How Does a Novel Monoplanar Pedicle Screw Perform Biomechanically Relative to Monoaxial and Polyaxial Designs?

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Abstract

Background Minimally invasive spinal fusions frequently require placement of pedicle screws through small incisions with limited visualization. Polyaxial pedicle screws are favored due to the difficulty of rod insertion with fixed monoaxial screws. Recently, a novel monoplanar screw became available that is mobile in the coronal plane to ease rod insertion but fixed in the sagittal plane to eliminate head slippage during flexion loads; however, the strength

of this screw has not been established relative to other available screw designs.

Questions/purposes We compared the static and dynamic load to failure in polyaxial, monoaxial, and monoplanar pedicle screws.

Methods Six different manufacturers' screws (42 total) were tested in three categories (polyaxial, $n = 4$; monoaxial, $n = 1$; monopolar, $n = 1$) utilizing titanium rods. An additional test was performed using cobalt-chromium rods with the monopolar screws only. Screws were embedded into polyethylene blocks and rods were attached using the manufacturers' specifications. Static and dynamic testing was performed. Dynamic testing began at 80% of static yield strength at 1 Hz for 50,000 cycles.

Results In static testing, monoaxial and monoplanar screws sustained higher loads than all polyaxial screw designs (range, 37%–425% higher; $p < 0.001$). The polyaxial screws failed at the head-screw interface, while the monoaxial and monoplanar screws failed by rod breakage in the static test. The dynamic loads to failure were greater with the monoplanar and monoaxial screws than with the polyaxial screws (range, 35%–560% higher; $p < 0.001$). With dynamic testing, polyaxial screws failed via screw-head slippage between 40% and 95% of static yield strength, while failures in monoaxial and monoplanar screws resulted from either screw shaft or rod breakage.

Conclusions All polyaxial screws failed at the screw-head interface in static and dynamic testing and at lower

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values than monoaxial/monoplanar screw designs. Monoplanar and monoaxial screws failed at forces well above expected in vivo values; this was not the case for most polyaxial screws.

Clinical Relevance Polyaxial screw heads slip on the screw shank at lower values than monoaxial or monoplanar screws, and this results in angular change between the rod and pedicle screw, which could cause loss of segmental lordosis. The novel monoplanar screw used in this study may combine ease of rod placement with sagittal plane strength.

Introduction

Minimally invasive spinal fusions are increasingly becoming standard procedures for many surgeons. These procedures frequently require the placement of pedicle screws through smaller incisions with less exposure. In the setting of such limited visualization, as well as with longer posterior constructs, polyaxial pedicle screws have become more commonly utilized because of the ease of rod insertion. In contradistinction, monoaxial screws with fixed heads are less favored because of the difficulty of rod insertion, especially when utilizing minimally invasive techniques. These techniques are often applied in circumstances of compromised anterior column load, including with traumatic burst fractures, after corpectomy, and with intervertebral cage placement. Under these circumstances, the ability of posterior pedicle screws to resist flexion and anterior compressive loads becomes more critical. However, previous studies have shown that the heads of polyaxial screws fail at a lower value than traditional monoaxial screws and that polyaxial screws typically slip at the screw-head interface [1, 13]. These studies also showed that different brands of polyaxial screws fail at different flexion loads [1, 13].

Biomechanical studies have shown that the highest load in vivo on a pedicle screw in the thoracolumbar spine is in flexion [10]. If this flexion load exceeds the polyaxial head's coupling device strength, loss of segmental lordosis can result. This has been seen with short-segment fixation of burst fractures, in which case, the more limited number of pedicle screws are prone to fail due to the loss of anterior column support [6]. This failure may, in part, be related to the polyaxial screw-head angular change, which is corroborated by the authors' own experience with radiographic

analysis of subsidence after interbody fusion and posterior fixation. The loss of segmental lordosis from such polyaxial screw-head slippage could ultimately contribute to sagittal malalignment. The possibility of a resulting sagittal imbalance has been associated with pain, pseudoarthrosis, and adjacent segment degeneration [2, 5, 8, 9, 16].

