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DOES THE SAGITTAL ORIENTATION OF THE PEDICLE SCREW AFFECT BIOMECHANICAL STABILITY? A FINITE ELEMENT STUDY

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ABSTRACT

Background: One of the most important steps in obtaining a successful outcome in spine surgery is appropriate placement of the pedicle screw. To test the hypothesis that sagittal angle of pedicle screw affects the biomechanical stiffness of the construct used in the spinal surgery, we evaluated the biomechanical results of different directions of pedicle screw on sagittal plane by performing a finite element (FE) analysis both in a single vertebral body and in a dual vertebral model.

Material/Method: Three-dimensional FE models of thoracic vertebrae (T10-T11) were used. The vertebral body was divided into three areas (upper, mid, lower one-third) in the sagittal plane. The entry points of the pedicle screw were same. The stiffness of different sagittal orientated screws in single vertebral body were evaluated in pull-out strength and in dual vertebral model strength of the screws were analyzed in flexion, extension and lateral bending movements.

Results: The screw at the upper one third of the vertebral body had the strongest pull-out load of 13.200N in single vertebral body model. The screw at the mid one-third of the vertebral body and lower-third of the vertebral body had 12.500N and 10.500N retrospectively. Flexion, extension and lateral bending tests had strongest loadings at upper one-third, mid one-third and lower one-third of vertebral body retrospectively.

Conclusion: The pedicle screw at the upper one-third of vertebral body in the sagittal was found to be more biomechanically stronger. This finding may be useful in clinical practice to prevent late complications of pedicle screw.

Keywords: Pedicle screw; sagittal plan; finite element; vertebral body

Level of Evidence: Biomechanical study, Level F.

INTRODUCTION

The use of pedicle screw is based on 1950s [(2)]. Pedicle screw is a surgical equipment, which has brought a new perspective to the spinal surgery after the use of Harrington rod and Luque's instrumentations. Especially in scoliosis surgery, the screw has enabled correction of more deformities with three-dimensional correction. Since it provides a more rigid stabilization compared to the other method, lower rates of pseudoarthrosis have been reported [(6,9,20,21)]. Pedicle screw has a wide area of usage regarding all fields of the spinal surgery such as spinal trauma, degenerative diseases, spinal tumor and infections of spinal region. Although this technique has such a wide area of use, probability of complications due to improper fixation of the screw is high because of its proximity to the neurovascular structures. Therefore, fixation of an appropriate and safe pedicle screw is the most important first step in the spinal surgery. Although it is predicted that being an experienced surgeon would decrease the complications due to pedicle screw, the rate of complications may reach up to 30 % even at experienced hands [(1,8,10,14,22)]. Complications due to improper fixation of the pedicle screw would lead to destructive results in early periods, although failures may also be seen in the chronic period due to the biomechanical structure of the implant in the screws considered of proper fixation [(5,16,19)].

We believe that, sagittal orientation of a pedicle screw fixed with free-hand

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method in the thoracic region affects biomechanical durability of the structure that is used in the spinal surgery. In order to test this hypothesis, we assessed biomechanical results of the orientation of the screw in sagittal plane in the vertebral body by performing the finite element analysis both in a single vertebral body, and in a dual vertebral model.

MATERIAL AND METHODS

T10 and T11 vertebral images of a healthy adult person aged 30 years were obtained on the computed tomography of the thoracal region. 3D image was transferred to the system. Finite element analysis model was produced for both vertebrae using Solidworks 2018 simulation program (Solidworks 2018, Dassault Systemes SE, France). Since the analysis was performed in both single and dual vertebra models, a disc model was created manually using Solidworks 2018 simulation program, because CT scan could not define the intermediate disk structure when the dual vertebra model was produced (Solidworks 2018, Dassault Systemes SE, France). The disc type was solid mesh, and so a curvature-based mesh model was used. Maximum element size was determined as 9.5229 mm and minimum element size as 1.90458 mm. "FEE plus iterative solver" was used for the finite element analysis. As the materials, High Density Polyethylene (HDPE) was chosen for the cortical bone, Low Density Polyethylene (LDPE) for the spongiosa bone, and Ti6A14V-ELI for the Screw-rod system.

