The Addition of Cerclage Wiring Does Not Improve Fixation of Mid-Diaphyseal Periprosthetic Humerus Fractures

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Introduction

The increased use of total shoulder arthroplasty (TSA) to treat osteoarthritis has led to an increase in periprosthetic fractures, especially in the elderly population. Reconstruction of humeral periprosthetic fractures continues to be a difficult task, with an overall complication rate ranging between 20% to 40%, and a nonunion rate up to 13%. Open reduction internal fixation with plates and screws is often used to address mid-diaphyseal fractures, but this poses a technical challenge because the stem of the TSA implant prevents bicortical screw purchase. Cerclage techniques (wrapping thin Kirchner wires around the plate and bone) have been used to provide additional support to a lateralized locking plate. Recently, the introduction of polyaxial locking screws has provided surgeons with the ability to gain improved screw purchase around the stem of the TSA implant without sacrificing rigidity of the screw-plate interface. It is currently unclear if this design feature eliminates the need for cerclage wiring to provide additional support. The purpose of this study was to determine the fixation strength of mid-diaphyseal periprosthetic fracture reconstructions using polyaxial locking plates with and without cerclage wires. We hypothesized that, although cerclage wiring provides limited support to the reconstruction in isolation, it is not a significant contributor to overall construct stability.

Methods

Eighteen synthetic (4th generation Sawbones) left humeri were implanted with a TSA implant by a fellowship-trained orthopaedic surgeon. Periprosthetic fractures were modeled by creating a ± mm osteotomy 10 mm distal to the tip of the humeral stem. Fractures were reconstructed with three separate techniques (n = 6): cerclage only (C), screws only (S) and cerclage with screws (CS) (Figure 1A). Specimens underwent dynamic axial compression and torsional test protocols. Compression tests modeled loaded motions, such as rising from a chair, which have transient centers of pressure on the humeral head. A custom jig that uses the humeral head as a fulcrum in a see-saw mechanism was designed for this study (Figure 1B). Controlled displacements were applied to the left side of the see-saw with a universal test frame and the opposite side was balanced by a spring. This setup resulted in humeral head contact forces between ~150-350 N. Because the crosshead of the test frame does not directly contact the humeral head, a thin film pressure sensor (Tekscan) was used to measure loads and 3D motion capture was employed to measure interfragmentary motions. Samples were cycled for 1000 cycles. For torsional tests, a custom mold was used to apply controlled moments of 2 Nm for 10 cycles, followed by a ramp to 10 degrees of valgus rotation relative to the shaft at 1 deg/s (Fig 1C). For both test modes, a bilinear fitting algorithm was used to determine stiffness in the toe and elastic regions of the stiffness curves. Differences between groups were determined by performing one-way ANOVAs with pairwise comparisons, with significance set to $p < 0.05$.

Results

Compression testing revealed no significant differences in stiffness between the C, S, and CS groups immediately after reconstruction (10 cycles) and after cyclic loading (1000 cycles) (Figure 2). Although stiffness values during compression were similar between groups, non-destructive torsional testing revealed significantly lower stiffnesses in the C group when compared to the other groups (Figure 3A&B). After 10 cycles the C group had a toe region stiffness of 0.816 ± 0.3 Nm/deg, while the S group (1.436 ± 0.3 Nm/deg, p = 0.0039) and CS group (1.268 ± 0.5 Nm/deg, p = 0.076) were substantially higher. For rotational stiffness in the elastic region, the C group demonstrated a decrease in stiffness (0.373 ± 0.1 Nm/deg) while the S and CS groups showed significantly higher elastic stiffnesses of 0.880 ± 0.1 Nm/deg and 1.069 ± 0.3 Nm/deg ($p < 0.001$ for both comparisons). Destructive torsional testing revealed differences in toe stiffness between the C group (0.607 ± 0.1 Nm/deg) and the S group (0.959 ± 0.2 Nm/deg, p = 0.0113), with similar findings compared to the CS group (0.928 ± 0.3 Nm/deg, p = 0.0415). Significant changes in

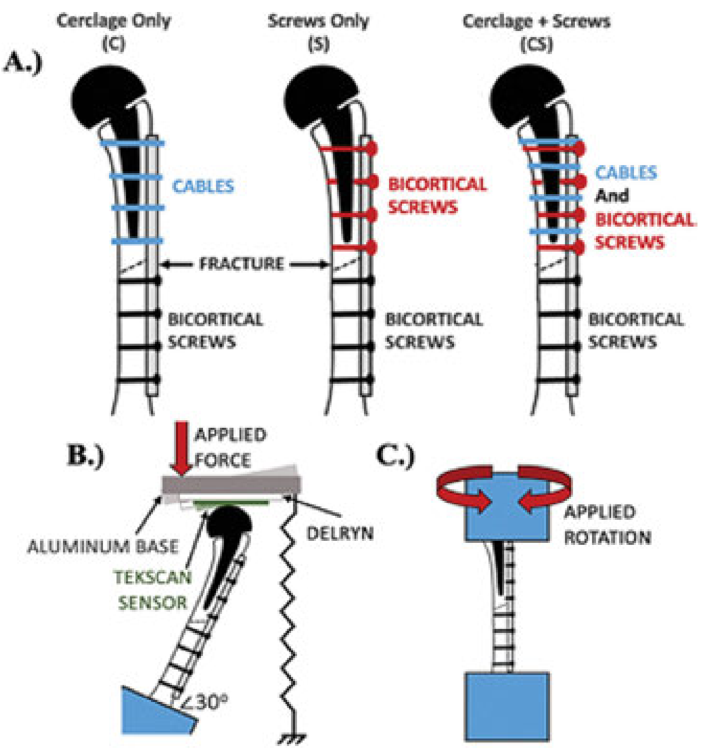

Figure 1. **(A)** Visualization of specimens in the C, S, and CS groups after fixation; **(B)** Axial see-saw test setup; **(C)** The torsional tests performed.

