An Original Study

Distal Humerus Internal Fixation: A Biomechanical Comparison of 90° and Parallel Constructs

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Abstract

The optimal plating configuration for open reduction and internal fixation of complex distal humerus fractures is controversial. We evaluated cyclic fatigue performance and biomechanical stiffness in a cadaveric model of distal humerus fractures: precontoured nonlocking parallel plates versus a 90° nonlocking construct. A paired design was used, and 8 matched pairs of cadaveric arms were evaluated for bone density and plated. An osteotomy gap was created to simulate comminution, and constructs were randomized to anteroposterior, mediolateral, and torsional stiffness testing. Finally, 350 N was applied cyclically until deformation was permanent. Although there was a trend toward more stiffness of the parallel construct in anteroposterior, mediolateral, and torsional testing, it was not significant (*P*>.05). Likewise, there was no significant difference in number of cycles to failure (*P*>.05).

Given these results, we suggest that it is reasonable to use a 90° construct or a parallel construct for internal fixation of distal humerus fractures. However, our findings are consistent with a trend in the biomechanical literature supporting use of a parallel construct.

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Distal humerus fractures, comprising less than 2% of all fractures in adults, can be particularly challenging to manage.[1] Operative management with open reduction and rigid internal fixation allows early range of motion and thus limits the elbow's tendency toward stiffness.[2-4] However, such factors as poor bone quality, comminution, and intricate anatomy are significant obstacles to achieving adequate fixation.[5-8]

For complex intra-articular fractures of the distal humerus, the AO (Arbeitsgemeinschaft für Osteosynthesefragen) study group for the treatment of trauma and disorders of the musculoskeletal system advocates fixation with 2 plates placed 90° in relation to each other. The plates may consist of two 3.5-mm reconstruction plates, or one 3.5-mm reconstruction plate (placed over the medial column) and one 3.5-mm low-contact dynamic compression plate (placed posterolaterally over the lateral column).[9] When these nonlocking constructs are used, they must be contoured during surgery to fit the distal humerus. Although significantly more expensive, a proposed alternative, the Mayo Clinic Congruent Elbow Plate System (Acumed, Portland, Oregon), has the options of locking and nonlocking precontoured parallel plate configuration.

> "We [sought] to compare the biomechanical stiffness—anteroposterior (AP) bending, mediolateral (ML) bending, and torsional load—as well as cyclic fatigue performance of these 2 constructs"

We conducted the present study to compare the biomechanical stiffness—anteroposterior (AP) bending, mediolateral (ML) bending, and torsional load—as well as cyclic fatigue performance of 90° and parallel nonlocking plating configurations for internal fixation of the distal humerus. We hypothesized that, in our cadaveric model, there would be no significant difference in cyclic load to failure in AP bending when comparing nonlocking parallel plating with conventional 90° nonlocking plating for internal fixation of the distal humerus.

Figure 1. Mediolateral stiffness loading of 90° construct.

Materials and Methods

Eight (5 male, 3 female) matched pairs of fresh-frozen human cadaveric humeri were used in this study. Mean age at death was 81.25 years (range, 58 to 94 years). We used 2 pairs in addition to the required sample size of 6 pairs, which was determined by using an α of 0.05, power of .90 using a paired Student *t* test, and estimates of mean and variance from another study that used comparable plating configurations at the distal humerus.[10] Our sample size calculation was based on a 50% difference in AP stiffness using constructs similar to our proposed protocol.

Specimens were stored at –20°C. For preparation, they were thawed at room temperature. Eight matched pairs of cadaveric arms were stripped of all soft tissues and then evaluated for bone quality using dual-energy x-ray absorptiometry (DXA) to ensure similarity within matched pairs. The proximal portion of each humerus was removed with an oscillating saw, leaving the most distal 20 cm of the bone for testing.

A paired design was used, and arms were allocated to either parallel or 90° constructs. We then created a metaphyseal comminution model similar to what other investigators have used.[10-13] A supracondylar fracture was created with an osteotomy made 4 cm proximal to the most distal extent of the trochlea. Provisional Kirschner-wire fixation was used while hardware was being contoured and applied to the right and left arms within each pair, as determined by alternating sides consecutively.

For the 90° construct, a 9-hole, 3.5-mm pelvic reconstruction plate was carefully contoured to the shape of the medial column using a plate bender. A 2.5-mm drill was used for all holes, a 3.5-mm tap was used for proximal holes, and three 3.5-mm cortical screws were inserted. Distal to the osteotomy, three 4.0-mm cancellous screws were inserted. The 3 most distal holes in the pelvic reconstruction plate were always used; there was usually 1 unused hole at the osteotomy site, and then the 3 more proximal holes were filled with cortical screws. A 7-hole, 3.5-mm low-contact dynamic compression plate was then contoured to fit the posterolateral aspect of the distal humerus with a plate-bending press. Proximal to the osteotomy, a 2.5-mm drill was used, then a 3.5-mm tap, and three 3.5-mm cortical screws were inserted. Depending on the size of the distal humerus, usually the most distal 6 holes of the plate were used; three 4.0-mm cancellous screws were inserted distal to the osteotomy.