The model type of High Density Polyethylene was defined as Linear Elastic Isotropic. Tensile strength was calculated as 2.21e+07 N/m^2, elastic modulus as 1.07e+09 N/m^2, Poisson's rate as 0.4101, Bulk density as 952 kg/m^3, and Shear modulus as 3.772e+08 N/m^2. The model type of Low Density Polyethylene was defined as linear elastic isotropic; tensile strength was calculated 1.327e+07 N/m^2, Elastic modulus as 1.72e+08 N/m^2, Poisson's ratio as 0.439, Mass density as 917 kg/m^3, and Shear modulus as 5.94e+07 N/m^2. The model type of Ti-6Al-4V-ELI was defined as linear elastic isotropic. Default failure criteria was defined as maximum von Mises Stress, yield strength as 8.27371e+08 N/m^2, Tensile strength as 1.05e+09 N/m2, Elastic modulus as 1.048e+11 N/m^2, Poisson's ratio as 0.31, Mass density as 4428.78 kg/m^3, Shear modulus as 4.10238e+10 N/m^2, and Thermal expansion coefficient as 9e-06 /Kelvin.

Same screw type and length, and same rod system were used for each analysis. The screws were of 6 mm diameter, 55 mm length, and polyaxial. The entry point of the pedicle screw was 2 mm caudal and 2 mm lateral to the junction of the lateral border of the superior facet joint and transverse process. Insertion point of each screw was defined as the same for each different region. The vertebral body was divided into 3 equal areas on the sagittal plane. The screws were inserted as to fit into 3 areas (1st, 2nd, and 3rd) on the sagittal plane. The first area was defined as the one third upper portion of the vertebral body, the second area as the one third medium portion of the vertebral body, and the third area as the one third lower portion of the vertebral body (Figure-1).

Each screw was subjected to pull-out load as to be parallel with the screw axis in the area of insertion in single vertebrae model (Figure-2.a-c).

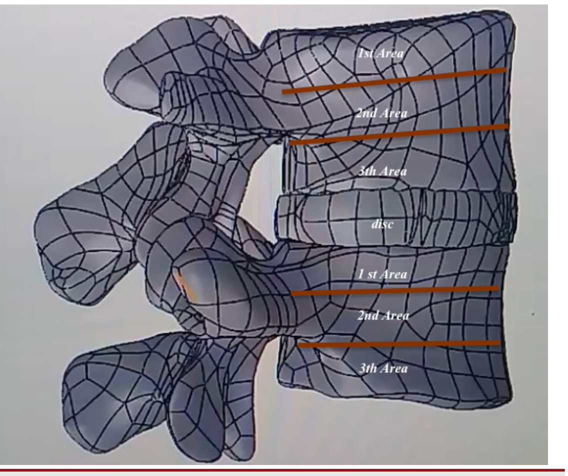

Figure-1. Illustration of areas 1, 2, 3 described in the study.

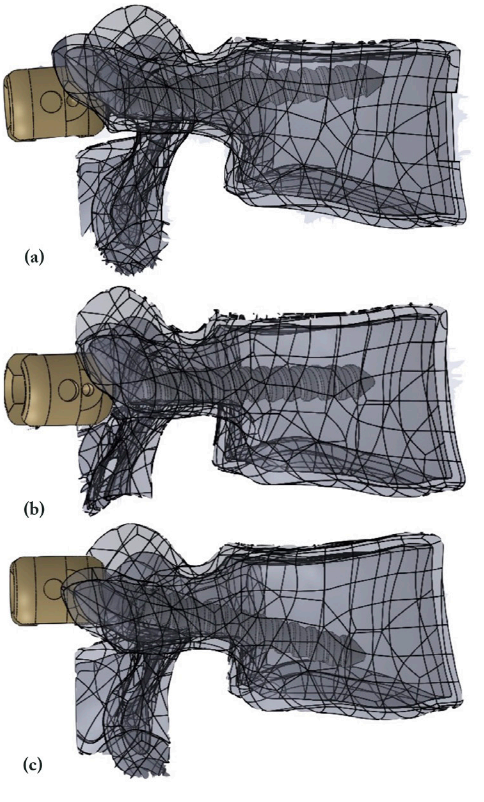

Figure-2.a-c. Description of pedicle screws in a single vertebral body model. **a)** Screw at first area, **b)** Screw at 2nd area, **c)** Screw at 3th area

In the dual vertebrae model, the screws inserted in the 1st, 2nd, and 3rd areas of each vertebra were combined with rod system, and three model of 4 screws inserted model was obtained (Figure-3.a-c).

