Biomechanical comparison of mono- and bicortical screws in an experimentally induced gap fracture

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Monocortical, bicortical, biomechanical, locking compression plate, tibia

Summary
Objectives: To compare the bending and torsional mechanical properties of mono- and bicortical locking screws in a canine cadaveric tibial gap ostectomy bridged by a locking compression plate (LCP).
Methods: A 10-hole 3.5 mm LCP was applied medially to the tibia with a gap ostectomy using locking screws in the two proximal and distal plate holes. One tibia of each pair was randomly assigned monocortical screws and the other bicortical screws. Constructs were tested non-destructively in mediolateral and caudocranial four-point bending and torsion, and then to failure in four-point bending. Stiffness, yield and failure variables were compared between screw lengths and load conditions using analysis of variance.

Results: Caudocranial and mediolateral four-point bending stiffnesses were not different between screw constructs. Torsional stiffness was greater and neutral zone smaller for bicortical constructs. Constructs were stiffer and stronger in caudocranial bending than in mediolateral bending. In caudocranial bending, bicortical constructs failed by bone fracture and monocortical constructs by screw loosening.

Conclusion: Bicortical constructs were stiffer than monocortical constructs in torsion but not bending. Bicortical screw constructs failed by bone fracture under the applied loads whereas monocortical screw constructs failed at the bone-screw interface.

Clinical relevance: Bicortical screw placement may be a safer clinical alternative than monocortical screw placement for minimally invasive percutaneous osteosynthesis LCP-plated canine tibiae with comminuted diaphyseal fractures.

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Introducción
Bone plates are commonly used for fracture fixation, limb sparing procedures, and angular limb deformity correction. Diaphyseal fractures of long bones are the most common fracture type encountered in companion animals. Diaphyseal fractures of the tibia represent 10% to 20% of fractures seen in veterinary practice (1). The goal of internal fixation is to maintain fracture stability to encourage bone union while maintaining a functional limb during healing.

With the recognition that biological osteosynthesis is important in addition to fragment stabilization, new implants for internal stabilization have been developed, such as interlocking nails and locking bone plates. Biological fracture fixation is characterized by minimal disruption of the fracture site during repair and sufficient fixation to achieve secondary bone healing. Healing of the fracture occurs under optimal biological conditions instead of absolute stability by minimizing soft tissue damage and preserving the blood supply to traumatized bone. Since the damage to the soft tissues and the blood supply is less extensive, more rapid bone healing can be achieved (1-5).

Bridge plating relies on biological osteosynthesis and secondary bone healing (1). Bridge plates can be applied using a minimally invasive percutaneous osteosynthesis (MIPO) or an “open but do not touch” approach to restore anatomical realignment through indirect manipulation of bone fragments without requiring anatomical reconstruction and interfragmental.
tary compression (1). Bridge plating techniques and biological osteosynthesis are characterized by a high plate to bone ratio (ratio between the length of the plate and the length of the bone), low screw to plate hole density (screws inserted divided by number of plate holes), and high plate to span ratio (plate length divided by fracture length), which decreases the strain on the plate. A high plate working length in turn decreases screw loading, thus fewer screws need to be inserted and the screw to plate hole density can be kept low (1, 6-8). It has been described that the plate span ratio should be more than three with a comminuted fracture and the screw to plate hole density should be less than 0.5 leaving two to three empty holes over or adjacent to the fracture (1, 8).

Traditional bone plating uses bone screws to compress the plate against the bone, achieving stabilization by creating friction between the bone and the bone plate (friction is maintained by compression generated by screw – bone purchase). Compression between the bone and the plate is maintained by the friction generated between the threads of the bone screws and the bone, also commonly referred to as bone purchase of the screws. Multiple factors have been shown to achieve maximum friction between the bone-plate interface, including perfect contouring of the bone plate and adequate screw-bone purchase requiring, in most cases, engaging both bone cortices and engaging six cortices in each main fragment.

The locking compression plate (LCP) can accommodate either a conventional screw or locking screw, allowing the plate to be used as a conventional compression plate, a pure internal fixator or a hybrid of both (6). When applied as an internal fixator, the LCP system functions as a fixed angle construct that does not rely on plate to bone compression to maintain stability, obviating the mechanical need of friction between the bone and the bone plate, however fixed angle constructs substantially increase the stress on the plate-screw interface (1).