Although polyaxial screw designs can include favored-angle modifications, with which there is a greater capacity of angular motion through the screw-tulip universal joint in a specific plane, the biomechanical properties of failure through this friction-limited joint are not improved. There is still a tendency for slippage through such a joint, when comparing it to monoaxial screw designs [15]. Uniplanar or monoplanar-type screws that are rigid in the sagittal plane through different screw-tulip interface designs may be advantageous in resisting compressive loads.

A novel monoplanar screw design was recently developed (Trinity Orthopedics, San Diego, CA, USA). Its head swivels freely in the axial plane (commonly referred to as coronal plane, although a misnomer) but is fixed in the sagittal plane, such that slippage of its screw-head interface cannot occur in the sagittal plane. Because of this orientation, the monoplanar screw can behave as a monoaxial screw in the sagittal plane, that is, sagittally stable. This design may combine the ease of rod insertion appreciated with a polyaxial screw with the sagittal plane strength of a monoaxial screw. In this study, we tested the static and dynamic biomechanical strength of the monoplanar screw and compared it to a monoaxial screw and four different brands of polyaxial screws.

Materials and Methods

Study Design

Four different brands of polyaxial pedicle screws, one monoaxial pedicle screw, and one monoplanar screw, all FDA approved, were studied. Six screws of each type were donated from different manufacturers (Aesculap, Center Valley, PA, USA; Nuvasive, San Diego, CA, USA; Phygen Spine, Irvine, CA, USA; Stryker, Kalamazoo, MI, USA; Theken Spine, Akron, OH, USA; Trinity Orthopedics, San Diego, CA, USA); the Aesculap was the monoaxial; Trinity, monoplanar; all others, polyaxial, except for the monoplanar screw, in which case a total of 12 screws were utilized to accommodate titanium and cobalt-chromium rod testing conditions. The four different types of polyaxial screws were tested to average the variability among different manufacturers [1, 15]. Only one monoaxial and one monoplanar screw type were utilized, as it was expected that their failure would be at higher loads and less variable than that seen with polyaxial screws based on prior

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research [1, 15]. The screws were 45 mm in length and ranged from 6.0 to 7.0 mm in diameter, which was based on the available diameter options from each manufacturer. All screws were made of titanium alloy and were of standard thread pitch. All screws were tested with 5.5-mm diameter titanium rods designed specifically for the particular screw, with the exception of 5.9-mm diameter titanium and cobalt-chromium rods used to test the monopolar screws (the only available diameter of rod for this screw type). An additional test was performed with the monopolar screws utilizing a cobalt-chromium rod to allow for greater likelihood of failure by the screw (rather than the rod).

The pedicle screws were placed in high-density polyethylene blocks to exclude the variability in bone density expected with cadaveric specimens, thus making the pedicle screw-rod construct the sole variable, as others have proposed [3, 10, 15]. The insertion depth was standardized in a manner that allowed for approximately 5 mm of exposed screw shank so as not to incur any abutment of the polyethylene block to the tulip head during angular changes (which could result in confounding variability in force transmission). Polyethylene blocks were utilized for testing in a manner consistent with the ASTM 1717 standard [15] (the standardized mechanical testing required for FDA 510(k) submission), and bending loads were calculated for each construct, based on the distance from where the loads were exerted relative to the distance from the central axis of the connecting rod within the receiver element (screw tulip).

The manufacturers' recommendations were followed precisely during implant assembly, including allowing for at least 1 to 2 mm of rod presence on either side of the set screw outer diameter. The rods were secured to the pedicle screws by set screws and tightened with each manufacturer's torque wrench and torque-limiter. The specific manufacturers' own tools were used so that assembly of the constructs was exactly to manufacturer recommendations.

