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The Effects of Screw Length on Stability of Simulated Osteoporotic Distal Radius Fractures Fixed with Volar Locking Plates

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Abstract

Purpose—Volar plating for distal radius fractures has caused extensor tendon ruptures secondary to dorsal screw prominence. This study was designed to determine the biomechanical impact of placing unicortical distal locking screws and pegs in an extra-articular fracture model.

Methods—Volar-locking distal radius plates were applied to 30 osteoporotic distal radius models. Radii were divided into 5 groups based on distal locking fixation: bicortical locked screws, 3 lengths of unicortical locked screws (abutting the dorsal cortex [full length], 75% length, and 50% length to dorsal cortex), and unicortical locked pegs. Distal radius osteotomy simulated a dorsally comminuted, extra-articular, fracture. Each constructs stiffness was determined under physiologic loads (axial compression, dorsal bending volar bending) before and after 1000 cycles of axial conditioning and prior to axial loading to failure (2mm of displacement) and subsequent catastrophic failure.

Results—Cyclic conditioning did not alter constructs stiffness. Stiffness to volar bending and dorsal bending forces were similar between groups. Final stiffness(N/mm) under axial load was statistically equivalent for all groups: bicortical screws(230), full-length unicortical screws(227), 75% length unicortical screws(226), 50% length unicortical screws(187), unicortical pegs(226). Force(N) at 2 mm displacement was significantly less for 50% length unicortical screws(311) compared to bicortical screws(460), full-length unicortical screws(464), 75% length unicortical screws(400), and unicortical pegs(356). Force(N) to catastrophic fracture was statistically equivalent between groups but mean values for pegs(749) and 50% length unicortical(702) screws were 16-21% less than means for bicortical(892), full-length unicortical(860), and 75% length(894) unicortical constructs.

Discussion—Locked unicortical distal screws of at least 75% length produce construct stiffness similar to bicortical fixation. Unicortical distal fixation for extra-articular distal radius fractures should be entertained to avoid extensor tendon injury since it does not appear to compromise initial fixation.

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Clinical Relevance—Biomechanical comparison of distal fixation techniques during volar locked plating for distal radius fracture.

Keywords

Bicortical; Biomechanical; Distal radius; Fracture; Volar locking plate

Introduction

Surgeons have trended away from dorsal plating of distal radius fractures and toward volar plating.⁽¹⁾ This transition was in part attributed to the perception of higher rates of extensor tendon complications following dorsal plating, ease of the volar surgical approach, and increased strength and stiffness of volar locking constructs.⁽²⁾

The increased use of volar locking plates for distal radius fixation has been accompanied by recognition of their complications. Extensor tenosynovitis, extensor pollicis longus rupture, extensor digitorum communis rupture, and second dorsal compartment tenosynovitis have been reported.⁽³⁻⁹⁾ These dorsal tendon injuries are likely multifactorial: extensor tendons may be injured by the initial trauma, drill penetration of the dorsal cortex, or dorsally prominent screws causing chronic irritation.

When using volar locking plates, the necessity of bicortical fixation in the metaphyseal and epiphyseal areas of the distal radius has not been proven. It is unknown whether or not it is necessary or advisable to obtain distal bicortical fixation and place the extensor tendons at potential risk, or whether unicortical fixation is sufficient to provide stability and fracture healing.

The primary aim of this study was to determine if unicortical distal locked fixation of extra-articular distal radius fractures is stable under simulated physiologic loads. Secondary aims were to quantify the biomechanical impact of varied unicortical screw lengths relative to the dorsal-palmar width of the radius and to compare the biomechanical data of fixation with locked pegs versus screws of equal length. Our central hypothesis was that unicortical fixation of at least 75% length would outperform screws of 50% length due to a lack of dorsal subchondral support for the distal radius articular surface. Our secondary hypothesis was that screw fixation would be mechanically superior to peg fixation.

Materials and Methods

Specimen construction

Thirty plastic radius models (Sawbones®, Pacific Laboratories, model #1027-130) were employed in this study. This Sawbone has been employed in several prior mechanical studies of distal radius fracture fixation.⁽¹⁰⁻¹²⁾ We chose the manufacturer's osteoporotic version (foam cortex, cancellous filling) to represent a common clinical scenario. Volar distal radius plates (Medartis, Basil, Switzerland -model #A-4750.17) were placed in a systematic and reproducible manner to ensure consistent subchondral screw (2.5mm) placement.

