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ORIGINAL PAPER

Parallel versus orthogonal plate osteosynthesis of adult distal humerus fractures: a meta-analysis of biomechanical studies

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Abstract

Purpose There are two widely used distal humerus fracture (DHF) fixation methods with either orthogonal or parallel double-plate osteosynthesis. However, biomechanical studies have shown inconsistent results on which technique is more effective. We performed a meta-analysis to compare these two fixation methods for adult DHF fixation.

Methods We searched the literature for entries discussing the biomechanical testing of orthogonal and parallel fixation techniques for DHFs. We then performed a meta-analysis of the following biomechanical outcome measures: axial/sagittal/coronal/torsional stiffness, load to failure, and torque to failure.

Results Seventeen studies comparing both constructs were included. The parallel configuration exhibited greater mechanical strength with respect to axial stiffness/load to failure, torsional stiffness, and posterior bending load to failure than the orthogonal constructs. Subgroup analysis revealed that parallel constructs also had higher torsional stiffness in supracondylar fractures.

Conclusions This meta-analysis shows that parallel constructs provide greater axial stiffness, axial strength, and torsional stiffness than orthogonal plate for DHF fixation. A subgroup analysis revealed that parallel constructs had better torsional stiffness in supracondylar fracture fixation.

Level of evidence IA

Keywords Biomechanical study · Distal humeral fracture · Fracture fixation · Meta-analysis · Parallel · Orthogonal

Introduction

Distal humerus fractures (DHFs), accounting for 2% of all fractures, are one of the most challenging elbow injuries to treat [1, 2]. Most DHFs require open reduction and stable internal fixation to permit early range of motion and avoid post-operative complications [3]. Implant failure, fracture non-union/malunion, and decreased range of motion are complications of inadequate fixation [4]. The reported dissatisfaction rate after surgical DHF stabilization may be as high as 20–25% in different studies [5, 6].

Surgical treatment principle is stable fixation of both medial and lateral columns [7, 8]. Double-plate constructs, including orthogonal [2, 8–10], parallel [2, 8, 11], and posterior two-plate constructs, are the popular methods [12–14]. Triple plating [15] and single plating [13, 16–18] are also options in selected fracture type. Other fixation methods include using tension band wires [19], crisscrossed screws [18, 20], and nails [21].

Double plating, either in orthogonal or parallel configuration, has superior biomechanical stability than single plating [22, 23] or triple plating [15]. Orthogonal configurations, suggested by AO group [10, 24], form a girder-like construct that provide resistance to the force generated in sagittal and coronal planes during elbow range of motion. Parallel configurations, proposed by Dr. O'Driscoll [25, 26], reconstruct the distal humerus by anchoring both columns to the base of

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humeral shaft with the connection at the arch crown via interdigitations of distal screws. In early clinical studies analyzing C-type DHF fixation [27, 28], both configurations yielded similar post-operative results; however, a recent clinical meta-analysis reported that parallel constructs have significantly lower complication rate than orthogonal ones [29]. In biomechanical studies, although both constructs allow for bicolumnar fixation [30, 31], it is still unclear which configuration offers better fixation strength or stability [1, 2, 7–10, 14, 24, 32–35]. Thus, through systemic review and an up-to-date search of currently available evidence, this meta-analysis aims to compare the biomechanical properties of orthogonal and parallel plate configurations.

Materials and methods

Inclusion and exclusion criteria

Our included studies are required to clearly describe specimen type, fracture type, fracture fixation, and mechanical testing protocol and meet the following criteria: (1) biomechanical studies with cadaveric or synthetic humeri; (2) comparing DHF fixation using orthogonal and parallel constructs; (3) intervention with loading test in axial, sagittal, or coronal direction; and (4) providing biomechanical data with calculated stiffness or failure loads. We excluded trials when they met the following criteria: (1) using pathologic fracture models; (2) paediatric fracture fixation models with pin fixation, isolated lateral or medial epicondylar fracture models; (3) models using animal cadaver specimens; (4) mechanical testing or measurement using unusual parameters; (5) using different plating systems in different constructs—locking plate in one group and non-locking plate in another. All abstracts were evaluated by two reviewers (CAS and TWT). When there were disagreements between the two authors, the third author (WRS) would join the discussion until the conclusion is made.

Search strategies and study selection

Relevant studies published before March 2017 were identified from Embase, PubMed, and Medline. We used the following Medical Subject Heading terms: humerus, fracture fixation, cadaver, synthetic bone, artificial bone, biomechanic, or mechanic. The search strategy is listed in the supplement file (Table 1, S). We also used the "related articles" function in PubMed to widen our search. All abstracts and citations were reviewed via a cross-reference check of all included articles. There was no language restriction.

Data extraction

Outcome data were extracted by two reviewers (CAS and WCL). Biomechanical outcomes (stiffness, load to failure, and torque to failure), numbers and type of specimens (cadaveric humeri or synthetic humeri), fracture model (supracondylar or intercondylar fracture), implant selection, and mechanical testing protocol were extracted. Individual reports were compared, and disagreements were solved by communication with a third reviewer (TWT). Authors of included studies were contacted for additional information.

Parameter selection

When several stiffness values were estimated in one study, the calculated axial stiffness that is closest to 50–150 N, the calculated lateral bending stiffness closest to 60 N, and posterior bending stiffness closest to 100 N were extracted. These loads reflect the stiffness values measured most frequently in other biomechanical studies. When a study compared one locking plate construct with another locking or non-locking plate construct, the strength of both locking plate constructs was chosen and extracted. When a study measured stiffness before and after cyclic loading, we extracted the stiffness value measured during pre-cyclic loading as pre-cyclic stiffness is the most common or the only stiffness value measured in other biomechanical studies.

Methodological quality appraisal

Two reviewers (TWT and CWL) assessed the methodologic quality using the Cochrane risk of bias tool, which included an evaluation of randomization, allocation concealment, blinding, incomplete outcomes, selective reporting, and other sources of bias. Any disagreement was resolved by a third reviewer (WRS).

Outcomes

The primary outcomes were the stiffness measured in axial, sagittal stiffness, coronal, and torsional directions. Secondary outcomes were the load to failure and torque to failure in different directions.

