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Biomechanical analysis of differing pedicle screw insertion angles

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Abstract

Background. Pedicle screw fixation to stabilize lumbar spinal fusion has become the gold standard for posterior stabilization. A significant percentage of surgical candidates are classified as obese or morbidly obese. For these patients, the depth of the incisions and soft tissue makes it extremely difficult to insert pedicle screws along the pedicle axis. As such, the pedicle screws can only be inserted in a much more sagittal axis. However, biomechanical stability of the angled screw insertion has been controversial. We hypothesized that the straight or parallel screw was a more stable construct compared to the angled or axially inserted screw when subjected to caudal cyclic loading.

Methods. We obtained 12 fresh frozen lumbar vertebrae from L3 to L5 from five cadavers. Schantz screws (6.0 mm) were inserted into each pedicle, one angled and along the axis of the pedicle and the other parallel to the spinous process. Fluoroscopic imaging was used to guide insertion. Each screw was then subjected to caudal cyclic loads of 50 N for 2000 cycles at 2 Hz. Analysis of initial damage, initial rate of damage, and total damage during cyclic loading was undertaken.

Findings. Average total fatigue damage for straight screws measured 0.398 ± 0.38 mm, and 0.689 ± 0.96 mm for angled screws. Statistical analysis for total fatigue damage ratio of angled to straight screws revealed that a significant stability was achieved in straight-screw construct ($P < 0.03$).

Interpretation. This study showed that straight screw insertion results in a more stable pedicle-screw construct. The angled screw insertion technique resulted in more scattered values of damage indicating that the outcome from the angled screw fixation is less predictable. This validates the use of this technique to implant pedicle screws across the axis of the pedicle (parallel to the mid sagittal line) rather than along the axis, and has broad implications in instrumented posterior lumbar spinal surgery.

Keywords: Pedicle screw; Cyclic loading; Biomechanical stability; Creep

1. Introduction

Pedicle screw fixation has become the mainstay of fixation for stabilization of the posterior lumbar spine. Originally described by Boucher in 1959, Roy-Camille popularized this technique in Europe in the 1960s, and his spinal plating system has been called the "predecessor of most modern pedicular screw–plate fixation systems" (Boucher, 1959; Roy-Camille et al., 1976; Roy-Camille, 1992). Pedicle screw fixation is now readily accepted for treatment of fractures, tumors, and degenerative disease. Loosening due to fatigue loading and screw breakage are commonly cited reasons for failure, and numerous studies have been conducted to determine which factors are most important in determining biomechanical stability of the pedicle screw (Esses and Bednar, 1989; Willet et al., 1993; Zdeblick et al., 1993). To date, biomechanical studies have for the most part examined pullout failure of the screw as the endpoint to determine stability (Barber et al., 1998; Law et al., 1993; Yerby et al., 1997). Even those few reports that used a cyclic loading model utilized a displacement control rather than load control mechanism to determine

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relative stability (Barber et al., 1998; Law et al., 1993; Soshi et al., 1991). While parameters studied included using bigger screws, drilling or probing the pilot hole, tapped and untapped screws, coupling, angular insertion, and augmentation with bushings and polymethylmethacrylate, fatigue failure based on the clinical scenario has rarely been reported.

