Influence of Different Fixed System in Osteoporotic Lumbar Vertebrae in Fusion—A 3D Finite Element Analysis

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Abstract

Background Posterior lumbar interbody fusion (PLIF) is the most common surgical method for lumbar fusion surgery, and pedicle screws (PS) can provide effective fixation strength in normal bone. However, pedicle screws are prone to loosening in osteoporotic patients. Cortical bone trajectory (CBT) screw fixation and cement augmentation pedicle screw (CAPS) fixation are often used to reduce the risk of loosening. Although several studies are reported that the pullout strength of CBT screws or CAPS screws are higher than PS, there are no relevant studies on different internal fixation stress analysis. The purpose of this study was to compare the stresses of different fixation methods and analyze the stresses of different internal fixation systems through finite element analysis.

Methods: Five finite element models were established and tested by simulating PLIF surgery at L4/5. They included: (1) normal model (2) osteoporosis model (3) pedicle screw model; (4) cement augmentation pedicle screw (CAPS) model; (5) cortical bone trajectory (CBT) model. The range of motion (ROM), stress of fusion cage and screw-rod system of different models were analyzed by simulating flexion, extension, left bending, right bending, left rotation and right rotation movements through software.

Results: The ROM of the osteoporosis model was increased compared with the normal bone, and the osteoporosis was increased by 5.3%, 17.6%, 11.5%, 11.3%, 7.5%, and 8.3% compared with the normal model during flexion, extension, left bending, right bending, left rotation, and right rotation. After fixation, the ROM decreased, but the difference was not significant between the different fixation models. The stress of the screw-rods is the minimum one in the CAPS group. The PS group is the middle one, and stress of the CBT group is max. The cage stress is the minimum in the CAPS group which is slightly smaller than the cage stress in the PS group. The cage stress was the max in the CBT group.

Conclusion: PLIF can decrease the ROM of lumbar vertebra and fusion segment in osteoporosis, but different fixation methods have little effect on ROM. However, there was a great difference in the stress of implant, with the least stress in the CAPS group, the second in the PS group, and the greatest stress in the CBT. The same is true for Cage stress. At the same time, the maximum stress zone to occur at the screw-rod junction, while the maximum stress of cage occurs at the position in contact with the endplate. The study has guiding significance for clinical practice.

Introduction

Posterior lumbar intervertebral fusion (PLIF) is a classic surgical method for the treatment of lumbar degenerative diseases, which was first proposed by Cloward in 1952 (1). Many clinical studies have shown that PLIF surgery can restore stable sagittal position and enhance immediate postoperative stability, while the fusion rate of surgery is high.

In spinal fusion, pedicle screw (PS) fixation is the most common and classic fixation implant. Pedicle screws provide effective fixation strength in normal bone. However, for elderly patients, the situation of

osteoporosis is very common. Current studies have found that internal fixation is more likely to produce loosening when osteoporosis occurs (2). Santoni et al first proposed the use of cortical bone trajectory (CBT) screws in 2009 and some studies have shown that CBT screw has better strength than conventional pedicle screws in normal bone quality situations. cement augmentation pedicle screw can also be used for fixation in patients with osteoporosis, and good clinical results have also been obtained (3). There are no studies to show the stress profile of CAPS and CBT in osteoporotic patients.

In this study, finite element analysis was used to simulate the PLIF surgical model in osteoporosis and analyze the stress of common PS, CAPS, and CBT screws in fixation. By establishing five different finite element models, these models are (1) intact model (2) osteoporosis model (3) PS model; (4) CAPS model; (5) CBT model (Fig. (Fig. 1), to compare the range of motion (ROM), screws-rod stress, and fusion cage stress of different fixation methods.

Methods

Design, location, and timing

The trial was completed in Affiliated Hospital of Hebei University from March 2021 to May 2021.

Participants

A male volunteer is selected, height 174 cm, weight 72 kg. CT and MRI is used to rule out spine-related diseases such as previous spinal dysplasia or degenerative disease (such as scoliosis, lumbar spondylolisthesis and other diseases). The volunteers were informed and signed the informed consent related documents according to the relevant requirements, and submitted to the ethics committee for relevant review and approval.

