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Effect of screw position on load transfer in lumbar pedicle screws: A non-idealized finite element analysis

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Abstract

Angled screw insertion has been advocated to enhance fixation strength during posterior spine fixation. Stresses on a pedicle screw and surrounding vertebral bone with different screw angles were studied by finite element analysis during simulated multidirectional loading. Correlations between screw-specific vertebral geometric parameters and stresses were studied. Angulations in both the sagittal and axial planes affected stresses on the cortical and cancellous bones and the screw. Pedicle screws pointing laterally (vs. straight or medially) in the axial plane during superior screw angulation may be advantageous in terms of reducing the risk of both screw loosening and screw breakage.

Keywords

Finite element analysis; non-idealized models; pedicle screw; screw orientation; stress analysis

INTRODUCTION

Excessive stress concentrations in bone after screw fixation in the spine can lead to localized bone failure and screw loosening, which can lead, in turn, to implant failures. Similarly, increased stresses on the screw can lead to screw breakage. Insertion of screws at an angle relative to the long axis of the spine during fixation of intervertebral stabilization systems (rods, plates) has been advocated as a way to enhance resistance to failure of the construct as a whole because it provides a more favorable stress distribution (Barber et al. 1998, Crawford et al. 2009, DiPaola et al. 2007, Santoni et al. 2009). When screws are angled inward, or "toed-in," during posteriorly directed loading, the screw-bone interface need only sustain a percentage of the force as shear at the threads, in addition to resisting some of the applied force as compression at the screw-bone interface. Loading bone under compression is favorable because the compressive modulus of bone is known to be at least 3 times greater than the shear modulus (Sanyal et al. 2012). Similar considerations allow interpretation of how stresses are better distributed during physiologic upright postural loading of spinal

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hardware, which should indicate that certain angles of inserted screws perform biomechanically better than others. Any method for improving resistance to failure of a fusion construct is especially important in poor-quality bone, which is known to perform poorly clinically during spinal screw fixation (Halvorson et al. 1994).

Finite element analysis (FEA) has been used to study stress distributions after screw fixation in the spine (Chen et al. 2012, Chen et al. 2008, Chen et al. 2003, Chevalier et al. 2008, Hussain et al. 2009, Kim et al. 2010, Wagnac et al. 2010, Wong et al. 2003). Using FEA, Hussain et al. (Hussain, Natarajan, Fayyazi, Braaksma, Andersson and An 2009) studied the effects of screw angle in the sagittal plane (rostrocaudal angle) in cervical spine fixation. Chen et al. (Chen, Lin and Chang 2003) used FEA to examine load transfer mechanisms within a screw/vertebra complex using different interface conditions and varying screw lengths, but did not include screw angulation. No published studies are available that describe FEA and the study of stresses in bone caused by varying screw angles in the axial plane, or screw angles in the lumbar spine (sagittal or axial plane).

Most studies involving FEA of the spine have included idealized models, built from the geometry of a single or representative anatomic structure, and incorporated material properties that are based on average values reported in the literature. However, as suggested many years ago by Halvorson (Halvorson, Kelley, Thomas, Whitecloud and Cook 1994), and more recently by Dreischarf et al.,(Dreischarf et al. 2014) Goel et al. (Goel et al. 2012), Meijer et al. (Meijer et al. 2011), Chevalier et al. (Chevalier and Zysset 2012), and Little et al. (Little and Adam 2012), FEA based on a non-idealized series of models, or patient-specific models, could be of greater value.

The hypotheses of this study were that varying pedicle screw orientations in the axial and sagittal planes would affect the peak stresses in the screw and in surrounding cancellous and cortical bone during loading, and that these variations in stress are dependent on different vertebral geometries as defined by a series of patient-specific models.

MATERIALS AND METHODS

Finite Element Analysis

Seven L4 vertebrae from human cadaveric spines were included (Table 1). Surface data were extracted from computed tomography images captured using a LightSpeed scanner (General Electric Medical Systems, Milwaukee, WI, USA) with axial slice spacing of 0.625 mm. Vertebral contours were detected by Scan-FE software (Simpleware, Exeter, UK), and vertebral geometries were meshed using ICEM (ANSYS, Inc., Canonsburg, PA, USA). Cortical bone was meshed with shell elements, whereas cancellous bone and the pedicle screw were meshed with 10-node tetrahedral elements. Different material properties were assigned to each type of bone and screw, with all 7 models having the same material properties (Table 2). The point of screw insertion for each model, which represented a clinically appropriate insertion point for a centered (sagittal plane) and straight (axial plane) pedicle screw, was determined by a neurosurgeon spine fellow. The same insertion point was used for all the different screw angles. Thus, the entry points did not vary and varying insertion angles were always relative to a centered and straight trajectory. The cortical wall thickness was set at 0.4 mm. The screw simulated a 40-mm × 6.5-mm pedicle screw made from titanium alloy, with 37.75 mm of the 40-mm threaded portion inside bone and 7.25 mm protruding from bone to point-of-load application. Soft tissues were not included in the models.

The resulting maximum stresses during a total of 9 screw orientations were compared (I1, I2, I3, M1, M2, M3, S1, S2, and S3), with each orientation (Fig. 1) represented by a combination of 1 of 3 angles in the sagittal plane (inferior [I], center [M], superior [S]) and 1 of 3 angles in the axial plane (lateral [1], straight [2], medial [3]). The actual maximum angles were defined by the geometry of each specimen, such that the screw threads would not penetrate the cortical wall.

A load of 500 N (approximate mass of the trunk above L4 for a person weighing 80 kg) was applied to the screwhead in 4 different directions; up (simulating flexion), down (simulating extension), and right and left (simulating axial rotation, Fig. 2). Only the left side of L4 was instrumented with a pedicle screw, simulating the inferior end of unilateral posterior screw-rod fixation. The maximum equivalent stress on the screw, cortical bone, and cancellous bone was analyzed using one-way repeated measures analysis of variance (followed by pairwise multiple comparisons using the Holm-Šidák method). The relationships among pedicle diameter, vertebral body height, vertebral width, pedicle angle (axial plane), total length of screw path (axial plane), and vertebral total volume (Table 1), and the maximum equivalent stress in cortical bone, cancellous bone, and screw for the various screw angles (with the same diameter screw in all cases) were analyzed using Pearson product moment correlations. Statistical significance was set at P=0.05.

