The effect of the combination of locking screws and non-locking screws on the torsional properties of a locking-plate construct

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Summary
Little is known about the torsional properties of bone-plate constructs when a combination of locking and non-locking screws have been used. Sixty cadaveric canine femurs were divided into three groups. In the first group, the plate was affixed using three non-locking screws. In the second group, only locking screws were used while a combination of one locking and two non-locking screws were used in the third group. All constructs were subjected to torsion until failure. Torque, angle of torsion, and work were all calculated at the maximum failure point, and at five degrees of plastic deformation. The locking group showed significantly higher torque and angle at the maximum failure point compared to the two other groups, and also for work at five degrees of plastic deformation. This study suggests that a construct composed of all locking screws will fail at a greater torque value, and sustain greater work to failure in torsion compared to a construct composed of all non-locking screws. The addition of a single locking screw to an otherwise non-locking construct will increase the torque at the offset failure point and may be of clinical value in constructs subjected to high torsional loads.

Introduction
Successful fracture fixation results when the various loading forces acting on the fracture fragments are resisted by the mechanical stiffness of the bone-implant construct. Conventional fracture plating relies on the friction between the plate and the bone to generate stability (1). Achieving such friction requires intimate contact between the plate and the bone, and therefore accurate pre-contouring of the plate to the bone. This extensive plate contact can negatively affect periosteal blood flow, resulting in bone necrosis and resorption under the plate (2, 3). Conventional dynamic compression plates have smooth, elongated screw holes allowing the placement of screws in a large range of orientations to match the needs of each specific fracture. Although the ability to vary screw angle is convenient for the surgeon, it may also represent a significant drawback of conventional plating. Since each screw can freely pivot in its hole, it therefore relies on two points of purchase (both cortices) to maintain its stability. Any loss of cortical purchase in either cortex due to bone resorption or microfracture may allow the screw to toggle in the plate hole, and the bone fragment to shift under the plate.

Locking plates were developed as a means to reduce these problems and improve bone healing, particularly in poor quality bone (4). In these systems, threads cut in the screw head engage with matching threads cut into the plate hole, effectively locking the screw to the plate. Because any angulation of the screw within the screw hole is rendered impossible, the construct acts as an internal fixator and does not rely upon friction for stability. Any screw toggling is effectively prevented, even with monocortical screws with only one point of purchase in the bone. Because this system does not rely on plate to bone friction to achieve stability, the plate does not have to be in direct contact with the bone and perfect plate contouring is no longer necessary. The space between the plate and the bone permit improved periosteal blood flow, resulting in reduced bone necrosis under the plate, and it may also decrease infection rates (4–7).

The holes of locking compression plates (LCP) are designed such that either a non-locking or locking screw can be used...
in them. The use of a traditional (non-locking) screw in a locking plate may be indicated when angulation of the screw is necessary, such as for the fixation of a butterfly fragment with lag screws, to assist with fracture reduction, or to avoid penetration of a joint surface. The use of non-locking screws may also be selected in order to decrease the overall cost of the fixation as locking screws are currently more expensive than regular screws. Although several studies have shown that LCP perform similarly to conventional plates when tested biomechanically, only a few studies have evaluated the effect of using a combination of locking and non-locking screws on the biomechanical properties of plate-bone constructs subjected to torsional loading (8–12). As locking screws create angle-stable constructs, we hypothesised that the addition of a locking screw to a non-locking construct would significantly affect the torsional strength of the bone-plate construct.

Materials and methods

Sixty femurs without gross evidence of orthopaedic disease were harvested from adult 20 to 25 kg dogs, which had been euthanatized for reasons unrelated to this study. All soft tissue was removed from the femurs, and the bones were wrapped in moist sponges, double bagged, labelled and stored at −20 °C in a conventional freezer until bone plate application and biomechanical testing. Before testing, the femurs were thawed at room temperature in a water bath (while still in the bags), and the distal and the distal end was potted in plastic cups using a polyester resinb to the level of the fabellae.