Statistical analyses

We used Review Manager, version 5.3 (The Cochrane Collaboration, Oxford, England) for data analysis. The meta-analysis was performed based on the PRISMA guidelines [36]. The outcome variables in this study are stiffness and the failure load/torque/cycles of each construct. The means and standard deviations are used for individual comparisons. Standard deviation was calculated via standard error or

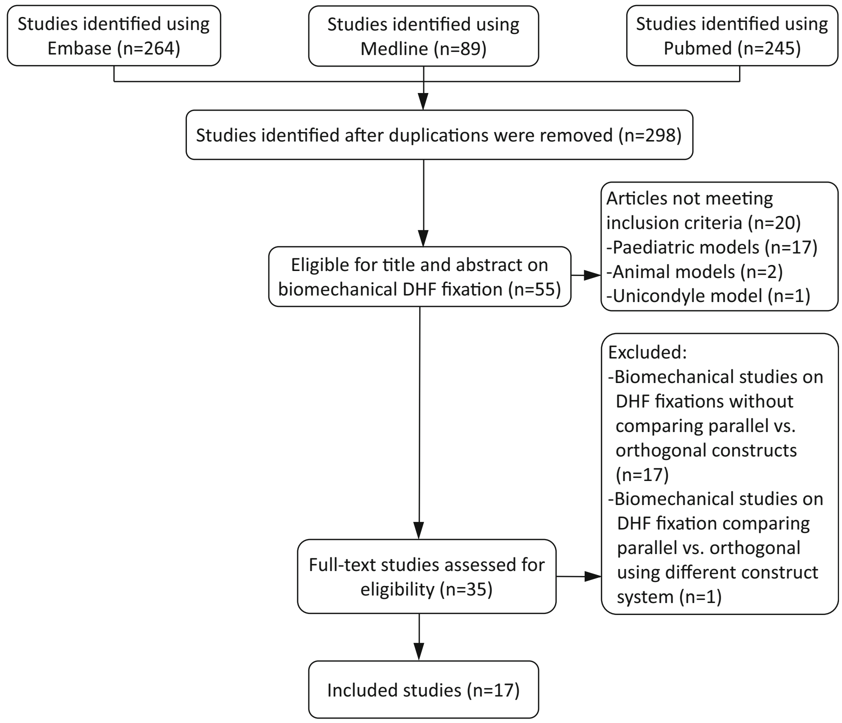


Fig. 1 Flowchart literature search according to the PRISMA 2009 guidelines

confidence interval. Cochrane Q tests and I^2 statistics were used to evaluate statistical heterogeneity and inconsistency, respectively, between studies. We set the statistical significance as $p < 0.10$ for the Cochrane's Q test. Statistical heterogeneity between studies was evaluated using the I^2 test, which quantifies the variability between different studies. We also performed subgroup analyses to evaluate differences between stiffness and failure load in supracondylar and intercondylar fracture models.

Results

Study characteristics

After proper screening and selection (Fig. 1), 17 studies were eligible for this meta-analysis (Table 1) [1, 2, 7–10, 14, 15, 22, 24, 30, 32, 33, 35, 37–39]. These included studies comparing orthogonal and parallel constructs published from 1900 to 2017 with sample sizes in each group that ranged from six to 28 specimens. Eleven studies used intercondylar fracture models [1, 7, 8, 14, 24, 30, 32, 33, 35, 37, 38], and six used supracondylar fracture models [2, 9, 10, 15, 22, 39]. Eleven studies performed their experiment with cadaveric humeri [2, 8, 9, 14, 15, 22, 24, 32, 35, 37, 39] while six used synthetic humeri [1, 7, 10, 30, 33, 38]. We summarized methodologic quality in Table 2, S. Selection bias (randomization method and allocation concealment) were unclear in two studies [22, 37] and low in others. In all studies, performance bias (blinding of participants and personnel) and detection bias (blinding of outcome assessment) were unclear, and attrition bias (incomplete outcome data), reporting bias (selective reporting), and other bias were low.

Meta-analysis on parallel versus orthogonal construct

Axial stiffness and load to failure in parallel and orthogonal constructs

The mean and standard deviation of axial stiffness in N/mm was recorded and compared in 11 studies on parallel and orthogonal configurations [1, 2, 7, 9, 14, 22, 30, 32, 33, 37, 38]. The pooled mean difference in axial stiffness was 240.88 (95% CI 148.13 to 333.63). Parallel constructs had a significantly higher stiffness than the orthogonal ones (Fig. 2). A subgroup analysis was performed based on the fracture type. The pooled mean difference in stiffness in the studies of intercondylar fracture models was 225.01 (95% CI 124.90 to 325.11), suggesting that stiffness was greater in the parallel group. The pooled mean difference in supracondylar fixation models was 403.05 (95% CI 29.41 to 776.67), which also favors parallel constructs. Axial load to failure was measured in one study, which failed to find significant difference between the two constructs [37].

Sagittal stiffness and load to failure in parallel and orthogonal constructs

Sagittal stiffness between parallel and orthogonal constructs was measured in 12 studies [1, 2, 8–10, 15, 22, 24, 30, 33, 35, 39]. Posterior bending stiffness was recorded in 11 studies [1, 2, 8–10, 22, 24, 30, 33, 35, 39], and one study was excluded

