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Biomechanical evaluation of an intramedullary clavicle screw in simple oblique and butterfly wedge fractures

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Abstract

This biomechanical study evaluates the performance of a solid titanium-alloy intra-medullary (*IM*) clavicular screw in torsion and cantilever bending in cadaveric clavicle specimens with simulated simple oblique and butterfly wedge midshaft fractures. Thirty-two fresh-frozen male clavicles were sorted into six experimental groups: Torsion Control, Torsion Simple Oblique Fracture, Torsion Butterfly Wedge Fracture, Bending Control, Bending Simple Oblique Fracture, and Bending Butterfly Wedge Fracture. The experimental groups were controlled for density, length, diameter, and laterality. All other samples were osteotomy-induced and implanted with a single 90 mm × 3 mm clavicle screw. All groups were tested to physiologically relevant cutoff points in torsion or bending. There were no statistically significant differences in the performance of the oblique and butterfly wedge fracture models for any torsion or bend testing measures, including maximum torsional resistance ($p = 0.66$), torsional stiffness ($p = 0.51$), maximum bending moment ($p = 0.43$), or bending stiffness ($p = 0.73$). Torsional testing of samples in the direction of thread tightening tended to be stronger than samples tested in loosening, with all groups either approaching or achieving statistical significance. There were no significant differences between the simple oblique or the butterfly-wedge fracture groups for any of the tested parameters, suggesting that there is no difference in the gross biomechanical properties of the bone-implant construct when the *IM* clavicle screw is used in either a simple midshaft fracture pattern or a more complex butterfly wedge fracture pattern.

Keywords

Biomechanical testing/analysis, orthopedic materials, biomaterials stress analysis, medical biomaterials [see slso biomaterials], musculo-skeletal mechanics

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Introduction

The clavicle is the most commonly fractured bone in the body and the most common location of fracture is the middle third, termed a midshaft fracture.[1] While nondisplaced and minimally displaced fractures are routinely treated nonoperatively, operative fixation has demonstrated improved functional outcomes and a lower rate of nonunion and malunion compared to a nonoperative group in a randomized clinical trial.[2] Many fixation options exist, including different forms of plate and screw constructs as well as various types of intramedullary (*IM*) fixation.[3–6] While plate fixation is currently the gold standard of operative treatment for clavicular fractures, *IM* fixation provides advantages of smaller operative incisions, reduced soft tissue dissection, and reduced implant profile, and is typically recommended for simple midshaft fractures.[7–9] One randomized control trial even found *IM* fixation to lead to shorter operative times, reduced infection rates, and reduced rates of hardware failure when compared to plate fixation.[9] As *IM* fixation has gained

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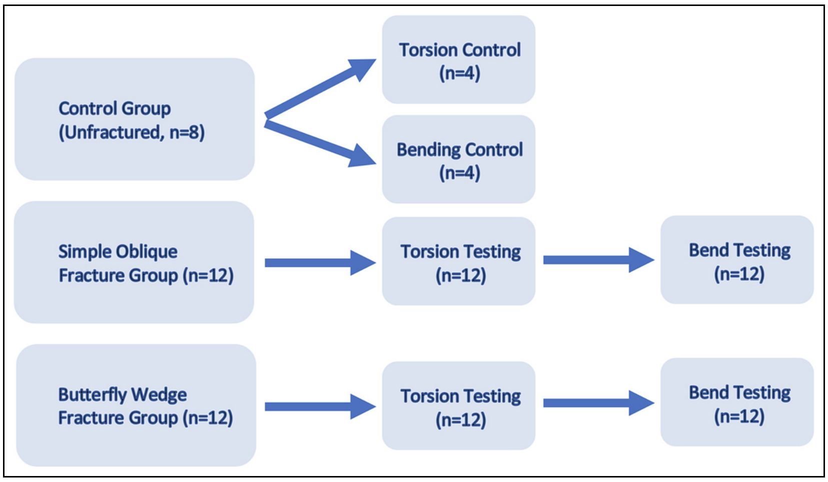