Initial testing consisted of testing all locking screw constructs and all non-locking screw constructs with eight paired femurs. This initial testing demonstrated that in order to detect a difference of 20% in the torque at failure with a power greater than 80%, 20 femurs would be needed in each group. The femurs were thus divided into three groups of 20 femurs each. The constructs used for the initial testing were included in their respective groups. The level of the osteotomy was marked on the bone using a standard template to ensure repeatability of the cut and to standardise the distance of the cut to the potting material. Depending upon group assignment, one-half of a six-hole, 3.5 mm LCP was attached to each femur using either three locking screws, three non-locking screws, or a combination of two non-locking and one locking screw (n = 20). The locking and non-locking groups were paired right and left femurs from the same dog, with the side decided randomly by coin toss. For the combination group, 20 femurs from 20 different dogs were used and the side was once again randomised with a coin toss. Given the straight nature of the canine mid-femur no contouring was needed for any of the plates.

All bone screws were inserted following standard AO principles. All non-locking screws were tightened by hand, as would be done clinically, by a surgeon experienced in orthopaedic surgery (NM). The locking screws were tightened using the recommended torque-limiting device. All screws were bicortical. For the combination group, the locking screw was placed in the most distal hole to the planned fracture site after the non-locking screws had been placed. Once the plate was secured to the bone with three screws, a transverse osteotomy was performed at the previously determined level and the cranial-caudal and medial-lateral diameters of the bone were measured at the osteotomy site using a caliper. Once the osteotomy was performed, the free half of the plate was bolted to a torsion jig using machine screws. No gaps were present between the end of the jig and the osteotomy, however, compression at the fracture site was not performed. Only one side of the plate was affixed to bone in an attempt to minimise the variability of having the plate attached to bone on both sides of the osteotomy. The torsion jig was designed to allow six degrees of freedom during testing. The constructs were tested in torsion to failure at a rate of 0.5°/s using a servohydraulic testing machine (Fig. 1).

Torque at failure, angle at failure and torsional stiffness were determined by analysis of the torque/ deformation curves (Fig. 2). Stiffness was calculated by creat-

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Fig. 1 (a) Potting fixture and torsional testing jig. The longitudinal axis was visually assessed. (b) Close-up of potted femur attached to testing jig.
ing a best-fit line in the initial linear region of the load-displacement curve for each sample. A second line parallel to the linear portion of the torque/ deformation curve, but offset by five degrees, was drawn on the graph, and the intersection between this line and the torque/ deformation curve was used to define clinical failure. This line represented five degrees of plastic deformation, which was thought to be more representative of clinical failure than the maximum torque. The torque and angle at failure at the intersection between the five degrees offset line and torque/ deformation curve were calculated and defined as the offset torque and offset angle. The work to clinical failure (the 5° offset failure point) and work to maximal failure were determined by calculating the area under the curve.

A general linear mixed model, which accounted for the random effect of ‘dog’ and the fixed effects of construct, side and their interaction, was run for the parameters measured. Residual analysis and a Shapiro Wilk test were used to determine if the data met the assumptions of the ANOVA. If the overall F-test was significant, pairwise comparisons were based on one-sided Tukey tests. Statistical analyses were performed with standard analysis software. Statistical significance was set at 0.05 for all tests.

Results

For all values for torque, angle, and work at the maximum and offset failure points, the greatest value was always from the all-locking construct group, the smallest value was always from the non-locking construct group, and the combination-construct group was always intermediate to the other two groups.

The mean ± standard deviation of the bone diameter (average of the medial-lateral and cranio-caudal diameters) for the locking group (15.20 ± 2.19 mm), non-locking group (15.05 ± 1.970 mm), and combination group (14.90 ± 0.904 mm) were not significantly different.

The mean ± standard deviation of the torsional stiffness for the locking construct (2.19 ± 0.381 Nm/degree), the non-locking construct (2.03 ± 0.569 Nm/degree), and the combination construct (2.31 ± 0.305 Nm/degree) were not significantly different. Representative curves for each construct are demonstrated in Fig. 3.