Table 1 Characteristics of the included biomechanical studies

Study	Injury type	Included specimen number in meta-analysis	Intervention	Stiffness testing	Axial stiffness (N/mm)	Sagittal stiffness (N/mm)	Coronal stiffness (N/mm)	Torsional stiffness (Nm/degree)	Load to failure (N)	Torque to failure (N)
Schemitsch	SC	16 Cadaveric bone	P: non-locking plates O: non-locking plates	A: 150 N S(P): 120 N *MAD (7.6–11 cm) C(L): 120 N T: 1.4 Nm	P: 3280 (940) O: 2080 (460)	N/A	P(L): 696.4 (206.2) O(L): 521.4 (159.5)	P: 5.5 (1.7) O: 3.5 (1.0)	N/A	N/A
Self	IC	12 Cadaveric bone	P: non-locking plates O: non-locking plates	A: 200 N	P: 127.2 (22.5) O: 180.3 (41.6)	N/A	N/A	N/A	P: 967.7 (407.6) O: 862.3 (234.6) *Axial loading	N/A
Jacobson	SC	10 Cadaveric bone	P: non-locking plates O: non-locking plates	S: unclear C: unclear T: 10 Nm	N/A	N/A	N/A	P: 2.1 (0.3) O: 1.8 (0.4)	N/A	N/A
Schwartz	IC	10 Cadaveric bone	P: locking plates O: locking plates	A: 2 Nm S(P): 100 N C(L): 100 N T: 2 Nm	P: 272.4 (37.0) O: 413.3 (173.3)	P(P): 37.8 (3.2) O(P): 42.1 (4.8)	P(L): 68.4 (12.9) O(L): 65.2 (13.4)	P: 37.9 (13.9) O: 51.5 (3.4) *Nm/rad	N/A	N/A
Arnander	SC	16 Synthetic humeri	P: non-locking plates O: non-locking plates	S(P): unclear	N/A	P(P): 214.9 (43.3) O(P): 138.3 (44.6)	N/A	N/A	P: 304.4 (63.5) O:234.9 (34.4) *Posterior bending	N/A
Stoffel	IC	24 Cadaveric bone	P: locking plates O: locking plates	A: 10–100 N T(I)(E): 2 Nm	P: 464 (82) O: 1073 (436)	N/A	N/A	P(I): 2.6 (1.2) O(I): 2.0 (0.9) P(E): 2.4 (0.9) O(E): 3.3 (1.3)	N/A	N/A
Kollias	IC	16 Cadaveric bone	P: non-locking plates O: non-locking plates	S(P): 120 N C(L): 120 N T: 1.5 Nm	N/A	P(P): 169.0 (72.1) O(P): 145.4 (82.5)	P(L): 200.6 (57) O(L):158.3 (98.6)	P: 6.14 (2.15) O: 5.16 (3.01)	N/A	N/A
Penzkofer	IC	12 Synthetic humeri	P: locking plates O: locking plates	A: 300 N S(P): 300 N	P: 1138.6 (130) O: 362.7 (30)	P(P): 115.6 (8.6) O(P): 201.7 (19)	N/A	N/A	N/A	N/A
Cai	IC	12 Cadaveric bone	P: non-locking plates O: non-locking plates	A: 100/200/300/400/50-0 N	P: 370 (12) O: 370 (11) *Stiffness measured at 100 N	N/A	N/A	N/A	N/A	N/A
Zalavras	IC	28 Cadaveric bone	P: locking plates with non-locking screws O: locking plates with non-locking screws	S(P): 20/40/60/80/100 N C(M): 20/40/60 N	N/A	P(P): 105.6 (24.5) O(P): 98.7 (24.0) *Stiffness measured at 100 N	P(M): 2.68 (0.43) O(M): 1.17 (0.6) *Stiffness measured at 60 N	N/A	P: 1287.8 (428.9) O: 800 (161.5) *Posterior bending	P(M): 20.7 (9.2) O(M): 15.9 (9.2)
Got	IC	20 Cadaveric bone	P: locking plates O: locking plates	S(P): 100 N T: 2.5 Nm	N/A	P(P): 156.74 (32.58)	N/A	P: 278.81 (37.41) O: 269.32 (34.16) *N/mm	N/A	P: 31.92 (16.23)

Table 1 (continued)

Study	Injury type	Included specimen number in meta-analysis	Intervention	Stiffness testing	Axial stiffness (N/mm)	Sagittal stiffness (N/mm)	Coronal stiffness (N/mm)	Torsional stiffness (Nm/degree)	Load to failure (N)	Torque to failure (N)
						O(P): 150.61 (47.69)				O: 44.07 (16.87)
Koonce	SC	20 Cadaveric bone	P: locking plates O: locking plates	A: 250 N S(A)(P): 4.5 Nm T: 1.6 Nm	P: 1558.0 (386.3) O: 1304.4 (416.8)	P(A): 1558.0 (386.3) O(A): 935.9 (164.8) P(P): 1236.9 (283.8) O(P): 994.9 (311.1)	N/A	P: 2.4 (0.8) O: 2.0 (0.7)	N/A	N/A
Caravaggi	SC	14 Cadaveric bone	P: locking plates O: locking plates	A: 250 N S(A)(P): 200 N	P: 540 (135) O: 319 (123)	P(A): 102(44) O(A): 81(26) P(P): 101(33) O(P): 108(43)	N/A	N/A	P: 781 (260) O: 553 (142) *Posterior bending	N/A
Kudo	IC	12 Synthetic humeri	P: locking plates O: locking plates	A: 50–150 N/100–200 - N	P(R): 153.4 (27.6) P(U): 112.0 (11.7) O(R): 90.5 (16.6) O(U): 107.1(11.7) *stiffness measured at 50–150 N	N/A	N/A	N/A	N/A	N/A
Taylor	IC	18 Cadaveric bone	P: locking plates O: locking plates	S(A)(P): 50 N T(I)(E): 2 Nm	N/A	P(A): 2.2 (1.8) O(A): 9.0 (1.5) P(P): 13.8 (2.1) O(P): 9.7 (1.3)	N/A	N/A	N/A	P: 10.5 (1.9) O: 8.4 (1.2)
Atalar	IC	16 Synthetic humeri	P: locking plates O: locking plates	A: 50 N S (A)(P): 50 N C(M)(L): 50 N	P: 2455.8 (1073.8) O: 2203.6 (831.6)	P(A): 545.5 (198.7) O(A): 516.5 (253.5) P(P): 473.5 (94.0) O(P): 627.4 (487.7)	P(M): 321.6 (68.1) O(M): 365.0 (59.3) P(L): 393.2 (36.0) O(L): 385.7 (44.8)	P(I): 2.9 (0.7) O(I): 2.3 (0.6) P(E): 3.2 (0.7) O(E): 2.4 (0.5)	P: 379.7 (35.7) O: 372.4 (26.5) *Posterior bending	N/A
Varady	IC	6 Synthetic humeri	P: locking plates O: locking plates	A: 300 N	P: 739.8 (81.3) O: 276.4 (65)	N/A	N/A	N/A	N/A	N/A