Figure 1. Flow-chart of experimental grouping and specimen testing.

popularity, newer *IM* implants have been developed to avoid complications associated with *IM* implants such as fracture shortening, lack of fracture compression, and implant prominence at the site of insertion.[7]

Some previous biomechanical studies have shown *IM* fixation of the clavicle performs comparably to plate fixation when tested in bending, while others have shown plate fixation to be superior in this regard.[10–15] However, all previous biomechanical studies acknowledge the particular susceptibility of *IM* clavicle fixation to weakness in torsion. As a single screw, it is relatively easy for the clavicle to rotate along the axis of the *IM* screw when torsional forces are applied.[10,14] Still, several clinical studies and randomized control trials have shown excellent clinical results following *IM* fixation, particularly in simple, mid-shaft fractures.[3,6,8,9,16]

Although *IM* fixation is currently only recommended for simple midshaft fracture patterns, certain clinical scenarios involving more complex fracture patterns may indicate the use of *IM* over plate fixation due to the previously mentioned clinical advantages of *IM* fixation. However, there is a lack of biomechanical data validating the performance of the implant in more complex fracture patterns.

The purpose of this study was to perform cadaveric biomechanical testing of a titanium-alloy, solid, dual-threaded clavicle screw in torsion and bending in two common fracture patterns, with the hypothesis that *IM* fixation would achieve comparable biomechanical fixation strength between a simple midshaft oblique fracture pattern and the more complex comminuted butterfly-wedge fracture pattern.

Materials and methods

Thirty-two fresh-frozen male cadaveric clavicles from 16 donors were visually and radiographically inspected for obvious osseous abnormalities. After recording the length, diameter, age, and laterality of each specimen, a Bruker Skyscan 1176 MicroCT was used to image the clavicles (Bruker, Billerica, MA). CT-Analyzer ("CTAn") software was used to analyze the reconstructed images to calculate the bone mineral density of each sample in Hounsfield Units (*HU*). The clavicles were then sorted into experimental groups and matched for bone mineral density, length, midshaft diameter, and laterality. One-way ANOVA analysis and Fisher's Exact Test were used to ensure that there were no statistically significant differences between groups for any of the controlled variables. The samples were divided into the following experimental groups: (1) Unfractured control group ($n = 8$), (2) Simple midshaft oblique fracture ($n = 12$), and (3) Midshaft midshaft butterfly wedge fracture ($n = 12$). A flow-chart depicting the experimental grouping and subsequent testing can be found in Figure 1.

For the midshaft simple oblique fracture groups, a custom machined device with a custom 45° cutting guide was used to secure the clavicle in the correct orientation as well as guide the reciprocating-saw to create the oblique fractures (Figure 2). For the midshaft butterfly wedge fracture groups, the same custom device with a separate custom insert was used to trace a wedge shape onto the anterior surface of the clavicle. A 27-mm wedge was then cut into the anterior-inferior midshaft location in order to mimic fractures seen by the authors in clinical examples. The wedge of the butterfly fragment reached an apex at 50% shaft thickness in each of the clavicles, with the apex pointing towards the posterior surface of the clavicle. The fracture patterns selected for the study were chosen to represent AO/ASIF simple oblique (B1.2), and bending wedge (B2.2) fracture patterns (Figure 3(a) and (b)). In order to standardize the medio-lateral location of the fracture for samples of differing length, all fractures were centered at 60% of clavicle length from the medial end. All fracture cuts were performed with a reciprocating orthopedic saw. The control group clavicles were left intact.

