Single cycle to failure in torsion of three standard and five locking plate constructs

J. B. Cabassu1; M. P. Kowaleski1; J. K. Shorinko2; C. A. Blake1; G. R. Gaudette2; R. J. Boudrieau1

1Cummings School of Veterinary Medicine at Tufts University, Department of Clinical Sciences, North Grafton, Massachusetts, USA; 2Worcester Polytechnic Institute, Worcester, Massachusetts, USA

Objectives: The biomechanical properties of standard plates and recently designed locking plates were compared in torsion. We hypothesized that titanium (Ti) constructs would have the greatest deformation, and String of Pearls (SOP) constructs the greatest strength and stiffness.

Methods: Dynamic compression plates (DCP), stainless steel (SS) limited contact (LC)-DCP, Ti LC-DCP, locking compression plate (LCP), 10 mm and 11 mm Advanced Locking Plate System (ALPS) 10 and 11, SOP and Fixin plates were applied to a validated bone model simulating a bridging osteosynthesis. Yield torque (strength), yield angle (deformation) and stiffness were compared using one-way ANOVA with post hoc Tukey (p <0.05).

Results: The ALPS 11 constructs had significantly greater elastic deformation than all constructs except for the ALPS 10. There were not any differences in strength observed except for the ALPS 10 constructs, which was less than that for the SOP, LCP, DCP and ALPS 11 constructs. No differences in construct torsional stiffness were observed with the SS LC-DCP, DCP, SOP and SPP constructs; however all had greater stiffness than all remaining constructs. The ALPS 10 construct had lower stiffness than all constructs.

Clinical significance: Modulus of elasticity of Ti explains the higher deformation and lower stiffness of these systems, with similar results for the Fixin due to its lower section modulus compared to all other plates. The SOP and standard constructs had surprisingly similar biomechanical properties in torsion. The rationale for selecting these implants for fracture repair likely needs to be based upon their differing biomechanical properties inherent to the diverse implant systems.

Introduction

During the last decade, advances in the understanding of bone healing have changed the philosophy of fracture stabilization. The internal fixation of fractures has evolved with a change of emphasis from mechanical to biological priorities, described as biological internal fixation (1–3). As a result, biological priorities have been added to mechanical design considerations. These new principles include the use of locked internal fixators (locking plates), which have reduced bone contact and introduced bridge plating across comminuted regions with longer plate span distances, while simultaneously using fewer screws for fixation (1,2,4). These recent developments in fracture management have led to the design of a number of new implants in veterinary surgery, particularly new locking bone plates that are manufactured from a variety of materials. Stainless steel (SS) remains the most commonly used material to manufacture these veterinary bone plates, but more biocompatible material such as titanium (Ti) has recently been advocated based upon its improved biocompatibility and demonstrated lower infection rates (1, 5, 6). The Dynamic Compression Plate® (DCP) has been the reference implant for decades of fracture management (2). This plate must, however, be contoured to the bone surface since stability relies on the friction generated from the compression produced at the plate-bone interface; however, this contact adversely alters the vascularity to the bone directly under the plate (6). In order to decrease the damage caused by the contact between the plate and the bone, a new implant was developed by the AO Foundation and manufactured by Synthes® with an undercut surface that decreased the plate-bone contact area, the Limited Contact Dynamic Compression Plate® (LC-DCP). This implant is available in both stainless steel (SS) and titanium (Ti). Stability of the fix-
Pearls® (SOP), Fixin system®, and Advanced Locking Plate System® (ALPS). The stability of these plates is obtained by the fixed angle construct as the screws lock to the plates; compression between the plate and bone is not required to attain stability as it is with standard plates. The different systems use a variety of locking mechanisms and are manufactured from SS, Ti, or a combination of both materials. They also are produced in a variety of shapes and thicknesses. Because of the varying implant designs, some can only be used as neutralization or bridging devices, while others can still be used to compress the bone fragments together in an identical manner to standard plate application. Additionally, some of these systems use dedicated locking screws while others use standard bone screws to obtain the fixed-angle locking interface with the plate. These different designs result in varying amounts of bone contact underneath the plate, which may affect bone vascularity to a greater or lesser extent. All of these design features also affect their mechanical properties.