Several factors have been shown to influence the stability of an LCP construct, such as the working length (the length of the plate between the two screws closest to the fracture), screw number and placement distance between the plate and the bone, and the size of the implants (9). It has been proposed, based on inherent biomechanical properties of the LCP construct, that monocortical screw placement is statically as secure as bicortical screw placement (2, 7, 8-10). Monocortical screw placement is useful for stabilization of vertebral fractures and instabilities, and for plate-rod stabilization of comminuted fractures to avoid interference with the neural canal and intramedullary implants (1, 9, 10). The authors are unaware of reports in the veterinary literature comparing monocortical and bicortical screw fixation in LCP constructs without an intramedullary rod. With the current trend towards biological osteosynthesis using MIPO approaches, the LCP construct using monocortical screws would add another option for repairing complex fractures. The aim of this biomechanical study was to compare unicortical and bicortical screws using an LCP construct in a simulated gap fracture model. Our hypothesis was that there would be no difference in mechanical properties between monocortical and bicortical screw tibial constructs when tested in four-point bending and axial rotation.

Materials and methods

Study design

An LCP plate was applied to bridge a middiaphyseal gap osteotomy in paired canine cadaveric tibia. Plates were attached to the tibia by either monocortical or bicortical screws in paired bones. The biomechanical stability of tibial constructs was compared between monocortical and bicortical screw placement during nondestructive mediolateral and caudocranial bending and torsion, and subsequently during failure in mediolateral or caudocranial bending.

Specimen preparation

Nine pairs of canine tibias from cadavers of medium to large breed (range of 20 kg to 30 kg), skeletally mature dogs euthanatized for reasons unrelated to orthopaedic disease were studied. Radiographs were taken to ensure absence of pre-existing orthopaedic pathology, and suitable size for application of a 10-hole plate. Paired tibias were harvested, manually debrided of soft tissues, wrapped in saline-soaked towels, and frozen at ~20°C until testing. Specimens were thawed at room temperature for at least 24 hours before testing.

One tibia from each pair was randomly selected for fixation using 12 mm long monocortical screws; the contralateral tibia was assigned 40 mm long bicortical screws. A 10-hole, 3.5 mm LCP was centred along the length of the bone and contoured to the medial surface of the tibia to minimize variation in the distance between the plate and the bone. The plates were applied to the medial surface of the tibia with four 3.5 mm diameter locking self-tapping screws, two screws in the most proximal, and two screws in the most distal, portions of the plate; resulting in a screw to plate hole density of 0.4.
order on statistical outcomes. Each pair was then tested to failure in either caudocranial bending (4 specimens) or lateromedial bending (5 specimens). The bending direction was alternated between consecutive pairs.

For caudocranial bending, a bending moment was created by contact of the inner load noses with lever arms affixed to the cups that held the bone ends (▶ Figure 3). Cups were held in low friction bearings allowing rotation in the desired plane during bending. Bending moment (BM) created by the axial load (F) was found by the equation BM = (FL)/2, where L is the moment arm length (from cup pivot to contact of load nose on lever arm). The construct was preconditioned by loading in bending for 30 cycles from 0 to 1 Nm load at 1 Hz under displacement control. The construct was then loaded in bending for five cycles from 0 Nm to 2 Nm at 1 Hz under displacement control while load and specimen rotation data was collected at 204 Hz. Preliminary tests showed 2 Nm was within the linear region of the load-deformation curve and no toe-region was observed. Each load unload cycle was followed by 0.5 seconds of zero load to approximate the timing of the walk gait cycle. Specimen rotation was measured using a rotary variable displacement transducer.

Torsion was performed by rigidly fixing the same fixation cups to the load frame (▶ Figure 4) and setting a compressive preload of 10 N. The bone was positioned so that the neutral axis of torsion was at the centre of the bone diaphysis. The construct was preconditioned in 30 cycles to ± 0.5 Nm at 0.25 Hz under angular rotational control. Then constructs were loaded in torsion for 10 cycles to ± 2.0 Nm at 0.25 Hz under angular rotational control while load and specimen rotation data were collected at 102 Hz.

After all nondestructive tests were run, pairs were loaded to failure in either caudocranial or lateromedial bending at 1 mm/sec of linear displacement of load noses, which corresponded to approximately 2 Nm/sec during the linear elastic portion of the test. A maximum load limit was set at 60 Nm to prevent damage to the load cell.