The most proximal diaphyseal screw was inserted 5.4cm proximal to articular surface. Then, the most ulnar hole in the distal row was filled (3mm proximal to ulnar-most juxta-articular bony prominence). An external drill guide was used for all distal and second row screws to eliminate variation in screw or peg angles. Distal screws/pegs were consistently placed immediately subchondral using this method. This placement was verified by visualizing the dorsal tip of bicortical screws and by fluoroscopic examination in all specimens (Figure 1).

Distal locking screws and pegs were distributed between proximal and distal rows (Figure 2).⁽¹³⁾ Two additional locking screws (oblong hole, and one hole distal) were inserted bicortically to maximize mechanical stability as recommended by Weninger et al.⁽¹¹⁾ The radii were divided into 5 groups (n=6 per group) based on type of distal locking fixation. For group 1 (bicortical), the distal screws (those in the distal radius fragment) were drilled and inserted bicortically and confirmed by visualization of the tips of the screws emerging from the dorsal cortex. In group 2 (full-length unicortical), the distal screws were drilled unicortically and placed to abut but not penetrating into or through the dorsal cortex. This was confirmed both by depth gauge measurement and fluoroscopy. Group 3 (75% length unicortical) was constructed with all distal screws lengths equaling 75% of the distance from the volar to the dorsal cortex. Group 4 (50% length unicortical) consisted of distal screws that were 50% of that distance. Group 5 had full-length unicortical locking pegs in the distal row with similar length screws in the second row. This configuration replicated the clinical use of distal row pegs to minimize complications associated with threaded intra-articular hardware. The locking pegs were partially threaded, allowing engagement in the volar cortex. All radii had 3 bicortical diaphyseal screws.

After plating radius osteotomy created an AO type 23 A3.2 extra-articular fracture, representing a dorsally comminuted distal radius fracture. A 1 cm dorsal wedge of bone was removed (based 1cm proximal to the distal aspect of Lister tubercle), and the osteotomy was completed by fracturing the volar cortex. We chose this dorsal fracture gap model for consistency with prior mechanical evaluations of distal radius fixation.^(11,13-16) Constructs were cut to 12.5 cm for potting. Prior to biomechanical testing the points of contact for load application were marked on all specimens in a systematic fashion to ensure identical load placement for all groups.

Biomechanical testing

A materials testing machine (Instron 5866) was used for testing. A custom-built device for load application was constructed, which consisted of a ball-bearing for the contact surface with the Sawbone.

The mechanical testing strategy for each radius consisted of non-destructive pre-testing to determine stiffness in each of 3 loading modes (axial compression, dorsal bending, volar bending; Figure 3), followed by 1000 cycles of axial loading, then non-destructive post-testing in the 3 modes, and finally destructive axial loading to failure. Non-destructive testing consisted of application of a 10N pre-load, followed by quasi-static axial compression loading of 250N, dorsal bending of 50N, and volar bending of 50N, each at a displacement rate of 0.5mm/sec to determine constructs' stiffness without causing irreversible changes in the construct. Force and actuator displacement were recorded by the Instron built-in transducers, while displacement across the osteotomy site was determined using a motion analysis system (see below). The osteotomy gap displacement was used for subsequent stiffness and motion calculations. All constructs were taken through 5 cycles of loading for each non-destructive test. The mean value of stiffness (slope of the linear portion of the load-displacement curve; N/mm) over the final 3 loading cycles was used for data analysis. Between pre- and post-tests the constructs were loaded with 1000 cycles of sinusoidal loading to 250N at 1 Hz to condition the construct, (i.e., to simulate a period of early active motion rehabilitation). Forces were consistent with previously published biomechanical testing of distal radius fixation and studies demonstrating that light active motion of the wrist and wrist with finger motion does not exceed 250N.^(12,15,17-24) Gondusky et al and Weninger et al identified in prior distal radius mechanical testing that all changes attributable to cyclic loading occurred in the first 200-500 cycles, thus we deemed 1,000 cycles sufficient to condition the constructs.^(11,13-16) Destructive testing was completed to determine load (N) to clinical failure which was defined as 2mm of

displacement at the osteotomy gap,⁽¹⁰⁾ and subsequent catastrophic failure, defined as complete closure of the 1 cm osteotomy gap and or fracture.

Two cameras with infrared light sources (Oqus system, Qualysis) were placed 1 m from the region of interest with their axes 60° apart. The system was calibrated with a 3-dimensional frame with 6 reflective markers at known coordinate locations. Hemispherical markers (2.5 mm diameter) were glued on the dorsal edges of the osteotomy gap and displacement of these markers was recorded.

