compared with Conventional Trajectory. Asian Spine J. 2015; 9:440-8.
8. Li H, Wang Z. Intervertebral disc biomechanical analysis using the FE modeling based on medical images. Comput Med Imaging Graph. 2006;30:363-70.
9. Smit TH, Odgaard A, Schneider E. Structure and function of vertebral trabecular bone. Spine (Phila Pa 1976). 1997;22:2823-33.
10. Alapan Y, Sezer S, Demir C, Kaner T, Inceoglu S. Load sharing in lumbar spinal segment as a function of location of center of rotation. J Neurosurg Spine. 2014;20:542-9.
11. Gong Z, Chen Z, Feng Z, Cao Y, Jiang C, Jiang X. FE analysis of 3 posterior fixation techniques in the lumbar spine. Orthopedics. 2014;37: e441-8.
12. Vadapalli S, Sairyo K, Goel VK, et al. Biomechanical rationale for using polyetheretherketone (PEEK) spacers for lumbar interbody fusion-A FE study. Spine (Phila Pa 1976). 2006;31: E992-8.
13. Lu YM, Hutton WC, Gharpuray VM. Do bending, twisting, and diurnal fluid changes in the disc affect the propensity to prolapse? A viscoelastic FE model. Spine (Phila Pa 1976). 1996;21:2570-9.
14. Matsukawa K, Yato Y, Imabayashi H, Hosogane N, Asazuma T, Chiba K. Biomechanical evaluation of lumbar pedicle screws in spondylolytic vertebrae: comparison of fixation strength between the traditional trajectory and a cortical bone trajectory. J Neurosurg Spine. 2016;24:910-5.
15. Ambati DV, Wright EK, Jr., Lehman RA, Jr., Kang DG, Wagner SC, Dmitriev AE. Bilateral pedicle screw fixation provides superior biomechanical stability in transforaminal lumbar interbody fusion: a FE study. Spine J. 2015;15:1812-22.
16. Dmitriev AE, Gill NW, Kuklo TR, Rosner MK. Effect of multilevel lumbar disc arthroplasty on the operative- and adjacent-level kinematics and intradiscal pressures: an in vitro human cadaveric assessment. Spine J. 2008;8:918-25.
17. Brown T, Hansen RJ, Yorra AJ. Some mechanical tests on the lumbosacral spine with particular reference to the intervertebral discs; a preliminary report. J Bone Joint Surg Am. 1957;39-a:1135-64.
18. Markolf KL. Deformation of the thoracolumbar intervertebral joints in response to external loads: a biomechanical study using autopsy material. J Bone Joint Surg Am. 1972;54:511-33.
19. Virgin WJ. Experimental investigations into the physical properties of the intervertebral disc. J Bone Joint Surg Br. 1951;33-b:607-11.
20. Li G, Wang S, Passias P, Xia Q, Li G, Wood K. Segmental in vivo vertebral motion during functional human lumbar spine activities. Eur Spine J. 2009;18:1013-21.

21. Renner SM, Natarajan RN, Patwardhan AG, et al. Novel model to analyze the effect of a large compressive follower pre-load on range of motions in a lumbar spine. J Biomech. 2007;40:1326-32.
22. Baluch DA, Patel AA, Lullo B, et al. Effect of physiological loads on cortical and traditional pedicle screw fixation. Spine (Phila Pa 1976). 2014;39: E1297-1302.
23. Perez-Orribo L, Kalb S, Reyes PM, Chang SW, Crawford NR. Biomechanics of lumbar cortical screw-rod fixation versus pedicle screw-rod fixation with and without interbody support. Spine (Phila Pa 1976). 2013;38:635-41.
24. Chao CK, Hsu CC, Wang JL, Lin J. Increasing bending strength and pullout strength in conical pedicle screws: biomechanical tests and FE analyses. J Spinal Disord Tech. 2008;21:130-8.
25. Oshino H, Sakakibara T, Inaba T, Yoshikawa T, Kato T, Kasai Y. A biomechanical comparison between cortical bone trajectory fixation and pedicle screw fixation. J Orthop Surg Res. 2015;10:125.
26. Takata Y, Matsuura T, Higashino K, et al. Hybrid technique of cortical bone trajectory and pedicle screwing for minimally invasive spine reconstruction surgery: a technical note. J Med Invest. 2014;61:388-92.