At the maximum failure point, the locking group (52.6 ± 8.43 Nm) had significantly higher torque than the non-locking group (38.7 ± 11.57 Nm, p = <0.001). Failure angle (47.8 ± 18.88° vs. 30.8 ± 13.46°, p = 0.001), and work to failure (1694 ± 886 Nm-degree, p = 0.002) were also significantly higher in the locking group than in the non-locking group (Table 1).

At five degrees of plastic deformation (clinical failure), the locking group had a significantly higher torque compared to the non-locking group (45.1 ± 7.26 Nm vs. 34.8 ± 9.20 Nm, p = <0.0001), and angle (25.2 ± 3.95° vs. 20.6 ± 4.69°, p = < 0.001) than the non-locking group. The comb-
Table 1 Summary of the data, which compares all locking, combination screw, and non-locking groups. Results are shown for both the maximum failure point and the five-degree offset failure point. Data represents the mean ± standard deviation (95% confidence interval). Data with the same superscripted letters denote significant differences between groups. (a) p = <0.0001; (b) p = 0.04; (c) p = 0.03; (d) p = <0.001; (e) p = 0.001; (f) p = <0.0001; (g) p = 0.02; (h) p = 0.001; (i) p = 0.04; (j) p = 0.002.

<table>
<thead>
<tr>
<th>Construct</th>
<th>Offset torque (Nm)</th>
<th>Offset angle (degrees)</th>
<th>Offset work (Nm, degree)</th>
<th>Maximum torque (Nm)</th>
<th>Maximum angle (degrees)</th>
<th>Maximum work (Nm, degrees)</th>
</tr>
</thead>
<tbody>
<tr>
<td>All locking</td>
<td>45.05 ± 7.26</td>
<td>25.19 ± 3.95</td>
<td>625.05 ± 149.8</td>
<td>52.63 ± 8.43</td>
<td>47.78 ± 18.88</td>
<td>1693.64 ± 886.48</td>
</tr>
<tr>
<td>Combination</td>
<td>40.71 ± 5.74</td>
<td>22.14 ± 3.28</td>
<td>525.81 ± 126.04</td>
<td>47.38 ± 5.74</td>
<td>36.59 ± 12.36</td>
<td>1118.6 ± 472.92</td>
</tr>
<tr>
<td>All non-locking</td>
<td>34.75 ± 9.2</td>
<td>20.58 ± 4.69</td>
<td>453.5 ± 178.55</td>
<td>38.7 ± 11.57</td>
<td>30.76 ± 13.46</td>
<td>841.99 ± 609.45</td>
</tr>
</tbody>
</table>

bination group had a significantly higher torque compared to the non-locking group (40.7 ± 5.74 Nm vs. 34.8 ± 9.20 Nm, p = 0.04), and a lower angle at the offset compared to the all locking construct (22.1 ± 3.28° vs. 25.2 ± 3.95°, p = 0.03).

Discussion

The current study found that the locking LCP construct was significantly stronger in torsion than the non-locking LCP construct. The combination construct of locking and non-locking screws was always intermediate to the other two groups for torque, angle, and work.

Bone plates are one of the most frequently used implants for fracture fixation. Traditional bone plates rely on the friction between the plate and bone, which is generated by the compression of the plate to the bone by the torque of the screws. The frictional force generated is equal to the product of force that presses the plate to the bone and the coefficient of friction between the plate and bone (13). The normal force of a conventional bone-plate construct is equal to the sum of the torques on each screw, which is related to the resistance to shear stress of the bone trapped within the threads of the screw (14, 15). Since non-locking plates are unable to prevent the screw from orienting in the direction of the applied force to the plate, the strength of the connection between the bone and the screws is only due to the shear stress between the screws and the bones (15). The weakest component of the plate-screw-bone construct is the screw-bone interface. The strength of this interface is equal to the stress resistance of the bone multiplied by the contact area between the screw and the bone (15). Locking plates do not rely upon the friction between the plate and the bone for stability. Locking screws result in an angle-stable device, in which the plate prevents the screws from orienting in the direction of the applied force to the plate, thus converting the shear stress at the screw-bone interface to a compressive stress, which bone is better able to resist. In order for a locking construct to fail, all screws must cut through the bone simultaneously, the bone must fracture, or the plate must fail (16).