Morphometric anatomic studies have determined that pedicles flare out laterally from the upper to lower lumbar spine (Ruland et al., 1991; Zindrick, 1991). Transverse pedicle angles of the lower lumbar spine range from 8.0–23.5° at L3 (mean 14.4°) to 19.0–44.0° at L5 (mean 29.8°) (Zindrick et al., 1987). Some studies have suggested that convergent screws are a stronger construct, and have recommended that screws be inserted axially within the lumbar pedicle (Barber et al., 1998). On the basis of these studies and those testing pullout strength (Law et al., 1993; Yerby et al., 1997), screw insertion technique along the axis of the pedicle has been described as superior, with increasing angular distance from the vertebral midline at lower levels of the lumbar spine (Cook et al., 2000). However, the patient population subjected to surgery includes a significant number of obese or morbidly obese individuals who present a challenge in exposure for pedicular insertion of screws. The incisions are deep and approaching the axial pedicle along its axis is difficult. As an alternative to pedicular insertion, in the technique described by Roy-Camille in 1976 and 1992, pedicle screws are inserted in a vertical fashion, crossing the axis of the pedicle rather than proceeding in line with it. While attempts have been made to determine the stability of these screws, most studies utilize a displacement control to determine pullout strength at the bone–screw interface, rather than examining them dynamically at sub-failure forces to determine relative stability based on amount of screw toggle acquired during fatigue testing (Barber et al., 1998; Brantley et al., 1994; Soshi et al., 1991).

This study was designed to determine the stability of pedicle screws that were inserted by both straight and angled techniques. Cyclic, sub-failure load control was used to simulate in vivo loading. The displacement of each screw was measured and compared with the contralateral screw that was inserted by the differing technique. Based on this work, the fatigue stability of the two different types of screw insertion technique was examined by answering two research questions: (1) is the rate of damage different between the two screw insertion methods? (2) is there a difference in stiffness and creep damage between the two methods?

2. Methods

2.1. Specimen preparation and screw implantation

Use of human tissue was approved by our hospital Institutional Review Board. Five fresh frozen cadaveric spines were obtained from a tissue bank. These specimens were procured from T6 to the sacrum with minimal soft tissue attachment and were stored at −32 °C. None had a history of metastatic disease. Fourteen total vertebral bodies were tested in this protocol. There were 4 males and 1 female with an average age of 67 years (range of 42–82 yrs). The vertebral bodies of L3–L5 were dissected free of soft tissue and were disarticulated from their corresponding segments. All disc material was removed and the end plates were cleaned. One of the L5 vertebral bodies had been damaged during cadaveric extraction and was removed from our sample group. Fluoroscopic imaging confirmed absence of pathologic process other than osteoarthrosis, which was evident in three of the five specimens. A total of five L3 vertebral bodies, five L4 vertebral bodies, and four L5 vertebral bodies were instrumented. All screws were alternated with respect to left and right pedicles and angled versus straight screw insertion technique. Absent scoliosis or congenital malformations, previous literature has documented right and left symmetry within the same vertebral body specimen (Zindrick et al., 1987). Fourteen straight and 14 angled screws were placed. Each specimen was labeled and stored at −32 °C until implantation and testing.

Each vertebral body was visualized with fluoroscopic imaging in axial, sagittal, and coronal planes, and the appropriate insertion point for each pedicle was identified and marked. Using fluoroscopy allowed tight control of vertebral body orientation to help insure a standardized insertion technique. Utilizing a 2.7 mm drill bit and a drill press, the starting hole was made in the pedicle. No specimens experienced cortical disruption through the pedicle wall or the anterior cortex of the vertebral body. For angled screws, a "Scotty Dog" was visualized and the screw inserted in the center of the pedicle. Under fluoroscopic guidance (Fig. 1), 6.0 mm Schantz screws (Synthes, Paoli, PA) were then inserted into the starting hole by hand. Anteroposterior, lateral and coronal images were obtained to confirm placement of the screws within the pedicle and the body during and after insertion. The screws

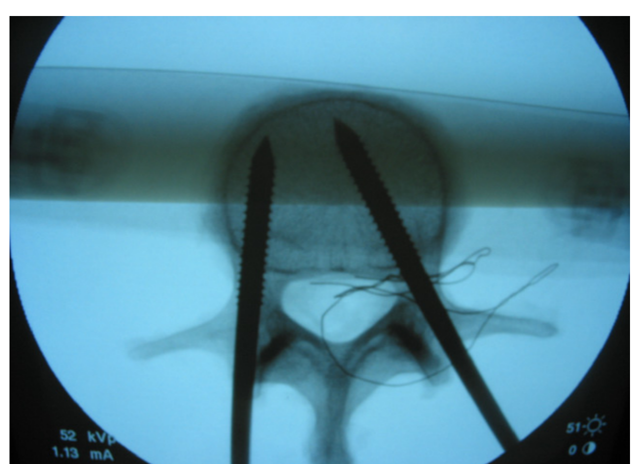

Fig. 1. Two different screw insertion techniques (fluoroscopy).