The proximal and distal epiphyses were potted in polymethylmethacrylate (PMMA) using a custom-made alignment jig that rigidly fixed the bone ends to aligned cups for attachment of the constructs to the mechanical testing system. Before potting, two perpendicular smooth Kirschner wires were placed transversely through each epiphysis of the bone to prevent rotation of the bone in the PMMA during mechanical testing. Care was taken to ensure that the PMMA did not encompass the bone plate (▶ Figure 2). The bones were wrapped in saline solution soaked paper towels until testing.

**Mechanical testing**

Because the tibia experiences tensile strain on the cranial surface and compressive strain on the caudal surface *in vivo* at a walk, plated tibial constructs were nondestructively tested in four-point bending (caudocranial with the caudal surface in compression and mediolateral with the medial surface in tension) and in torsion (internal and external rotation about the long axis) (▶ Figure 3, ▶ Figure 4) (11). The order of nondestructive tests was the same within tibial pairs and varied among tibial pairs to minimize the effect of test order on statistical outcomes. Each pair was then tested to failure in either caudocranial bending (4 specimens) or lateromedial bending (5 specimens). The bending direction was alternated between consecutive pairs.

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Data reduction

Non-destructive tests

Bending data were analysed for the third cycle to 2 Nm to avoid artefacts associated with the beginning and end of cyclic tests. Construct behaviour was visualized on bending moment versus specimen angle (moment arm angle) curve plots. Construct stiffness was calculated as the slope of the linear fit of the data from 0.3 Nm to 1.7 Nm bending moment. A small residual specimen angular displacement was apparent at 0 Nm bending moment between cycles. This residual displacement was captured as the difference between the angle at 0 Nm bending moment prior to the third cycle and the angle at 0 Nm bending moment after the third cycle.

Torsional data were analysed for internal and external rotation to 2.0 Nm from the fifth cycle to avoid artefacts associated with the beginning and end of cyclic tests. Construct behaviour was visualized on torque versus angular rotation curve plots. The torque-angular deformation curves also displayed hysteresis with a neutral zone centred on 0 Nm torque where relatively large angular deformations were associated with small changes in torque. The range of the neutral zone was captured by determining the maximum internal rotation (although only external rotation stiffnesses are illustrated) to capture non-linear features.

The torque-angular deformation curves were calculated for two regions of the curves. Initial stiffness was calculated for the 0.5 to 0.9 Nm torque range and terminal stiffness was calculated for the 0.9 to 1.6 Nm torque range. Stiffness was calculated as the slope of the linear fit of the data within the respective torque range.

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<p>| Table 1 | Stiffnesses and residual displacements (mean ± SD) for monocortical and bicortical screws, locking compression plate plated, tibial constructs tested non-destructively in mediolateral or caudocranial four-point bending. |</p>
<table>
<thead>
<tr>
<th>Construct mode</th>
<th>Monocortical</th>
<th>Bicortical</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Bending mode</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>Mediolateral</td>
<td>2.45 ± 0.87</td>
<td>3.39 ± 1.57</td>
<td>0.18</td>
</tr>
<tr>
<td>Caudocranial</td>
<td>4.80 ± 1.53</td>
<td>5.03 ± 1.19</td>
<td>0.59</td>
</tr>
<tr>
<td>Residual displacement (*)</td>
<td></td>
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<td></td>
</tr>
<tr>
<td>Mediolateral</td>
<td>0.149 ± 0.230</td>
<td>0.053 ± 0.093</td>
<td>0.26</td>
</tr>
<tr>
<td>Caudocranial</td>
<td>0.098 ± 0.120</td>
<td>0.031 ± 0.054</td>
<td>0.15</td>
</tr>
</tbody>
</table>
Failure tests

Data from the monotonic four-point bending to failure tests were visualized on bending moment versus construct angle curve plots. Construct yield was determined using a 0.1° angle offset from the line-fit of the linear elastic region of the curve using a custom program\(^b\). Stiffness was defined as the slope of the least-squares linear fit of the middle one-third of the data between 0 N and yield loads. The failure point was defined as the maximum bending moment.

Stiffnesses and neutral zone angles (mean ± SD) for monocortical and bicortical screw, locking compression plate plated, tibial constructs tested non-destructively in torsion.