The polyethylene block was attached to the vertical actuator arm of the testing station (Fig. 1), with freedom of motion only allowed in the flexion-extension plane. The vertical actuator arm was only allowed vertical compression motion with polyethylene block rotation occurring through its attachment to the actuator arm. The distance between the center of the load cell and the central axis of the connecting rod within the screw tulip was measured for each screw using a digital micrometer and recorded for moment arm calculations.

Static Yield

The static yield test was performed using the Instron 8500 hydraulic testing station (Instron, Norwood, MA, USA)

Fig. 1 The experimental setup is shown. Dotted lines indicate center of load and center of screw head. The red line indicates moment arm measurements.

equipped with a 1000-pound (453.6-kg) load cell. For each pedicle screw type, a single screw was utilized for initial static yield value determination by applying compression at a rate of 25 mm/minute as done by previous authors [15]. The force versus displacement curve was recorded to determine the initial drop in load. The displacement included screw slippage, screw breakage, or dissociation between some components of the construct. The yield moment was calculated from the yield load multiplied by the moment arm (Fig. 1).

Dynamic Testing

Initial dynamic testing was conducted at 80% of static yield value determined in the static testing in a load-controlled manner. If at least two screw constructs survived the initial loading condition, the load was increased by 10% and the dynamic test was repeated. If failure occurred, the load was reduced sequentially by 10% intervals, and the dynamic test repeated, in which case the load was reduced twice (to 60% of static yield strength) to achieve survival of the construct under dynamic loading conditions (Fig. 2). Dynamic load was applied in a sine waveform at a rate of 1 Hz for 50,000 cycles. At every 100 cycles, peak displacement and force output were recorded. Using the moment arm measurements of each screw, vertical displacement was converted into angular changes in the pedicle screw construct.

Measurements

The pedicle screw construct was photographed before the start and after the end of each test to measure angular

Fig. 2 A polyaxial head slipping through angular change relative to the screw shank is shown.

Fig. 3A–C The method utilized for determining angular changes occurring between the screw head and shank is shown. (A) Four points are defined. (B) Two lines are defined. (C) Angles are then measured.

changes between the top and bottom testing blocks. Angle measurements were performed using custom image analysis scripts in MATLAB[®] (The MathWorks, Natick, MA, USA). The change in angle between the testing blocks corresponds clinically to sagittal plane change or segmental angular change (Fig. 3).

Throughout the duration of testing, vertical separation distance between the test blocks was recorded every 100 cycles using the integrated Instron linear variable differential transformer (0.05-mm resolution). This distance corresponds clinically to loss of anterior column height. At the end of each screw testing, the connecting rod position within the test fixture was noted and its linearity measured. Loss of fixation or any deformation of the connecting rod was recorded and photographed.

Statistical Analysis

One-way ANOVA was used to test for significant differences in static and dynamic torque before failure among groups (SPSS[®] Version 12; IBM Corp, Armonk, NY, USA). Bonferroni-corrected post hoc pairwise comparisons

were performed to detect significant differences between individual groups. Statistical significance was defined as a *p* value of less than 0.05.

Results

The monoaxial and monoplanar screws sustained greater bending moments before failure than the polyaxial screws, and the screw designs failed by different mechanisms. During static testing, all polyaxial screws failed at the head-screw interface (Fig. 2) at loads ranging from 4 Nm to 16 Nm (Fig. 4). The monoaxial (S2) and monoplanar (S1, S1 CoCr) screws failed by rod breakage in the static test at a higher bending moment (range, 37%–425%) than any of the polyaxial screws ($p < 0.001$) (Fig. 5). The monoplanar screw with the titanium rod failed at 23 Nm and the same screw with the cobalt-chromium rod failed at 24 Nm, both higher than the monoaxial screw ($p < 0.001$), which failed at 20 Nm.