were advanced to the anterior cortex of the body, but did not pierce the cortex. For the straight screws, the pedicle insertion point was identified utilizing fluoroscopy (Fig. 1). The starting point was marked and the starting hole made utilizing the drill press. The 6.0 mm Schantz screw was then inserted in similar fashion under fluoroscopic guidance. Once both screws were placed, the specimen was again stored at −32 °C until testing could be performed.

2.2. Biomechanical testing

Biomechanical testing was undertaken once the specimens had again thawed to room temperature. We fashioned a grip to hold the vertebral body and polymethylmethacrylate (Zimmer, Warsaw, IN, USA) bone cement was utilized to pot the specimen. The cement was used to ensure uniform contact between the grip and the uneven vertebral body to prevent motion during testing and to distribute the forces being transmitted to the vertebral body by the loaded pedicle screw. The vertebra was fixed when the cement was in the doughy state and post testing analysis did not demonstrate cement intrusion into the vertebral body (Pfeiffer et al., 1996). The grip was firmly secured. Each specimen was placed in an upright anatomic orientation, and care was taken to prevent contact of the inferior or superior facets with the grip. To minimize the moment arm that existed outside of the vertebral body specimen, each pedicle screw was marked at 2 cm distance from their point of initial bony contact as a marker for the load contact point.

Once the bone cement had hardened, the construct was placed in a servohydraulic materials test machine (8501M, Instron, Canton, MA, USA) for fatigue testing. Specimens were maintained at room temperature and were kept moist throughout testing. A cylindrical loading rod attached to a load cell was then aligned with the screw such that the point of contact was at the 2 cm mark. The cylinder was brought down to the screw until initial contact was made. The load control testing protocol was programmed at 2000 cycles at 2 Hz, with peak load of 50 N with a load ratio $R = 0.1$ (min 5 N/max 50 N), directed in a caudal fashion in sinusoidal pattern of compression; neither preconditioning nor tensile load was utilized. The anatomical right pedicle was tested first in all cases whether the screw was oriented straight or angled. Data acquisition was performed with acquisition rate set at 100 Hz. Following the conclusion of testing, the screws were carefully removed and a depth gauge was used to determine depth of screw insertion from the point where bony contact was initially made. Further, the coronal fluoroscopic images were used to calculate angles from the midline for each pedicle screw. The mid vertebral line was drawn at the measured center distance of the neural canal and tip of the spinous process.

Information from one specimen (L5, Specimen A) was lost due to an error in the data acquisition setup, and the specimen was removed from our sample set. The second specimen (L4) in our test group was used for preliminary tests. This left a total of 12 samples and 24 total pedicle screws within our test group for data analysis.

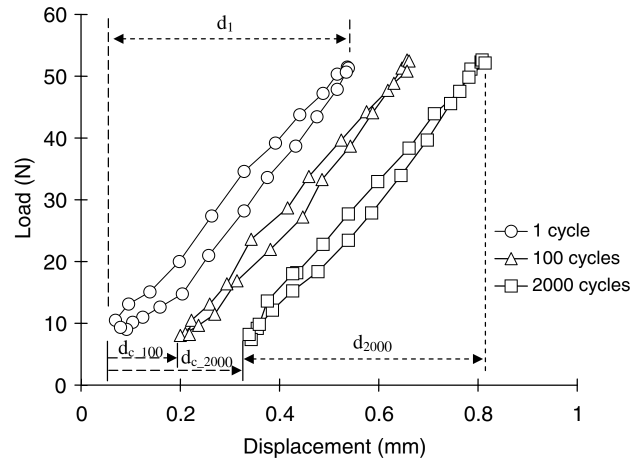

Fig. 2. A typical fatigue load–displacement curve. Stiffness damage ($d_{s\_N} = d_N - d_1$), creep damage (d_{c_N}), and total damage ($d_{t\_N} = d_{s\_N} + d_{c\_N}$).