Figures

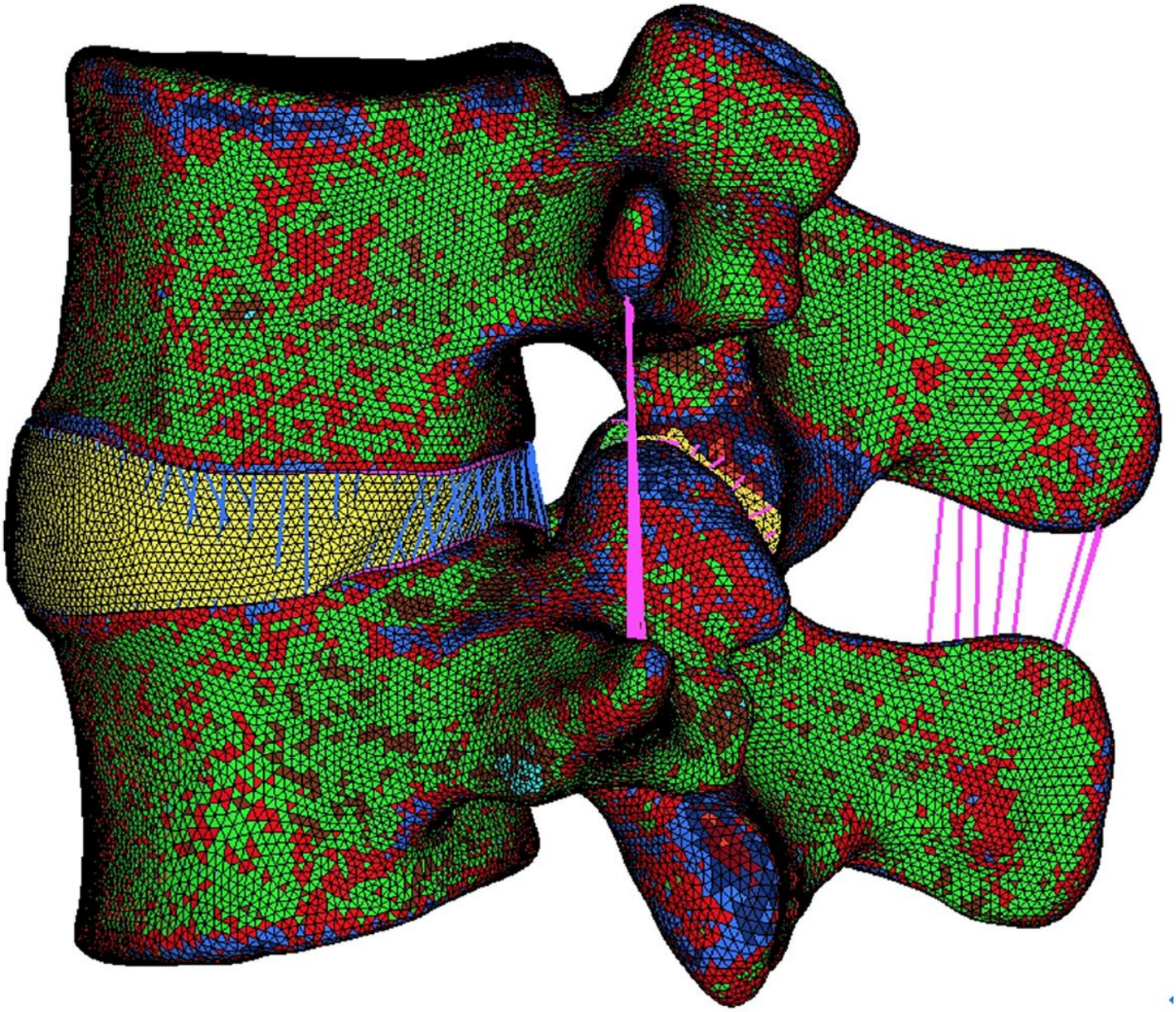

Figure 1

FE model of L4–L5 segment.