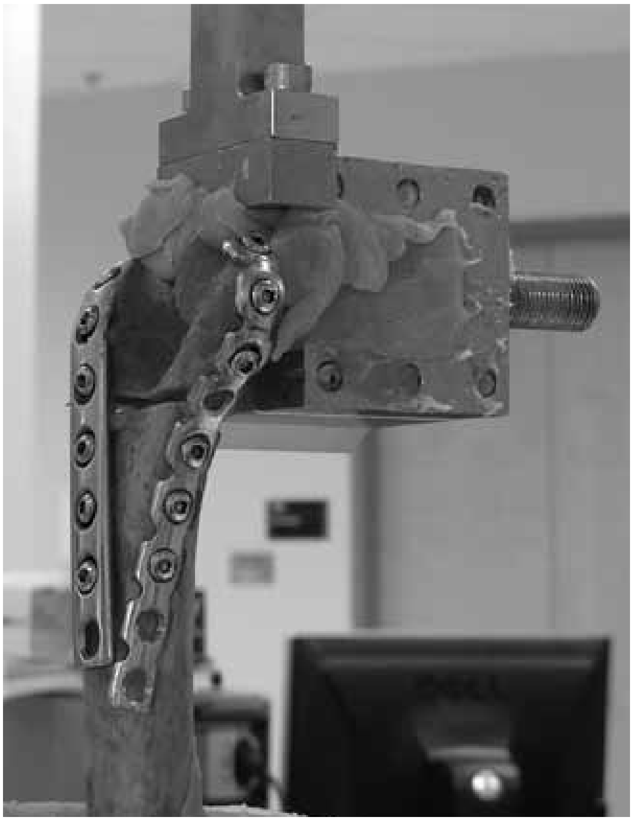

Figure 2. Torsional loading of 90° construct.

The parallel plating technique was carried out in accordance with recommended principles from the manufacturer.[14] Depending on the size of the distal humerus specimen, 7- or 8-hole plates were used for the medial column, and 9- or 10-hole plates were used for the lateral column. These precontoured parallel plates usually had to be bent slightly to conform to the exact shape of the medial column. The time required to do this was minimal compared with that required to contour conventional plates. The plates were fixed to bone using cortical screws, with the bone drilled and tapped consistent with operative technique. Three screws were placed distal to the osteotomy and 3 proximal. All attempts were made to achieve interdigitation of screws.

After the respective hardware was applied to each distal humerus, a second osteotomy was made, proximal to the first, and the sliver of bone was removed to create a 4-mm osteotomy gap. Necessary screws were backed out to allow this, and then reinserted (but this was seldom required in either group, and its frequency was similar

Figure 3. Anteroposterior loading of parallel construct.

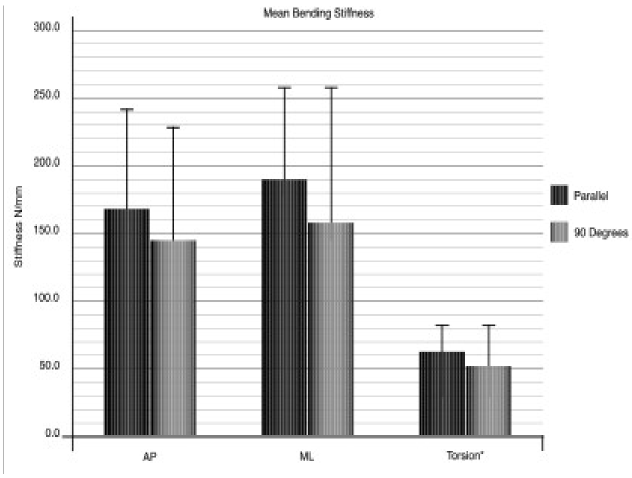

Figure 4. Mean (SD) bending stiffness for AP, ML, and torsional loading. *Torsional stiffness is quantified in Nm/degree. Abbreviations: AP, anteroposterior; ML, mediolateral.

between the groups). Although the manufacturer indicates a need for compression across fracture lines, we maintained the osteotomy gap to simulate comminution and maintain consistency with the 90° construct. This osteotomy gap fracture model is entirely consistent with that used by other investigators.[10-13,15]

Each construct was then cemented in a square aluminum bone pot using dental monomer (Pronto II; CMP Industries, Albany, New York). The proximal diaphysis was potted up to 10 cm from the distal end of the bone, and the distal articular surface was secured using a custom aluminum bone grip secured to the articular surface using dental monomer. Anodized aluminum pot and bone grip were designed specifically to allow loading along 3 orthogonal axes, aligned with the AP, ML, and proximodistal axes of the distal humerus (Figures 1–3).