All failures in the dynamic tests for polyaxial screws occurred through screw-head slippage resulting in a change of at least 10° in sagittal plane alignment (Fig. 6). Two of the polyaxial screw types (S4 and S5) did not fail until challenged dynamically at 95% of static yield strength. One type of polyaxial screw (S3) failed at 80% of its static yield strength when tested dynamically, while another (S6) severely underperformed the others by failing dynamically at only 40% of static yield strength. This was the same screw that underperformed the others in static testing as well, failing at 4 Nm. In contrast, dynamic failures in the monoaxial and monoplanar screws resulted from either screw shaft breakage or rod breakage at 70% of static load failure. The mean dynamic bending moment sustained by the monoplanar screws (S1 and S1 CoCr) was also higher (89%–560%) than that sustained by any of the polyaxial screws (S3 to S6) ($p < 0.001$) (Fig. 6). The monoaxial screw (S2) also sustained a higher dynamic bending moment (35%–122%) than the polyaxial screws (S4, S5, and S6) ($p < 0.001$). Notably, the 6.0-mm diameter screw was polyaxial and yet still failed at the universal joint, rather than at the screw shank, and at lower values than the monoaxial or monoplanar screws.

Fig. 4 A graph shows the static yield testing results. S1 is the monoplanar screw; S2 is the monoaxial screw; S3, S4, S5, and S6 are polyaxial screws. All were tested with titanium rods except S1 CoCr for which cobalt-chromium rods were used. All polyaxial screws in static testing failed at the head-screw interface at loads ranging from 4.6 Nm to 16 Nm. The monoaxial and monoplanar screws failed by rod breakage, the monoaxial screw at 20.2 Nm and the monoplanar screws at 23.2 Nm (titanium rod) and at 24.2 Nm (cobalt-chromium rod).

Fig. 5 A monopolar screw shank that had broken during testing is shown.

Discussion

Pedicle screw instrumentation is used routinely in posterior spinal reconstruction to maintain stability for fusion procedures. It is also used as a powerful reduction tool for correction of scoliosis, kyphosis, and spondylolisthesis. With the advent of minimally invasive surgery, the pedicle screw has become a more common method of fixation. In the thoracolumbar spine, maintenance of lordosis by pedicle screw instrumentation can be critical in maintaining overall sagittal balance. Sagittal alignment has been shown to be the most important radiographic variable responsible for pain and disability [2, 5]. Previous studies [13] have demonstrated that polyaxial screws fail at the screw-head interface and at a value that is less than that for monoaxial screws. Also, the greatest load experienced by a pedicle screw is in flexion [10]. Thus, failure of a polyaxial screw

in flexion can result in loss of segmental and global lordosis. We therefore compared the static and dynamic biomechanical strength of a new monoplanar screw design, which seeks to minimize loss of lordosis, to that of a monoaxial screw and four different brands of polyaxial screws.

This study had several limitations. First, we utilized only one type of monoaxial and monoplanar screw to compare with the polyaxial screws. We also did not include favored-angle screws. Without testing different manufacturers' monoplanar, monoaxial, or favored-angle screws, no statements regarding how the designs we tested might fare against others can be made. We also had a small sample size of screws; while the sample size was adequate to detect important differences, and some indeed were detected, smaller (but still potentially clinically important) differences might have appeared as no-difference findings in this report. In addition, the testing protocol was limited to a single screw/rod system, rather than a segmental construct. Previous *in vivo*/segmental construct studies have shown polyaxial screws to have stronger construct stiffness because of the ease of rod reduction *in vivo* [13, 14]. Although a single screw/rod system gives valuable information regarding specific components, a perfect correlation between these biomechanical results to *in vivo*/segmental construct clinical results is not possible.

Although screw diameters did vary slightly based on manufacturer availability, it does not appear that this variance had any meaningful impact on our results. In fact, the smaller-diameter polyaxial screws still failed at much lower loads through their universal joints, rather than through screw shank fracture seen with the monoplanar and monoaxial screws. Likewise, screw thread pitch or type does not appear to have had any impact on the results of our study, given the lack of pullout or failure within the