Table 2

<table>
<thead>
<tr>
<th>Variable</th>
<th>Screw placement</th>
<th>p-value</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Monocortical</td>
<td>Bicortical</td>
</tr>
<tr>
<td>External torsion - Stiffness (Nm/°)</td>
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<td></td>
</tr>
<tr>
<td>Initial</td>
<td>0.28 ± 0.11</td>
<td>0.44 ± 0.05</td>
</tr>
<tr>
<td>Terminal</td>
<td>0.30 ± 0.11</td>
<td>0.44 ± 0.04</td>
</tr>
<tr>
<td>Internal torsion - Stiffness (Nm/°)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Initial</td>
<td>0.28 ± 0.11</td>
<td>0.44 ± 0.05</td>
</tr>
<tr>
<td>Terminal</td>
<td>0.31 ± 0.09</td>
<td>0.44 ± 0.05</td>
</tr>
<tr>
<td>Neutral zone (0 Nm torque)</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Range (°)</td>
<td>1.84 ± 1.06</td>
<td>0.38 ± 0.23</td>
</tr>
</tbody>
</table>

Non-destructive tests

Bending stiffness was not significantly different between monocortical and bicortical screw constructs in either caudocranial or mediolateral bending (Table 1). Caudocranial bending stiffness was approximately 51% and 67% greater (p = 0.03 and p = 0.01) than mediolateral bending stiffness for monocortical and bicortical screw constructs, respectively. Residual angular displacement at 0 Nm bending moment was not different between screw constructs for mediolateral or caudocranial bending, nor between bending modes. No significant interaction between screw placement and bending direction was found for any variable.

For torsion, the shape of the torque-angular rotation curves was typically non-linear for monocortical constructs and linear for bicortical constructs (Figure 5). Monocortical screw constructs were 30–37% less stiff compared to bicortical screw constructs for initial and terminal, external and internal rotation, stiffnesses (Table 2). Further, monocortical constructs developed a neutral zone that was

* Constructs did not fail in mediolateral bending because the failure bending moment was greater than 60 Nm (the maximum load limit for tests). NA: not applicable.

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D. Demner et al.: Comparison of monocortical and bicortical screws
over 4.5 times larger than that of bicortical constructs.

Failure tests
There were no significant differences between monocortical and bicortical screw constructs for failure variables measured for constructs tested in mediolateral or in caudocranial bending (Table 3).

Constructs were stiffer and stronger at yield in caudocranial bending than in mediolateral bending. Stiffness and yield bending moment were 39% and 25% greater, respectively for caudocranial bending constructs than for mediolateral bending constructs (p <0.001, p = 0.042, respectively). Failure bending moment was greater for mediolateral bending than for caudocranial bending; all caudocranial bending constructs failed at bending moments between 19 and 44 Nm, and all mediolateral bending constructs failed at a bending moment >60 Nm (Fischer’s Exact Test, p <0.001).

Failure mode
Bicortical constructs failed in caudocranial bending by longitudinal fracture through the screw holes in the proximal fragment in three of four constructs (Figure 7) and by longitudinal fracture through the distal screw hole of the distal fragment in the remaining construct. Monocortical constructs failed by screw pull-out of the distal fragment in three of four constructs and by loosening of one screw in the proximal bone fragment in the remaining construct. Failure was not apparent in mediolateral bending constructs because failure loads exceeded the 60 Nm limit of the load cell, however residual bending was apparent in all specimens after unloading (Figure 8).

Discussion
Bending and torsional mechanical properties of gap ostectomized tibia, bridged with a medial LCP plate were compared between monocortical and bicortical screw plate fixations. Construct stiffness was not different between monocortical and bicortical screw type in mediolateral and caudocranial bending directions; however, bicortical constructs were stiffer than monocortical constructs in torsion. Screw placement affected caudocranial bending failure mode. Bicortical constructs failed by bone fracture through screw holes, whereas monocortical constructs failed by screw pull-out. Screw constructs were stiffer and had higher yield strength when failure occurred in caudocranial bending compared to mediolateral bending.

Physiological loading conditions in canine tibias with a fixation plate are unknown, but the tibia experiences tensile strain on the cranial surface and compressive strain on the caudal surface in vivo at a walk (11). In the current study, because of the nature of the gap model construct (plate on the medial side and gap creating loss of support on the lateral side), loading probably creates bending with the medial side in tension and the lateral side in compression. Therefore, plated tibias were tested to induce tensile strain on the cranial surface (caudocranial bending) and on the medial surface (mediolateral bending). Tibias were also loaded in internal and external torsion to simulate another loading mode which these constructs could experience.