Locked plating methods have been designed to function as internal fixators, that is the fixed angle constructs are similar to external skeletal fixators but they are implanted internally. The principle of internal fixators has been shown to be an efficient technique to obtain fracture healing and also decrease the potential for vascular compromise, and thus also lower infection rates (1). Consequently, a variety of locking bone plates have been developed and are currently marketed for veterinary surgery: Locking Compression Plate®, String of Pearls® (SOP), Fixin system®, and Advanced Locking Plate System® (ALPS). The stability of these plates is obtained by fixed angle constructs; compression between the plate and bone is not required to attain stability as it is with standard plates. The different systems use a variety of locking mechanisms and are manufactured from SS, Ti, or a combination of both materials. They also are produced in a variety of shapes and thicknesses. Because of the varying implant designs, some can only be used as neutralization or bridging devices, while others can still be used to compress the bone fragments together in an identical manner to standard plate application. Additionally, some of these systems use dedicated locking screws while others use standard bone screws to obtain the fixed-angle locking interface with the plate. These different designs result in varying amounts of bone contact underneath the plate, which may affect bone vascularity to a greater or lesser extent. All of these design features also affect their mechanical properties.

Mechanical testing of orthopaedic implants is a commonly used method to provide some insight into the differing mechanical properties of bone plates; the standard of comparison usually has been the DCP and LC-DCP, as these implants have been commonly used for many years in veterinary orthopaedics (10). Although previous studies have reported the mechanical properties in bending or torsion of various bone plates, most of them have focused on bending properties and evaluated only some of the available bone plates (DCP, LC-DCP, LCP and SOP) (11–15). None of these studies have evaluated the biomechanical properties of the more recently designed

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**Key:** DCP = Dynamic compression plate; SS LC-DCP = stainless steel, limited contact dynamic compression plate; Ti LC-DCP = titanium, limited contact dynamic compression plate; LCP = locking compression plate; SOP = String of Pearls; ALPS 10 and 11 = 10 mm and 11 mm advanced locking plate system; 316L SS = 316L stainless steel; CP Ti = commercially pure titanium (grade 4); Ti-6Al-4V = titanium alloy (grade 5 [chemical composition includes 6% aluminium, 4% vanadium]).

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4 String of Pearls® (SOP®): Orthomed, Halifax, West Yorkshire; UK
5 Fixin system: TraumaVet SRL, Rivolito Italy
6 Advanced Locking Plate System (ALPS): Kyon, Zürich, Switzerland

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Materials and methods

Bone model

A previously validated bone model of fourth generation short-fibre filled epoxy (SFE) hollow cylinders, with 3 mm wall thickness and 20 mm outer diameter was used to simulate bone segments (16).

Plates and screws

Eight different types of plates were tested. These included three conventional plates (3.5 mm DCP, 3.5 mm LC-DCP in both SS [SS LC-DCP] and Ti [Ti LC-DCP]), and five locking plates (3.5 mm LCP, 3.5 mm SOP, 3.0/3.5 mm Fixin system, and 10 mm and 11 mm Advanced Locking Plate System [ALPS 10 and ALPS 11 respectively]). Each plate was secured with its corresponding screws as per the manufacturer’s recommendations (Table 1).

Construct assembly

The SFE bone model was cut into 152 mm long cylinders, simulating two bone segments. Both segments were aligned using a custom designed jig. A polypropylene spacer was then removed, resulting in final plate-SFE cylinder constructs with a 25 mm wide gap simulating a bridging osteosynthesis (Fig. 1).

All plates were placed on the bone model to simulate a bridging osteosynthesis, and loaded to failure in torsion. Based upon the material manufactured (Ti or SS) and physical characteristics of the different plate designs, our study hypotheses were that Ti plate constructs (Ti LC-DCP, ALPS 10 and 11) would sustain the highest elastic deformation (yield angle), and the SOP would be the strongest (yield torque) and stiffest construct.

Biomechanical testing

Four specimens per group were tested. Constructs were tested in angular displacement control using a servo-hydraulic testing machine. Torque and displacement data were recorded at a rate of 10 Hz using commercially available software. The tubes were centred in the testing machine with the ends of the constructs mounted in custom designed grips, and additionally secured using the transverse bolt fixed to the grips (Fig. 2). The distance between the end of the jig and the last screw of the plate was the same for all of the different constructs (38 mm).