Both groups of fractured clavicles were then implanted each with a single 90 mm × 3 mm clavicle screw using corresponding implantation devices and techniques, with the screw pointing medially (Figure 3(c) and (d)). A custom-machined device was

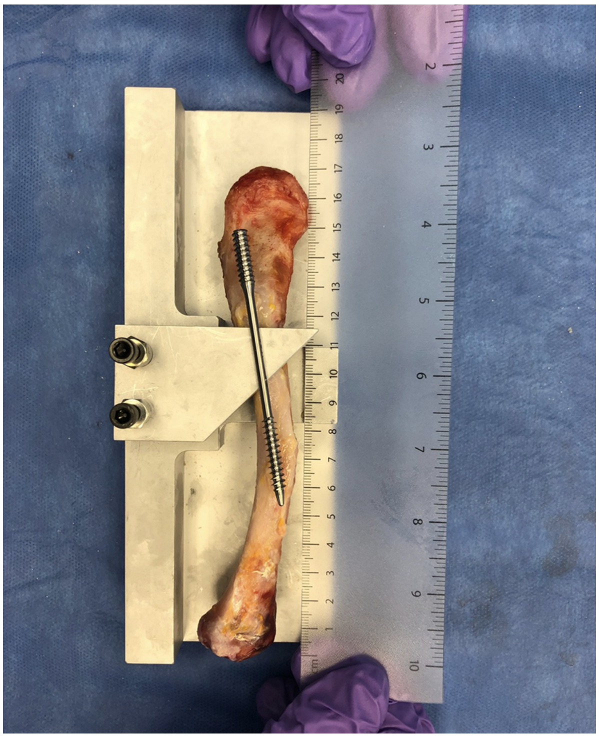

Figure 2. The custom-machined device seen here was used to hold the clavicle in place as well as guide the fracture cut. The angled plate resting superiorly on the clavicle created a guide to make consistent 45° cuts. This setup was used to create all midshaft oblique fractures for the study.

used to hold the fractured clavicles steady in a fully reduced and aligned position during implantation. The control samples were un-implanted. All fractures and implantations were performed by a single Shoulder fellowship-trained attending surgeon (SAP) or Orthopedic Sports Medicine Fellow (JC).

The medial and lateral 3 cm of each sample was potted in custom-machined aluminum tubes using Goldenwest R1 Fast Cast #891 Resin (Goldenwest Manufacturing, Grass Valley, CA). A drill press was used to align the samples along the mechanical axis of the bone within the aluminum tubes as the samples were potted. The clavicles were then allowed to set in plastic tubes to ensure straightness of the aluminum tubes over the clavicles.

The mechanical axis of the clavicle was defined as the line that extends from the central medial articular surface to the central lateral articular surface as defined in previous studies.[18]

All mechanical testing was conducted on an Instron Hydraulic Mechanical Testing Frame collecting data at 500 Hz (Model 8874, Instron, Norwood, MA). Due to the fragility of the samples in torsion testing, and relative non-destructive loading mode, all implanted groups were tested first in torsion and then in bending, as there was no permanent deformation of the bone-implant construct for any of the samples following torsion testing. With only a small sample size of 32 specimens available, this allowed for the inclusion of 12 samples in these experimental groups instead of 6, resulting in greater statistical power for all analyses.

For torsion testing, the specimens were oriented vertically, with the medial end of the clavicle held stationary in a grip inferior to the lateral end where torsion was applied (Figure 4(a)). Torsion was applied in the clockwise direction for left clavicles and in the counterclockwise direction for the right clavicles to simulate shoulder flexion at a rate of 0.5°/s as seen in previous studies.[17] The control group samples were first tested to fracture in order to establish a physiologically relevant cutoff point for the implanted samples. Testing was stopped for the control specimen after they structurally yielded or there was a significant decrease in measured torsional load. Maximum torsional resistance (Nm) was calculated directly from output from the Instron, while torsional stiffness was calculated from the line of best fit applied to the torsion (Nm) versus rotation (°) curve collected from each sample (for individual specimen graphs, see Supplemental Appendix Figure 1(a)).

Since the implanted samples were pre-fractured, they did not have an obvious biomechanical fracture point when mechanically tested, therefore a physiologically relevant cutoff point established by the control specimens was necessary. The average angle of rotation (°) at failure for the control specimens was used as the physiologically relevant cutoff point for the implanted samples. In the judgment of the authors, rotation past this value was taken to represent unacceptably significant disruption of the clavicle and surrounding anatomy, and often was observed to cause significant plastic deformation of the implants. Testing of the implanted torsion samples was performed identically to the unfractured samples, except that testing was stopped shortly after the cutoff point was reached.

