Biomechanical Comparison of Endplate Forces Generated by Uniaxial Screws and Monoaxial Pedicle Screws

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abstract

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Current surgical treatment of idiopathic scoliosis involves the use of various segmental instrumentation. Various pedicle screws have allowed for improved correction. Although monoaxial screws have improved rotational control compared with polyaxial screws, their use may increase screw–bone interface or vertebral endplate forces if not inserted in an exactly straight trajectory. Uniaxial screws potentially decrease these forces while retaining the advantages of monoaxial screws with respect to better rotational control. The purpose of this study was to compare the vertebral endplate forces with monoaxial or uniaxial screws when being reduced to a rod. Thirty-two plastic, surrogate T11 vertebrae were prepared with monoaxial (n=16) or uniaxial (n=16) screws. Screw angles relative to inferior vertebral endplates were assessed with lateral radiographs. The vertebrae were fixed to a load cell, and loads were measured as the screw was reduced to a rod. Monoaxial screws demonstrated a linear progression of endplate force with increasing screw angle. Uniaxial screws demonstrated minimal endplate force until approximately 20°, coinciding with screwhead excursion angle. As this maximum excursion angle was passed, uniaxial screws demonstrated a force slope similar to the monoaxial screws.

The measured endplate forces should be equivalent to forces at the screw–bone interface. The reduced force with uniaxial screws is expected to have less cranial–caudal plow potential as the screw is coupled to a rod for deformity correction. This could have potential implications for screw failure, especially in less dense bone.

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The surgical treatment of idiopathic scoliosis has evolved to encompass various potential implants, including sublaminar wires, hooks, and pedicle screws. The goal of spinal instrumentation is to achieve control and correction of a deformity. Recent advances in spinal instrumentation have allowed for improved 3-dimensional corrections with posterior-only approaches. These techniques have largely been advanced by the use of pedicle screw constructs and the techniques associated with their use.\textsuperscript{1-3}

Polyaxial screws are the predominant type of fixation currently used for degenerative spine work. A multidirectionally variable angle between the screw shaft and tulip facilitates coupling of the screw with a longitudinal rod. This type of screw has the advantage of facilitating attachment of the screw to a rod independent of the screw angle (within the tolerance of head variability allowed by the screw). The disadvantage of this type of screw is that it does not allow for rotation control or correction of the vertebra with newer vertebral body derotation techniques that are often now used for deformity correction.

Monoaxial screws have a fixed angle between the screw shaft and tulip. This allows for potentially better rotational control of the vertebra.\textsuperscript{4,5} However, despite the potential advantages of such screws, the straight-ahead trajectory for their placement to allow coupling with a rod may be technically challenging. If not placed in an ideal trajectory, coupling of the screw to the rod has the potential to increase forces at the screw–vertebra interface, which may be associated with screw plowing. This mode of failure has been seen clinically in anterior and posterior methods of deformity correction and has been demonstrated biomechanically with anterior instrumentation.\textsuperscript{6,7}

To accommodate this limitation of monoaxial screws while maintaining rotational control, uniaxial screws that allow for cranial–caudal variability between the screw shaft and tulip have been developed.\textsuperscript{8}

The touted advantage of this type of screw is that it allows for rotational control of the vertebra while still allowing for some variability in the cranial–caudal placement angle of the screw.

Nonetheless, no current studies analyze the touted potential advantages of uniaxial screws over monoaxial screws for deformity surgery. The current biomechanical study was performed to compare the force generated when coupling monoaxial and uniaxial screws to a longitudinal rod with variable cranial–caudal angulation.

**MATERIALS AND METHODS**

A total of 32 surrogate T11 clear plastic vertebrae were obtained (Pacific Research Laboratories, Inc, Vashon, Washington). They were a hard plastic in which significant movement was not allowed between potential instrumentation and synthetic bone material. Vertebrae were divided into 2 groups of 16. Half were prepared with stainless steel Xia monoaxial 5.5×40-mm screws (n=16), and the other half were prepared stainless steel Xia uniaxial 5.5×40-mm screws (n=16) (Stryker, Mahwah, New Jersey). Only 1 screw was placed per vertebra to allow accurate radiographic analysis of the specimens; for consistency, the left pedicle was always instrumented.

