SCIENTIFIC ARTICLE

Study of stress distribution in pedicle screws along a continuum of diameters: a three-dimensional finite element analysis

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Objective: To create a three-dimensional finite element model to identify the biomechanically optimal diameter of pedicle screws for placement in L1 vertebral bone.

Methods: The effects of pedicle screws with different diameters on the maximum Von Mises stress in L1 vertebra bone were evaluated by a finite element method. Pedicle screw diameters ranging from 4.0 to 6.5 mm were assessed.

Results: The simulation showed that, under axially oriented pullout forces, stress was decreased in all models when the diameter of screws was in the range of 4.0 mm to 6.5 mm. With a standard external load and a 6.5 mm diameter screw, load transferred to cortical and cancellous bone was reduced by 47.24% and 34.28%, respectively, and displacement of the screw was reduced by 21.35%. When the diameter was ≥ 5.0 mm, the variable of stress was stable in all models.

Conclusion: When the diameter of the screws is in the range of 4.0 mm to 6.5 mm, increasing the diameter of pedicle screw can improve the distribution of axial pullout stress on the screws, cortical bone and cancellous bone. Provided the bone mass permits it, pedicle screws with a diameter of not less than 5.0 mm should be chosen.

Key words: Finite element analysis; Mechanical; Pedicle screw; Stress

Introduction

Pedicle screws play an important role in the treatment of orthopedic diseases by providing stability, support and connection between fractured bones. Although pedicle screws successfully restore long-term stability of spinal segments in over 90% of cases, screw loosening, fracture, and pullout still contribute to a substantial failure rate^{1,2}. Excessive local loading on the vertebral body due to unfavorable biomechanical properties of the screw is the principal reason for surgical failure^{3–5}. To ensure long-term stability, not only should the material of the pedicle screw be biocompatible, but the design of the screw should also have biomechanical compliance. Many studies have indicated that the bone-screw interface plays an important

role in fixation. After the initial fixation, subsequent fusion is expected. Several characteristics affect the fixation stability of pedicle screws, including length, diameter, implant position, implant direction, implant technology and the quality of the host bone. The diameter is a particularly important mechanical factor which partially determines pullout strength and stability⁶. A complete understanding of how these characteristics affect the success of surgery is important in order to make effective clinical decisions about the diameter of the screws to be implanted.

Nowadays, the characteristics of pedicle screws that affect fixation stability are assessed mainly by biomechanical experiments. Several previous researchers have proved that increasing diameter of pedicle screws enhances the strength of screw fixation and decreases the stress on the vertebral segment. However, because all biomechanical analyses have only examined screws of a single diameter⁷, the effects of implant diameter on stress distribution and screw fixation stability remain unclear. The optimal range of screw diameter is hard to define. It is necessary to understand the role of screw diameter in regions with

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Table 1 Deep pedicle screw thread size from China National Standards (mm)

Diameter and code of thread	d_1		d_2		e ≈	p	r_3 ≈	r_4 ≈
	Basic dimensions	Deviation tolerances	Basic dimensions	Deviation tolerances				
HA3.5	3.5	0–0.15	1.8	0–0.15	0.1	1.75	0.8	1.2
HA4.0	4.0	0–0.15	1.9	0–0.15	0.1	1.75	0.8	0.3
HA4.5	4.5	0–0.15	2.3	0–0.15	0.1	1.75	0.8	0.3
HA5.0	5.0	0–0.15	2.4	0–0.15	0.1	1.75	0.8	0.3
HA6.5	6.5	0–0.15	3.0	0–0.15	0.2	2.75	1.2	0.8

poor quality bones. A variety of diameters are available to surgeons and the biomechanical properties of different diameters of screws need to be assessed.

The finite element method for the analysis of screw biomechanics provides many advantages over other methods in that it simulates the complexity of clinical situations. It offers a unique computational method for the evaluation of pullout strength related to different diameters of pedicle screws. Therefore, the purpose of this study was to use Ansys Workbench Design Xplorer finite element analysis to explore the relationship between pedicle screw pull-out strength and screw diameter. The results of this study could enhance clinical decisions in regard to selecting appropriate screw sizes.

Materials and methods

Three-dimensional model design

An L1 vertebral segment was modeled on a personal computer (CPU 2.99 G, RAM 4 G, ATI 4890) by using a CT image of the spinal segment from a 32-year-old healthy man. One mm slices were obtained and stored in DICOM (Digital Imaging and Communications in Medicine) format. Then, all images were imported into Mimics 11.1 software to build a three-dimensional (3-D) model of the L1 segment. The model was then imported into Pro/E software (Pro/E Wildfire, Parametric Technology Corporation, Needham, MA, USA) for further analysis. The L1 model contained a thick layer of cortical bone surrounded by dense cancellous bone. The thickness of the cortical bone was simulated at 1 mm⁸ and the cancellous bone was simulated mainly in the vertebral body⁹.