For bend testing, the samples were oriented horizontally, with the medial end held stable in a grip while load was applied to the lateral end (Figure 4(b)). Load was applied in a superior-to-inferior direction at a rate of 1 mm/s to simulate physiological loading conditions as seen in previous studies.[11] Load and displacement data were collected directly from the Instron. Maximum bending moment was then calculated using the lengths of the individual specimen. Bending stiffness was calculated from the line of best fit applied to the bending Moment (Nm) versus displacement (mm) curve (for individual specimen graphs see Supplemental Appendix Figure 1(b)). Similar, to the torsional testing, the bending control specimens were tested first in order to establish a physiologically relevant cutoff point for the testing of the implanted specimens. Failure for the control samples was defined as fracture of the intact bone or a significant decrease in resistance to bending. The average lateral displacement of the control specimens at failure was used as the failure point for the implanted specimens. Testing of the implanted samples was performed identically to the unfractured control group. Testing was stopped for the implanted samples

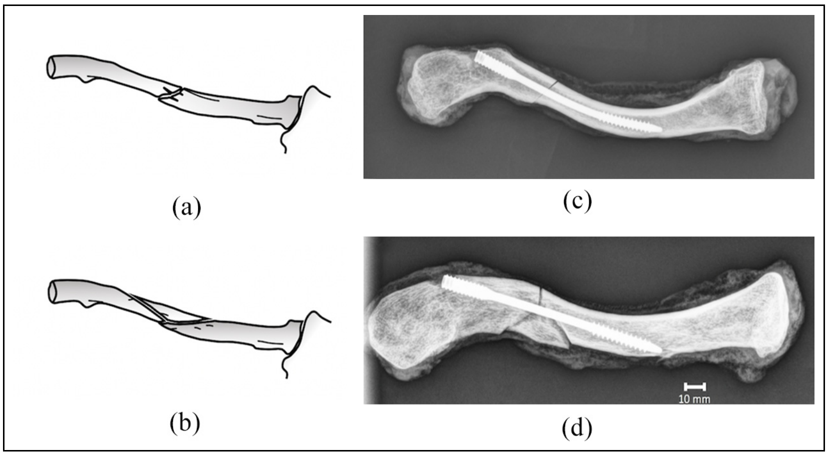

Figure 3. (a and b): AO/ASIF classification of midshaft clavicle fractures used in this study: (a) oblique (b1.2), and (b) bending wedge (b2.2); (c and d): *X*-ray images of implanted samples demonstrating fracture patterns used in the study as well as the implanted screws.

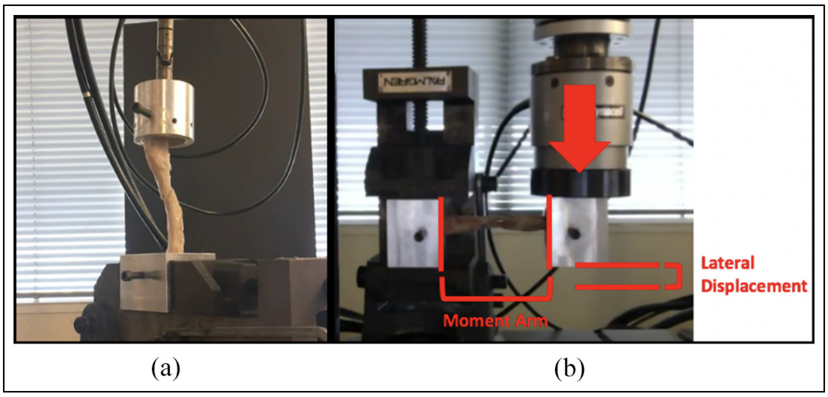

Figure 4. (a) Image of torsion testing setup. The potted samples were held inferiorly by a vice grip and superiorly by a fixture directly connected to the Instron mechanical testing device and (b) image of bend testing setup with relevant biomechanical parameters defined. The potted samples were held medially by a vice grip as the Instron applied force to the lateral end in the superior-to-inferior direction.

shortly after the specimens fractured, or the specimen reached the established cutoff point.