2.3. Data analysis

Undamaged initial secant stiffness was measured by dividing the applied force (50 N) by maximum displacement at the first cycle. The stiffness damage ($d_{s\_N} = d_N - d_1$) was determined by subtracting the amount of cyclic displacement at the first loading cycle (d_1) from that for the current cycle (d_N) (Fig. 2). For example, the stiffness damage at 2000 cycles was calculated as $d_{s\_2000} = d_{2000} - d_1$. The creep damage ($d_{c\_N}$) was defined as the permanent displacement at the minimum (5 N) applied load for each cycle. The total damage ($d_{t\_N} = d_{s\_N} + d_{c\_N}$) was then calculated as the sum of the stiffness and creep damage. Differences between the straight and the angled screw types were examined using paired t-tests. To account for the high inter-specimen variability, the ratio of measured parameters from the angled-screw configuration to those from the straight-screw configuration within the same specimen was tested against a mean value of 1 using a one-sample t-test. All statistical analyses were performed using Statview software (SAS, NC, USA) and statistical significance was set as $P < 0.05$.

3. Results

Preliminary testing revealed that a 200 N force would cause traumatic fracture of the pedicle within the first and second cycles. Further testing at 50 N demonstrated the first and second phases of the standard three-phase response of fatigue could be obtained. A specimen was tested at 25 N but revealed no measurable damage, and the decision was made to proceed with testing the remainder of the specimens at 50 N. At these small loads, no preliminary or experimental specimen was cycled to failure despite early specimens being run to 10,000 cycles.

Table 1
Comparison of parameters between straight and angled screw types

Parameters	Straight	Angled	Paired *t*-test
Angle (°)	7.0 ± 3.2	26.3 ± 6.9	$P < 0.001$
Depth (mm)	51.3 ± 4.6	54.2 ± 5.3	$P < 0.020$
Initial stiffness (N/mm)	1.533 ± 0.075	1.690 ± 0.272	$P = 0.139$
Total damage at 2000 cycles (d_{t_2000}) (mm)	0.398 ± 0.38	0.690 ± 0.96	$P = 0.286$
Creep damage at 2000 cycles (d_{c_2000}) (mm)	0.430 ± 0.37	0.673 ± 0.92	$P = 0.341$
Stiffness damage at 2000 cycles (d_{s_2000}) (mm)	-0.032 ± 0.051	0.0167 ± 0.049	$P = 0.063$
Initial damage rate ($\Delta d_{t\_10}/\Delta t$) (mm/sec)	0.023 ± 0.0004	0.048 ± 0.0064	$P = 0.275$
Secondary damage rate ($\Delta d_{t\_2000}/\Delta t$) (mm/sec)	$8.50 \times 10^{-5} \pm 6.55 \times 10^{-9}$	$15.31 \times 10^{-5} \pm 18.90 \times 10^{-9}$	$P = 0.104$

Average ± standard deviation. $n = 12$.