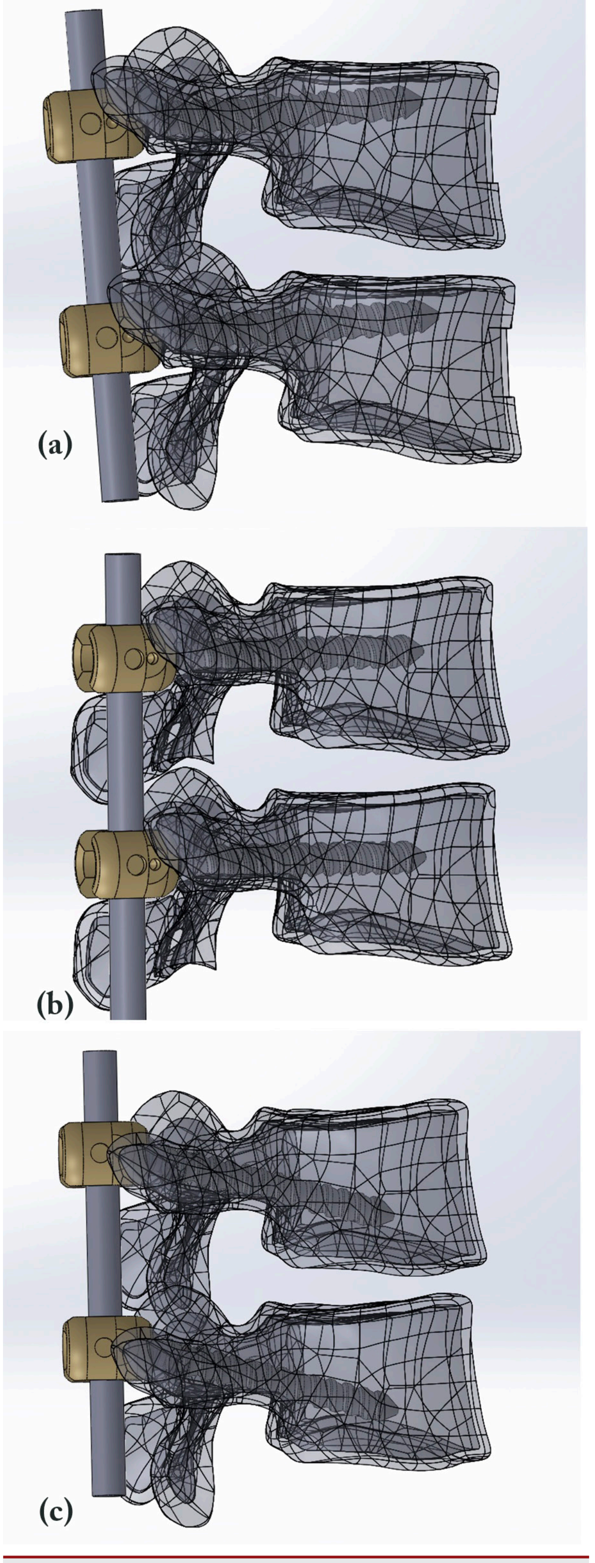

Figure-3. a-c. Description of pedicle screws in a dual vertebral body model, a) Screw at 1st area, **b)** Screw at 2nd area, **c)** Screw at 3th area

In these model, the lower vertebra was fixed with "surface boundary condition definition", and the upper vertebra was subjected to the forces that will provide vertebra to make flexion, extension, and lateral bending movements (Figure-4.a-c).