Each screw was inserted after drilling and tapping the pedicle. Screws were inserted at variable angles from straight ahead (parallel to the inferior vertebral endplate) to anatomic (angled cephalad to caudal) (Figure 1). Vertebrae were visually inspected to insure that no fracture or violation of the pedicle wall or vertebral endplate existed. The screw–bone interface was noted to be exceedingly rigid, thus allowing extrapolation of the endplate forces measured to be proportional to what would be clinically observed at the screw–bone interface. Screw angle was subsequently confirmed and quantified from a lateral radiograph (Figure 2). Measurements for these angles were made with digital imaging analysis software (Synapse PACS 3.2.1; Fujifilm, Stamford, Connecticut).

Each specimen was attached perpendicularly to a 445-N capacity unidirectional load cell (S-Beam Load Cell, LC101-100; Omegadyne, Inc, Sunbury, Ohio) and a spring via its inferior endplate. An additional spring was attached to the superior endplate (Figure 3). An initial resting force was recorded using LabView version 8.5 software (National Instruments, Austin, Texas) (Figure 4).

A 5.5-mm rod (stiffness 4000 N/mm; yield force, 3000 N) was brought into contact with the screw tulip. The screw was pulled to the rod using a persuader, and the force on the load cell was recorded (Figure 5). The total force at the endplate was calculated by subtracting the resting force from the force recorded after pulling the screw to the rod.

Statistical analysis was performed using SPSS version 17.0 software (SPSS, Inc, Chicago, Illinois). Mean angles and

![Figure 1: Variations in screw insertion angle ranged from straight ahead to anatomic trajectories.](image)

![Figure 2: Lateral radiograph used to measure the angle of screw insertion relative to the inferior vertebral endplate.](image)
mean forces for the uniaxial and monoaxial data were compared using t tests. The uniaxial data was divided at insertion points above and below 20°, corresponding with the machined excursion distance of the uniaxial screw head. Results were scatter-plotted, and linear regression analysis was performed. Significance was determined by P values of .05 or less.

RESULTS
Screw angles as measured from the long axis of the screw shaft to the inferior endplate of the vertebrae ranged between 2° and 32° (mean, 15.3° ± 9.07°) for the monoaxial screws and between 1° and 30° (mean, 15.38° ± 8.73°) for the uniaxial screws (P > .99) (Table).

Overall endplate forces for the monoaxial and uniaxial screws were 4.70 ± 2.80 and 1.10 ± 1.40 N, respectively (P < .001) (Table). Mean endplate forces were also compared for angles of insertion above and below 20°. Endplate forces above 20° of insertion were 7.30 ± 2.60 and 2.60 ± 1.80 N for the monoaxial and uniaxial screws, respectively (P = .021). For angles of insertion less than 20°, the mean endplate forces were 3.60 ± 1.70 and 0.48 ± 0.43 N for the monoaxial and uniaxial screws, respectively (P < .001).

Results were scatter-plotted, and linear regression analysis was performed (Figure 6). The monoaxial screw plot demonstrated a 0.28 N/degree slope (R² = 0.79). This linear relationship of force to screw insertion angle was constant over the range of screw insertion angles tested.

The uniaxial data were divided in 2 ranges. The first range was screw placement angles from 0° to 20° with a 0.02 N/degree slope (R² = 0.09). In this range (below the excursion limit of the uniaxial screws), large forces were not seen. The second range was screw placement angles above 20° and had a 0.39 N/degree slope (R² = 0.96). In this range (above the excursion limit of the uniaxial screws), the force/angle relationship was close to that of the monoaxial screws.

DISCUSSION
Particularly in deformity surgery, different types of vertebral body fixation may be considered. Pedicle screws have evolved to have varying properties, many of which are attributed to the articulation of the screw shaft with the screw tulip. Although the biomechanics of pedicle screws have been studied,7-10 the potential advantage of uniaxial pedicle screws over monoaxial pedicle screws has not been previously reported.