Average angle and depth of insertion (relative to vertebral midline) were significantly different between the straight and the angled screw ($P < 0.001$ for angle and $P < 0.02$ for depth, respectively) (Table 1). Although none were cycled to the final third phase of failure, our fatigue test data followed a standard model for a two phase loading cycle fatigue test, as demonstrated previously: a rapid early phase and a constant second rate phase (Fig. 3) (Kim et al., 2004a,b). We observed stiffening of the construct, rather than loss of stiffness, for 15 out of 24 tests. We attribute this to the compaction of failed trabeculae at the screw–bone contact surface. Stiffness changes, however, were about 70 times less than creep damage. Thus, it is concluded that creep (d_c) dominated the cyclic behavior for both screw types. The magnitudes of the initial stiffness, total damage at 2000 cycles (d_{t_2000}), creep damage at 2000 cycles (d_{c_2000}), the stiffness damage at 2000 cycles (d_{s_2000}), initial damage rate ($\Delta d_{t\_10}/\Delta t$), and secondary damage rate ($\Delta d_{t\_2000}/\Delta t$) were measured higher for the angled screw than those for the straight screw. However, the difference of those magnitudes between the screw types turned out to be not significant ($P > 0.104$ and $P = 0.063$) for the stiffness damage at 2000 cycles (Table 1). This result was attributed to the significantly large variability (standard deviation) of the values of total damage, creep damage and secondary rate from the angled-screw construct than those from the straight-screw construct (*F*-test, $P < 0.01$).

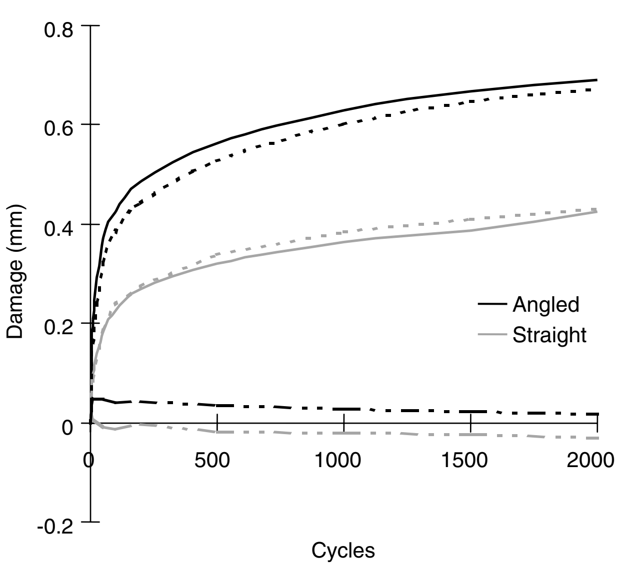

Fig. 3. The response curve of the averaged damage vs. cycle. Black lines are for the angled- and gray lines are for the straight-screw constructs. Total damage (—), creep damage (- - -), and stiffness damage (-·-·-). Creep dominated the fatigue damage caused by screw insertion.

Analysis of ratios demonstrated that the total damage in the angled-screw system was indeed significantly (1.6-fold) greater than in the straight-screw system at 2000 cycles ($P < 0.034$) but other parameters remained to be non-significant ($P > 0.08$).

4. Discussion

Multiple studies have been conducted to examine stability characteristics of lumbar pedicle screw systems that have been developed for posterior fixation. Zindrick et al. (1986) performed a thorough biomechanical study performing axial pullout and cyclic loading modes (displacement control) with multiple screw designs at various depths. The construct was assumed failed when 50% of the initial force was required to displace a total of 6 mm (3 mm caudad, 3 mm cephalad). Further, screws were inserted along the axis of the pedicle for the lumbar spine, though the authors did examine sacral fixation at medial and lateral angles. Authors concluded that screws inserted to a greater depth achieved better stability. However, they did not specifically compare the fatigue behavior of constructs with axial versus angular insertion of screws. Ruland et al. (1991) put forth that axial pullout during forward bending is the main mode of failure, despite the paucity of clinical literature to support this. Authors since that time have demonstrated linear correlation between bone density and pull-out strength. Soshi et al. (1991) assessed inline pedicle pullout, which may not represent in vivo behavior accurately, and found a linear relationship with osteoporosis. Willet et al. (1993) also studied pullout strength in Schanz screws, but again did so using displacement control modalities to show that the 6.0 mm Schantz screw was a better biomechanical construct than the 5.0 mm construct. Zdeblick et al. (1993) correlated insertional torque with increased pullout strength. Part of this study also looked at bone mineral density and found this to be a less effective predictor of pedicle screw stability.