During cyclic testing 2 specimens sustained catastrophic failure. One construct failed at an incompletely seated locking screw. A second fractured at the bone and potting interface likely secondary to mal-alignment of the construct in the testing apparatus. Both failed constructs were discarded prior to data analysis, and new specimens were substituted in their place.

Data Analysis—The stiffness under dorsal bending volar bending, axial loading and load to failure was determined for each fixation group. Descriptive statistics for each group defined mean values ($\pm$ SD). Paired Student' t-test analyzed differences between elastic stiffness before and after cyclic conditioning for each fixation group. One-way analysis of variance compared the groups' mean stiffness under each loading condition, load to 2mm displacement, and load to catastrophic failure (fracture).

Statistical significance was set at $P < 0.05$. Based upon previous analyses, 6 specimens per experimental group were determined sufficient to adequately detect a 200 N difference in failure force between the constructs with 80% power.^(10-12,18)

Despite planning and conducting this study to be powered consistent with prior investigations, several analyses were underpowered.^(10-13,15,25-28) Post-hoc analysis demonstrated appropriate power when analyzing axial compression force at 2mm displacement; however, we would have needed 22 specimens per group to detect the difference between our most rigid and least rigid constructs for axial stiffness or force at fracture. Thus, we have detailed relative changes between groups that might be of potential clinical importance in-vivo despite statistical equivalence.

Results

Mean stiffness for each group was similar before and after cyclic loading (Figure 4) with no significant changes in stiffness under axial compression, dorsal bending, or volar bending. Mean stiffness to bending forces was statistically similar between fixation constructions ($P > 0.4$) with mean volar bending stiffness consistently equaling or exceeding dorsal bending stiffness (Table 1).

Final mean stiffness in axial loading for bicortical screws, full-length unicortical screws, 75% length, and 50% length unicortical screws was 230 (± 38) N/mm, 227(± 49) N/mm, 226(± 46) N/mm, and 187 (± 25) N/mm respectively. Unicortical fixation with locked pegs abutting the dorsal cortex reached 226 (± 48) N/mm stiffness. Comparing all groups, there was no statistical difference in stiffness ($P = .60$). However, the mean axial stiffness of unicortical screws with at least 75% of length to the dorsal cortex was 98% of the mean for bicortical screws, whereas 50% length screws averaged only 82% of the stiffness of bicortical fixation.

Table 2 presents the average axial loads to clinical failure. Bicortical screws, full-length unicortical screws, and 75% length unicortical screws sustained at least 400 N force. In this case, 50% length unicortical screw fixation failed at 68% (311 N) of the mean load required

for 2mm displacement in constructs fixed with bicortical screws. This decrease in force to failure was statistically significant ($P=0.04$). The only construct with a minimum force to failure below 200N was 50% length unicortical screw fixation.

The average force required for catastrophic failure was $892(\pm 314)$ N with bicortical fixation, $860(\pm 257)$ N with full-length unicortical screws, $894(\pm 151)$ N for 75% length unicortical screws, and $702(\pm 267)$ N for 50% length unicortical screws. Unicortical pegs failed at $749(\pm 215)$ N. These difference were not statistically significant ($P=0.56$).

In all specimens clinical failure with 2mm displacement occurred through plastic deformation of the Sawbone and plate while catastrophic failure occurred via comminution of the distal fragment along lines of screw placement and complete closure of the 1cm dorsal defect. There was no observed failure of the plates, screws, or pegs.

Discussion

In an osteoporotic model of extra-articular fractures of the distal radius, we demonstrated that full-length and 75% length unicortical distal locked fixation are similar mechanically to bicortical fixation. This finding suggests that the stiffness of locked subchondral support of the distal radius articular surface is produced by the fixed-angle screw-plate interface. Although the dorsal cortex imparted limited or negligible fixation rigidity, it is potentially of critical importance as a barrier between the hardware and the dorsal extensor tendons.

We believe that the mechanical equivalence (clinically and statistically) of unicortical and bicortical distal fixation should be viewed as evidence in favor of placing unicortical fixation in clinical practice. Unicortical distal fixation of distal radius fractures is potentially advantageous to bicortical fixation for several reasons. First, bicortical fixation places the extensor tendons at risk for irritation, synovitis, and rupture as a result of dorsally prominent screws.⁽³⁻⁹⁾ Second, extensor tendons can be injured with drill penetration during fracture fixation. This inciting injury may progress to tendinitis or rupture. Thus, the current model replicated the clinical practice of the senior author in which drilling for unicortical screws is completed when the dorsal cortex is abutted and not penetrated during fixation of extra-articular distal radius fractures.