The results from this biomechanical comparison of monoaxial and uniaxial screws demonstrate increasing force at the endplate with increasing angles of screw insertion for both monoaxial and uniaxial screws. On average, monoaxial screws demonstrated a significantly higher force at the endplate than did uniaxial screws. Furthermore, the force at the uniaxial endplate was negligible until approximately 20° of insertion, corresponding with the maximum excursion of the screw tulip. After this, the relationship of endplate force to screw insertion angle followed the same relationship as that seen for the monoaxial screws.

Although forces in this study were measured at the endplate, all pedicle screws demonstrated a rigid, intimate fit at the screw–vertebra interface in the surrogate plastic bones used. Accordingly, the forces measured at the endplate were believed to be proportional to cranial–caudal plow forces within the vertebral body that would exist in the clinical setting. This may have implications for screw failure, especially in less dense bone. Furthermore, in the clinical setting, only forces tolerated by the vertebral body would then be expected to be transferred to the endplates. This may have implications on adjacent segment breakdown at the end of a construct.
Although this is the first reported biomechanical study on the differential forces of monoaxial and uniaxial screws at the bone interface, previous studies have examined various properties of pedicle screws. Barber et al\(^9\) reported that paired pedicle screws placed in 30° of convergence improved pullout strength when compared with those placed parallel in lumbar vertebrae. Fogel et al\(^10\) reported that the coupling of the polyaxial head to the screw rather than the screw itself was the first to fail when loaded cyclically.

Lehman et al\(^11\) reported that straight-forward trajectory insertion of thoracic pedicle screws was superior to anatomic trajectory with regard to pullout strength and torque. Furthermore, pullout strength and torque were dependent on bone mineral density. McKinley et al\(^12\) delineated the importance of placing pedicle screws in the center of the sagittal plane of the pedicle and not using screws less than 40 mm in length so as not to increase intrapedicular bending moments.

To the current authors’ knowledge, this is the first biomechanical study comparing monoaxial with uniaxial pedicle screws; however, the study had some limitations. Isolated vertebrae were used, thus removing the influence of other vertebrae, intervertebral disks, muscles, and ligaments that may affect forces in the vertebrae. Thus, force magnitudes required to bring the screws to the rod in the current study are not representative of in vivo loads. In addition, the springs connecting the load cell to the frame provided resistance when fixing the screw to the rod, which was incorporated into the total endplate force. Also, rigid plastic vertebrae were used. However, this constraint had to be accepted because cadaveric specimens are generally from elderly, osteoporotic patients that do not model the younger, denser bone of adolescents, for whom this type of fixation is regularly considered. By using the denser plastic bone, all effects of the plow forces on the screws could be measured at the vertebral endplates, as opposed to having poorly quantified distribution of force dissipated within the vertebral body and at the vertebral endplates.

**Conclusion**

This study demonstrates a linear increase in endplate forces with increasing angles of insertion from straight ahead to anatomic trajectories for monoaxial screws, and minimal endplate forces with increased insertion angles until the excursion limit of the screw tulip was reached for uniaxial screws. From a clinical perspective, this suggests that the cranial–caudal plow forces may be less with uniaxial screws than with monoaxial screws because they are coupled to a longitudinal rod. However, af-

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<th>Mono Screw</th>
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<th>Uni Screw</th>
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**Abbreviations:** deg, degrees; Mono, monoaxial; Uni, uniaxial.

\(^a\)Measured from lateral radiographs.

\(^b\)Measured from the load cell.

\(^c\)Mean, 15.38±9.077.

\(^d\)Mean, 4.70±2.807.

\(^e\)Mean, 15.38±8.733.

\(^f\)Mean, 1.1±1.4.

\(^g\)t test.

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**Figure 6:** Scatter plot with linear regression analysis showing biomechanical comparison of uniaxial (Uni) and monoaxial (Mono) screws.
ter the angle of insertion has exceeded the excursion angle of the screw head, uniaxial and monoxial screws may have similar plow potential.

REFERENCES