Bicortical screw constructs were stiffer in torsion than monocortical screw constructs, and bicortical screw construct stiffness approached that of normal intact tibias. One other study found a mean maximum torsional stiffness of 0.53Nm/° for normal intact canine tibias tested to failure (12). In the current study with an LCP plate applied to bridge an ostectomy gap, bicortical screw constructs had 83% of the stiffness, while monocortical screw constructs had 57% of the stiffness of the normal intact tibias described previously by others (12). Although the construct torsion curves appeared to reach a linear terminal stiffness, construct tibias were not tested to failure; therefore construct stiffness may be a conservative estimate of maximum torsional stiffness as reported by Tyler and colleagues (12). Greater torsional stiffness for bicortical constructs is probably related to greater working length for screws an-
chored in both cortices (13). Non-destructive cyclic torsional loading also resulted in a larger neutral zone between internal and external torsion for monocortical constructs than bicortical constructs. Therefore, bicortical screw constructs are likely to have greater rotational stability and less motion during physiological loading circumstances.

Although no differences were observed between monocortical and bicortical screw constructs in bending mechanical properties, the mode of failure for caudocranial bending was different between screw constructs. Bicortical screw constructs failed by bone fracture through the screw holes. Monocortical screw constructs failed by screw loosening or screw pull-out. Screw-plate pull-out is a common failure mode for LCP constructs (14, 15). The pull-out resistance of monocortical locking screws is about 70% of that of bicortical locking screws (16). This finding is consistent with the mode of failure results in the current study, where monocortical screws were more likely to pull-out than to fracture the bone. In contrast, bicortical constructs failed by bone fracture through screw holes. Apparently, the stress-riser at the screw holes was the weakest interface of the construct.

Characteristics of the locking screws, including screw length and thread height, may play important roles in screw pull-out. Appropriate selection of monocortical screw length relative to bone diameter is critical to overcome a pitfall with self-tapping locking head screw placement and screw length relative to bone diameter (16). If the screw is too short, the threads in the cis-cortex may not have enough purchase and the implants will be prone to failure by pull-out. In contrast, if the monocortical screw is too long, the screw tip may push off from the trans-cortex, thus destroying the thread in the cis-cortex (16). The monocortical screws used in our mechanical study were chosen to be a single standard length for all bones based on macroscopic and radiographic observations, not on objective measurements of the cis-cortex. The 12 mm screw length ensured that full threads engaged the cis-cortex in the current study. Screw security in the cortex is optimized when screws are long enough so that the screw flute and tapered threads are entirely within the medullary cavity, as in the current study (16). Furthermore, locking screws have a shallow thread profile to maximize screw core diameter for bending and shear strength (1, 3, 7, 17). It is possible that screw pull-out could be minimized for monocortical screws by optimizing thread pitch and thread profile design to balance the need to maximize screw bone interface while maintaining sufficient core diameter to resist bending and shear of the screw in the fixed angle construct of a locking compression plate system.

Constructs were stiffer and stronger in caudocranial bending compared to mediolateral bending. This is probably due to the greater moment of inertia of the plate in the caudocranial direction. Mediolateral bending of the plate could be minimized by addition of an intramedullary pin or a lateral buttress plate.

This study represents an ex vivo model of a canine tibial fracture, and as such, it has limitations. In vivo factors such as bone remodeling were not taken in consideration. Physiological loading cannot be replicated because it is unknown, therefore important features were captured by simplifying loading to bending and torsion. In a clinical setting, construct failure is often due to cyclic fatigue, therefore fatigue testing would give additional insight.

In summary, monocortical screw constructs had similar in vitro mechanical stability to bicortical screw constructs in low bending loads, such as may be seen at a walk; however, bicortical constructs were stiffer in torsion. Only the monocortical locking screw system can be used when the medullary canal is partially filled with an intramedullary rod or when self-tapping, self-drilling screws are used (16). A clear contraindication for unicortical screw placement in humans is in anatomical locations exposed to high rotational forces (9, 16). The use of a monocortical screw system avoids traversing the medullary canal and the intramedullary pin. Also the sharp tips of the self-drilling screws may cause neurovascular or soft tissue damage if they traverse across the far cortex. Bicortical screw placement is preferred over monocortical screws in metaphyseal bone or poor quality cortical bone (1, 7, 16, 18, 19). In fact, the only advantages of using unicortical screws may be to avoid intramedullary implants or penetration of the bone and periosteum on the far side of the bone, such as in the vertebral bodies, in which bicortical purchase may produce life-threatening complications.

Conflict of interest

K. Hayashi, effective since June 2014, will provide consulting services for DePuy Synthes, and will serve as an instructor for the DePuy Synthes Lab at the Association for Veterinary Orthopedic Research and Education meeting on January 15, 2015.

References