IC intercondylar, *SC* supracondylar, *P* parallel group, *O* orthogonal group, *A* axial, *S* sagittal, *C* coronal, *T* torsional, *(A)* anterior bending (sagittal), *(P)* posterior bending (sagittal), *(M)* medial bending (coronal), *(L)* lateral bending (coronal), *(I)* Internal rotation loading (torsion), *(E)* external rotation loading (torsion)

*Data are presented as the mean (SD)

Study or Subgroup	Parallel Mean	Parallel SD	Parallel Total	Orthogonal Mean	Orthogonal SD	Orthogonal Total	Weight	Mean Difference IV, Random, 95% CI
1.1.1 Intercondylar model (N/mm)								
Atalar 2017	2,455.8	1,073.8	8	2,203.6	831.6	8	0.9%	252.20 [-688.94, 1193.34]
Cai 2011	370	12	6	370	11	6	14.0%	0.00 [-13.03, 13.03]
Kudo 2016	132.7	21.2	6	98.8	14.4	6	14.0%	33.90 [13.39, 54.41]
Penzkofer 2010	1,138.6	130	6	362.7	30	6	11.9%	775.90 [669.15, 882.65]
Schwartz 2006	272.4	37	5	413.3	173.3	5	10.1%	-140.90 [-296.22, 14.42]
Self 1995	180.33	41.57	6	127.17	22.47	6	13.8%	53.16 [15.35, 90.97]
Stoffel 2008	1,072.7	436.3	12	463.6	81.8	12	6.9%	609.10 [357.94, 860.26]
Varady 2017	739.8	81.3	3	276.4	65	3	11.5%	463.40 [345.61, 581.19]
Subtotal (95% CI)			**52**			**52**	**83.1%**	**225.01 [124.90, 325.11]**
Heterogeneity: Tau² = 15305.36; Chi² = 285.65, df = 7 (P < 0.00001); I² = 98%								
Test for overall effect: Z = 4.41 (P < 0.0001)								
1.1.2 Supracondylar model (N/mm)								
Caravaggi 2014	540	135	7	319	123	7	10.8%	221.00 [85.71, 356.29]
Koonce 2012	1,558	386.3	10	1,304.4	416.8	10	4.6%	253.60 [-98.62, 605.82]
Schemitsch 1994	3,280	940	8	2,080	460	8	1.5%	1200.00 [474.81, 1925.19]
Subtotal (95% CI)			**25**			**25**	**16.9%**	**403.05 [29.41, 776.69]**
Heterogeneity: Tau² = 71600.84; Chi² = 6.77, df = 2 (P = 0.03); I² = 70%								
Test for overall effect: Z = 2.11 (P = 0.03)								
Total (95% CI)			**77**			**77**	**100.0%**	**240.88 [148.13, 333.63]**
Heterogeneity: Tau² = 15905.61; Chi² = 305.40, df = 10 (P < 0.00001); I² = 97%								
Test for overall effect: Z = 5.09 (P < 0.00001)								
Test for subgroup differences: Chi² = 0.81, df = 1 (P = 0.37), I² = 0%								

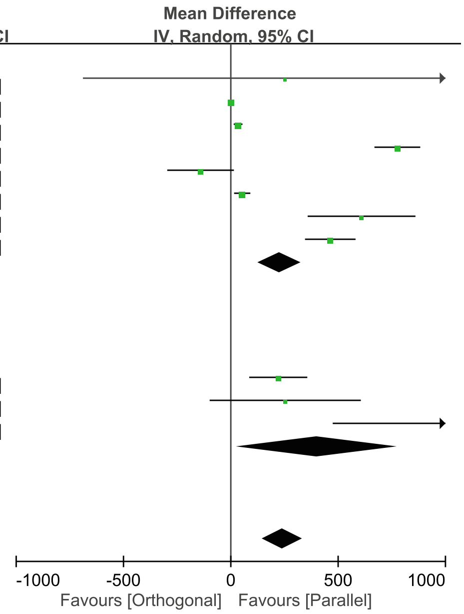


Fig. 2 Meta-analysis on construct axial stiffness

because it used a different measuring unit (Ncm/°) [22]. Ten studies were included in a meta-analysis of posterior bending stiffness measured in N/mm. No difference in sagittal stiffness was noted between the two constructs (mean difference −6.41, 95% CI −22.71 to 9.88). The excluded study revealed higher posterior bending stiffness in parallel construct [22]. Subgroup analysis found no difference between the posterior bending stiffness of either intercondylar or supracondylar fractures (Fig. 3a). Posterior load to failure was recorded in six studies [1, 2, 8–10, 39], posterior bending median fatigue limit (MFL) in one study [33], and posterior bending cycles to failure in one study [24]. The pooled results showed no difference between the two constructs (Fig. 3b), even after subgroup analysis. With respect to MFL, Penzkofer et al. found a greater load to failure in the parallel group. However, Kollias et al. [24] revealed that there was no significant difference regarding cycles to failure within two groups.

Anterior bending stiffness was calculated in four studies [1, 2, 9, 39]. This meta-analysis did not find a difference between the two constructs even after subgroup analysis (Fig. 3c).

Torsional stiffness and torque to failure in parallel and orthogonal constructs

Torsional stiffness in Nm/° or Nm/rad was recorded and compared in eight studies involving parallel and orthogonal constructs [9, 15, 22, 24, 30, 32, 35, 39]. One study was excluded because of the use of different measuring units (N/mm) [35]. Meta-analysis of the seven remaining studies found a higher torsional stiffness in parallel constructs (Fig. 4a). In the subgroup analysis, parallel constructs had higher stiffness when used in supracondylar but not intercondylar fracture models (Fig. 4a). The excluded study showed no significant difference in stiffness between the two constructs [35].

Torque to failure was recorded in two studies [32, 35]. No significant difference was noted between the two constructs (Fig. 4a).

Coronal stiffness and load to failure in parallel and orthogonal constructs

Coronal stiffness was recorded in six studies [1, 8, 15, 22, 24, 30]. One study was excluded because of the use of different units (relative stiffness) [15]. Four studies evaluated lateral bending (valgus loading) stiffness [1, 22, 24, 30], and two studies measured medial bending (varus bending) stiffness [1, 8]. Neither medial nor lateral bending stiffness showed a difference between the two constructs (Fig. 5a,b), even after subgroup analysis (Fig. 5a,b). The only excluded study found a significantly greater coronal stiffness in orthogonal constructs for supracondylar fracture fixation [15].