Fig. 6 A graph plots flexion moment versus vertical displacement during dynamic testing. S1 is the monoplanar screw; S2 is the monoaxial screw; S3, S4, S5, and S6 are polyaxial screws. All were tested with titanium rods except S1 CoCr, for which cobalt-chromium rods were used. All failures in polyaxial screws were screw head

slippage resulting in a change of at least 10° . Two of the polyaxial screws did not fail until 95% of static yield strength. One failed at 80% and the other screw failed at only 40%. Failures in the monoaxial and monoplanar screws resulted from either screw shaft breakage or rod breakage at 70% of static load failure.

polyethylene-screw interface. However, since our study only used polyethylene blocks, our results are limited to the mechanical properties of the screws, rather than biomechanical challenges presented by varying bone quality. Certainly, screw failure in osteoporotic bone, for instance, might occur at lower loads and at the screw-bone interface.

Our static testing demonstrated that monoaxial and monoplanar screws had greater resistance to failure under flexion loads, when compared to polyaxial screws. Additionally, we found a considerable variability in flexion load resistance among different types of polyaxial pedicle screws. Monoaxial and monoplanar screws always failed with screw shank or rod breakage during static testing. In contrast, all polyaxial screws failed through head-screw universal joint slippage. Our study and those of Fogel et al. [1] and Stanford et al. [15] similarly found that, under sagittal bending loads, polyaxial screws (and screw-connecting rod constructs) initially failed at the universal joint, which provides for polyaxial adjustability. Regardless of design, in all polyaxial screws tested in our study and those of Fogel et al. [1] and Stanford et al. [15], the universal joint was less resistant to sagittal bending loads than the other constituent elements of the screw and connecting rod. Consistent with the findings of Fogel et al. [1] and Stanford et al. [15], the polyaxial screws failed at a range of loads in our study. This substantial range of static failures included one type of polyaxial screw that underperformed the others by a wide margin. This particular screw failed at 4.6 Nm under static testing conditions, which is well below the expected load in vivo (9 Nm) [11, 12]. Two others failed at 10.5 and 10.8 Nm, which is only marginally above the expected in vivo load. One of the polyaxial screws

maintained its strength up to 16 Nm. In contrast, the monoplanar and monoaxial heads did not fail but the titanium rods deformed at 23.2 and 19.8 Nm, respectively. The monoplanar screw was also tested with a cobalt-chromium rod to compare the impact of rod type on failure, and static testing demonstrated rod failure at loads similar to the titanium rods.

Dynamic testing resulted in mixed results in the polyaxial screws. As expected, the screw that performed poorly under static testing also performed poorly in the dynamic testing and failed at only 40% of its static strength. The monoaxial and monoplanar screws failed by screw shaft or rod breakage at 70% of static load. Our findings were similar to those of Fogel et al. [1] and Stanford et al. [15] in this regard. Interestingly, in vivo studies have not shown radiographic slippage of polyaxial screws [4, 7] and construct stiffness to actually be stronger with polyaxial screws than monoaxial screws secondary to ease of rod reduction [13, 14]. Although our study shows polyaxial screws to fail dynamically and statically at values under monoaxial and monoplanar screws, it is possible that they are adequately designed to withstand real-life situations.

As spine surgery and minimally invasive techniques advance, polyaxial pedicle screws are becoming more commonly utilized. The ease of rod reduction makes polyaxial screws more favorable when compared to traditional fixed-head screws. However, according to our results, polyaxial screw heads slip on the screw shank at lower loads than monoaxial or monoplanar screws, and such failure results in angular change between the rod and pedicle screw. This study showed significant variability in the forces required for polyaxial screw failure. Most

failures occurred at loads that are approximate to expected in vivo human forces, and one screw failed at loads lower than those expected in vivo. This failure could potentially lead to loss of segmental sagittal alignment. The monoplanar screw used in this study might aptly combine the ease of rod reduction seen in polyaxial screws with the sagittal plane strength valued in monoaxial screws. Future studies should be performed to analyze performance of monoplanar screw designs with regard to clinical utility, as well as radiographic and clinical outcomes.