Scanning CT and Reconstruction of Three-dimensional Models of the Lumbar Spine

The volunteers were placed supine on a CT machine and scanned using CT (GE, Sensation16Siemens, Germany) with scanning parameters: tube voltage 120 kV, tube current 125 mV, slice thickness 0.625 cm, interslice distance 0, no interval scanning, and CT data of L1-5 were obtained and saved as DICOM format files. Import DICOM file in Mimics 20.0 software (Materialise, Belgium), select "bone" threshold selection tool, select vertebral bone in relevant range, remove bone substance and fill vacant bone substance, reconstruct L1-5 vertebral three-dimensional model, then perform wrap, smoothing and other related processing for the model, remove excess triangular plane, generate and save it as STL format file. GeomagicStudio12.0 software (Geomagic Company, NC) was applied for materialization.

Hypermesh13.0 (AltairEngineering, Michigan, USA) was used to grid the intervertebral disc, bone, and ligament structures. The intervertebral disc was composed of annulus fibrosus matrix, nucleus pulposus, annulus fibrosus fibers, and upper and lower endplates. The nucleus pulposus accounts for 43% of the total disc (4), and trusselements with a diameter of 0.8 mm are applied to simulate annulus fibrosus fibers, which and the upper and lower endplates are between 15 and 30 ° in a horizontal plane, totaling 8 layers (5). The outer cortex was set at 0.5 mm, the endplate at 0.6 mm (6), and facet contact was set as plane-to-plant, where the coefficient of friction was defined as 0.1 (7). According to Polikeit et al., the elastic modulus of the normal model is indicated in the table, while the elastic modulus of all structures in the osteoporosis model is decreased: the elastic modulus of cancellous bone is decreased by 66%, and that of cortical bone, endplate, and posterior structures is decreased by 33% (8). All ligaments were modeled as non-linear connections. The material properties were confirmed according to the previous literatures (9, 10). See Table 1 for details.

Construction of finite element model

The surgical model was processed in Hypermesh13.0 software. L4/5 posterior lumbar fusion surgery was performed to remove the posterior partial lamina of L4, the medial 1/2 bone of inferior facet of L4 and the medial 1/2 bone of superior facet of L5, and the L4/5 intervertebral disc and annulus fibrosus.

Pro/Engineer 5.0 software was used to make screw-rods and fusion cage. The diameter of pedicle screw and bone cement pedicle screw was 6.5 mm, the length was 45 mm, the bone cement in CAPS was 2ml in each vertebra, the diameter of cortical bone trajectory screw was 5.0 mm, the length was 35 mm. The size of fusion cage is 28 * 10 * 11 mm. Assembly was performed in hypermesh13.0 software. Pedicle screws and cement-augmented screws were placed at the medial facet and midpoint of the transverse process on the vertebral body, and cortical trajectory screw placement methods were based on the method of Santoni et al (11). A total of five models were constructed, which were (1) normal model (2) osteoporosis model (3) PS model; (4) CAPS model; and (5) CBT model (Fig.2) .

Loading conditions and simulation of lumbar motion

Abqus 6.13 software was applied to load conditions to simulate lumbar motion. Assuming that the lumbosacral region is fixed, displacement and rotation do not occur under any loading conditions. Axial load was applied on the superior surface of the L1 vertebral body, while torsional moment was applied at the center of the L1 vertebral body. According to the previous literature records and the bearing capacity of human body as a reference, the axial load is 200N, and a torque load of 7.5 Nm is applied to simulate the spinal flexion and extension, lateral bending and rotation, respectively.

Assessment

The mechanical properties of model fixation were evaluated from three aspects: 1.L1-L5 range of motion (ROM) 2.L4/5 range of motion 3. Stress of internal fixation 4. Stress of fusion device

Results

Validation of the model

The finite element scheme consists of a model of 797,989 points and 132,131 units. We compared ROM data from the full model with previous cadaver and FEA studies (12). Our results are close to the data from previous studies (Fig.2).