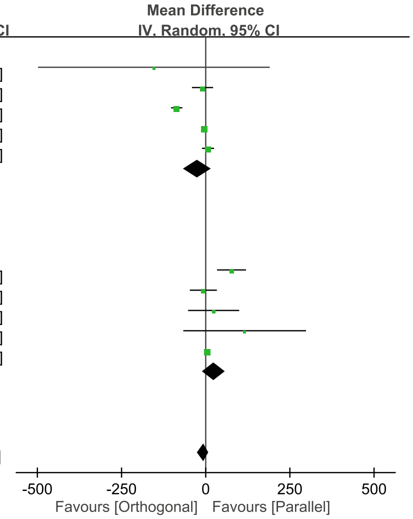
Medial bending load to failure was calculated in one study [8], which measured a higher load to failure in parallel constructs for intercondylar fracture fixation.

Assessment of publication bias

We assessed the publication bias of the included studies. The funnel plot of studies that measured axial stiffness and posterior bending stiffness showed a low publication bias (Fig. 6).

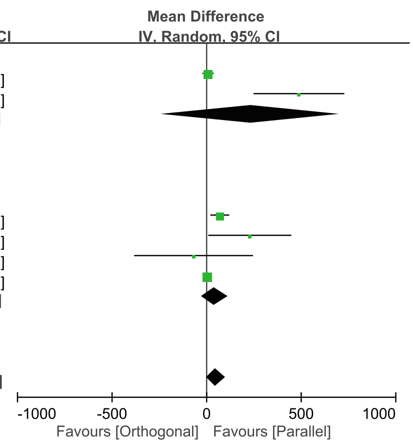
a

Study or Subgroup	Parallel Mean	Parallel SD	Parallel Total	Orthogonal Mean	Orthogonal SD	Orthogonal Total	Weight	Mean Difference IV, Random, 95% CI
2.1.1 Intercondylar model (N/mm)								
Atalar 2017	473.5	94	8	627.4	487.7	8	0.2%	-153.90 [-498.07, 190.27]
Got 2012	269.32	34.16	10	278.81	37.41	10	11.0%	-9.49 [-40.89, 21.91]
Penzkofer 2010	115.6	8.6	6	201.7	19	6	15.6%	-86.10 [-102.79, -69.41]
Schwartz 2006	37.8	3.2	5	42.1	4.8	5	18.3%	-4.30 [-9.36, 0.76]
Zalavras 2011	105.6	24.5	14	98.7	24	14	15.2%	6.90 [-11.07, 24.87]
Subtotal (95% CI)			**43**			**43**	**60.2%**	**-25.15 [-65.50, 15.20]**
Heterogeneity: Tau² = 1617.79; Chi² = 88.84, df = 4 (P < 0.00001); I² = 95%								
Test for overall effect: Z = 1.22 (P = 0.22)								
2.1.2 Supracondylar model (N/mm)								
Arnander 2008	214.9	43.3	8	138.3	44.6	8	8.1%	76.60 [33.53, 119.67]
Caravaggi 2014	101	33	7	108	43	7	8.7%	-7.00 [-47.15, 33.15]
Kollias 2010	169	72.1	8	145.4	82.5	8	3.7%	23.60 [-52.32, 99.52]
Koonce 2012	1,051.6	243.7	10	935.9	164.8	10	0.8%	115.70 [-66.64, 298.04]
Taylor 2016	13.8	2.1	9	9.7	1.3	9	18.5%	4.10 [2.49, 5.71]
Subtotal (95% CI)			**42**			**42**	**39.8%**	**24.34 [-10.04, 58.71]**
Heterogeneity: Tau² = 847.67; Chi² = 12.85, df = 4 (P = 0.01); I² = 69%								
Test for overall effect: Z = 1.39 (P = 0.17)								
Total (95% CI)			**85**			**85**	**100.0%**	**-6.41 [-22.71, 9.88]**
Heterogeneity: Tau² = 372.02; Chi² = 133.91, df = 9 (P < 0.00001); I² = 93%								
Test for overall effect: Z = 0.77 (P = 0.44)								
Test for subgroup differences: Chi² = 3.35, df = 1 (P = 0.07), I² = 70.1%								



b

Study or Subgroup	Parallel Mean	Parallel SD	Parallel Total	Orthogonal Mean	Orthogonal SD	Orthogonal Total	Weight	Mean Difference IV, Random, 95% CI
5.1.1 Intercondylar model (N)								
Atalar 2017	379.7	35.7	8	372.4	26.5	8	30.1%	7.30 [-23.51, 38.11]
Zalavras 2011	1,287.8	428.9	14	800	161.5	14	3.8%	487.80 [247.73, 727.87]
Subtotal (95% CI)			**22**			**22**	**33.9%**	**232.20 [-237.72, 702.11]**
Heterogeneity: Tau² = 107815.24; Chi² = 15.14, df = 1 (P < 0.0001); I² = 93%								
Test for overall effect: Z = 0.97 (P = 0.33)								
5.1.2 Supracondylar model (N)								
Arnander 2008	304.4	63.5	8	234.9	34.4	8	25.3%	69.50 [19.46, 119.54]
Caravaggi 2014	781	260	7	553	142	7	4.5%	228.00 [8.54, 447.46]
Koonce 2012	1,548.5	422.4	10	1,617.6	280.5	10	2.4%	-69.10 [-383.37, 245.17]
Taylor 2016	10.5	1.9	9	8.4	1.2	9	33.9%	2.10 [0.63, 3.57]
Subtotal (95% CI)			**34**			**34**	**66.1%**	**43.43 [-24.96, 111.82]**
Heterogeneity: Tau² = 2491.52; Chi² = 11.23, df = 3 (P = 0.01); I² = 73%								
Test for overall effect: Z = 1.24 (P = 0.21)								
Total (95% CI)			**56**			**56**	**100.0%**	**47.89 [-2.10, 97.88]**
Heterogeneity: Tau² = 1917.03; Chi² = 27.05, df = 5 (P < 0.0001); I² = 82%								
Test for overall effect: Z = 1.88 (P = 0.06)								
Test for subgroup differences: Chi² = 0.61, df = 1 (P = 0.44), I² = 0%								