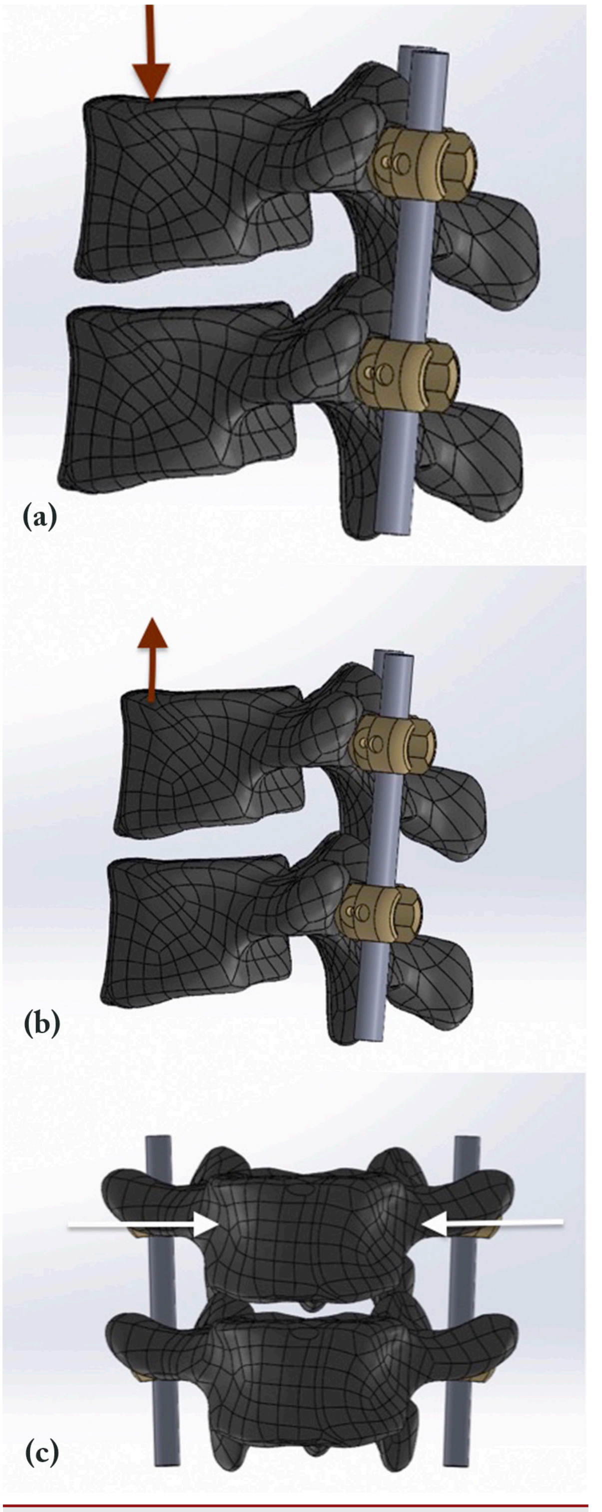

Figure-4.a-c. Illustration of forces providing; **a)** flexion, **b)** extension, **c)** lateral bending movements.

Loadings were started with 0 N, and the loads where the implants begin to deform were recorded. Each force was applied from the same point in each model, and durability of the screws against the forces applied were studied in the dual vertebra model.

RESULTS

"Pull-out" tests of screws in the single vertebral body model were performed with a static load applied at the screw axis for all three conditions. According to the results obtained, the screw which was delivered to the upper one third area (1st Area) led to deformation in the vertebra model at a pull out load of 13,200 N. Whereas the screw inserted in the middle one third area (2nd Area) caused deformation in the vertebra model at a pull out load of 12,500 N. On the other hand, the screw inserted in the lower one-third area (3rd Area) led to deformation in the vertebra model at a pull out load of 10,500 N. Accordingly the strongest attachment area of the screw is the upper one-third area of the vertebral body. Less strong areas are the middle and lower one third of the vertebral body, respectively (Table-1).