The three torsion experimental groups were compared using two-tailed, independent sample, equal variance *t*-tests. The three bending experimental groups were compared identically. A bonferroni correction was applied for both groups, resulting in a value of $p < 0.0167$ indicating significant difference between groups.

Finally, the torsion data was re-analyzed in order to determine the effect of laterality on torsional strength, since torsional strength of threaded devices can depend on the direction of applied torque. Two-tailed, independent sample, equal variance *t*-tests were again used to determine significance, with $p < 0.05$ indicating significance. All statistical analysis for the study was performed using SPSS Statistics Software, Version 24 (IBM, Armonk, NY, USA).

Results

One-Way ANOVA analysis between the three torsion groups showed no significant differences in bone mineral density ($p = 0.98$), length ($p = 0.72$), or midshaft diameter ($p = 0.99$), while Fisher's Exact Test showed no difference in the proportion of left and right samples between groups ($p = 0.88$). Identical testing for the bending groups showed no difference in bone mineral density ($p = 0.99$), length ($p = 0.36$), midshaft diameter ($p = 0.95$), or laterality ($p = 0.20$). This data is summarized in Table 1.

For the four control specimens tested in torsion, the average load at failure was 14.9 ± 3.4 Nm and occurred at an average of 22.9 ± 6.9 degrees of rotation. This rotation value was taken as the physiologically relevant cut-off point for the testing of the implanted samples. The oblique and butterfly fracture groups performed similarly to each other in both maximum torsional resistance ($p = 0.66$) as well as torsional stiffness ($p = 0.51$), differing by only 6.8% and 13% in these values on average, respectively (Table 2). However, both fracture groups yielded values lower than those of the control group in regard to both maximum torsional resistance and torsional stiffness (Table 2). The simple oblique fracture specimen showed only 20% of the maximum torsional resistance as the control specimens

Table 1. Experimental group characteristics and controlled variables.

	Control ($n = 4$)	Oblique ($n = 12$)	Butterfly ($n = 12$)	p Value
Torsion testing groups				
Density (HU)	3465 ± 103	3451 ± 114	3460 ± 175	0.98
Length (mm)	162 ± 6.3	165 ± 7.1	166 ± 6.9	0.72
Midshaft diameter (mm)	13.3 ± 2.8	13.1 ± 1.7	13.2 ± 2.3	0.99
Laterality (# right:# left)	1:3	5:7	6:6	0.88
Bend testing groups				
Density (HU)	3457 ± 270	3451 ± 114	3460 ± 175	0.99
Length (mm)	160 ± 7.2	164 ± 7.1	166 ± 6.9	0.36
Midshaft Diameter (mm)	13.5 ± 1.7	13.1 ± 1.7	13.2 ± 2.3	0.95
Laterality (# right:# left)	4:0	5:7	6:6	0.20

HU: Hounsfield units; mm: millimeters.

Table 2. Biomechanical testing data and statistical analysis.

Torsion testing				
Experimental group	Average rotation at failure (°)	Maximum torsional resistance (Nm)		Torsional stiffness (Nm/°)
Control	22.9 ± 6.9	14.9 ± 3.4		0.47 ± .14
Oblique	–	3.02 ± 1.5		0.14 ± .07
Butterfly	–	3.24 ± .96		0.16 ± .06
Bend testing				
Experimental group	Average displacement at failure (mm)	Maximum bending moment (Nm)		Bending stiffness (Nm/mm)
Control	20.5 ± 3.9	15.6 ± 2.1		0.85 ± .16
Oblique	–	6.92 ± 1.5		0.35 ± .08
Butterfly	–	6.39 ± 1.7		0.33 ± .11
Pairwise statistical analysis				
	Maximum torsional resistance (p value)	Torsional stiffness (p value)	Maximum bending moment (p value)	Bending stiffness (p value)
Control versus oblique	0.005*	0.015*	0.001*	0.005*
Control versus butterfly	0.006*	0.020	0.001*	0.003*
Oblique versus butterfly	0.658	0.507	0.433	0.725

Nm: Newton-meter; mm: millimeter.