c

Study or Subgroup	Parallel Mean	Parallel SD	Parallel Total	Orthogonal Mean	Orthogonal SD	Orthogonal Total	Weight	Mean Difference IV, Random, 95% CI
2.2.1 Intercondylar model (N/mm)								
Atalar 2017	545.5	198.7	8	516.5	253.5	8	1.1%	29.00 [-194.20, 252.20]
Subtotal (95% CI)			**8**			**8**	**1.1%**	**29.00 [-194.20, 252.20]**
Heterogeneity: Not applicable								
Test for overall effect: Z = 0.25 (P = 0.80)								
2.2.2 Supracondylar model (N/mm)								
Caravaggi 2014	102	44	7	81	26	7	25.3%	21.00 [-16.86, 58.86]
Koonce 2012	1,236.9	283.8	10	994.9	311.1	10	0.8%	242.00 [-19.00, 503.00]
Taylor 2016	12.2	1.8	9	9	1.5	9	72.8%	3.20 [1.67, 4.73]
Subtotal (95% CI)			**26**			**26**	**98.9%**	**12.35 [-18.52, 43.22]**
Heterogeneity: Tau² = 377.52; Chi² = 4.06, df = 2 (P = 0.13); I² = 51%								
Test for overall effect: Z = 0.78 (P = 0.43)								
Total (95% CI)			**34**			**34**	**100.0%**	**9.92 [-13.66, 33.50]**
Heterogeneity: Tau² = 198.28; Chi² = 4.11, df = 3 (P = 0.25); I² = 27%								
Test for overall effect: Z = 0.82 (P = 0.41)								
Test for subgroup differences: Chi² = 0.02, df = 1 (P = 0.88), I² = 0%								

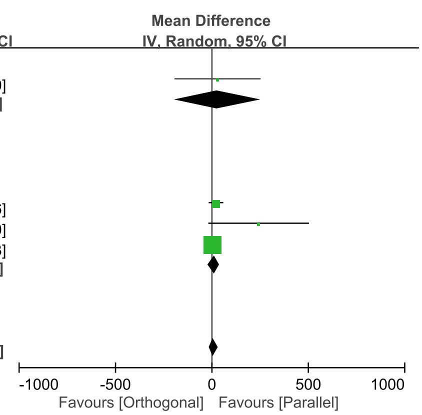


Fig. 3 Meta-analysis on construct sagittal strength: (**a**) posterior bending stiffness, (**b**) posterior bending load to failure, and (**c**) anterior bending stiffness

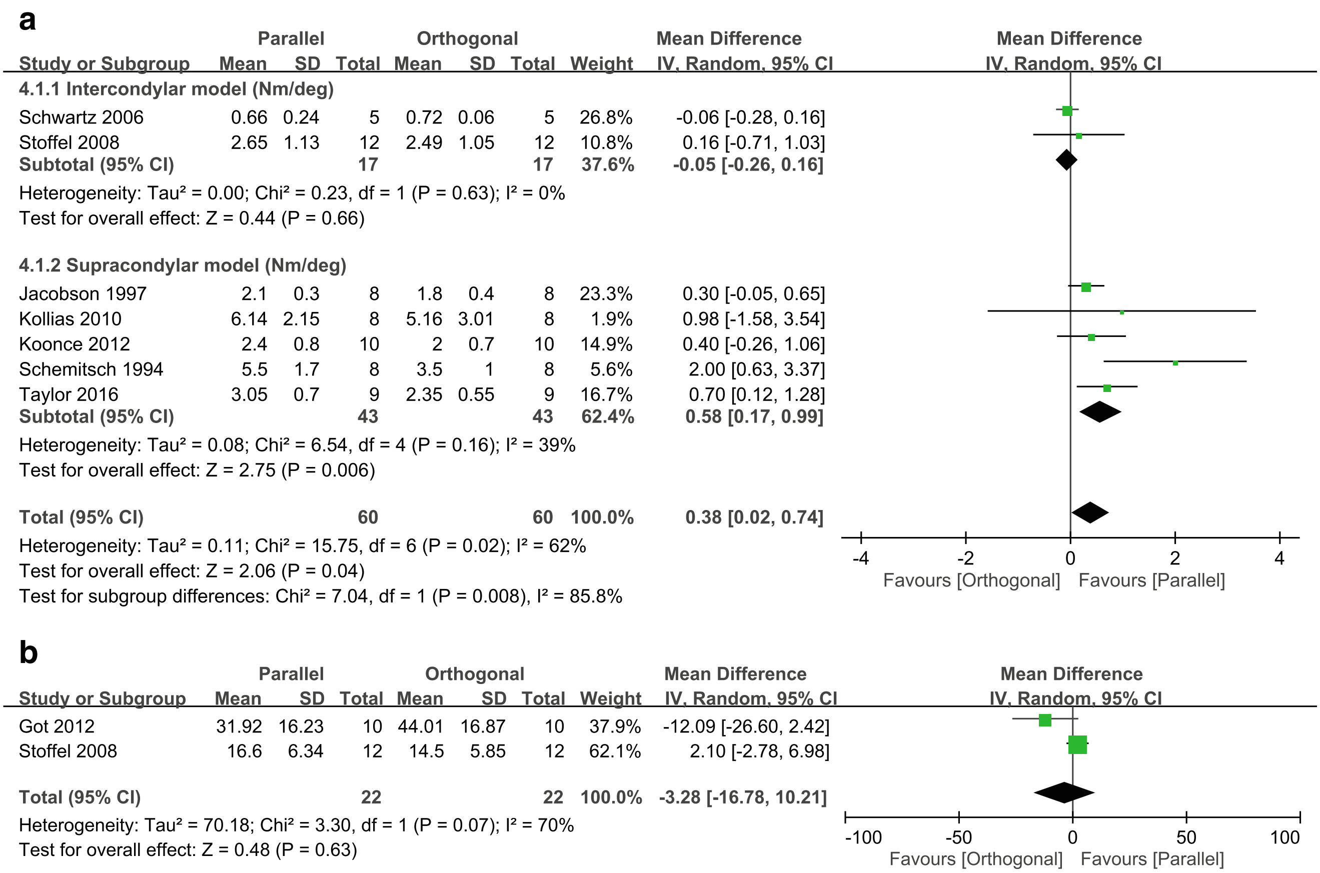


Fig. 4 Meta-analysis on construct torsional strength: (**a**) torsional stiffness and (**b**) torque to failure

Discussion

This is the first meta-analysis to evaluate the biomechanical properties of parallel and orthogonal constructs. Parallel constructs have significant higher axial stiffness than the orthogonal ones. In a subgroup analysis, parallel constructs have higher axial stiffness for both supracondylar and intercondylar fixation. Torsional stiffness is only greater in supracondylar fractures than in intercondylar fractures. The characteristics and testing protocols of the included biomechanical studies are summarized in Table 1. The rationale for both construct fixation techniques in the literature is summarized in Table 2.