Changes in the ROM of L1-S1 in different models

The ROM of the osteoporosis model was increased compared with the normal bone, and the osteoporosis was increased by 5.3%, 17.6%, 11.5%, 11.3%, 7.5%, and 8.3% compared with the normal model during flexion, extension, left bending, right bending, left rotation, and right rotation. However, after fixation, the ROM decreased, but the difference was not significant. The ROM of PS screw fixation was reduced by 14%, 17.4%, 11.1%, 10.9%, 14.8% and 19.1% compared with osteoporosis model in flexion, extension, left bending, right bending, left rotation and right rotation; the ROM value of CAPS screw fixation was reduced compared with PS screw fixation, and reduced by 15.4%, 17.8%, 12.5%, 11.9%, 16.9% and 21.1% compared with osteoporosis model in flexion, extension, left bending, right bending, left rotation and right rotation; the ROM value of CBT screw fixation was reduced by 13.9%, 19.3%, 11.3%, 10.6%, 12.7% and 17.1% compared with osteoporosis model in flexion, extension, left bending, right rotation and right rotation.

Change in the ROM of L4/5

The ROM of L4/5 in osteoporosis model increased compared with the intact model, by 2.7%, 1.9%, 11.4%, 11.1%, 4.6%, and 4.8% in flexion, extension, left bending, right bending, left rotation, and right rotation, respectively. After fusion, the ROM values of L4/5 decreased significantly, with CAPS decreasing the most, by 86.5%, 91.7%, 90.1%, 83.5%, 81.9%, and 86.1% in flexion, extension, left bending, right bending, left rotation, and right rotation, respectively, CBT decreasing the least (except extension), and by 83%, 91.4%, 86.7%, 74.5%, 72.4%, and 78.1% in flexion, extension, left bending, right rotation, and right rotation, respectively. PS was between the two models, and by 85.9%, 91.2%, 81.8%, 81.8%, and 84.1% in flexion, extension, left rotation, and right rotation, respectively.

The VMS of screw-rods

The model showed that the stress of screw-rods in CAPS group was the smallest, and the maximum stress VMS was 219.2Mpa, 55.49Mpa, 126.4Mpa, 236Mpa, 233.5Mpa, and 160.9Mpa in flexion, extension, left bending, right bending, left rotation, and right rotation, respectively; the stress in CBT group was the largest, and the maximum stress VMS was 304.4Mpa, 66.8Mpa, 215.9Mpa, 299.7Mpa, 395.5Mpa, and 318.5Mpa in flexion, extension, left bending, right bending, left rotation, and right rotation, respectively, and the maximum stress VMS of PS group was between the two models. the maximum stress VMS was 258.2 Mpa, 62.16 Mpa, 138.8 Mpa, 283 Mpa, 265.3 Mpa, and 167.9 Mpa in flexion, extension, left bending, right bending, left rotation, and right rotation, respectively.

The VMS of Cage

In the CAPS group, the cage had the smallest stress, and the VMS was 86.71Mpa, 10.04Mpa, 43.69Mpa, 32.26Mpa, 42.29Mpa, and 34.61Mpa during flexion, extension, left bending, right bending, left rotation, and right rotation; in the PS group, the VMS was slightly higher than that in the CAPS group, and the maximum stress VMS was 119Mpa, 11.61Mpa, 49.96Mpa, 33.73Mpa, 51.54Mpa, and 37.15Mpa during flexion, extension, left bending, right bending, left rotation, and right rotation; and in the CBT group, it was the highest, and the maximum stress VMS was 119.9Mpa, 12.25Mpa, 58.23Mpa, 38.29Mpa, 58.51Mpa, and 45.24Mpa during flexion, extension, left bending, right bending, left rotation, and right rotation, respectively.