*Indicates statistical significance of $p < 0.016$ with Bonferroni correction factor applied.

($p = 0.005$), while the butterfly fracture samples showed 22% of the maximum torsional resistance as the control specimens ($p = 0.006$). The oblique fracture specimen showed only 30% of the torsional stiffness as the control specimens ($p = 0.015$), while the butterfly fracture samples were 34% as stiff in torsion as the control specimens ($p = 0.020$). With significance set at $p < 0.0167$, all fractured torsion group differences were statistically significant when compared to controls except for control versus butterfly wedge (Table 2).

For the four control specimens tested in bending, the average load at failure was 15.6 ± 2.1 Nm and occurred at an average of 20.5 ± 3.9 mm displacement. This displacement value was taken as the physiologically relevant cut-off point for the testing of the implanted samples. Mechanical testing data for all bend testing trials demonstrated that the oblique and butterfly fracture groups yielded similar values for both maximum bending moment ($p = 0.43$) as well as bending stiffness ($p = 0.73$), differing by only 7.7% and 5.7% in these values on average, respectively (Table 2). However, all fracture groups yielded values significantly lower than those of the intact control group (Table 2). The simple oblique group showed only 44% of the maximum bending moment as the control samples ($p = 0.001$) and only 41% of the bending stiffness ($p = 0.005$). The butterfly wedge group showed only 41% of the maximum bending moment of the control group ($p = 0.001$) and only 39% of the bending stiffness ($p = 0.003$). With significance set at $p < 0.0167$, all fractured bending group differences were statistically significant when compared to controls (Table 2).

Analysis of the effect of laterality on torsional strength for simple oblique and butterfly wedge fracture groups tested for both maximum torsional resistance and torsional stiffness revealed that left clavicles tended to be

Table 3. Torsion data comparison between left and right clavicle specimens.

	Fracture pattern	Left samples	Right samples	p Value
Maximum torsional resistance (Nm)	Oblique	3.62 ($n = 7$)	2.17 ($n = 5$)	0.092
	Butterfly	3.85 ($n = 6$)	2.63 ($n = 6$)	0.019*
Torsional stiffness (Nm/°)	Oblique	0.18 ($n = 7$)	0.10 ($n = 5$)	0.056
	Butterfly	0.20 ($n = 6$)	0.12 ($n = 6$)	0.257

*Indicates Statistical Significance ($p < 0.05$).

stronger in both measures when compared to right clavicles. However, the only experimental group to reach significance was the butterfly wedge fracture group tested for maximum torsional resistance ($p = 0.019$). All other groups only approached significance (Table 3).

Discussion

While several previous biomechanical and clinical studies have evaluated the performance of *IM* fixation compared to plate fixation, no previous study has evaluated *IM* fixation in differing fracture patterns. Many types of *IM* fixation have been utilized in treating midshaft clavicle fractures in recent decades, including Rockwood or Hagie pins, titanium elastic nails, screws, and newer generation clavicle-specific *IM* fixation devices which have varied diameters, material compositions, sizes, and complication profiles.[7] Additionally, several biomechanical studies evaluating clavicle fractures and their fixation devices are available. However, there is no standard and validated biomechanical model for testing clavicle fractures. The two dominant methods of biomechanical evaluation are three or four point bending and cantilever bending with some studies also including torsional and axial load testing. We chose torsion and cantilever bending as the testing methods in this study, as we consider these values to be the most relevant and accurate physiologic representation of the forces acting on the clavicle.