Validation

A three-axis rosette strain gauge (350 ohm) was attached to the base of the left pedicle on an L3 vertebral body obtained from a fresh frozen cadaveric spine (61-year-old man, bone mineral density 0.866 g/cm^2). Compared to 11 other sites on a vertebral body, the base of the pedicle has been shown to be a site with increased strain during compressive loading.(Hongo et al. 1999). The left side of the L3 body was instrumented with a self-tapping 6.5 × 45-mm polyaxial pedicle screw (Lanx, Inc, Broomfield, CO) using a standard pedicle screw trajectory (i.e., centered and straight), and the majority of the body (excluding the screw and strain gauge) was embedded in a block of fast-curing resin (Smooth-Cast 300Q, Smooth-On, Inc., Easton, PA) (Fig. 3). The block was clamped in an angled vise such that compressive loads could be applied to a short rod attached to the head of the screw, similar to the loading conditions used during the FEA. Strain and applied load were recorded vs. time during tests loading the screwhead up, the screwhead down, and the screwhead to the left. Assuming homogenous isotropic material conditions (Hooke’s law, E = 12 GPa, v=0.3) and plane stress, we used the rosette strain measurements to calculate principal and von Mises stresses on the cortical bone at the base of the pedicle.(Shah et al. 2012) The experimentally determined stresses during loading were compared to the ranges of stress obtained from the seven geometrically different models during loads simulating flexion, extension, and left axial rotation.

RESULTS

With a centered (sagittal plane) and straight (axial plane) screw trajectory, the ranges of cortical bone stress at the base of the pedicle as found by FEA were similar to the experimentally obtained cortical bone stresses (Fig. 4), thus validating the models. The mean (±SD) screw angles in the sagittal plane, as defined by the geometry of each specimen and the left pedicle, were 17.2±4.5 degrees with the screw in a superior trajectory (S1, S2, S3), 13.6±6.0 degrees in a center trajectory (M1, M2, M3), and 11.0±4.0 degrees in an inferior trajectory (I1, I2, I3). Mean angles (±SD) in the axial plane were 16.1±3.5 degrees in the lateral trajectory (S1, M1, I1), 19.1±2.3 degrees in the center trajectory (S2, M2, I2), and 20.0±2.7 degrees in the medial trajectory (S3, M3, I3).

Disregarding screw angle, we found the mean cumulative stress (sum of maximum equivalent stresses from all 4 directions of loading) to be significantly greater in cortical bone (718.0±41.3 MPa) than in the screw (655.5±17.2 MPa, $P<0.001$). The mean cumulative stress in cancellous bone (96.9±5.9 MPa) was significantly less than stress in both the screw and cortical bone ($P<0.001$).

Cortical Bone

The stress contour maps generated during the FEA (Fig. 5) showed that the stresses on cortical bone at the base of the pedicle vary with the direction of loading and screw trajectory. A comparison of all 9 screw trajectories in terms of cumulative stress in cortical bone did not show a significant difference ($P=0.397$). When the cumulative values of maximum stress (all 4 loading directions combined) were grouped for all 7 specimens, the range of maximum equivalent stress (25th to 75th percentiles) in cortical bone varied the most with the screw angled inferior and lateral (I1, Fig. 6B) and the least when the screw was at center and straight (M2). However, analysis of maximum stress during individual loading directions showed a significant difference during flexion ($P=0.001$), with S2 stress > I2 stress ($P=0.003$), S3 stress > I2 stress ($P=0.008$), and S2 stress > I1 stress ($P=0.025$). All other pairwise comparisons during flexion were statistically insignificant ($P>0.05$), and analyses for other loading directions did not show any differences between screw trajectories ($P>0.06$). Ignoring the axial plane angle (i.e., comparing S, M, and I) showed that a superiorly angled screw caused significantly greater stress in cortical bone than an inferiorly angled screw (Fig. 7B, $P=0.035$), but neither a superiorly nor an inferiorly angled screw caused significantly different cortical stress than a centered screw (in the sagittal plane, $P>0.2$). There were significant negative correlations (Table 3, $P\leq0.042$) between the maximum stress in cortical bone and pedicle angle (S1), screw path length (I3 and M3), vertebral body height (I1 and M3), and vertebral body volume (M1, M3, and S3).

Cancellous Bone

Analysis of cumulative maximum stresses in cancellous bone during the 9 individual loading directions did not show any significant differences ($P=0.540$). Similarly, comparisons of maximum stresses during individual directions of loading did not show any significant differences ($P>0.1$). Ignoring the sagittal plane angle (i.e., comparing 1, 2, and 3) showed that screw trajectory had a significant effect on the maximum equivalent stress in cancellous bone during flexion, with medial screw angulation causing significantly greater stresses in the cancellous bone than lateral angulation (P=0.013, Fig. 7C). However, neither lateral nor medial screw angle caused a significantly different stress in cancellous bone compared with a screw with a straight trajectory (in the axial plane, P>0.09). Similarly, during axial rotation, a screw angled inferiorly (ignoring axial plane angle) caused significantly greater stress in cancellous bone than a screw angled superiorly (P=0.030, Fig. 7D), but neither an inferiorly nor a superiorly angled screw caused a significantly different stress in cancellous bone as compared to a centered screw (in the sagittal plane, P>0.14).