Discussion

With the aggravation of the aging problem of society, lumbar degenerative diseases have gradually increased, and spinal surgery has also gradually increased (13). Therefore, internal fixation-related complications, such as internal fixation loosening, fracture, cage subsidence and other problems, have gradually become a research hotspot. Osteoporosis often occurs with spinal degenerative diseases, so the rates of related complications increases when patients have osteoporosis. Studying the stress of different internal fixation methods helps surgeons to select fixation methods, The suitable fixation choices can reduce the occurrence of complications.

At present, the most common method in spinal fusion is the application of pedicle screws for fixation. Pedicle screws were first designed and manufactured by King in 1949 (14), Boucher et al first applied it to the fixation of three columns of the spine (15), and Roy-Camille et al used pedicle screws to fix thoracic fractures with good results (16). Current studies have shown that there is a significant correlation between axial pullout stress of screws and bone mineral density (17, 18), and studies by Okuyama et al have shown that pedicle screws do not achieve effective fixation strength in vivo when bone density is less than 80 mg/cm² (19). The failure rate of pedicle screw fixation is as high as 12% when patients have osteoporosis (20). Current studies have found that the stability of pedicle screws is directly proportional to the screw diameter and inversely proportional to the thread spacing, therefore, increasing the screw diameter or reducing the thread spacing can increase the stability of pedicle screws (21).

The use of CBT screws was first proposed in 2009 by Santoni et al (11). Although the diameter and length of CBT screws are smaller than those of pedicle screws, they contact the cortical area due to the trajectory of the screws. Matsukawa et al. compared common pedicle screws with CBT screws, the mean torque was 1.58 ± 0.44 Nm for pedicle screws and 2.71 ± 1.36 Nm for CBT, and CBT screws increased 1.7 times the anti-pullout stress by 30% compared with common screws (22, 23). Liu et al. (24) compared CBT screw and PS screw and found that the greatest zone of screw stress was at the screw-rod junction for both CBT and PS screws, with CBT screws having significantly greater stress than PS screws. While in the case of dynamic fixation, the stress will further increase, and when the lumbar spine is flexed, the stress is the largest, about 800Mpa. Liu's study was mainly in normal subjects, while in osteoporosis models, our study also found similar results. CBT group increased by 119Mpa, 11.61Mpa, 49.96Mpa, 33.73Mpa, 51.54Mpa, 37.15Mpa in flexion, extension, left bending, right bending, left rotation, right rotation than PS group. Compared with pedicle screw fixation, it has been shown that CBT screws have less ROM in fusion segment, which indicates that CBT screws have better stability (25). In the adjacent segment, the ROM value of CBT screws was greater than that of common screw fixation (25). In our study, the result is a little different. The CBT group's ROM of L4/5 (the fusion segment) were similar with PS group. The difference may be due to different CBT screw sizes and osteoporosis. Our study also found that the CBT group had more cage stress than the PS group in osteoporosis. The result may be related to the smaller diameter and length of CBT screws compared with pedicle screws. When spine receives vertical stresses, shorter screws do not bear the stresses in the front of spine, and more stresses are concentrated in the cage. However, the maximum stress of the cage tends to be concentrated at the contact zones between the cage and the upper and lower endplates, which is also similar to previous studies (25).

Cement augmentation pedicle screws were first used in cadaveric and animal experiments as biomechanical research tests (26, 27). Subsequently, CAPS were gradually used in revision surgery after internal fixation failure. Wuisman et al. (28) used CAPS for degenerative scoliosis surgery and achieved good results. As so far, CAPS have been widely used in patients with osteoporosis (29–32). In a cadaver study, Tan et al. (33) found that CAPS significantly increased strength (69%) than PS, which in turn reduced the risk of cage subsidence. Liu et al. (34) found that the pullout strength and the energy to failure of internal fixation with CAPS were twice as high as with PS by applying different fixation in L1-4 cadaveric vertebrae. Sarzier's study (35) found that axial pullout forces increased 1.8 times to 2.1 times with the use of CAPS compared with PS. A study by Guo et al (36) found that the maximum VMS stress increased by 40.79%-54.66% when PS were applied compared with CAPS. While the maximum stress of cage occurred in flexion position, it increased by 19.43%-47.24% compared with CAPS. In our study, the maximum VMS stress was smaller when CAPS were applied, which decreased by 4.2%-16.6% compared with PS. The difference between previous studies and our study may be due to osteoporosis. The area of maximum stress is located at the screw-rod junction, which is also a common part of internal fixation fracture. In different internal fixation models, the maximum stress of cage occurred during spinal flexion, and the VMS of cage was smaller in the CAPS group than in the PS group, which may be related to the