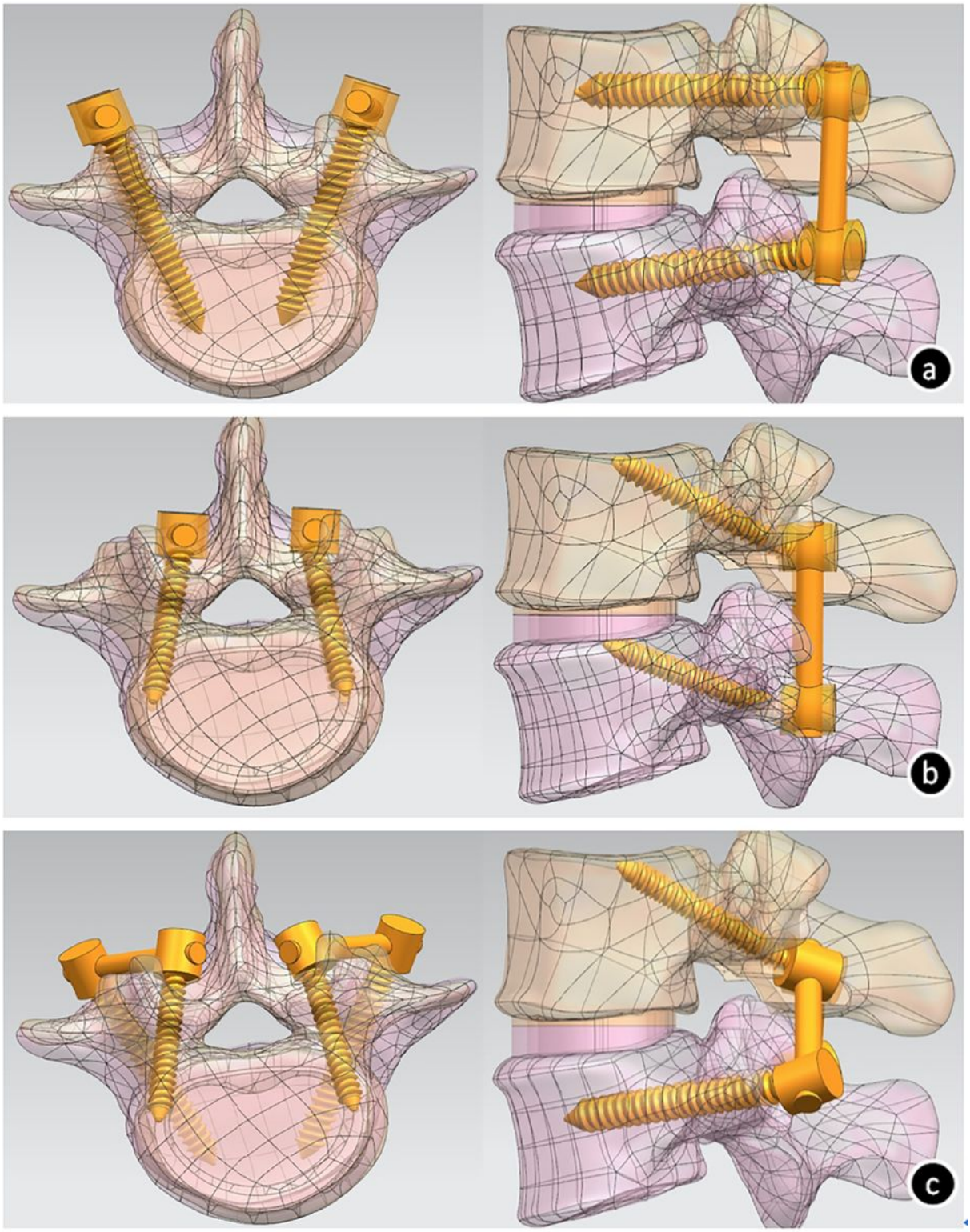

Figure 2

FE models. (a) TT screw fixation, (b) CBT screw fixation, (c) CBT + TT screw fixation.