For stiffness (our secondary outcome measure), loads were applied using a servohydraulic materials tester (858 Minio Bionix II; MTS, Eden Prairie, Minnesota) in the AP and ML directions using 1 df unidirectional force control (120 N). Torsional loads were applied using 2 df by specifying 1.4 N-m of torsion (internal and external rotation) along with 20 N of compression for the other df. These magnitudes were consistent with those applied by other investigators.[10] Stiffness testing involved 10 applications of the load in which all points were plotted when the load was increasing. The slope of the resulting line was used to determine stiffness, which was calculated as newtons per millimeter (N/mm) in the ML and AP directions. Torsional stiffness was quantified in newton-meters/degree (Nm/degree) by measuring the load-elongation curves at 1 Hz using a sine wave pattern. The order of the stiffness testing of the constructs was randomized.

Finally, 350 N was applied to the 16 constructs in a cyclical manner in the AP plane at 1 cycle per second until 0.9 mm of deformation of the construct was achieved. This represented a mechanically significant shift in the construct (achieved in most specimens), indicating failure of the screw–bone interface. We selected 350 N because our preliminary investigations at 200 N did not show evidence of failure, even when taken to 20,000 cycles (this had been unexpected, as other investigators have demonstrated failure at 3000 to 4000 cycles for 200 N of load[10]). In contrast, when 500 N was trialed, specimens failed quickly, at between 50 and 100 cycles. All specimens used for these trial analyses were not included in our study of 8 pairs of arms and were not used in the analysis. We decided that 500 N is beyond physiologic loads that can be expected in vivo in such patients during the rehabilitation period. Therefore, we chose 350 N for AP bending load to failure, with the primary outcome measure being number of cycles until failure or until the specimen reached 20,000 cycles in the AP direction.

Paired *t* testing was used for DXA data. Rigidity and load-to-failure data for the parallel and 90° constructs were analyzed with the Wilcoxon signed-rank test for a paired analysis at a level of significance of .05.

Results

DXA demonstrated similar bone mineral densities within pairs of distal humeri (P=.19), thereby validating use of our paired testing model, with a post hoc power of 0.99. There was a trend toward more stiffness of the parallel construct compared with the 90° construct in AP, ML, and axial torsion (internal-external rotation) testing, but this trend was not significant (P>.05). Mean stiffness in the AP direction was 169.0 N/mm (SD, 72.1 N/mm) for the parallel construct and 145.4 N/mm (SD, 82.5 N/mm) for the 90° construct; the difference was not significant (P=.25). In ML testing, mean bending stiffness was 200.6 N/mm (SD, 57.0 N/mm) for the parallel construct and 158.3 N/mm (SD, 98.6 N/mm) for the 90° construct; again, the difference was not significant (P=.31).

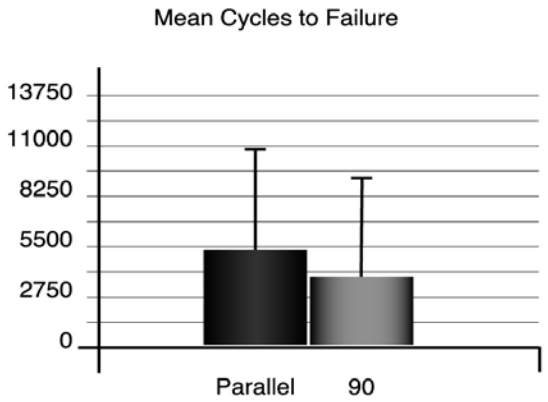

Figure 5. Mean (SD) cycles to failure of 90° and parallel plated constructs.

Torsional testing showed a similar, nonsignificant (P=.46) trend; mean stiffness was 6.14 Nm/degree (SD, 2.15 Nm/degree) for the parallel construct and 5.16 Nm/degree (SD, 3.01 Nm/degree) for the 90° construct (Figure 4).