It is difficult to assess the prominence of bicortical screws dorsally during fracture fixation. The dorsal surface of the distal radius has irregularities that create difficulty assessing the screw tip relationship to the dorsal cortex fluoroscopically.⁽²⁹⁾ Semi-pronated and semi-supinated fluoroscopic views have been described; however, these images may still fail to demonstrate screw tips within the depression of the third extensor compartment adjacent to Lister tubercle.⁽³⁰⁾ For this reason the authors pursued the secondary aim of determining the effect of varying the unicortical screw lengths and also the effect of smooth pegs as compared to screws of identical length. The mechanical performance of 75% length unicortical screws (compared to full-length unicortical screws abutting the dorsal cortex or bicortical screws) indicated that there is length that may be subtracted from absolute anteroposterior radius width if the surgeon is concerned about hardware prominence without causing a clinically or statistically relevant loss of fixation strength. A gradual decrease in axial load to failure to clinical failure was observed when shortening screws to 50% of the distance to the dorsal cortex. Despite the observed mean failure force that in all groups was above 250N, the minimum force for failure in the 50% length group was 193 N. This is potentially less than the force produced with active motion of the fingers and wrist prior to bony healing.⁽¹⁹⁾ This low value places this 50% screw length construct at risk for failure with early active wrist and digit motion.

At this time, there is no universally validated ex-vivo model for the mechanical evaluation of distal radius fracture fixation. We chose to place distal fixation in both the distal and second row of the plate to improve stiffness. Replicating clinical practice, we filled 3 distal holes and 2 second row holes. If all holes in the distal rows had been filled we anticipate that stiffness may have increased in all groups. Our study used a Sawbone model in order to ensure consistency between specimens and to obviate the need for computed tomography to assess for bone quality or significant structural differences between specimens. The Sawbone model has been used in previously published studies evaluating the biomechanics of distal radius fracture fixation techniques.^(10-12,14,15) This osteoporotic model was chosen to best replicate the clinical situation when fixation stiffness is most reliant on subchondral locked support in fractures at risk for collapse. In this study, all specimen failures during catastrophic testing were by collapse of the bone, and not by either plate or screw failure. This mode of failure has been observed following mechanical testing of elderly cadavera with fractures of the distal radius.^(16,25) The values of stiffness under axial compression and loads to catastrophic failure obtained in this study were similar to those observed in elderly cadaveric models, further supporting the validity of our Sawbone model (Table 3).^(13,16,25) While models such as the fourth generation Sawbones (Pacific Research Labs, Vashon, WA) have excellent reliability and uniform mechanical properties, they are manufactured to be comparable to young adult diaphyseal bone. The rigidity far exceeds that of the elderly distal radius metaphysis as evidenced by the following study.⁽³¹⁾ Sokol et al evaluated hybrid versus all-locked distal radius fixation in a synthetic osteoporotic model.⁽¹⁴⁾ They used fourth generation Sawbones and reported that despite decreased medullary foam density (to simulate osteoporosis) distal radius constructs maintained stiffness equivalent to the non-osteoporotic specimens and catastrophic failure was universally by plate bending.⁽¹⁴⁾ In their experiment, the stiffness of the cortical shell was unchanged from their non-osteoporotic model and both the Sawbone cortex and screw-bone interface exceeded plate stiffness.

Although the range of anticipated forces across the wrist with unresisted active digital and wrist motion are likely in the range of 90-250 N, the exact force transmission to the radius is understood incompletely. Light active wrist motion is anticipated to result in 100N of force with digital flexion increasing this up to 250N based on load created by extrinsic finger flexors in a position of grip.^(19-21,23) Despite these widely referenced values, the distribution of forces between the radius and ulna are expected to vary depending on radioulnar variance, and the forces across the radius during rehabilitation likely vary between axial load, volar bending, dorsal bending, and torsion forces. We focused on evaluating failure under axial compression rather than bending stiffness, as axial loads against the 250N standard are believed to be the major failure force in vivo.⁽³²⁾