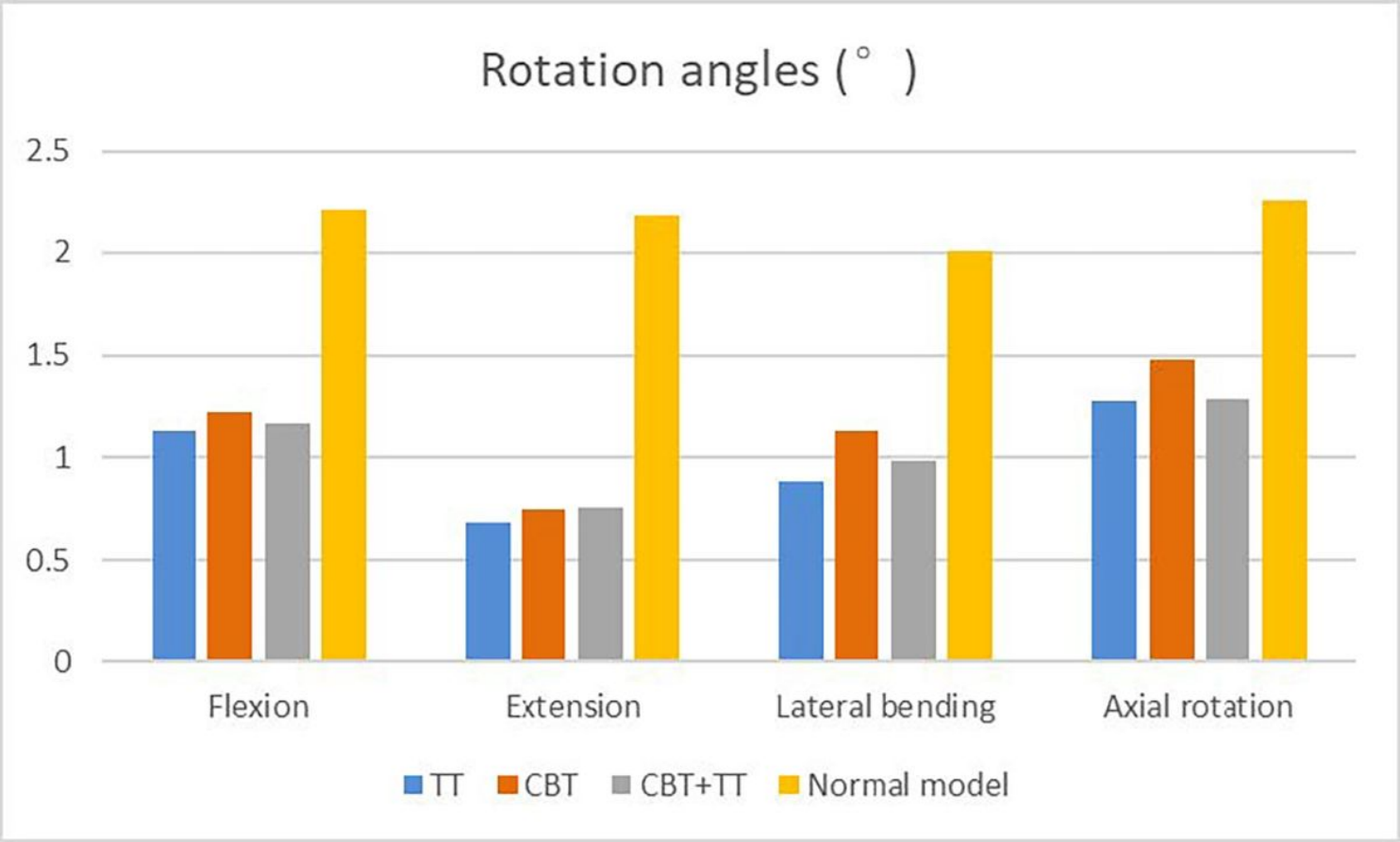

Figure 3

Changes in rotation angles during flexion, extension, lateral bending and axial rotation in the FE models with 3 different types of fixation methods.