Screw

The range of maximum equivalent stress in the screw varied the most with the screw angled center and lateral (M1, Fig. 6C) and the least when it was angled superior and straight (S2). The cumulative values of maximum stress in the screw varied the most with the screw angled inferior and lateral (I1, Fig. 6A) and the least when it was angled inferior and straight (I2). A comparison of all 9 screw trajectories did not show a significant difference in cumulative maximum equivalent stress in the screw (P=0.632), or during individual directions of loading (P>0.15). Ignoring sagittal screw angle showed that a laterally angled screw caused a significantly greater stress in the screw than a medially angled screw (P=0.046) during left axial rotation (Fig. 7A). However, neither a medially nor a laterally angled screw caused a significantly different stress than a screw with a straight trajectory (in the axial plane, P>0.15). Correlation analysis of maximum stresses during different directions of loading (Table 3) showed that there were significant negative correlations (P≤0.047) between the maximum stress in the screw and pedicle diameter (M3, S2, and S3), vertebral body height (S1), vertebral body width (M3, S2, and S3), and vertebral body volume (I2, M3, S2, and S3).

There were significant positive correlations (Table 3, P≤0.048) between the maximum stress in cancellous bone and pedicle diameter with left axial rotation (I1, I3), screw path length (M2), vertebral body width (I1), and vertebral body volume (M2, S2, S3). There were also significant negative correlations between the maximum stress in cancellous bone and pedicle angle during extension (I3, S2) and, during right axial rotation, between the cancellous bone and the screw path length (I3, M3), and vertebral body volume (S2, S3). A negative correlation was also noted between the maximum stress in cancellous bone and vertebral body volume during flexion (S2).

DISCUSSION

The use of surface strain to validate finite element models in biomechanics is not uncommon,(Hao et al. 2011) and analyses of surface strain in the spine using rosette strain gauges have also been reported.(Hongo, Abe, Shimada, Murai, Ishikawa and Sato 1999, Kayanja et al. 2004, Kayanja et al. 2005, Szivek et al. 2002) Calculations of stress from strain recorded by a rosette strain gauge are less common in spinal biomechanics, but have been reported in a study involving the femur.(Shah, Bougherara, Schemitsch and Zdero 2012) Although the stresses calculated from strain in our in vitro study fell a bit outside the ranges of the stresses calculated by the finite element models during axial rotation, they were very near the mean values in two of the three directions of loading (Fig. 4), suggesting that the plane stress assumption for cortical bone at the site in question (base of pedicle) is valid during simulated flexion-extension. The slight deviation during axial rotation may be related to the slightly deeper vertebral body used during in vitro testing vs. those scanned by computed tomography for FEA (Table 1). As noted, the compressive modulus of bone is at least three times greater than the shear modulus of bone, and therefore, loading scenarios in which bone is loaded by a screw under compression are preferred to scenarios in which bone is loaded by a screw under shear for resisting failure of a screw. Another approach to the current study might therefore have been to assess how different loading patterns shift the relative amounts of compressive loading vs. shear loading along the screw. Instead, Von Mises equivalent stresses were reported because they are a combination of the normal stresses and therefore naturally take into consideration the relative amounts of compression and shear at each node. In terms of the findings in the current study, increased stresses in cancellous bone are perhaps of most interest and relevance, considering the relative magnitudes of the calculated maximum stress in the three materials studied (Fig. 6) and the corresponding Young’s moduli (Table 2), with the difference being the smallest for cancellous bone. It is theorized that excessive stresses in cancellous bone caused by load transmission via a load-bearing implant (such as a pedicle screw) can lead to cancellous bone failure and subsequent screw loosening. Our data show that a superiorly angled screw may be better than an inferiorly angled screw in preventing excessive levels of stress in cancellous bone (as found during one direction of loading), and a laterally oriented screw may result in less stress in cancellous bone than a medially angled screw (as found during another direction of loading). Our results also indicate that certain vertebral dimensions can have an effect on the maximum stress in cancellous bone during various screw trajectories (Table 3). For instance, the maximum stress in cancellous bone seems to increase, whereas the maximum stress in the screw decreases when the pedicle is relatively wide compared with when the vertebra in general is relatively large, at least during some directions of loading. This stress variation can, in part, be explained by more stress shielding being provided by cortical bone in narrow pedicles (i.e., more bone purchase by the screw into the cortical wall of the pedicle and therefore greater resistance to motion of the screw). Maximum stresses in the screw increased when the screw was large relative to the pedicle and vertebra in general, and mainly when the screw was oriented superiorly (sagittal plane) and straight or medial (axial plane). This finding is similar to what has been reported in the literature. In an in situ study, using synthetic vertebrae and pedicle screws instrumented with a strain gauge internally, McKinley et al.(McKinley et al. 1997) found that screw moment increased when the pedicle length increased or when the pedicle height decreased. They reported that pedicle width did not affect screw loads. The pedicle height was not measured in our study, whereas pedicle width was measured. The differences in results with respect to the effects of pedicle width (or screw diameter) are most likely due to differences in the mode of loading. McKinley et al. only included offset axial compressive loading (flexural moment), whereas our study included multidirectional loads.

When the pedicle screw was centered and straight, the maximum stresses in cancellous bone increased with increasing vertebral body height, suggesting that centered and straight screw may not be ideal in a relatively tall vertebral body. Our data also suggest that it may not be of interest to direct a screw medially (axial plane) and center it (sagittal plane) when the pedicle angle is large or when the vertebral body is narrow (short screw path), as this trajectory seems to increase the maximum stresses in cancellous bone, which could lead to failure of cancellous bone and subsequent screw loosening.

The effects of pedicle screw angle have been investigated by others. Cook et al. (Cook et al. 2000, Youssef et al. 1999), Crawford et al. (Crawford, Yuksel, Dogan, Villasana-Ramos, Soto-Barraza, Baek, Porter, Marciano and Theodore 2009), Inceoğlu et al. (Inceoglu et al. 2011), Kilincer et al. (Kilincer et al. 2007), Patel et al. (Patel et al. 2010), and Sterba et al. (Sterba et al. 2007). Sterba et al. (Sterba, Kim, Fyhrie, Yeni and Vaidya 2007) compared, in vitro, pedicle screws inserted at an angle along the axis of the pedicle (i.e., straight in the axial plane, as defined in our study) vs. parallel to the spinous process (i.e., laterally in the axial plane, as defined in our study) in cadaveric human bone; they found that laterally placed screws were more stable (resulted in less damage) than straight screws after cyclic loading. Similarly, Inceoğlu et al. (Inceoglu, Montgomery, St Clair and McLain 2011) studied screw angle using pullout tests and found that laterally angulated screws (with medial starting point) had greater pullout strength than screws inserted through a standard starting angle (presumably straight, as defined in our study). Our data, which show that the maximum stress in cancellous bone was significantly greater with medially vs. laterally placed screws, support these findings. Other authors have reported no effect of varying screw angles in the axial plane in terms of pedicle screw pullout strength (Crawford, Yuksel, Dogan, Villasana-Ramos, Soto-Barraza, Baek, Porter, Marciano and Theodore 2009, Kilincer, Inceoglu, Sohn, Ferrara and Benzel 2007).