Zdeblick's study also examined probed versus drilled screws and found no significant difference in insertional torque, cycles to failure or ultimate failure load. The pullout test was also combined with caudad/cephalad toggling by offsetting the force vector from the screw axis. High forces leading to destructive failure were used, rather than non-catastrophic forces observed clinically (Yerby et al., 1997). Zdeblick's data was confirmed by Myers et al. (1996) who demonstrated that quantitative CT imaging combined with stiffness and insertion torque were the strongest predictors of pull-out strength. The authors used a ramped sinusoidal load, again utilizing destructive failure methods. Law et al. (1993) examined pedicle fill with augment bushings to increase cortical contact within the pedicle. This study used caudad/cephalad loading and noted toggle loosening with a fulcrum at the base of the pedicle. These screws were given high loads up to 200 N, and the screw construct was only able to withstand three cycles. Large displacements up to 8 mm were observed, which certainly would be considered failure. Our preliminary testing experienced fracture of the pedicle at 200 N loads within the first two to three cycles, confirming earlier testing by Law. Brantley et al. (1994) examined non-destructive mechanical testing to mimic in vivo forces to examine the effect of screw size on stability. Again, displacement control was used rather than load control. Most of these specimens were inserted in line with the pedicular axis and data analysis of angular orientation was not performed. Finally, Cook et al. (2000), in studying an expansible pedicle screw design in cases of compromised bone quality, utilized an axial pullout model of screws inserted inline with the axis of the pedicle.

We feel that the loading protocol employed in the current study is more relevant to the progressive failure of vertebra/screw constructs. It was indicated that, for cortical bone, fully reversed cyclic loading to one half of the yield strain caused fatigue fracture in 1000 cycles (Carter et al., 1981). However, to date, no fatigue failure characteristics of human cancellous bone have been reported. Recently, Lu et al. (2004) found that cyclic loading with 30% of the yield load of human vertebrae significantly increased microcrack density in the vertebral trabeculae at 20,000 cycles of loading but the vertebrae did not fail. In the preliminary phases of the present study, we found that the pedicle screw system failed at 150 N. Therefore, the load level of 50 N we used for cyclic loading is about 30% of the system failure load. This level of cyclic load may be insufficient to cause failure of the entire pedicle screw system in a vertebra but sufficient to cause damage in bone adjacent to the screw.

To the author's knowledge, fatigue testing utilizing load control methods at non-destructive levels has not been utilized to better understand the behavior of pedicle screw stability. Further, though pullout testing has become the main in vitro predictor for stability, this mode of failure in vivo is rare. Converging pedicle screws have been advocated in osteoporotic bone, but the axial pullout method does not truly test the stability of these screws, and significant "butterfly shaping" within the vertebral body have been demonstrated (Law et al., 1993). Our two-phase fatigue test of angled versus straight pedicle screws revealed that total damage of the angled screw was higher when compared to the straight screws. As a component of the total damage, stiffness loss at the initial phase of loading cycles is likely attributed to the local compressive damage in trabecular bone around the metal screw. The locally damaged trabeculae at the contact surface between the screw and the bone compacted in progression with increasing cycles of loading. The compaction of failed trabeculae seemed to maintain the stiffness at the second-rate phase of fatigue. Overall, creep displacement was approximately 70 times as large as the displacement change associated with fatigue. This observation indicated that creep was the major cause of fatigue failure of pedicle screw systems, consistent with the results of other bone-interface fatigue testing that showed creep damage as the primary mode of failure (Kim et al., 2004a,b). This finding suggested that loosening between the pedicle and the vertebral bone observed in clinical situations (Pihlajamaki et al., 1997) could be a consequence of the increase in creep displacement with in vivo fatigue cycles. Other notable differences between the two sample groups occurred. For example, depth of screw insertion was significantly different between angled and straight screws. Though the straight screws were inserted shallower with less bone to distribute load and resist strain, the average deformation was less than the more deeply implanted angular screws. Angular screws demonstrated greater variability of results compared to straight screws. These results indicate that the angled screw system is less predictable than the straight-screw insertion.