The pedicle screw was created using Pro/E software. A 3-D solid screw model was established that was visually identical to a real screw. Thread pitch was inputted according to the standard depth of pedicle screw thread size provided by the national YY 0018–2008 standards in China by Pro/E software (Table 1, Fig. 1). Screw diameter (D) was a changeable variable. D ranged from 4.0 mm to 6.5 mm. Screw length was fixed at 45 mm.

The vertebral body and pedicle screw models were assembled as a simplified screw-bone 3-D solid model by applying the Pro/E assembly function¹⁰. During the simulation, the relationship between contact surfaces of all models was set as “bond” in the software. The dimensions of the screw-bone complex model are shown in Fig. 2. All models were meshed and analyzed by Ansys Workbench

Figure 1 Deep pedicle screw thread size.

d_1 , thread diameter; d_2 , core diameters; e , crest width; p , thread pitch; r_3 , leading edge radius; r_4 , trailing edge radius.

Figure 2 A cross-sectional view of the model.

D, thread diameter of pedicle screw; L, length of the pedicle screw.

Table 2 Mechanical properties of the materials used in the 3-D finite element models

Materials	Young's modulus (Mpa)	Poisson ratio
Cortical bone	12,000	0.3
Cancellous bone	100	0.2
Screw (titanium alloy)	110,000	0.3

10.0 (SAS IP, Canonsburg, PA, USA). It is important to note that this study focused only on a single cortex of bone with fixation by pedicle screw. All materials used in the models were considered to be isotropic, homogeneous, and linearly elastic. The elastic properties were taken from the literature¹¹ and are shown in Table 2.

Elements and nodes

Models were meshed with 10-node tetrahedron elements. A refined mesh was generated around the screw-bone complex model; models were composed of 100,000 elements and 25,000 nodes on average (Fig. 3).

Constraints and loads

Biomechanical studies have shown that the pullout strength of lumbar pedicle screw ranges from 2000 to 3000 N^{12–15}. The purpose of this study was to evaluate the mechanical characteristics of the screw(s)-bone complex under a static load, the load applied to the pedicle screws being about 1/2 of minimum pull force (about 1000 N) along the long axis, which ensured that the screw would not be pulled out of the bone.

Convergence test

In the current study, convergence tests with mesh refinement were performed. The equivalent (EQV) stress

Figure 3 A cross-sectional view of the meshed model. The elements are of high quality.

observed in cortical bones was used for convergence monitoring and a tolerance of 3% was established. Changes of less than 3% in EQV stress in cortical and cancellous bones were indicative of convergence. An adaptive convergence was achieved when the mesh refinement loop was set to three and the detailed depth was set to two.

Results

The distributions of EQV stress in cortical and screw displacements were similar to that reported previously¹⁶ (Fig. 4). All figures in this study were automatically generated by the Ansys Workbench Design Xplorer program.

The result showed that, as the diameter of the screw increased within the range of 4 mm–6.5 mm, the maximal stress on the screw, cortical and cancellous bone and the maximum displacement of the screw all decreased (Table 3).

As the diameter of the screw increased, the response curves analysis revealed that the maximum EQV stress in the bones and maximal displacement of the screw decreased under an axially-oriented pullout force (Table 4). This figure, which represents the response curves, was formed by the median value of 45 mm for single input length parameter, and illustrates that the optimum value of single input length could be confirmed, according to the slope rate of the response curve, to be between –1 and 1. This was true in most cases, since D appeared to have a negligible effect on the maximum EQV stress in cancellous bone under the axially-oriented pullout force load. These data suggest that the most stable

Figure 4 The EQV stress distribution in bone and pedicle screw under axial pullout force load (D = 5.0 mm and L = 45 mm). (A) EQV stress distribution in the bone; (B) EQV stress distribution in the pedicle screw. The stress concentration is in the screw head.

Table 3 Maximum equivalent stresses in screw-bone complex sample (MPa) and maximum displacement of screw (mm)

L (mm)	D (mm)	Maximum equivalent stresses on screw	Maximum equivalent stresses on cortical bone	Maximum equivalent stresses on cancellous bone	Maximum displacement of screw
45	4.0	387.352	82.545	6.933	1.92E-01
45	4.5	275.724	58.475	5.032	1.83E-01
45	5.0	190.791	53.012	4.329	1.72E-01
45	6.0	103.372	45.883	3.545	1.67E-01
45	6.5	85.696	43.551	3.505	1.51E-01

and minimal level of stress and displacement value is achieved when $D \geq 5.0$ mm.