A monotonic, single ramped test in torsion was performed on all constructs at a rate of one degree per second, and was stopped after 100° of torsion or catastrophic failure of the construct (when an abrupt drop in torque occurred).

Data analysis

Torque versus angle curves were generated using commercially available spreadsheet software. Yield torque (strength in N-m) and yield angle (elastic deformation, i.e. angular displacement in degrees) were determined at the yield point. The linear portion of the torque versus angular displacement curve was determined using linear regression analysis for a best fit; the slope of this portion of the curve evaluated construct stiffness (N-m/rad), and construct yield point was then ascertained with the

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0.2% offset method. The ultimate failure mode was recorded at the end of each test as either plastic deformation of the implants (plates or screws) or fracture of the bone model.

Statistical analysis was performed using a one-way ANOVA with post hoc Tukey Honestly Significant Difference and Bonferroni tests when appropriate. Significance level was set at p < 0.05.

Results

Strength

Yield torque results are summarized in Table 2 (Fig. 3). The differences between the strengths of the constructs were not significant, except that the ALPS 10 construct had a significantly lower yield torque than the SOP, LCP, DCP and ALPS 11 constructs.

Elastic deformation

Yield angle results are summarized in Table 2 (Fig. 4). The yield angle of the ALPS 11 construct was significantly greater than the other seven constructs. In addition, the ALPS 10 construct had a significantly greater yield angle than the SS LC-DCP and DCP constructs.

Torsional stiffness

Stiffness results are summarized in Table 2 (Fig. 5). The SS LC-DCP, DCP, LCP and SOP constructs had the highest mean stiffness, and they were significantly stiffer than the other four constructs. The Ti LC-DCP, Fixin, ALPS 11 constructs were significantly stiffer than the ALPS 10 construct.

The ALPS 10 construct was significantly less stiff than all other constructs.

Discussion

Constructs that were completely made with Ti implants (ALPS 10, ALPS 11, Ti LC-DCP) sustained the greatest elastic deformation, although only the ALPS 11 construct had a deformation that was significantly greater than all constructs made with SS implants. In addition, the ALPS 10 construct had significantly more deformation that the two reference SS implants (DCP and LC-DCP), supporting our first hypothesis. However, the finding that the strengths of the constructs were not significantly different, except from the ALPS 10 constructs, did not support our second hypothesis. We found that the SOP construct was not significantly stronger or stiffer than the other SS constructs.

Construct elasticity and stiffness are important factors likely to influence type and duration of fracture healing. Titanium has a lower modulus of elasticity than SS,
making Ti plates more flexible. The modulus of elasticity of pure Ti is approximately one-half that of stainless steel, with a result that an implant made of Ti will deform twice as much with the same force applied compared to a SS implant of the same design and geometry (17, 18). This material characteristic explains the very large elastic deformation of the constructs stabilized with implants made completely of Ti (ALPS 10, ALPS 11, Ti LC-DCP) prior to plastic deformation. The low modulus of elasticity of Ti is illustrated best by the ALPS 10 constructs, which reached the second greatest yield angle at the lowest yield torque. It would be expected that the Ti LC-DCP would have one-half the stiffness of the SS LC-DCP, as these implants are identical, except for their material properties; however, despite this supposition, this was not the case. Although there was a significant difference in stiffness between the SS and Ti LC-DCP constructs, this was not reflected in the deformation, as they were not significantly different. The Ti LC-DCP constructs reached a 33% higher deformation than the SS LC-DCP constructs; this finding is in agreement with another ex vivo biomechanical study (19). Similar to our results, the magnitude of the difference in deformation between the SS LC-DCP and Ti LC-DCP is less than the difference in elastic modulus of the material themselves. This finding can be explained with the model, in which the construct was tested and not simply the plates; therefore, the interface of the screws with the bone model must play a role in the overall interaction (19, 20).