Patel et al. (Patel, Shepherd and Hukins 2010) studied screw angle during in vitro pullout tests of cervical screw–plate constructs in polyurethane foam blocks of various densities. They concluded that although there was a correlation between screw angle and pullout force, the effect of screw insertion angle on pullout strength should not be considered in isolation from other parameters (e.g., bone quality, geometry).

Studies of screws in spinal vertebrae involving finite element models are ideal for studying the effects of isolated parameters that can be easily controlled and modified. However, many times the models are idealized and comparisons of different configurations of instrumentation are made using an “average” vertebra (Chen, Lin, Tsai, Wang and Chao 2012, Chen, Tai, Lin, Hsieh and Chen 2008, Chen, Lin and Chang 2003, Chevalier, Charlebois, Pahra, Varga, Heini, Schneider and Zysset 2008, Hussain, Natarajan, Fayyazi, Braaksma, Andersson and An 2009, Kim, Park, Kim and Lee 2010, Wagnac, Michardiere, Garo, Arnoux, Mac-Thiong and Aubin 2010, Wong, Gehrchen, Darvann and Kiaer 2003). Meijer et al. (Meijer, Homminga, Veldhuizen and Verkerke 2011) made note of this limitation while studying L3-4 motion and the effects of intervertebral geometry. While focusing on spinal stiffness without considering stresses in bone, they concluded that the sensitivity for individual differences in tissue properties (ligaments and disc) are much smaller than the sensitivity for differences in geometry, and that the actual physiologic range of mechanical properties remains unclear. We did not vary the material (mechanical) properties (i.e., Young’s moduli) in our study, but believe that this is one of the first studies involving FEA of pedicle screws in bone using multiple non-idealized models of vertebrae, each with unique geometries. By studying multiple geometries, it then becomes possible to add statistics, including standard deviation as a measure of variability, to the modeling outcomes. However, the standard deviation as recorded here has unique properties that deserve mention. Unlike standard deviation from a benchtop experiment, which represents variability due to material differences, anatomical differences, and experimental error among test subjects, this measure of standard deviation relates only to the variability caused by the anatomical differences. Standard deviation among modeling outcomes can therefore give additional insight into the sensitivity of different outcome parameters to anatomical variation at the level of variability expected among a typical group of individuals.

Limitations of Study

Model Assumptions—The vertebra used for the in vitro validation tests were within many of the limits set by the seven vertebrae used in the FEA (age, pedicle width, vertebral body height) (Table 1). However, it is possible that its slightly narrower pedicle angle, wider vertebral body width, and screw path length, possibly related to lumbar levels (L3 vs. L4), could have had an effect on the results. Given the variations in vertebral geometry along the lumbar spine, the findings from this study may not apply to all lumbar levels.

Pedicle screwhead design and the screw-rod interface were not included in the models. However, given that pedicle screws generally fail just inside the bone, away from the screwhead,(Youssef, McKinley, Yerby and McLain 1999) we believe that it is justified to exclude the screwhead and attached rod from the model. It is also possible that different screw insertion points would have produced different results, as could have been true if different amounts of cortex had been engaged. Parameters such as insertion point and cortical engagement are of interest for future studies.

Material Properties—The bony architecture of vertebrae is inherently non-homogenous, and it is possible that having more detailed assignment of material properties throughout the vertebrae could have an effect on the results. The main objective was to study the effects of screw angles and geometric variations of bone on stress distributions, and therefore we kept the material properties the same for all models and assumed homogeneity. In an experimental study involving strain gauged screws in synthetic bone models with a varying modulus for the portion representing cancellous bone, McLain et el.(McLain et al. 1997) showed that the changes in modulus had no measurable impact on pedicle screw-bending moment. Furthermore, including additional specimen-specific material properties would have made the analyses more convoluted and complex.

Applied Loads—On the basis of an approximate mass of a trunk and upper body above L4 for a person weighing 80 kg, we used a load of 500 N. The load was applied 7.25 mm distal to the screw entry point into bone, resulting in a bending moment on the screw of 3.62 Nm at the point of entry. Schultz et al. (Schultz et al. 1982) reported predicted forces of 470 N and 500 N (using two methods of calculations) on the L3 motion segment during upright standing. Rohlmann et al. (Rohlmann et al. 1997, Rohlmann et al. 1999) measured loads on a spinal fixator (positioned more or less vertically) during walking and various body positions, in vivo, and reported peak vertical loads of up to just over 300 N on one side in one patient (during walking down stairs), and peak bending moment between 3 and 7.5 Nm. Note that these patients were instrumented bilaterally, whereas our study involved only one side. Nonetheless, in terms of vertical load, the load used in this study is within both a predicted and an experimentally determined physiologic range. The load in our study was applied to the screwhead in four different directions, simulating flexion, extension, and right and left axial rotation. Rohlmann et al. (Rohlmann, Bergmann and Graichen 1999) measured in vivo loads on an internal fixator; however, these did not include twisting positions or axial rotation. Schultz et al. (Schultz, Andersson, Haderspeck, Ortengren, Nordin and Bjork 1982) reported predicted compressive loads at L3 during upright standing and resisting 10 kg twist to be between 730 N and 860 N, a bit higher than the 500 N used in our study. However, as pointed out by Chen et al.(Chen, Lin and Chang 2003), who used finite element to analyze load transfer mechanisms within a screw-vertebra complex under varying interface conditions and varying screw lengths, it is important to view results from a study like this in a relative sense rather than as absolute values for each case.