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References

1. Fogel GR, Reitman CA, Liu W, Esses SI. Physical characteristics of polyaxial-headed pedicle screws and biomechanical comparison of load with their failure. *Spine (Phila Pa 1976)*. 2003;28:470–473.
2. Glassman SD, Berven S, Bridwell K, Horton W, Dimar JR. Correlation of radiographic parameters and clinical symptoms in adult scoliosis. *Spine (Phila Pa 1976)*. 2005;30:682–688.
3. Kim YJ, Bridwell KH, Lenke LG, Rinella AS, Edwards C, 2nd. Pseudarthrosis in primary fusions for adult idiopathic scoliosis: incidence, risk factors, and outcome analysis. *Spine (Phila Pa 1976)*. 2005;30:468–474.
4. Kuklo TR, Potter BK, Polly DW Jr, Lenke LG. Monoaxial versus multiaxial thoracic pedicle screws in the correction of adolescent idiopathic scoliosis. *Spine (Phila Pa 1976)*. 2005;30:2113–2120.
5. Lazennec JY, Ramare S, Arafati N, Laudet CG, Gorin M, Roger B, Hansen S, Saillant G, Maurs L, Trabelsi R. Sagittal alignment in lumbosacral fusion: relations between radiological parameters and pain. *Eur Spine J*. 2000;9:47–55.
6. Liu T, Zheng WJ, Li CQ, Liu GD, Zhou Y. Design and biomechanical study of a modified pedicle screw. *Chin J Traumatol*. 2010;13:222–228.
7. Lonner BS, Auerbach JD, Boachie-Adjei O, Shah SA, Hosogane N, Newton PO. Treatment of thoracic scoliosis: are monoaxial thoracic pedicle screws the best form of fixation for correction? *Spine (Phila Pa 1976)*. 2009;34:845–851.
8. Park P, Garton HJ, Gala VC, Hoff JT, McGillicuddy JE. Adjacent segment disease after lumbar or lumbosacral fusion: review of the literature. *Spine (Phila Pa 1976)*. 2004;29:1938–1944.
9. Pateder DB, Park YS, Kebaish KM, Cascio BM, Buchowski JM, Song EW, Shapiro MB, Kostuik JP. Spinal fusion after revision surgery for pseudarthrosis in adult scoliosis. *Spine (Phila Pa 1976)*. 2006;31:E314–E319.
10. Pienkowski D, Stephens GC, Doers TM, Hamilton DM. Multicycle mechanical performance of titanium and stainless steel transpedicular spine implants. *Spine (Phila Pa 1976)*. 1998;23:782–788.
11. Rohlmann A, Bergmann G, Graichen F. Loads on an internal spinal fixation device during walking. *J Biomech*. 1997;30:41–47.
12. Rohlmann A, Graichen F, Weber U, Bergmann G. 2000 Volvo Award winner in biomechanical studies: monitoring in vivo implant loads with a telemeterized internal spinal fixation device. *Spine (Phila Pa 1976)*. 2000;25:2981–2986.
13. Serhan H, Hammerberg K, O'Neil M, Sturm P, Mardjetko S, Crawford A. Intraoperative techniques to reduce the potential of set-screw loosening in long spinal constructs: a static and fatigue biomechanical investigation. *J Spinal Disord Tech*. 2010;23:e31–e36.
14. Shepard MF, Davies MR, Abayan A, Kabo JM, Wang JC. Effects of polyaxial pedicle screws on lumbar construct rigidity. *J Spinal Disord Tech*. 2002;15:233–236.
15. Stanford RE, Loeffler AH, Stanford PM, Walsh WR. Multiaxial pedicle screw designs: static and dynamic mechanical testing. *Spine (Phila Pa 1976)*. 2004;29:367–375.
16. Umehara S, Zindrick MR, Patwardhan AG, Havey RM, Vrbos LA, Knight GW, Miyano S, Kirincic M, Kaneda K, Lorenz MA. The biomechanical effect of postoperative hypolordosis in instrumented lumbar fusion on instrumented and adjacent spinal segments. *Spine (Phila Pa 1976)*. 2000;25:1617–1624.