Although the Fixin construct was considered a SS plate, its stiffness was similar to that of the Ti constructs. The section modulus of this plate was considerably smaller than all the other implants; additionally, they were secured to the tubes with Ti screws, which in turn were secured to the plate with Ti inserts. These factors likely account for the lower stiffness. However, based upon the lack of identifiable failure

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**Fig. 3**
Yield torque per construct; whisker box plots (median, standard deviation around mean, minimum and maximum); constructs that share the same lower case letter are significantly different. Details of the eight different plates, and their application, are described in Table 1.

**Fig. 4**
Yield angle per construct; whisker box plots (median, standard deviation around mean, minimum and maximum); constructs that show the same lower case letter are significantly different. Details of the eight different plates, and their application, are described in Table 1.

**Fig. 5**
Yield stiffness per construct (whisker box plots (median, standard deviation around mean, minimum and maximum); constructs that show the same lower case letter are significantly different. Details of the eight different plates, and their application, are described in Table 1.

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a The Fixin implant was composed of a stainless steel (SS) ‘support’ (terminology for the plate as per TraumaVet) with titanium (Ti) screws; Ti inserts indirectly secured the screws to the plate. Because the plate was SS, this construct was classified as SS.
of the screws or inserts, the smaller dimension (smaller section area) of this plate was likely most responsible. Despite the absence of any observed screw failure, the interaction of the Ti screws and inserts, as a contributing factor to the lower stiffness, cannot be discounted. Another possible concern with the Fixin implants is the potential for galvanic corrosion resulting from the mixed SS and Ti implant compositions. Despite the general recommendations to avoid mixing of different metals in orthopaedic implants, recent studies have demonstrated that the corrosion effect is minimal and even lower than between two SS interfaces (21, 22).

The ALPS 10 and ALPS 11 plates were secured with a combination of locking monocortical and standard bicortical screws. Although monocortical locking screws might provide better screw to plate rigidity than standard bicortical screws due to their rigid fixed angle design, the monocortical locking screws will provide less screw to bone model resistance to torsion than bicortical locking screws, thus resulting in tube failure (Fig. 7). This failure, however, did not occur in the ALPS 11 until there was significantly greater deformation than all other constructs. This finding does illustrate, nevertheless, a potential weakness to this fixation. When comparing all Ti plate constructs, the deformation behaviour also is consistent with the section area of these plates.

The angular deformation of the LCP constructs at yield was not significantly different from the SS LC-DCP and DCP constructs. This finding differs from another ex vivo biomechanical study in torsion, where the mean failure twist angle was significantly higher for the LCP constructs than the SS LC-DCP constructs (14). However, in that study, failure was not considered as the yield point but rather when an abrupt drop in torque was observed, or after 130° of torsion (12). In the authors’ opinion, failure of a construct should be considered when the construct undergoes plastic deformation (yield point) rather than catastrophic failure.

The finding that the strength of the SOP constructs was not significantly different from the other constructs was somewhat at variance with the findings of a previous study comparing the strength of various bone plates in bending (11). Indeed, in bending, the SOP was significantly stronger than the DCP, the LCP and the SS LC-DCP (11). However, that study evaluated the plate by itself and it did not take into consideration the interaction between the plate and the bone model, and the plate and screws, i.e., the construct (11). As previously noted, the strength of the construct represents a combination of the strength of the plate and screws, and the strength of the bone model, potentially attenuating the difference in strength between the different plates; this difference in plate alone versus construct behaviour has been demonstrated in bending (20). Certainly, the section area, or more specifically the tension constant (which is a measure of torsional stiffness) and the area moment of inertia (which is a measure of bending stiffness), will differ depending upon plate geometry; thus, the geometric shapes of the plates, and the tension constant, may play a major role in the differences we observed. However, these specific differences in plate geometry were not investigated, only the biomechanical behaviour of the plate constructs, which we believe to be the more clinically relevant information as opposed to the plates themselves. Furthermore, although all the SS plates are made of 316 L SS, differences in cold working of bar stock might have an influence on the torsional properties of the various plates. Another explanation could be that the screws used to fix the SOP plate to the bone model are standard 3.5 mm screws with a 2.4 mm core diameter. These screws are identical to those used to secure both the 3.5 mm DCP and LC-DCP to the bone models, as opposed to the LCP, which is fixed to the bone model with 3.5 mm locking screws with a core diameter of 2.9 mm. The core diameter of the screws also influences the strength of the construct, thus lowering the strength of the SOP constructs relative to the LCP constructs while also attenuating any difference with the DCP constructs. This probably explains the lack of significant difference identified between the SOP and some other constructs due to the differing screw core diameters, despite the SOP greater individual plate strength.