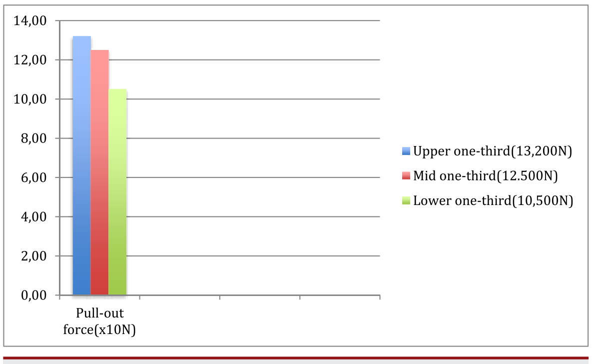

Table-1. Values of static load led to deformation in the single vertebra model for all three-pedicle screw conditions

Flexion tests in the dual vertebra model were performed with a flexion load created on the vertebra for all three conditions Test assembly consisted of 2 vertebra model, 4 screws and 2 rods. The system assembled with the screws delivered to the upper one third area (1st Area) was deformed at a flexion load of about 520 N. The deformation began in the screw neck. The system assembled with the screws delivered to the middle one third area (2nd Area) was deformed at a flexion load of about 500 N. In this case also the deformation began in the screw neck. The rate of deformation in the rods and screw necks was higher compared to the vertebra model with the screws in the first area. The system assembled with the screws delivered to the lower one third area (3rd Area) was deformed at a flexion load of about 400 N (Table-2).

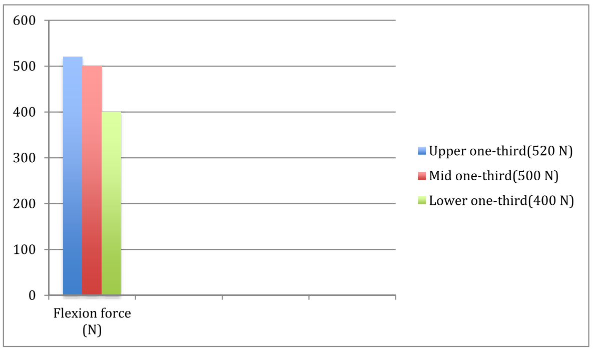

Table-2. Values of Flexion load lead to deformation in the dual vertebra model for all three pedicle screw conditions.

The deformation began in the rods. The rate of deformation in the screw necks and rods was significantly higher compared to the other two areas.

The extension tests applied in the dual vertebra model were performed with an extension load produced on the vertebral for all three conditions. Test assembly consisted of 2 vertebra model, 4 screws and 2 rods. The system assembled with the screws delivered to the upper one third area (1st Area) was deformed at an extension load of about 520 N. The deformation began in the screw neck. The system assembled with the screws delivered to the middle one third area (2nd Area) was deformed at an extension load of about 500 N. In this case also the deformation began in the screw neck. The rate of deformation in the rods and screw necks was higher compared to the vertebra model with the screws in the first area. The system assembled with the screws delivered to the lower one third area (3rd Area) was deformed at an extension load of about 400 N (Table-3).

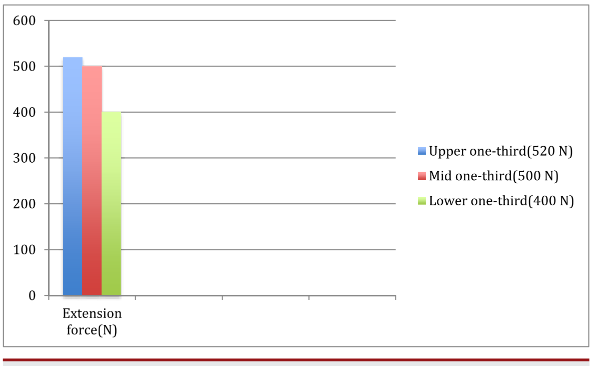

Table-3. Values of Extension load lead to deformation in the dual vertebra model for all three pedicle screw conditions.

The deformation began in the rods. The rate of deformation in the screw necks and rods was significantly higher compared to the other two areas.

Bending tests were performed with the lateral load created on the dual vertebra model for all three conditions. Test assembly consisted of 2 vertebra model, 4 screws and 2 rods. The system assembled with the screws delivered to the upper one-third area (1st Area) was deformed at a bending load of about 540 N. The deformation began in the screw neck. The system assembled with the screws delivered to the middle one-third area (2nd Area) was deformed at a bending load of about 440 N. In this case, the deformation began in the rods and screw necks. The rate of deformation in the rods and screw necks was higher compared to the vertebra model with the screws in the first area. The system assembled with the screws delivered to the lower one-third area (3rd Area) was deformed at a bending load of about 360 N (Table-4).

