Injury, Int. J. Care Injured 43 (2012) 921–925

Contents lists available at SciVerse ScienceDirect

Injury

journal homepage: www.elsevier.com/locate/injury

Biomechanical analysis of locked and non-locked plate fixation of the clavicle

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ARTICLE INFO

Article history:
Accepted 8 February 2012

Keywords:
Clavicle fracture
Osteosynthesis
Internal fixation
Biomechanics
Locking plate
Brachial plexus

ABSTRACT

Introduction: During plate fixation of clavicular fractures the brachial plexus and subclavian vessels are vulnerable to injury beneath the clavicle. Locking plate fixation allows for mono-cortical fixation, theoretically reducing the risk of injury to these structures. Biomechanical analysis of the performance of such fixation is limited, and this study was designed to explore this further as a treatment option in clavicle fractures.

Materials and methods: Fixation of fifteen simulated mid-shaft fractures was undertaken using a combination of mono-cortical locked, bicortical locked and bicortical non-locked plating methods in cadaveric clavicles. Samples were then tested via three-point bending to destruction, and the performance of each with respect to failure load, bending stress, bending stiffness and Young's modulus was then analysed. The influence of the number of cortices engaged and locking was also assessed.

Results: Clavicles fixed with monocortical locking plates displayed a significantly lower bending stress (12 ± 1 MPa) than both the bicortical locking (28 ± 3 MPa, $p = 0.015$) and non-locking specimens (24 ± 3 MPa, $p = 0.002$). Engaging two cortices with the fixation produced a significant increase in failure load (291 ± 28 N vs 138 ± 48 N, $p = 0.018$) and bending stress (26 ± 2 MPa vs 9.9 ± 3.5 MPa, $p = 0.002$) compared to single cortex fixation.

Discussion: The greatest influence upon the performance of the fixation was the number of cortices engaged, with bicortical fixation performing significantly better than mono-cortical. Whether or not the fixation device was a locking one did not have a significant bearing upon the performance.

Conclusion: This in vitro biomechanical analysis demonstrates that mono-cortical locked plating fails at significantly lower levels of load and stress than bicortical locked and non-locked plating in mid-shaft fractures of the clavicle, and caution would therefore be advised in its use as a fixation modality for these injuries.

Introduction

Fractures of the clavicle are a common injury representing 2.6% of all fractures and 44% of those in the shoulder girdle, with an overall incidence of 64 per 100,000 per year.[1,2] Of these, 69–81% are mid-shaft fractures.[1,3,4] In the majority of cases, they can be managed non-operatively with the expectation of the patient achieving a good outcome.[2,5–9] Certain factors, including displacement and comminution, are associated with a higher risk of symptomatic non-union, although not inevitably so.[9]

When fixation is indicated, the use of a plate would still appear to have the strongest weight of evidence in its favour.[2,10–13] This is not without risk as the brachial plexus and subclavian vessels lie immediately beneath the clavicle. They are vulnerable to injury, both at the time of fracture, and during operative fixation. Over penetration by drill bits or screws through the inferior cortex is one such method of producing neurovascular injury.[14,15]

The use of intramedullary fixation has been proposed as one method of avoiding neurovascular injury. Initially, K-wires were employed in this fashion but there have been serious consequences reported secondary to their migration.[16–20] Threaded pin devices avoid such problems, but still demonstrate complication rates in excess of 35%, including infection, re-fracture, delayed union and non-union.[21] More recently interest in percutaneous elastic nail fixation has increased, with the perceived advantages of less invasive surgery, and avoidance of the neurovascular structures. However, these have again been associated with high complication rates, and a need for open reduction in two thirds of cases.[22]

From a biomechanical perspective, locking plates are identical to external fixators in their action because the angular stability between plate and screws means that the plate does not need to be in contact with the bone.[23] This creates the option to use mono-cortical

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0020–1383/$ – see front matter.
doi:10.1016/j.injury.2012.02.007

screw fixation with locking plates, and if this performs at least as well as bicortical non-locked fixation in mid-shaft clavicular fractures, then the risk of injury to the brachial plexus and subclavian vessels at the time of fixation would be negated, as the need to penetrate the inferior cortex would be obviated.

There have been a number of recent studies looking to assess locking plate fixation of clavicular fractures, both in terms of in vitro analysis of biomechanical performance, and clinical outcome.[24–27] However, there appear to be no studies simultaneously directly comparing mono-cortical locking, bicortical locking, and bicortical non-locking plate fixation of mid-shaft clavicular fractures in vitro, and this study was therefore undertaken to address this.

Materials and methods

Fifteen cadaveric clavicles were obtained from the Department of Anatomy of the University of Glasgow. All of the specimens were from adult corpses embalmed with formaldehyde, and in each case all soft tissues were removed, leaving only the clavicle itself. Prior to fixation and testing, the specimens were stored in a sealed, darkened environment moistened with physiological saline.

There were eight left sided, and seven right-sided specimens, consisting of four pairs from the same donor, and seven from individual bodies. Details of donor past medical history were not available, but in all cases, the clavicles appeared macroscopically sound prior to the start of the experiment.