CONCLUSION

Orientation of a pedicle screw may play a role in the risk of screw loosening or screw breakage in vivo, and angulations in both the sagittal and axial planes can have an effect. Placing a pedicle screw pointing laterally (vs. straight or medially) in the axial plane during superior screw angulation may be advantageous in terms of reducing the risk for both screw loosening (decreased cancellous bone stress) and screw breakage (decreased stresses on the screw).

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ABBREVIATION

FEA	finite element analysis

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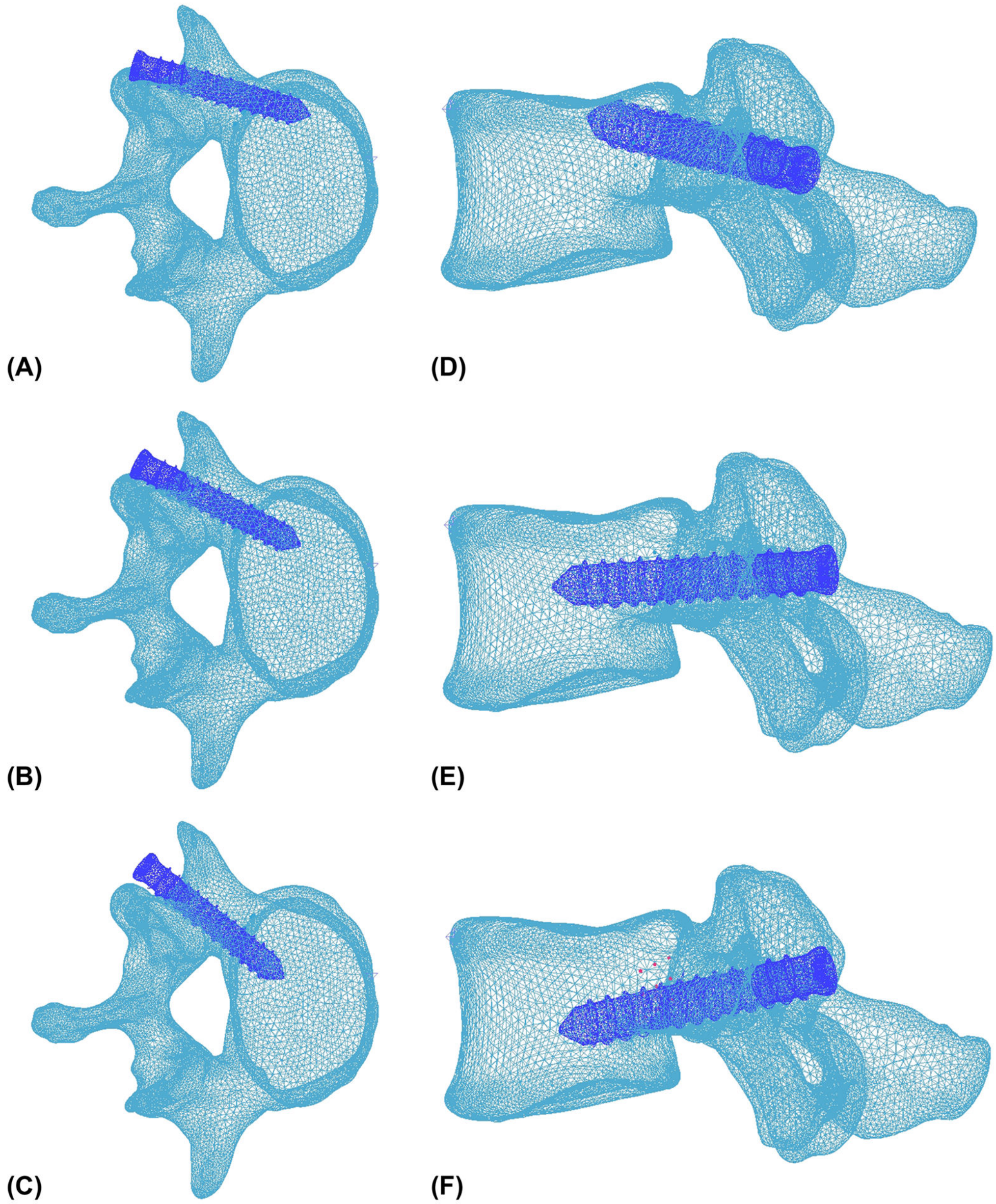

Figure 1.
Screw angles in axial **(A–C)** and sagittal planes **(D–F). (A)** Lateral, **(B)** straight, **(C)** medial, **(D)** superior, **(E)** center, **(F)** and inferior. *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

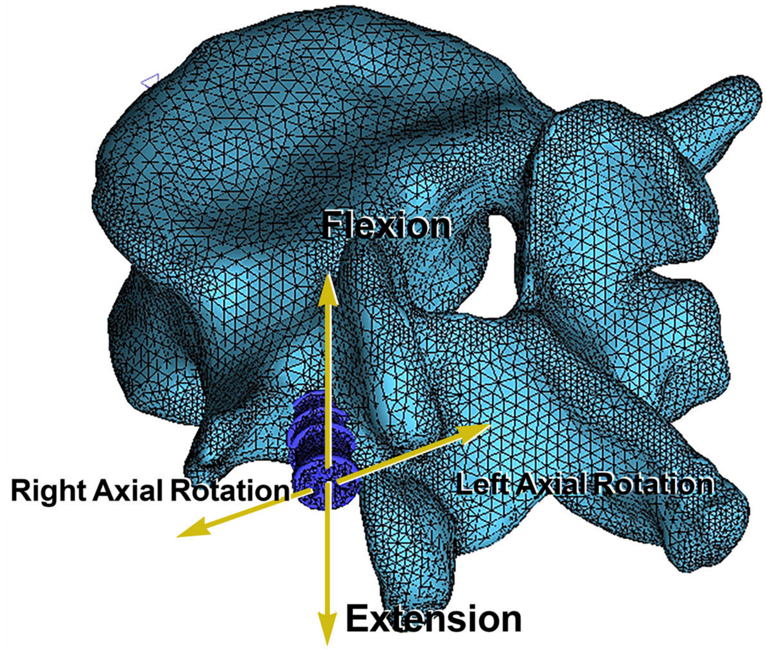

Figure 2.