The results of the study showed that the two fracture patterns tested comparably in all testing scenarios, with no statistically significant difference between the two fracture groups in any of the experimental groups (Table 2). These results suggest that simple midshaft fracture patterns and more complex midshaft fracture patterns that have undergone *IM* fixation behave comparably under common and physiologically relevant biomechanical testing scenarios. However, there are several other variables other than biomechanical stability to take into consideration when deciding which implant to use in clinical scenarios, and further study will be necessary to determine both the biomechanical and clinical performance of *IM* fixation compared to plate fixation in more complex fracture patterns.

Although each experimental fracture group tested comparably, all fracture groups tested inferiorly to the unfractured control specimens with the exception of the butterfly fracture versus control in torsional stiffness, which closely approached significance (Table 2). This was to be expected, as a previous biomechanical study found that neither plate nor screw fixation achieves the same strength as intact bone.[11] The fractured groups were also particularly weak in torsion compared to intact specimens. This was also to be expected, as previous biomechanical studies have consistently shown *IM* fixation to be very weak in torsion compared to both plate fixation and intact bone.[10–15] This weakness of *IM* fixation must be recognized by surgeons and appropriate patient counseling and immobilization is necessary to minimize rotational forces on the clavicle in the early post-operative phase when *IM* fixation us used.

The results of this study also showed that the strength of *IM* clavicle fixation is at least somewhat dependent on the direction of applied torque when threaded devices such as the clavicle screw are used (Table 3). In this study, left clavicles were tested clockwise, and right clavicles were tested counterclockwise to simulate shoulder flexion. Since all screws were inserted facing medially, the left clavicles were tested in the direction of screw tightening, and the right clavicles were tested in the direction of loosening. This explains why left clavicles tended to have higher torsional strength in this study compared to right, with all groups either approaching or achieving significance in favor of left samples (Table 3). However, *IM* fixation is still significantly weaker in torsion no matter the direction of applied torque when compared to plate fixation, and whether the effect of laterality on *IM* fixation has any clinical significance will require further study.

Several previous biomechanical studies have evaluated the performance of *IM* fixation compared to the gold standard of plate fixation, with most showing that *IM* fixation comes with reduced biomechanical stability, particularly in torsion.[10–15] Despite this, clinical investigations have shown that *IM* fixation can still be used to achieve excellent clinical results when compared to plate fixation.[3,6,8,9,16] While this study did not include a plated group for biomechanical analysis and instead included an intact, un-implanted control group, several other studies have directly compared plate and *IM* fixation, and therefore a summary of these studies and comparison of results is particularly pertinent.

Golish et al. compared a 3.8 or 4.5 mm Rockwood pin with a 3.5 mm dynamic compression (DC) plate placed superiorly on the clavicle and found that the DC plating demonstrated less displacement at fixed loads and greater loads at fixed displacement compared to the IM screw. A vertical, midshaft osteotomy was used for their fracture model. Renfree et al. compared the Rockwood pin with a precontoured unicortical locking plate and a precontoured bicortical non-locking plate in a Sawbones model and assessed cantilever bending and found no difference between the plate constructs, however the Rockwood pin performed poorly in comparison with significantly inferior resistance to cantilever bending. Their fracture model was formed by creating the Sawbones in two separate pieces, simulating an oblique, non-comminuted fracture pattern. They also found that in terms of resistance to rotational torque, "the lateral bone fragment rotated on the intramedullary pin at a very low force" during cantilever bending resulting in poor resistance to cantilever bending. No other formal testing of rotational stiffness was performed.