Osteosynthesis with a double plate construct is the most accepted method for DHF fixation [12]. An orthogonal configuration, originally proposed by the AO group, allows both sagittal and coronal DHF fixation, forming a girder-like arrangement [10]. Early biomechanical studies recommended the use of orthogonal double-plate fixation. However, the complication rate of orthogonal constructs may be as high as 35% and includes implant failure or malunion in osteoporotic bone or poor patient compliance [32]. Later proposed by Dr. O'Driscoll, a parallel plate configuration permits the medial and lateral column distal screws to interdigitate, creating an arch-like structure [25, 26]. In addition to direct intercondylar fracture fixation, parallel constructs permit the maintenance of a compression force in the supracondylar region, facilitate anatomical reduction, and strengthen distal humerus reconstruction [26]. An increasing number of studies have showed that parallel constructs provide better biomechanical stability than the orthogonal ones [2, 7, 33, 38]. This biomechanical advantage to restore distal humerus anatomy also yielded good clinical outcomes [40]. Thus, in the clinical study, some authors suggested the use of parallel constructs to treat comminuted DHFs [3]. Moreover, a recent clinical meta-analysis showed that parallel configuration had significantly lower complication rate than the orthogonal ones in C-type DHFs [29].

This meta-analysis showed that axial stiffness, in consistent with many previous studies [7, 32, 33, 38], was greater in parallel construct, while sagittal stiffness was not [1, 2, 8, 35]. Plate placement at different anatomical positions could affect greatly on construct stability [33]. In axial loading, the second moment of inertia is lower in posterolateral plate of orthogonal construct than in medial or lateral plate, resulting in lower stiffness values in orthogonal configuration [33]. However, flexion moment arm is larger in sagittal bending, and it acts differently in axial loading mode, therefore, generating different results [33, 38]. Unlike flexion or extension,

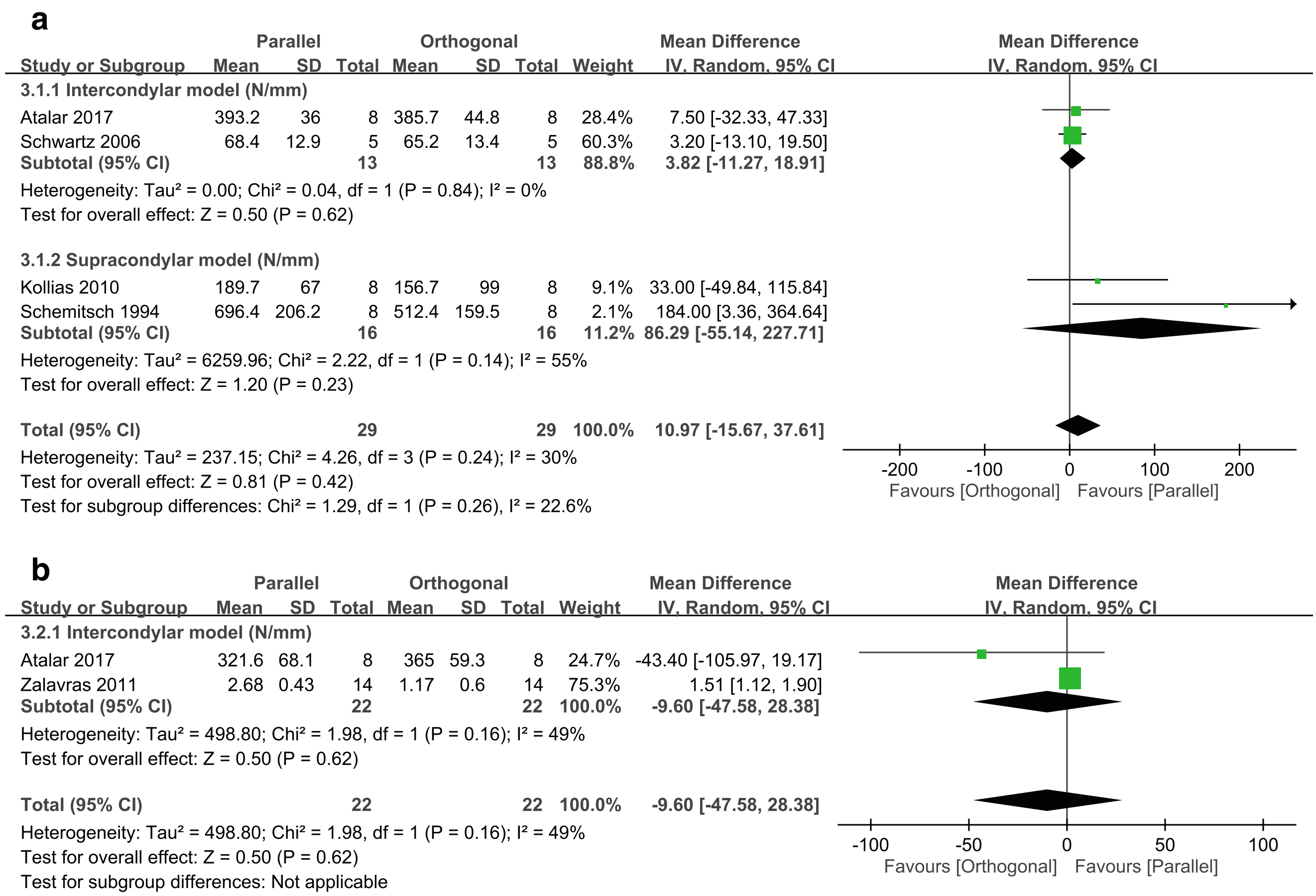


Fig. 5 Meta-analysis on construct coronal strength: (**a**) lateral bending stiffness and (**b**) medial bending stiffness

elbow coronal bending force mainly derives from unintended load application [38]. The coronal bending stiffness showed no significant difference in this meta-analysis and other biomechanical studies [1, 22, 24, 30]. Regarding torsional stiffness, parallel construct is significantly higher in this study and others [22, 39]. The reason could be explained by that parallel configuration had greater moment arm, thus, sustaining more resistance to torsional loads [30]. These results suggest that parallel configuration, especially in axial and torsional loading, may provide greater biomechanical stability to allow early elbow ROM during rehabilitation.