Finite element model of L4 vertebra and pedicle screw showing directions of simulated loading conditions (flexion, extension, right axial rotation, and left axial rotation). *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

Figure 3.
Photograph showing cadaveric vertebra being tested with a vertical load simulating flexion. The specimen is upside down, and the vertical compressive load is applied to a rod attached to the screwhead. Strain in three directions was recorded during loading with a rosette strain gauge attached to the base of the left pedicle. *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

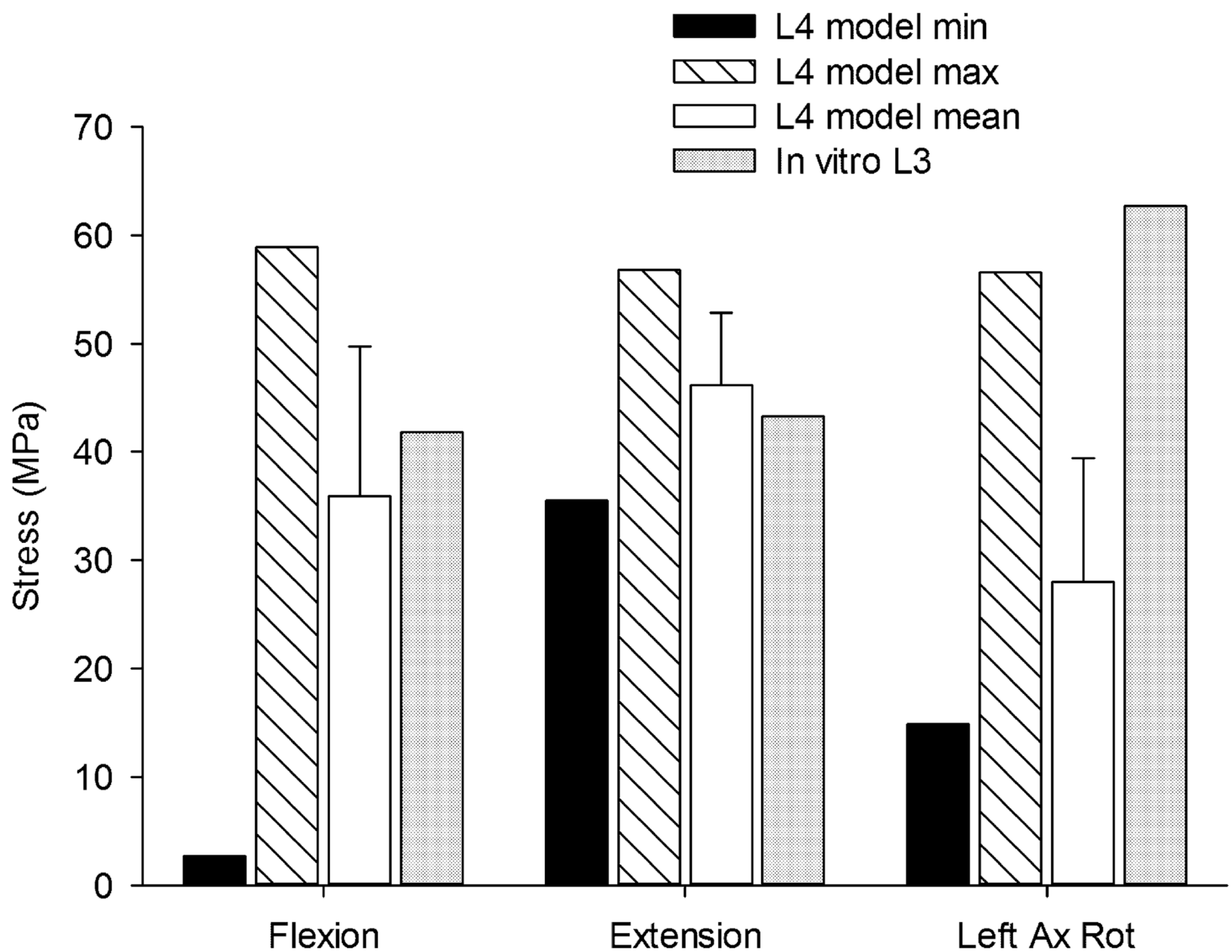

Figure 4.

In all directions of loading, the stress at the base of the pedicle derived from the recorded strain (in vitro L3) was in good agreement with stresses found using the finite element model. The error bars for the L4 model mean values indicate one standard deviation. Ax Rot indicates axial rotation. *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