There have been several biomechanical studies comparing the properties of locking versus non-locking constructs and these studies often have conflicting results. While some studies demonstrated a significant improvement in the strength of locking constructs in axial compression, torsion, and bending when compared to non-locking constructs (11, 16–25), others did not find any significant differences between the two (8, 10, 26–33). Reasons for the conflicting results found in these studies include the use of a variety of locking systems used in a variety of bones, both cadaveric and synthetic, as well as the use of different species, and the use of different methods for the testing.

A number of studies have specifically examined the torsional properties of LCP. Aguila et al compared LCP using only locking screws to limited contact dynamic compression plate (LC-DCP) in cadaveric canine femurs and found that the mean twist–to-failure of the LC-DCP was significantly lower than that for the LCP, however the torsional stiffness of the two constructs was not significantly different (8). This is in contradiction with Sod et al who compared 4.5 mm LCP to 4.5 mm LC-DCP in equine third metacarpal bones. In their model they used a combination of locking and non-locking screws in the LCP and found torsional stiffness was significantly greater for the locking plates (17).

Since non-locking and locking constructs employ different biomechanical principles to impart stability to the construct, it is generally not recommended to use combinations of locking and non-locking screws in the same construct. However, there are a number of situations in which the use of a combination of fixation techniques would be beneficial. If compression at the fracture site is desired, then non-locking screws must be used. Non-locking screws allow surgeons the freedom to angle screws to engage bone fragments that locking screws would be unable to engage sufficiently, also non-locking screws are sometimes used to reduce fractures by drawing the bone to the plate. Finally, non-locking screws are significantly less expensive than locking screws, and thus may be selected in order to decrease the cost of the construct. When non-locking and locking screws are used in the same construct it is recommended that the non-locking screws should be used first to establish the friction fit between the plate and bone, and then followed by locking screws to protect the fixation (4, 34).
The addition of a locking screw to a non-locking construct should theoretically change the construct into an angle stable construct. Thus, the addition of a single locking screw should significantly increase the torque at failure and work as demonstrated in this study, in which a 17% increase in torque at five degrees offset was observed after addition of a locking screw to a non-locking construct. However, the addition of a single locking screw was not equivalent to an all-locking construct (Table 1).

There are very few studies examining the biomechanical effects of using a combination of locking and non-locking screws in the same construct. Gardner et al. studied the effects of using a combination of two locking and one non-locking screw in a LCP in a synthetic human humeral-os- teoporotic model on cyclic torsion testing. That study did not find any significant differences between the all-locking and the combination group, but both of these groups were significantly stiffer than the non-locking group throughout all phases of cyclic testing. After 1000 cycles, the locking and combination constructs had declined to 80.0 ± 10.2% and 79.2 ± 9.5% of their initial stiffness, while the non-locking group had declined to 22.3 ± 12.6% (11).

In our study, the non-locking plates failed at a lower torque and at a lower angle than the combination plates or the locking plates, and the all-locking plates failed at a higher torque and angle compared to the other two groups of plates. Since all of the plates tested were the same type (LCP), it was the attachment of the plate to the bone that was being tested and any differences seen would be due to how the plate was secured to the bone. The fact that the same type of plate was used in all three groups also explains why the stiffness of the constructs was similar between groups as long as the torsional force did not exceed the screws’ holding power or overcome the frictional forces between the bone and the plate. Similar stiffness in torsion between locking and non-locking constructs using LCP was also found in both synthetic and cadaveric human humeral fractures and in cadaveric human ulnar fractures (28, 32). This is in contradiction with Gardner’s study which found a significant difference between the stiffness of the locking versus the non-locking constructs. The difference found in Gardner’s study may be due to the difference in testing material (synthetic humeri versus cadaveric femurs), but is more likely due to the fact that Gardner et al. overdrilled the screw holes to simulate reduced screw purchase in osteoporotic bone while our study used normal canine bones, and no attempts were made to stimulate osteoporosis (11).