anchoring effect of bone cement, while the VMS of cage also occurred at the junction of cage and upper and lower endplates, which may be responsible for cage subsidence.

This study presented some limitations. First, finite element analysis idealizes the material properties, and future studies should further refine the material properties. Secondly, the construction of the model cannot finely reduce the degenerative diseases of the vertebral body and intervertebral disc, such as osteophytes, vertebral height loss and so on; finally, osteoporosis behaves differently in different parts of the bone, which makes the model preparation of osteoporosis still have some deviations. Even so, this test still reflects the current fixation stress situation. Through the study, we found that fixation leads to a decrease in the lumbar ROM and the ROM of fusion segment in the osteoporosis model, but different fixation methods have little effect on the ROM value. Among them, the stress of screw and rod is very different, in which the stress of CBT screw is the largest, the stress of PS screw is the second, the stress of CAPS is the smallest, and the stress of cage is the same. At the same time, the VMS tends to occur at the screw-rod combination position, while the VMS of cage occurs at the position in contact with the endplate. It is found that it has certain guiding significance for clinical surgery. In future studies, the accuracy of the model should be further improved, so as to more truly simulate the surgical situation.

Conclusion

PLIF can decrease the ROM of lumbar vertebra and fusion segment in osteoporosis, but different fixation methods have little effect on ROM. However, there was a great difference in the stress of implant, with the least stress in the CAPS group, the second in the PS group, and the greatest stress in the CBT. The same is true for Cage stress. At the same time, the maximum stress zone to occur at the screw-rod junction, while the maximum stress of cage occurs at the position in contact with the endplate. The study has guiding significance for clinical practice.

Abbreviations

PLIF: posterior lumbar interbody fusion, PS: pedicle screws, CBT: cortical bone trajectory, CAPS: cement augmentation pedicle screw, ROM: range of motion, VMS: Von Mises stress

Declarations

Ethics approval and consent to participate

The study had been approved by the ethical committee of the participating hospitals. All subjects signed informed consent by each patient. All clinical investigations had been conducted according to the principles expressed in the Declaration of Helsinki.

Availability of data and materials

Please contact the corresponding author for data requests.

Competing interests

The authors declare that they have no competing interests.

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Authors' contributions

Miao Jun contributed to the data collections. Han Ye studied the design. Wang Xiaodong wrote the manuscript, Wu Jincheng analyzed the data. All authors read and approved the final manuscript.

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Consent for publication

Written informed consent for publication was obtained from all participants

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Table

Due to technical limitations, table 1 is only available as a download in the Supplemental Files section.

Figures

Figure 1

Finite element models of fixation constructs: a. intact model; b. osteoporosis model; c. pedicle screw (PS) model; d. cement augmentation pedicle screw (CAPS) model; e. cortical bone trajectory (CBT) model

Figure 2

a. In vitro and finite element total segmental rotation ROM b. In vitro and finite element total segmental rotation ROM c. In vitro and finite element total segmental rotation ROM d. The IDP of axial compressive load IDP intradiscal pressure ROM range of motion

ROM of the models

Figure 3

the range of motion of L1-S1

ROM of L4-5

Figure 4

the range of motion of L4-5

Figure 5

The VMS of rods and screws VMS= von Mises stress

Figure 6

The Von Mises stress distribution at rods and screws in different positions

The VMS of Cages

Figure 7

The Von Mises stress of Cages

Figure 8

The Von Mises stress distribution at Cages in different positions

Supplementary Files

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