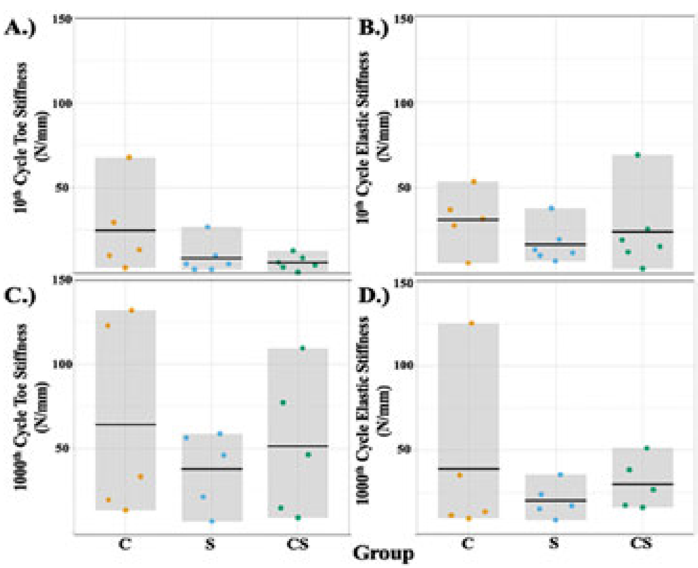

Figure 2. **(A)** Toe stiffness after 10 cycles; **(B)** Elastic Stiffness after 10 cycles; **(C)** Toe stiffness after 1000 cycles; (D) Elastic stiffness after 1000 cycles.

rotational stiffness were not present in the elastic region of the curve. Finally, there were significant differences in the maximum torque achieved. Group C showed a maximum torque of 6.331 ± 0.8 Nm, and specimens in group S (8.057 ± 0.5 Nm) and CS (7.527 ± 1.0 Nm) were significantly higher than C, but not compared to each other.

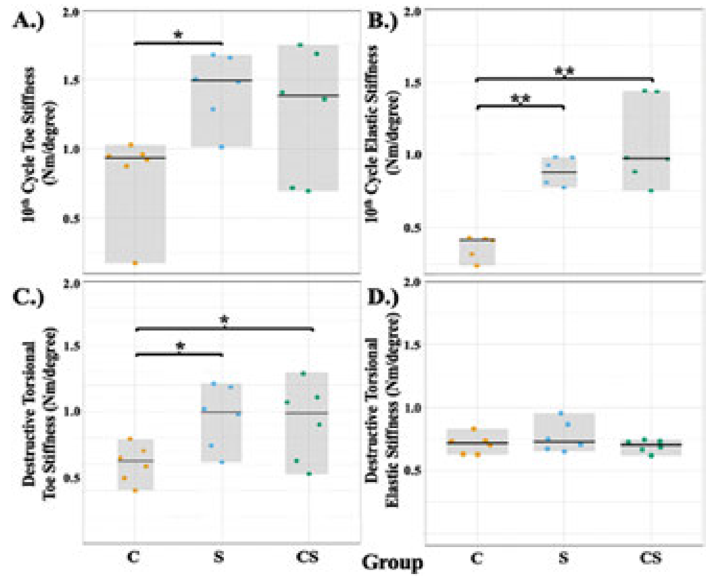

Figure 3. **(A)** Toe stiffness after 10 cycles of non-destructive torsion; **(B)** Elastic Stiffness after 10 cycles non-destructive torsion; **(C)** Toe stiffness during destructive torsion; **(D)** Elastic stiffness during destructive torsion. (*) represents a statistical difference of $p<0.05$ and (**) represents a statistical difference of $p<0.001$.

Discussion

Results from this study indicate that the addition of cerclage wiring does not improve initial fixation of polyaxial locking plates in humeral periprosthetic fracture reconstruction. Interestingly, the isolated use of cerclage wiring provided remarkably strong fixation during compression tests. However, qualitative assessments during experimentation indicated that this group experienced settling as specimens were loaded into the jig. Similarly, torsional ramp to failure testing indicates significant differences in toe stiffness, but not in the elastic region. This may be explained by some toggling of the cerclage-supported proximal fragment at low torques followed by stiff behavior as the bone engages with the wire support. This study was limited to the analysis of synthetic bones and further testing should be conducted with a cadaveric model.

Significance/Clinical Relevance

Fixation of periprosthetic humerus fractures continue to be challenging. Although it may be tempting to believe that initial fixation strength can be improved with the addition of cerclage wires, this technique provides no biomechanical advantage.

Acknowledgements

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