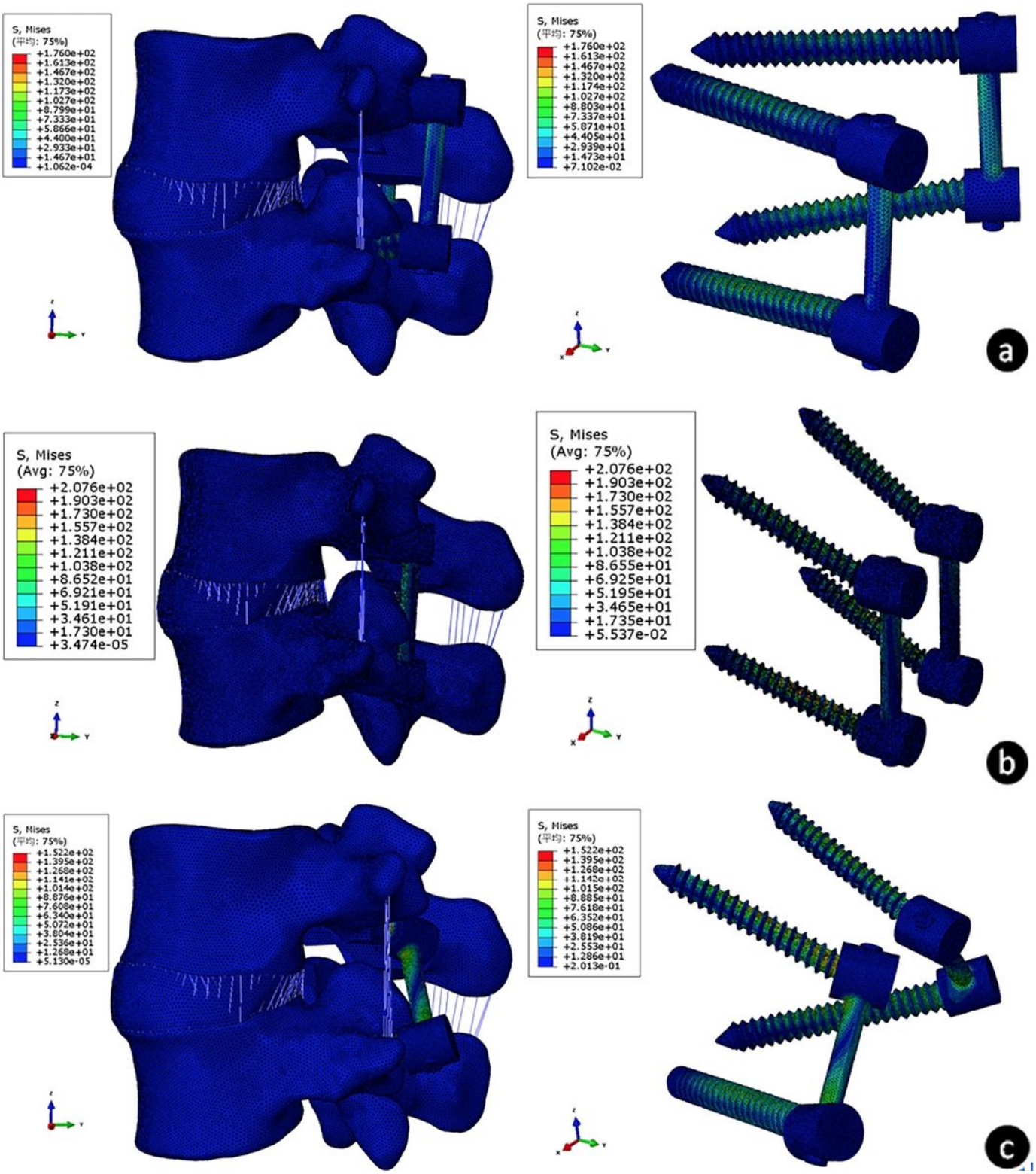

Figure 4

The stress distribution of three different internal fixation methods. (a) TT screw fixation, (b) CBT screw fixation, (c) CBT + TT screw fixation.