The statistical analysis suggested that the study was underpowered to make strong conclusions without specimen pairing. Average total damage of the angled screw was higher than that of the straight screw, and the ratio analysis demonstrated that this difference was significant. All of the damage indices tested in this study, which are total damage at 2000 cycles, creep damage at 2000 cycles, the stiffness damage at 2000 cycles, initial damage rate, and secondary damage rate (Table 1), showed higher values for the angled screw construct than those for the straight screw construct indicating that the angled screw construct is inferior in fatigue performance compared to the straight screw technique. One would expect that the greater bone interface would yield a more stable construct, but this did not hold true. Previous studies have put forth that angular screws, contrary to our results, have more resistance to pullout strength and therefore should be used preferably, especially with osteoporotic bone (Cook et al., 2000). However, this was based on pullout of the pedicle screw in which the angular screw caused fracture of the pedicle as it was continuously loaded in tensile stress in the direction parallel to the midline of the vertebral body. Failure of the pedicle screw – vertebral body interface does not typically occur in vivo by this mechanism. From our experience with removal of failed screws, we felt that the majority of them are loose within the vertebra but that the vertebra does not break. Therefore, dynamic fatigue testing over thousands of cycles would yield more clinically relevant data regarding the stability of these constructs. In addition, we think that the stability of the straight screw system can be justified by the three-point fixation that holds the straight screw closer to the cortical part of the vertebra in three regions; in the insertion point, across the pedicle and at the end point. The straight screw end point is closer to the edge where there is less cancellous and more cortical type hard bone. On the other hand, angled screws pass through the middle of the pedicle and rely on size for cortical contact; placing the larger screw in the safe zone is technically more difficult. The end point of the angled screw sits in cancellous bone, relying more on the weaker cancellous structure of the vertebra.

We identified several limitations within this study. In vivo testing is theoretically more likely to give accurate information regarding frequency of forces about the pedicle screw, displacement of the screw, and evidence for mode of failure. In vitro testing, however, is inevitably without the presence of the body's immune response or ability to heal and adapt to load changes in the environment. In our specimens, the visual appearance of failure was not examined. Although the pedicle is generally cylindrical providing a straight path for screw insertion, it is possible that there is pivoting at some instances of loading and that toggling is involved in the damage mechanism. Nonetheless, if its presence is significant, this is an inherent part of the screw-technique and should be represented in engineering definitions of damage. Another limitation was that the quality of specimens likely varied between ages of donor and between levels of vertebra. Although taking bone mineral density into consideration could account for this variability, we felt that a side-to-side comparison study would obviate the need for knowing what the density of these specimens truly was. However, in retrospect, bone mineral density may have shown a separate correlation with how each specimen behaved. It is possible that screws interacted in the paired configuration, however, by alternating the order of testing of straight and angled screws between vertebrae, the potential effect of this interaction was equally distributed between groups. This probably increased the variability in the data, however, this matching was deemed necessary given the more difficult task of matching properties between bones from different sources if separate groups were used.

5. Conclusions

This study showed that straight screw insertion results in a pedicle-screw construct that has a better fatigue performance. From a clinical perspective, insertion of the pedicle-screws in a straight fashion is certainly more practical as it does not require extensive dissection, retraction, or excision of paraspinal musculature to achieve screw insertion along transverse pedicle angles that can range up to 38° from the midline. Further, this technique, though with less support from the literature, is likely more widely practiced already than currently reported. In large patients or those in whom minimally invasive techniques are attempted, insertion along the pedicular axis is particularly difficult and may require percutaneous screw placement. Our results support this method of pedicle screw insertion.

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