Another study examined the effects of replacing the end screws of a locking unicortical construct with bicortical (locking or non-locking) screws and compared these constructs to a non-locking bicortical construct. During torsional testing, they found that the unicortical locking group was significantly less stiff than the other three groups, and that replacing the end screws of a locking unicortical construct with bicortical screws significantly increased the construct stiffness. Both of the hybrid constructs provided equivalent stability in torsion compared to the non-locking bicortical group. The locked hybrid group was significantly stiffer than the other three groups in anteroposterior bending. The authors concluded that the major factor in determining torsional stability was likely the increased working length of the bicortical screws, rather than the use of locking or non-locking screws (12). This latter study is difficult to compare to our study because of the confounding effect of the locking screw location or the placement of additional locking screws.

The non-locking screws were hand tightened by an experienced orthopaedic surgeon. This was done in an attempt to mimic the clinical setting. It should be noted that different clinicians will tighten screws to different degrees. This variation could potentially alter the results of this study. However, experienced clinicians will generally apply a consistent level of torque (38). The amount of torque applied to the screw will directly affect the amount of compression achieved between the plate and the bone, and thus the friction between the two (14).

One of the major limitations of this study was its in vitro nature. The femurs were stripped of all surrounding soft tissues which could contribute to fracture stability in vivo. A simple, transverse fracture model with no bone loss was used, which is rarely found in clinical situations. It has been shown that bone resorption occurs around implants when even minute instability exists. This resorption could result in loose implants which would then affect the stability of the construct (39). For this resorption to occur, active blood supply is required and therefore resorption cannot be accurately reproduced in vitro, even with cyclic testing. For this reason, we tested the specimens by applying a single load to failure. Static testing is often used as a first step for testing constructs and provides information regarding the levels of force that a construct can stand prior to failure (30, 40). The behaviour of these constructs in the presence of bone resorption or in poor quality bone warrants further investigation.
We chose to use a cadaveric femur model to most closely approximate the clinical situation. While using a synthetic model would have eliminated inter-specimen variability, these models do not have all of the same biomechanical and frictional properties as bone, nor do they always have the same screw holding power as cadaveric femora (41). Many of these models have similar biomechanical properties to bone when tested whole (42–46). However, the strength of these surrogate models can become significantly different from bone when subjected to a notch test, which is used to determine fracture toughness (41). Several studies have shown that screw holes can significantly weaken bone and that the screw hole will influence the behaviour of the fracture (47–50). Screw holes can cause bone surrogates to have significantly different fracture toughness than real bone, which may affect the load that the constructs fail at, and may also cause them to fail in a different pattern compared to real specimens (41). This was further demonstrated in a study in which locking plates were compared to non-locking plates in both cadaveric humeri and synthetic humeri tested in torsion. This study demonstrated that the locking plates had significantly less angular displacement than the non-locking plates in cadaveric specimens, while the two constructs were not significantly different in the synthetic specimens (23).

From previous studies as well as the current study, it can be seen that LCP used in a locking fashion are as strong, if not stronger than conventional plates or LCP used in a non-locking fashion. Although the torsional stiffness of the constructs in the initial phase of loading was similar, non-locking constructs failed at lower torque than the locking and combination constructs. Adding a single locking screw to an all non-locking construct increased the torque to the offset failure point by 17%. In the current study, no significant differences were found between the combination and all-locking groups for the offset torque and angle. This would suggest that the addition of just one locking screw is able to provide an angle-stable construct and it is unnecessary to replace non-locking screws, which have been used to initially reduce the fracture, with locking screws to create an all-locking construct. This study also suggests that an all-locking construct be used when the greatest implant strength is needed, and that adding locking screws to a non-locking construct can increase the torsional strength of that construct.

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