When a straight line is drawn at a tangent to a curve, the slope rate of the straight line represents the changing frequency of the curve. When the slope rate in our study ranged from -1 to 1 , it indicated a slight change in the maximum EQV stress in relation to the diameter of the pedicle screw. Based on the biomechanical assessment, the optimum implant parameters were determined to in the range where the maximum EQV stress reached its minimal value (Fig. 5).

Discussion

The success of spinal operations depends on the stability of the vertebrae after implantation of pedicle screws. The fixation strength of pedicle screws is dependent on many factors, the diameter undoubtedly being the most important of these. It is also the most easily modified variable in the clinical setting.

In addition to the length, the diameter of the pedicle screw plays a key role in enhancement of screw stability. McIlain *et al.* suggested that the resistance of the pullout force on the pedicle screw can be enhanced by increasing the diameter of the screw¹⁷. Based on the results from this study, it appeared that the stresses on all samples were decreased as the diameter was increased. This finding highlighted the importance of screw diameter in reducing bone stress and enhancing pedicle screw stability. However, bigger does not always translate to better, and if the diameter of the screw was more than 7 mm, the pedicle was predicted to fracture easily¹⁸.

Among the related pedicle screw parameters, diameter plays a key role in the success of pedicle screw implantation because it is directly related to implant stability. Until now, most studies have evaluated only independent parameters of diameter to determine their influence on the stability of the pedicle screw, and have not considered the influence of changing the parameters along a continuum. Exploring the idea of changing diameters along a continuum may enhance the stability of implants. Based

on our findings, the ideal screw should be at least 5.0 mm in diameter, keeping in mind that upper limits are also important.

The density of cancellous bone in the vertebral body is relatively low, making the evaluation of optimal diameter exceptionally important. In this study, increasing diameter resulted in a decrease in bone stress and a concomitant increase in screw stability. The diameter played a more significant role in reducing the stress on cancellous bone, indicating that more attention should be paid to screw diameter when considering the cancellous region of the vertebral bone.

Although finite element analysis is an important complement to research methods in orthopedic biomechanics, many simplifications and assumptions are made in the process of carrying out this type of analysis¹⁹. Complex human body movements are difficult to simulate precisely, even with the most advanced finite element analysis tools²⁰. The human spine model is idealized in using finite element analysis technology, resulting in certain mechanical factors, such as friction, not being considered. Therefore, finite element analysis can only partially reflect the biomechanical elements, and the results should be cautiously interpreted. Future physical experimental protocols should be used to validate the results of finite element simulation.

This report describes a preliminary study that utilized finite element modeling to examine the relationship between stability and the diameter size of pedicle screws in a single thickness cortical bone model. The biomechanical changes predicted with double the thickness of cortical bone and the mechanics of fixation with double cortex should be examined in future studies.

Conclusion

Several meaningful biomechanical conclusions can be drawn from these results. First, stress in the lumbar vertebrae is influenced by the diameter of the screw. Second, from a biomechanical perspective, screw diameters exceeding 5.0 mm are optimal for internal fixation of

Table 4 Response curves of input variables to Max Equivalent stresses stress in bones and response curves of input variables to Max displacement in the screw

Item	Maximum EQV stress on cortical bone	Maximum EQV stress on cancellous bone	Maximum displacement of screw
D:4.0–6.5 mm L:45 mm
Decrease Percentage*	47.24%	34.28%	21.35%
Optimum Selection†	D ≈ 5.0 mm		/
Decreased Percentage* = $(\text{Stress}_{\text{Max}} - \text{Stress}_{\text{Min}}) / \text{Stress}_{\text{Max}} \times 100\%$			
Decreased Percentage† = $(\text{Displacement}_{\text{Max}} - \text{Displacement}_{\text{Min}}) / \text{Displacement}_{\text{Max}} \times 100\%$			
Optimum Selection‡ = optimum diameter of pedicle screw

Figure 5 Chart depicting the site of optimum selection on a curve of changing slope rate.

lumbar vertebrae. While this study provides doctors and patients with important insights into the prognosis of fixation by pedicle screws, our results are limited by assumptions about the properties of materials and by the simplified models used in finite element analysis. These results should be considered, then, as a preliminary guide to selecting screws, since prospective clinical studies are required to confirm the results.

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Disclosure

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