The deformation began in the rods. The rate of deformation in the screw necks and rods was significantly higher compared to the other two areas.

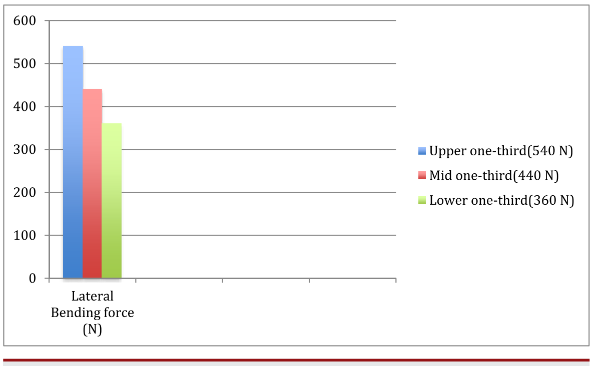

Table-4. Values of Lateral bending load created in the dual vertebra model for all three pedicle screw conditions.

DISCUSSION

There are a lot of accessory equipment to use for increasing the correctness and conformity of pedicle screws. Among the examples are C-arm fluoroscopy and computer aided navigation. The leading surgical techniques used especially by experienced spinal surgeons is insertion of pedicle screws with free hands. There are different entry points for pedicle screws in the free hand technique that have been described by authors [15,18,23]. Each technique define medialization of the pedicle screw according to the entry axis and orientation of the screw on the sagittal plane, but to apply this in the practice requires experience. Indeed, one of the main starting point of our study was to evaluate our pedicle screw insertion technique which we used in the operations [11]. We have previously stated that being perpendicular to the lamina in front of the pedicle with screw inserted would be helpful for adjustment of the sagittal orientation because in the free hand pedicle screw insertion defined in the literature, it may not always possible to adjust the sagittal orientation.

Especially, mistakes in the medialization may give rise to highly catastrophic outcomes for patients, differences in the orientation of the screw on the sagittal plane may cause outcomes that are not recognized in the early period, but may lead to implant failure in the advanced periods, such as loss of correction in the deformity and failed knitting in the fusion area [5,12,19,24].

Pull-out power of the pedicle screw has been evaluated in many studies in the literature, and the proper location of the screw in the vertebral corpus on the sagittal plane has been subjected to debate [3-4,16,19]. In this study, we evaluated 3 different orientation of the pedicle screw in the vertebral corpus on the sagittal plane. The finite element models that we created for this purpose revealed that attachment of the screw inserted in the upper one third area was stronger than the other areas in the evaluations made both at a pull-out power of the pedicle screw, and of two pedicle screws together. Our results show similarity with those of the literature. Newcomb et al. reported that superior screw angulation may be advantageous in reduction of the loosening and breaking of the screw [17]. Matsukawa et al. stated that lateral-cranial screw orientation a has higher pull-out strength compared to other orientations [16].

In a study from the literature comparing anatomic trajectory and straight-forward trajectory, it was emphasized that caudal orientation of the screw which was inserted in the anatomic trajectory on 22 degree sagittal plane provided the pedicle screw more bone channel contact [7]. In a biomechanical study by Lehman et al. comparing the two technique, straight-forward technique was reported to be superior over anatomic technique in terms of maximum insertion torque and pull-out strength [13].

Limitation of our study may be that bone mineral density was not included in the evaluation criteria. However, we studied vertebra model obtained from CT imaging of a 30-year-old healthy young adult patient. If we would studied in vertebra models obtained from more then one person we had to take into account bone mineral density since it would affect screw fixation strength (comparison of pedicle). In addition, as our study not an in vitro experimental trial, material properties defined in the modelling could cause us to obtain different results because of nonhomogenous structure of the spine. Therefore, we taken the material properties used in the modelling on a single vertebra identical and produced our own homogeneity.

CONCLUSION

The finite element analysis we performed revealed that insertion of the pedicle screw end in the upper one-third area of the vertebral body is better for the attachment.

Taking care to fit the screw in the upper one-third area would reduce complications such as late period fusion failure and loss in the correction of deformity.

Conflict of interest: None
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