Many screws from the SOP constructs were bent at the level of the screw head (just below the plate) (Fig. 6); in addition, the SOP is not compressed to the bone surface, whereas all other plates with the same standard 3.5 mm screws (DCP, LC-DCP) are compressed to the bone surface. The latter will increase the overall construct stability with a well-contoured plate, which is mimicked in this model. We believe that these differences confirm that the screws also affect construct strength. Additionally, the mechanism of locking between the LCP and SOP, for example, is different; the threaded interface of the LCP screw is
stronger than the interference fit of the SOP (23). Once again, the lack of difference between these constructs can be due to the differing screw properties, such as the locking mechanism, despite the variations in individual plate strength.

There are a number of limitations to any such *in vitro* study. First, although we used a previously validated bone model instead of cadaveric bones, the latter would remain the ideal testing material (16). We elected to use the bone model in lieu of cadaveric bone due to the limited availability, handling problems and specimen variability of cadaveric bone (16). The bone models were therefore cost effective due to decreased number of specimens needed, and because the uniformity of the constructs permitted greater repeatability of construct preparation and testing. The latter was evident in the consistently similar standard deviations obtained in our study.

Standard specification and test methods for testing bone plates only describes a standard single cycle test method in bending (24). There are currently no standardized methods of testing bone plates and assessing their biomechanical properties in torsion. Our protocol was based on previously reported studies, where testing was performed in torsion, under displacement control at one degree per second or a similar rate (13, 14, 25–28). The use of the bone model permitted all plates to be fixed in an identical manner. However, despite these efforts, there was slight variation in the points of fixation and number of screws. Regardless of these variations, the plate positioning was considered comparable, and mimics what would be performed clinically, as no two plates from different manufacturers will have identical points of attachment.

The fracture configuration was selected to mimic a bridging osteosynthesis because this is the most appropriate use of a locked construct spanning a zone of comminution. Furthermore, this is a more severe test of the differences in plate geometry and attachment methods. Because of the test setup, the centre of rotation was aligned with the axis of the tube, which will introduce a translation effect with the plate under load, which probably would not occur *in vivo*. If the experiment centred the torsion along the axis of the plate it could be argued to be more physiological. However, muscle forces would also very probably exert a combination of bending, torsional and compressive forces rather than a pure torsion of the plate. Centring the torsion along the axis of the plate also would have changed the relative positions of the torsional axis between the different specimens due to the diverse physical dimensions of these plates; direct comparisons would then be difficult to interpret. Furthermore, such centring on the implant would heavily favour the plate as opposed to the overall construct geometry. We believe that better uniformity between specimens was obtained by centring the test on the tube axis, although this may have overestimated construct stiffness due to the offset of the plate from this axis. Lastly, previous studies performing similar *in vitro* and *ex vivo* torsional studies similarly centred the axis of rotation on the tube or bone; thus, we believe the precedent exists for the testing method used herein (13, 14, 25–28).

This study reports single cycle torsional loading to failure of constructs. Although the magnitude and specific combination of forces present at the level of a particular fracture *in vivo* are unknown, it has been shown that a combination of bending and torsional forces represent more than 90% of the physiological loads acting on a repaired fracture (29, 30). Therefore, the results of both bending and torsional testing should be interpreted in concert (20). Clinically, the fixation must resist both acute and cyclic loading conditions until bone healing occurs. Fatigue testing studies are currently being performed at our institution to complete the full picture of the plate and construct evaluation. In addition, clinical studies reporting fracture fixation using these implants will provide complementary data that will help define specific indications for this diverse group of implants.

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Conflict of interest
Two of the authors (R. J. Boudreau and M. P. Kowaleski) have received honorariums (course instruction and product development) from two of the companies (Kyon and Fixin) that donated the implants that were used in the study, and these two authors have also acted indirectly through the AO Foundation as consultants on product development to Synthes Vet (as members of the Veterinary Expert Group for the AO Technical Commission). There are no conflicts of interest to declare for the other authors.

References