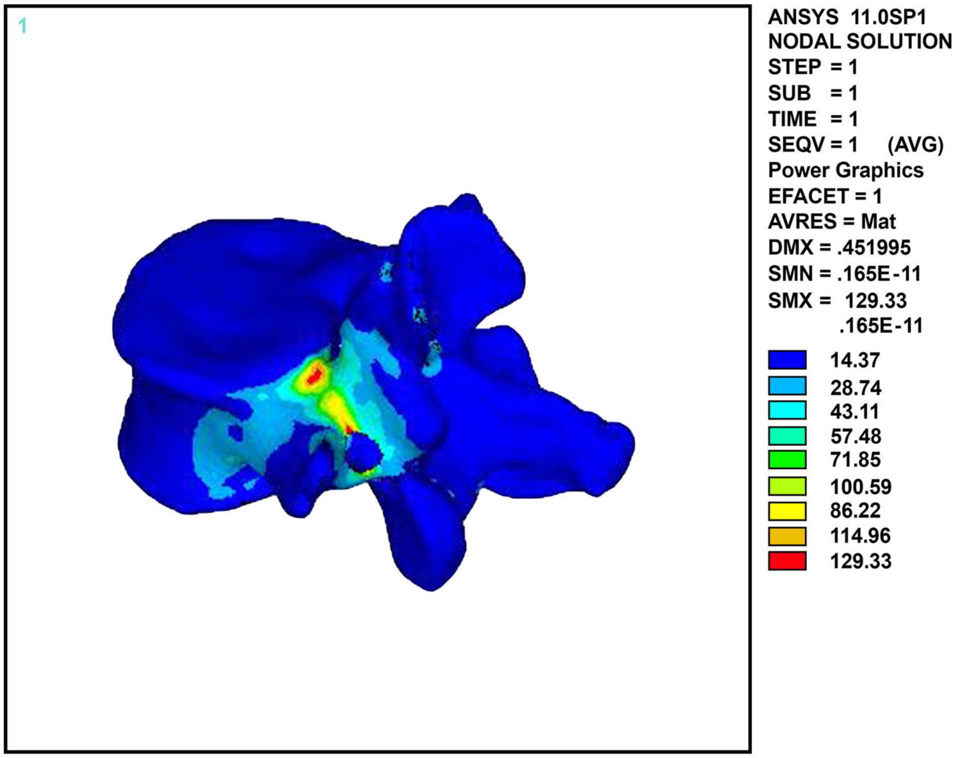

Figure 5.

Graphic representation of the calculated distribution of maximum equivalent stress, measured in GPa, on cortical bone during loading simulating axial rotation. *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

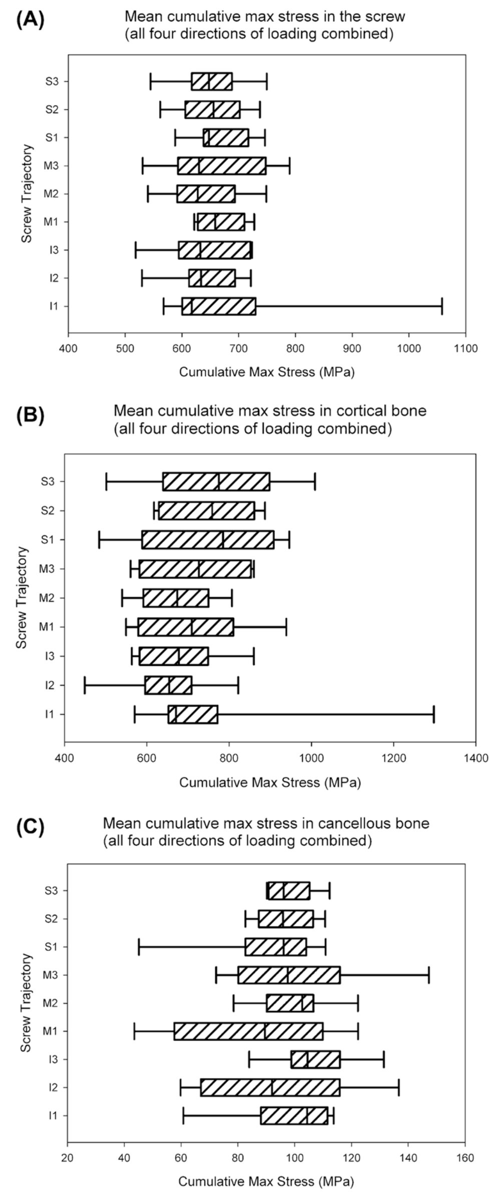

Figure 6.

Box plots showing 25th to 75th percentiles (plus median and error bars for 10th and 90th percentiles) of maximum equivalent stress for different screw orientations with all specimens and load directions combined on **(A)** the screw, **(B)** the cortical bone, and **(C)** the cancellous bone. *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