The type of plates was also different between the included biomechanical studies. Some used locking plates [7, 30, 32, 35, 38] while others used non-locking plates [10, 14, 15, 22,

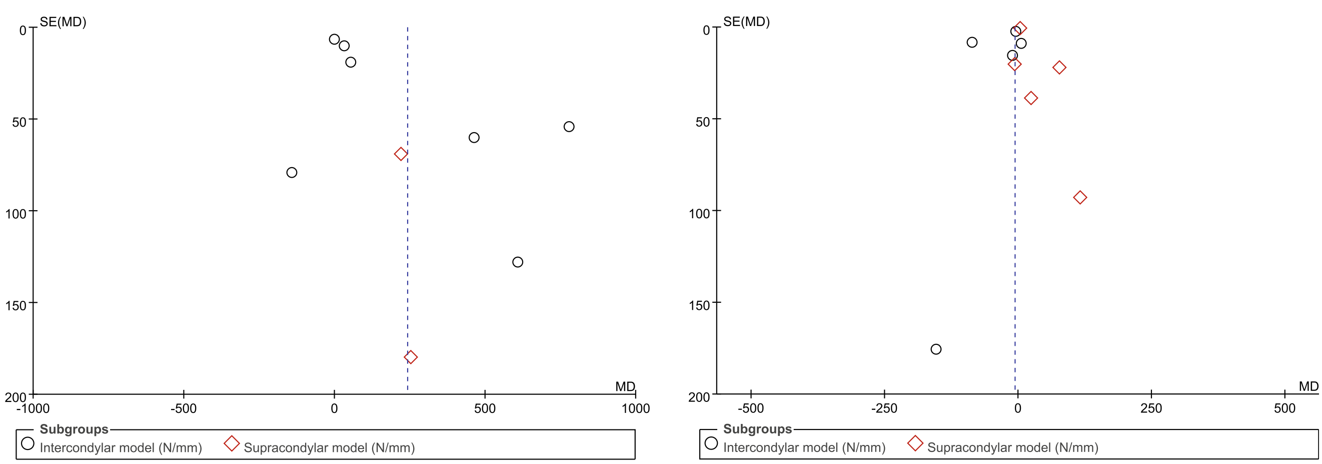


Fig. 6 Funnel plot comparing (**a**) axial stiffness comparison and (**b**) posterior bending stiffness comparison of the parallel and orthogonal constructs

Table 2 Literature on adult biomechanical distal humerus fixation in parallel and orthogonal systems

Type of fixation	Rationale
Parallel plate fixation	(1) Advantages: reconstruct structure of both columns in distal humerus with distal screw interdigitation and anchor both columns to the shaft of the humerus, forming the arch keystone (2) Disadvantages: require extensive soft tissue dissection posteriorly and medially; lateral plate skin irritation not uncommon
Orthogonal plate fixation	(1) Advantages: provide resistance to the force during elbow flexion arc and form a strong "girder-like arrangement"; easier posterolateral plate placement than lateral plate placement with less muscle and ligament attachment (2) Disadvantage: less resistance to varus loading and to the compression force between metaphysis and shaft

24, 37]. Few studies compared the biomechanical strength of locking versus non-locking double plate constructs. Korner et al. [12] showed no significant differences in axial, sagittal, and torsional stiffness between locking and non-locking plates in either a posterior two-plate construct or an orthogonal plate construct for supracondylar fracture fixation. Koonce et al. [9] also showed that there was no difference in the axial, sagittal, and torsional stiffness of orthogonal locking plates compared with the non-locking plates. It is therefore still unclear whether locking plates provide better biomechanical properties over non-locking plates in DHF fixation. A subgroup analysis was also not performed because of inadequate data collection in this study.

Clinically, the need for hardware removal after surgical fixation may be a potential concern. Two randomized controlled studies compared hardware removal rates after surgical treatment for C-type DHFs. The reported incidence rates were 29.4% (5/17) [28] and 25.0% (8/32) [4] in orthogonal constructs and 33.3% (6/18) [28] and 34.2% (13/38) [4] in the parallel ones. The incidence rates had no significant difference between groups in both studies ($p = 0.803$ [28] and $p = 0.402$ [4]). However, the reasons for hardware removal, which may be prominence of either olecranon implants or humeral ones, were not provided separately to make subgroup analysis [4, 28].

Limitations

There are some limitations to this study. First, heterogeneity exists in the included studies. However, despite the various biomechanical factors that may lead to heterogeneity, the direction of mechanical loading in each study is similar. We also performed a subgroup analysis between supracondylar and intercondylar fixation to further evaluate these factors.

Second, the sample size in the included biomechanical studies is generally small; however, the differences between the two groups were statistically significant. This is also the first and largest study that compares the biomechanical properties of parallel and orthogonal constructs.

Unlike clinical studies, biomechanical studies may fail to analyze clinical and soft tissue outcomes. However, a biomechanical meta-analysis based on similar biomechanical protocols and testing can provide more consistent results with fewer errors, allowing surgeons to pay more attention to construct stiffness, failure load, and stability.

Conclusion

In conclusion, this meta-analysis shows that parallel constructs provide greater axial stiffness, axial strength, and torsional stiffness than the orthogonal plate for DHF fixation. A subgroup analysis revealed that parallel constructs have better torsional stiffness in supracondylar fracture fixation.

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Author contribution CAS, WCL, WRS, and TWT: study conception and design

CAS and WCL: acquisition of data

CAS, WRS, WCL, TWT: analysis and interpretation of data

CAS and TWT: drafting of the article

All authors have read and approved the final submitted manuscript.

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Compliance with ethical standards

Conflict of interest None.

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