Using a double-blinded randomisation technique, the clavicles were assigned to three fixation groups, mono-cortical locked, bicortical locked or bicortical non-locked. The length of each clavicle was then measured, and the midpoint of the bone was identified. Using digital Vernier callipers, the width and depth of the clavicle at this point were noted, and the cross sectional area calculated. Using a 0.61 mm thick blade and an electric oscillating bone saw an osteotomy was made in each clavicle at the previously marked midpoint. This was perpendicular to the long axis of the bone, and was chosen in order to replicate a fracture of the mid-shaft of the clavicle, as this is the fracture pattern that is most frequently seen.

All of the specimens were then fixed using a PERILOC anatomical clavicular plate (Smith and Nephew, Warwick) along their superior aspect. An identical eight-hole Superior Medial plate was used for each, with three screws securing each fragment. No attempt was made to contour any of the plates, and the screws were placed in a 'best fit' fashion within three of the four holes available according to what appeared to achieve optimal fixation. Sections of aluminium trough were cut to facilitate potting of the clavicles, and the specimens were fixed into these aluminium bases using a styrene-based polyester paste (David's Isopon P38, Wellingborough, Northants). A spirit level placed superiorly onto the plate ensured that the clavicles were positioned parallel with the bases of the aluminium troughs. A period of forty-eight hours was then given to allow full hardening of the polyester resin and security of the potting, prior to testing.

Testing was undertaken using an Instron machine with a 5 kN load cell (Instron, High Wycombe, Bucks). A rig was designed based upon three rollers of 10 mm in diameter, which were all adjustable with reference to the long axis of the clavicle. One roller was positioned directly at each end of the clavicle, beneath the aluminium holders. The third was sited superiorly, attached to the load cell and positioned to apply a downward force at an offset of 0.02 m to one side of the site of the mid-clavicular osteotomy. This was deliberately chosen to position the roller away from a screw head or empty screw hole, and directly over a solid section of plate. The specimens were then tested to failure via a single three-point bend test, without any preconditioning. The loading was continued in each case until the construct was seen to visibly fail, and the amount of deflection of the crosshead (mm) and load required to produce failure (N) were recorded. Using Euler–Bernoulli beam theory, the bending stress (σ_b) at the surface, the bending stiffness (EI), and the composite Young's modulus (E) for each of the constructs was then calculated.

Initially, the three fixation groups were compared using one-way ANOVA and subsequent Bonferroni correction where a significant difference was apparent. The parameters compared were the displacement of the crosshead at failure (m), the failure load (N), the bending stress (MPa), the bending stiffness (N m^2) and the aggregate Young's modulus (GPa) of the clavicle and plate. Independent ANOVAs were then performed to assess the effect of whether the plate was locked, or whether the screws passed through one or two cortices with respect to the same parameters. Given the small numbers involved in each of the data sets only the main effects of the independent variables were analysed and not their interactions. For all analysis, a significance level of 5% was chosen.

Results

The clavicles fixed with monocortical locking plates displayed a significantly lower bending stress (12 ± 1 MPa) than both the bicortical locking (28 ± 3 MPa, $p = 0.015$) and the bicortical non-locking specimens (24 ± 3 MPa, $p = 0.002$). The locked bicortical specimens also demonstrated a higher failure load (340 ± 49 N) than the monocortical locked (188 ± 23 N) and bicortical non-locked groups (242 ± 41 N), but this difference was not statistically significant. The differences between the fixation groups with respect to crosshead displacement at failure, bending stiffness at failure or composite Young's modulus were minimal (Table 1).

Whilst all of the variables (displacement, load, stress, stiffness and Young's modulus) increased with locking the screws into the plate, none of these differences achieved statistical significance. However, the number of cortices that the screws passed through did have a significant effect upon some of the failure parameters. Engaging two cortices with the fixation produced a significant increase in failure load (291 ± 28 N vs 138 ± 48 N, $p = 0.018$) and bending stress (26 ± 2 MPa vs 9.9 ± 3.5 MPa, $p = 0.002$) compared to single cortex fixation. Bending stiffness was in fact the only variable that was higher for single cortex fixation (2.9 ± 0.5 N m^2 vs 2.8 ± 0.3 N m^2), but this difference was not statistically significant (Tables 2 and 3).

Table 1
Comparison of the three fixation methods.

	Mean	SEM	Lower 95% CI	Upper 95% CI
Crosshead displacement (m)				
Non locked bicortical	0.004	0.002	0	0.007
Locked bicortical	0.004	0.002	0	0.007
Locked monocortical	0.003	0.001	0	0.006
Failure load (N)				
Non locked bicortical	241.548	40.844	161.493	321.602
Locked bicortical	339.946	49.464	242.996	436.895
Locked monocortical	187.791	23.451	141.827	233.756
Bending stress (MPa)				
Non locked bicortical	23.698	3.472	16.893	30.503
Locked bicortical	28.314	3.417	21.617	35.011
Locked monocortical	12.174	1.002	10.209	14.139
Bending stiffness (N m^2)				
Non locked bicortical	2.355	0.25	1.865	2.844
Locked bicortical	3.205	0.394	2.433	3.977
Locked monocortical	3.337	0.531	2.296	4.379
Young's modulus (GPa)				
Non locked bicortical	1.042	0.152	0.745	1.339
Locked bicortical	1.115	0.084	0.951	1.28
Locked monocortical	0.778	0.135	0.515	1.042