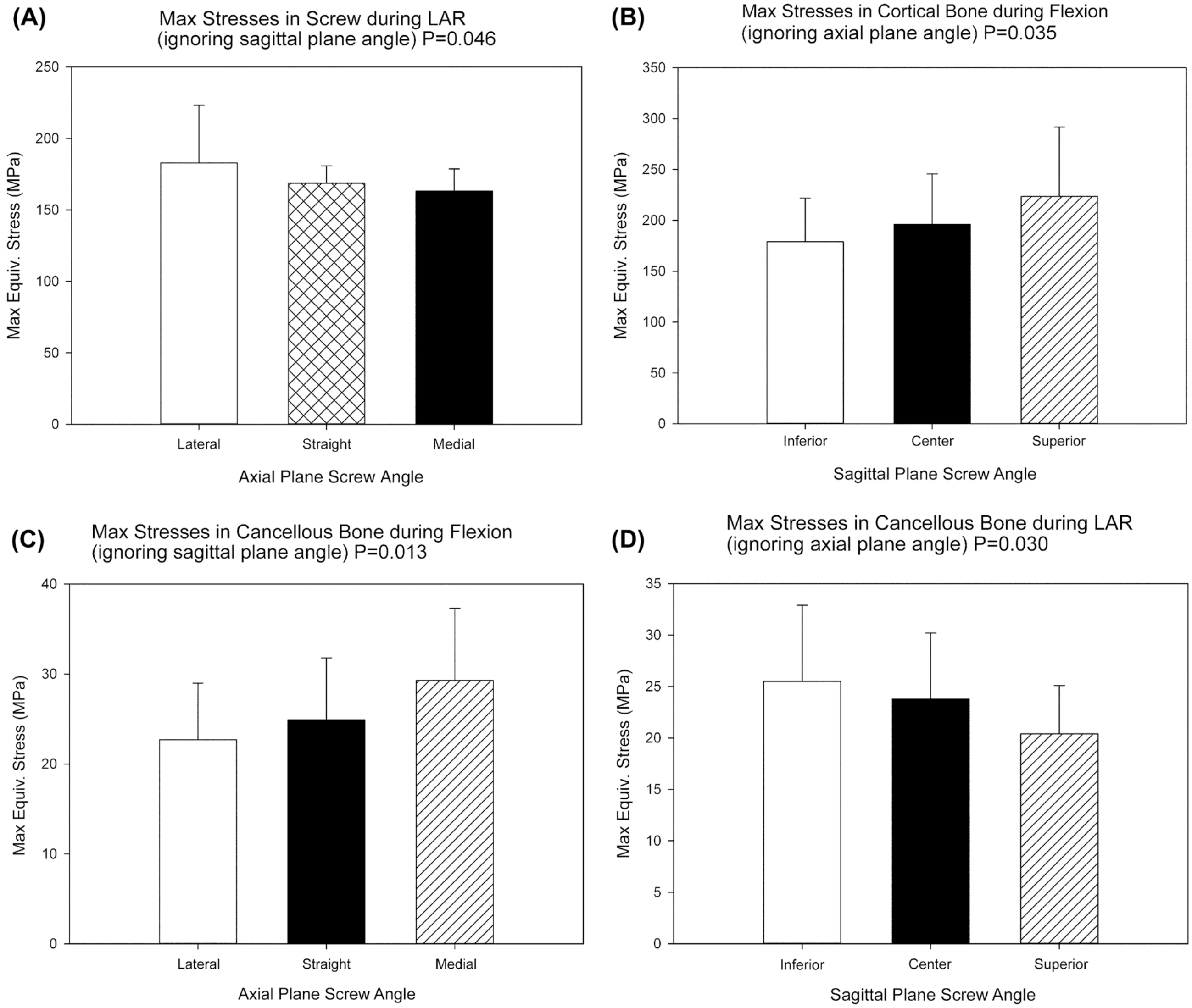

Figure 7.

Mean maximum stresses. **(A)** Screw during left axial rotation (LAR) (ignoring sagittal plane angle), **(B)** cortical bone during flexion (ignoring axial plane angle), **(C)** cancellous bone during flexion (ignoring sagittal plane angle), and **(D)** cancellous bone during LAR (ignoring axial plane angle). Equiv. indicates equivalent. *Used with permission from Barrow Neurological Institute, Phoenix, Arizona.*

Table 1

Specimen information including vertebral dimensions

Variable	Vertebrae Used for Finite Element Models Mean ± 1 SD (Range)	Vertebra Used for Validation Testing
Age (yrs)	54.6±15.9 (33–85)	61
Sex	4M/3F	M
Pedicle angle (degrees)	27.9±2.2 (24.8–31.5)	23
Pedicle width (mm)	12.2±1.8 (10.5–15.5)	11.4
VB height (mm)	27.9±2.4 (24.6–30.5)	30.0
VB width (mm)	33.0±2.2 (30.2–36.3)	47.5
Screw path length (mm)	56.2±2.8 (51.1–58.4)	64.4
L4 VB volume (cm^3)[a]	65.1±10.0 (52.1–80.5)	NA

Abbreviation: NA, not available; VB, vertebral body.

[a]From Z-Corp. software.

Table 2

Model properties

Component	Young's Modulus (MPa)	Poisson's Ratio
Cortical bone	12,000	0.3
Cancellous bone	100	0.2
Pedicle screw (titanium-alloy)	120,000	0.33

Table 3

Statistically significant correlation coefficients and *P* values for relationships between maximum equivalent stress and different vertebral geometry parameters, by screw angle and media during individual directions of loading[a]

Parameter and Screw Angle	Screw		Cortical Bone		Cancellous Bone	
	R value	*P* value	R value	*P* value	R value	*P* value
Pedicle diameter						
(I1)					0.814 LAR	0.026
(I2)						
(I3)					0.797 LAR	0.032
(M1)						
(M2)						
(M3)	−0.872 LAR	0.010				
(S1)						
(S2)	−0.875 FL	0.010				
(S3)	−0.772 FL −0.921 LAR	0.042 0.003				
Pedicle angle						
(I1)						
(I2)						
(I3)					−0.763 EX	0.046
(M1)						
(M2)						
(M3)						
(S1)			−0.795 RAR	0.033		
(S2)					−0.758 EX	0.048
(S3)						
Screw path length						
(I1)						
(I2)						
(I3)			−0.916 LAR	0.004	−0.766 RAR	0.044
(M1)						

Parameter and Screw Angle	Screw		Cortical Bone		Cancellous Bone	
	R value	*P* value	R value	*P* value	R value	*P* value
(M2)					0.859 LAR	0.013
(M3)			−0.847 LAR	0.016	−0.876 RAR	0.010
(S1)						
(S2)						
(S3)						
Vertebral body height						
(I1)			−0.795 LAR	0.033		
(I2)						
(I3)						
(M1)						
(M2)						
(M3)			−0.790 FL	0.034		
(S1)	−0.799 FL	0.031				
(S2)						
(S3)						
Vertebral body width						
(I1)					0.823 LAR	0.023
(I2)						
(I3)						
(M1)						
(M2)						
(M3)	−0.874 LAR −0.762 RAR	0.010 0.047				
(S1)						
(S2)	−0.815 LAR	0.026				
(S3)	−0.811 LAR	0.027				
Vertebral body vol.						
(I1)						
(I2)	−0.779 FL	0.039				

Parameter and Screw Angle	Screw		Cortical Bone		Cancellous Bone	
	R value	*P* value	R value	*P* value	R value	*P* value
(I3)						
(M1)			−0.813 FL −0.776 RAR	0.026 0.040		
(M2)					0.785 LAR	0.036
(M3)	−0.787 EX	0.036	−0.959 LAR	0.001		
(S1)						
(S2)	−0.870 FL	0.011			−0.777 RAR 0.760 LAR	0.040 0.048
(S3)	−0.913 FL	0.004	−0.771 FL	0.042	−0.850 FL −0.874 RAR 0.768 LAR	0.015 0.010 0.044

Abbreviations: EX, extension; FL, flexion; LAR, left axial rotation; RAR, right axial rotation.

[a] Combinations of angles in the sagittal plane (inferior [I], center [M], superior [S]) and axial plane (lateral [1], straight [2], medial [3]).