The primary outcome measure was fatigue performance, number of cycles to failure in the AP direction. The means were 3884 cycles (SD, 5275 cycles) for the 90° constructs and 5376 cycles (SD, 5436 cycles) for the parallel constructs. Interestingly, 2 of the specimens from different pairs did not fail when cycled at 350 N, even when taken to 20,000 cycles. In the other 6 pairs, there was no significant difference (P=.22) between constructs in cyclic load to failure (Figure 5). A post hoc power calculation revealed a power of 0.22 for an α of 0.05. Of the 2 constructs that did not fail, 1 was 90°, and 1 was parallel. Displacement of these 2 constructs, even at 20,000 cycles, was minimal, between 0.4 and 0.6 mm. These particular humeri had bone mineral densities above the group mean, but there were other specimens with higher bone densities that failed. Therefore, for the failure analysis, we included the 6 pairs of arms in which both constructs reached at least 0.9 mm of deformation. Mode of failure for all constructs was loosening of the screw–bone interfaces and not deformation of the plates themselves. Screw backout was not a notable feature; in constructs that experienced massive failure, particularly in our trial analysis at 500 N, failure tended to occur from bone pulling through the screws in the metaphyseal area just proximal to the osteotomy.

Discussion

Results from this cadaveric biomechanical comparison of internal fixation for distal humerus fractures support the hypothesis that there is no significant difference in number of cycles to failure between a 90° nonlocking construct and a parallel precontoured nonlocking plate construct. Although we found a consistent trend toward more stiffness of the parallel construct, it was not statistically significant.

In 1990, Helfet and Hotchkiss[11] conducted some of the earliest work that supports perpendicular plating, but they made no comparisons with parallel constructs. Several other investigators have compared the biomechanical rigidity of nonlocking 90° plating constructs and parallel constructs in cadaveric specimens.[10,12,16] Schemitsch and colleagues[10] found that, compared with 90° constructs, a parallel plating technique had more AP, axial, torsional, and lateral bending. Self and colleagues,[16] evaluating axial rigidity and one-time load to failure in axial compression, found no significant difference between 90° constructs and parallel plating when bolts were not used. Jacobson and colleagues[12] also compared parallel and 90° constructs but did not find a significant difference in torsional stiffness. None of these 4 studies[10-12,16] used a paired design or evaluated bone density, and only Jacobson and colleagues used radiographs to confirm absence of bone lesions or defects. Two other studies[15,17] have compared nonlocking parallel and 90° constructs in epoxy resin humeri. Arnander and colleagues[15] found that parallel constructs had significantly more stiffness, whereas Schwartz and colleagues[17] found no significant difference in stiffness. In another study, using locking plates and a cadaveric model, Stoffel and colleagues[18] found that parallel constructs had significantly more stiffness in axial compression and external rotation, but there was no significant difference in load to torsional failure between perpendicular and parallel constructs.

Although the present study can be criticized for not evaluating locking constructs, we suggest that the biomechanical superiority of locking over nonlocking plates for the distal humerus has not yet been clearly demonstrated in the literature. Korner and colleagues[13] compared locking 90° constructs and nonlocking plate 90° constructs in a cadaveric distal humerus model and found that, though the locking construct showed a trend toward more stiffness in axial compression and anterior bending, this trend was not significant. There was no significant difference in load-to-failure testing between the 2 constructs. Parallel plating was not evaluated. Furthermore, many treatment centers may not have access to these costly locking plates for treatment of distal humerus fractures, and, therefore, we believe it is relevant to discuss nonlocking fixation.

Despite the increased expense of implantation, use of precontoured plates can be more time efficient, as a significant amount of surgical time can be spent precisely bending pelvic reconstruction and low-contact dynamic compression plates. With a parallel plating construct, longer screws can engage across the trochlea and capitellum because of the wider ML distance as compared with the AP width required for the posterolateral plate in the 90° construct. New locking plate technology has combined 90° and parallel screw orientation with a posterolaterally placed plate along with a medial plate.[19]

In the present study, we expected that a paired design would help limit the variation even more than in previous studies. However, despite use of a paired design and

DXA, which confirmed very similar bone densities within pairs, the variation was huge. Although the low power is reflective of this variation, it is relevant, as it indicates that 32 pairs of specimens would be required to achieve a β of 0.90 if estimates of mean and variance from our study were used along with an α of 0.05. The cost of such a study would be enormous, and it would be of questionable benefit, as we would expect that any effect size found would be relatively small and would not justify use of one implant over another from a biomechanical standpoint.

Conclusions

There is increasing biomechanical evidence supporting use of a parallel construct in open reduction and internal fixation of distal humerus fractures. When all the literature is considered, we suspect that actual differences between parallel and 90° constructs may be quite small in the clinical setting of osteoporotic fractures. Perhaps a more useful discussion involves primary total elbow arthroplasty versus open reduction and internal fixation in the trauma setting, especially in light of recent publications indicating improved functional outcomes in elderly patients who undergo arthroplasty for complex distal humerus fractures.[20-24] In the future, there likely will be a shift toward performing more primary arthroplasties in these patients. In the interim, when arthroplasty is not the chosen treatment, surgeons should be aware of the methodologic challenges of treating complex distal humerus fractures and should recognize that reduction is arguably more important than which plates are used.

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