Locked peg fixation demonstrated a range of forces to failure (2mm displacement and fracture) similar to that of full-length unicortical screws during failure testing. Notably, our model likely represents a best case scenario for peg performance as poorer bone quality minimizes screw purchase. With no advantage to peg use when drilling unicortically and increased clinical operative difficulty removing pegs compared to screws, we prefer use of locked screws in the distal radius as opposed to pegs.⁽³³⁾

Several limitations are apparent in this study. First is our use of the synthetic bone model. As addressed previously, we chose this model to maintain consistency between tested constructs and allow reproducible placement of internal fixation. These constructs might not reflect the in-vivo state that has intact secondary stabilizers (e.g., periosteum), and our experiment does not assess any differences between groups introduced during healing. Second, we did not test constructs under torsional or translational applied forces. While dorsal and volar loading was done to recreate wrist flexion and extension, physiologic

motion may also impart torsional forces to the construct. Putnam et al did not identify a clear torsion moment in their study of forces across the wrist, and we felt that evaluations of these forces would not have contributed substantially to fulfilling our aims.⁽¹⁹⁾ Finally, our conclusions of this study are limited to an extra-articular fracture model in which no coronal plane fracture exists between the volar and dorsal cortices. Extrapolation to intra-articular fracture fixation (e.g., those fractures with dorsal lunate facet fragments) cannot be recommended.

Our data indicate that unicortical fixation with either pegs or screws, provides adequate stiffness and strength for the internal fixation of extra-articular distal radius fractures under physiologic loads associated with early composite wrist and digit flexion-extension. We recommend unicortical fixation to minimize extensor tendon complications since there is a negligible effect of decreasing screw length on construct strength. As screw lengths of at least 75% of the distance from the volar plate to the dorsal cortex maintained mechanical properties approximating bicortical fixation we recommend drilling to (but not through) the dorsal cortex and placing screws 2mm shorter than measured length in cases when concern exists regarding dorsal screw prominence. Alternatively, if a comminuted dorsal cortex is penetrated during drilling and the distance to extensor tendons is uncertain, then placing screws 4 mm short of measured length should protect extensor tendons while providing necessary articular support for the distal radius.

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Figure 1. Representative fluoroscopy image of subchondral screw placement. (Image from Group 2: unicortical distal screw fixation abutting dorsal cortex).

Figure 2.
Diagram of fixation placement according to group.

Figure 3.

Photographs of biomechanical testing apparatus (A: axial compression, B: dorsal bending, C: volar bending)

Figure 4.
Mean construct stiffness before and after cyclic conditioning by fixation group.

Table 1

Mean dorsal and volar bending stiffness.*

Fixation Construct	Dorsal Bending Stiffness (N/mm) ($\pm$ SD)	Volar Bending Stiffness(N/mm) ($\pm$ SD)
Bicortical	59 ($\pm$ 47)	68 ($\pm$ 47)
Unicortical Screws	55 ($\pm$ 38)	82 ($\pm$ 59)
75% Unicortical Screws	66 ($\pm$ 48)	63 ($\pm$ 50)
50% Unicortical Screws	54 ($\pm$ 31)	104 ($\pm$ 60)
Unicortical Pegs	61 ($\pm$ 48)	114 ($\pm$ 54)
P values**	0.99	0.42

* All post cyclic conditioning

** ANOVA

Table 2

Forces to failure by construct group.

Construct	Failure Force (N $\pm$ SD)	Minimum	Maximum
Bicortical Screws	460 $\pm$ 140	321	700
Full-length Unicortical Screw	464 $\pm$ 86	363	580
75% Length Unicortical Screws	400 $\pm$ 61	306	493
50% Length Unicortical Screws	311 $\pm$ 82	193	446
Full-length Unicortical Peg	356 $\pm$ 61	302	450
P value *	0.04		

* ANOVA

Table 3

Comparison of mechanical properties between current study and elderly cadavera models.

Study	Fixation	Axial Stiffness (N/mm $\pm$ SD)	Catastrophic Failure Load (N $\pm$ SD)
Current Study	Volar Locking Plates	229 $\pm$ 38	892 $\pm$ 314
Gondusky	Volar Locking Plates	183 $\pm$ 70	1107 $\pm$ 445
Kandemir	Volar Locking Plates	199 $\pm$ 80	1238 $\pm$ 530
Mehling	Volar Locking Plate	208 $\pm$ 60	228 $\pm$ 56 [*]

^{*} This fracture model included gap dorsal and volar with 2 distal row screws, 2 proximal row screws