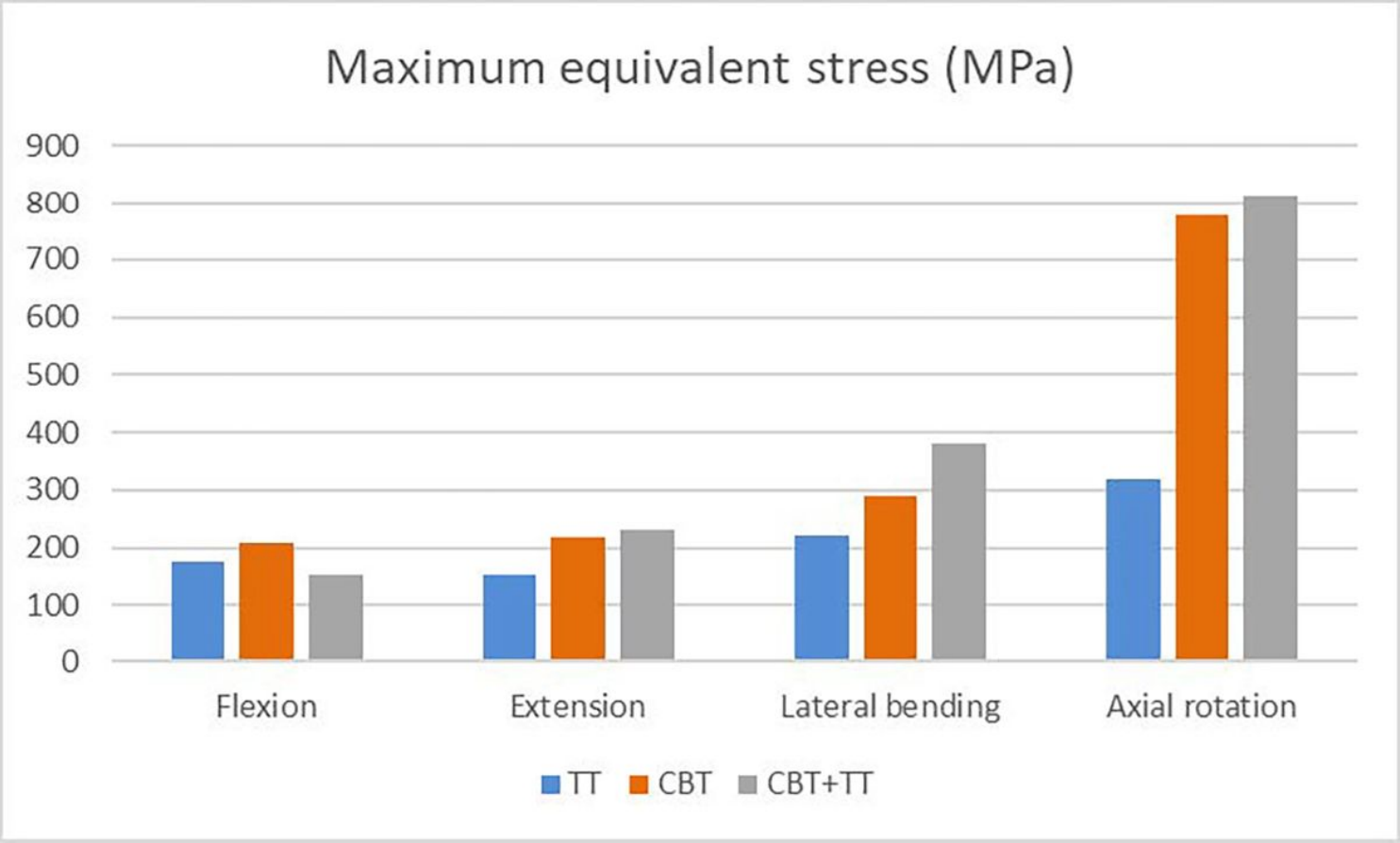

Figure 5

Comparison of maximum stress during flexion, extension, lateral bending and rotation in the FE models with 3 different types of fixation techniques.