Drosdowech et al. used fresh cadaveric clavicles to compare the 4.5 mm Rockwood pin with a 3.5 mm reconstruction plate, a 3.5 LCDC plate and a 3.5 locking compression plate. The testing model included cantilever bending and torsion with three different fracture patterns similar to our study including intact, transverse, and comminuted (1 cm triangular cortical defect inferiorly). Similar to Renfree et al, Drosdowech found that the Rockwood pin demonstrated significantly reduced resistance to rotational torque including 40% less than of that of both the intact and plated clavicle in the inferior cortical defect model, and less than 50% resistance in the transverse model. The Rockwood pin demonstrated reduced bending stiffness compared to the plate constructs with only 56% and 57% compared to an intact clavicle for the simple fracture and inferior cortical defect model, respectively. All plate groups were stronger than the intact clavicles. By comparison, our study found that the screw demonstrated between 0.14 and 0.16 Nm/° rotational stiffness in the transverse and comminuted fracture pattern, respectively, which is superior to that of the Rockwood screw.

Wilson et al. performed combined axial and rotational biomechanical testing of a second generation clavicle-specific IM device (Sonoma CRx, Sonoma Orthopedic Products, Inc, Santa Rosa, CA, USA) on fourth generation sawbones with a transverse fracture pattern. No cantilever bend testing was performed. The Sonoma CRx can be implanted with a flexible medial aspect that can later be "actuated" to cause it to become rigid once appropriately seated. This also causes talons to expand outward medially to increase the rotational purchase. In addition, a 2.7 mm interlocking screw placed in the lateral aspect provides a second point of rotational purchase. During testing the Sonoma demonstrated 0.51 Nm/° (60%) rotational stiffness in comparison to the superior locking plate construct (0.84 Nm/°). The 60% of rotational stiffness for the Sonoma is higher than the Rockwood pin (30% as measured in the Drosdowech study and 30%–34% as measured in this study. Coincidentally, the authors later reported on two early failures of this implant in clinical use, highlighting the need for proper patient and fracture selection.

There were several limitations to this study. As previously discussed, there was no plated group included in this study. We felt that with a limited number of cadaver specimens available and with the main focus of the study being to compare *IM* fixation in different fracture patterns, maximizing the size of each IM experimental group was a higher priority than including a plated group. Previous studies have elucidated the biomechanical advantages of plate and *IM* fixation in simple fracture patterns, but a direct biomechanical comparison of plate and *IM* fixation in more complex fracture patterns will require further study. Limitations of this study related to the use of cadaveric bone as opposed to biomechanically equivalent synthetic bone were sought to be mitigated through careful matching of bone density, length, midshaft diameter, and laterality of the cadaveric samples. Another limitation is that only two fracture patterns were included in this study. However, the authors feel that these two fractures represent two of the most commonly encountered patterns in clinical scenarios, but further study with other fracture patterns should be considered. Another limitation is that the fractured samples were not tested to complete biomechanical failure. Due to the lack of obvious failure point in the fractured specimens as well as the extreme and clinically impractical amount of deformation required to get to that point, the authors felt that it was appropriate to establish a clinically relevant cut-off point for the fractured samples. One final limitation is that the fractured samples were tested in torsion before again being tested in bending. This was done to increase statistical power by maximizing the number of samples in each group due to limited specimen availability. Loading and unloading of the samples in torsion did not demonstrate that any of the samples had been tested beyond the elastic range, nor that any load hysteresis had occurred. Therefore, the authors do not feel that this testing method had a significant impact on the experimental results.

Despite these limitations, this study was the first to biomechanically compare *IM* clavicle fixation in different fracture patterns. Further studies should look to biomechanically evaluate plate versus *IM* fixation in comminuted fracture patterns. If validated, further studies should also evaluate the clinical and functional outcomes of IM fixation in more complex fracture patterns.

Conclusions

There was no difference in the biomechanical performance of a clavicle screw in a simple oblique or a comminuted butterfly wedge fracture pattern when tested in physiologically relevant torsion and bending. With *IM* fixation biomechanically validated in a comminuted fracture pattern, further studies should look to biomechanically evaluate the performance of *IM* compared to plate fixation in more complex fracture patterns. If further validated by biomechanical studies, *IM* fixation of the clavicle in certain clinical scenarios could show to be advantageous in a wider variety of fracture patterns than what is currently indicated.

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Declaration of conflicting interests

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Supplemental material

Supplemental material for this article is available online.

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