Comparison of the mechanical behaviors of semicontoured, locking plate–rod fixation and anatomically contoured, conventional plate–rod fixation applied to experimentally induced gap fractures in canine femora

Clara S. S. Goh, BVSc; Brandon G. Santoni, PhD; Christian M. Puttlitz, PhD; Ross H. Palmer, DVM, MS

Objective—To compare the mechanical behaviors of a semicontoured, locking compression plate–rod (LCP-rod) construct and an anatomically contoured, limited-contact dynamic compression plate–rod (LC–DCP-rod) construct applied to experimentally induced gap fractures in canine femora.

Sample Population—16 femora from 8 cadaveric dogs.

Procedures—8 limbs from 8 dogs were assigned to the LCP-rod construct group or the LC–DCP-rod construct group. In each femur, a 39-mm mid-diaphyseal ostectomy was performed at the same plate location and the assigned construct was applied. Construct stiffness and ostectomy gap subsidence were determined before and after cyclic axial loading (6,000 cycles at 20%, 40%, and 60% of live body weight [total, 18,000 cycles]). Three constructs from each group further underwent 45,000 cycles at 60% of body weight (total, 63,000 cycles). Following cyclic loading, mode of failure during loading to failure at 5 mm/min was recorded for all constructs.

Results—After 18,000 or 63,000 cycles, construct stiffness did not differ significantly between construct groups. No implant failure occurred in any construct that underwent 63,000 cycles. In both construct groups, ostectomy gap subsidence similarly increased as axial load increased but did not change after 18,000 cycles. Mean ± SEM loads at failure in the LCP-rod (1,493.83 ± 200.12 N) and LC–DCP-rod (1,276.05 ± 156.11 N) construct groups were not significantly different. The primary failure event in all constructs occurred at the screw hole immediately distal to the ostectomy.

Conclusions and Clinical Relevance—Biomechanically, the semicontoured LCP-rod construct is similar to the anatomically contoured LC–DCP-rod system. (Am J Vet Res 2009;70:23–29)

Abbreviations

<table>
<thead>
<tr>
<th>Abbreviation</th>
<th>Definition</th>
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<tr>
<td>BMD</td>
<td>Bone mineral density</td>
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<tr>
<td>IMR</td>
<td>Intramedullary rod</td>
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<tr>
<td>LC–DCP</td>
<td>Limited-contact dynamic compression plate</td>
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<td>LCP</td>
<td>Locking compression plate</td>
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<tr>
<td>MIPO</td>
<td>Minimally invasive plate osteosynthesis</td>
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Complex, nonreconstructable fractures of the femur are one of the most commonly encountered and challenging types of fractures in dogs. Currently, biological bridging osteosynthesis is the treatment of choice for many comminuted nonreconstructable fractures. Bridge plating preserves biological potential for healing while creating an internal splint and stable fixation. Spatial alignment of the bone is restored without precise anatomic reconstruction of bone fragments, thereby preserving blood supply from soft tissue attachments. Newer techniques of percutaneous plate application (termed MIPO) may further hasten bone healing as a result of improved periosteal perfusion. From a mechanical perspective, bridge plating demands maximal performance from the bridging implants, which may be subject to plastic deformation or breakage early in the postoperative period or fatigue failure over time, depending on the healing potential of the patient. Addition of an IMR to the construct reduces stress applied to the plate; as much as a 10-fold extension of the fatigue life of the bone plate can be achieved. The IMR can also assist in reestablishment of the spatial alignment of the limb and main fracture segments. Requirements of the IMR and screws for available bone stock often dictate the

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need for unicortical placement of many of the plate screws.

The use of conventional bone plates in bridging osteosynthesis has several biological, mechanical, and surgical application limitations, especially when MIPO techniques are used. Construct stability is reliant on compression between the conventional plate and underlying bone that is generated by adequate screw fixation. When achieved, this compression of the plate against the bone disrupts periosteal vascular supply and can retard healing of the underlying bone. Where insufficient screw-to-bone fixation is available (because of poor bone quality, short bone fragment in juxta-articular fractures, or unicortical screw purchase), cyclic loading of conventional plate-bone constructs can lead to secondary loss of fracture alignment and stability. Additionally, conventional screws are primarily stabilized by their bony purchase. Bicortical purchase of screws imparts greater angular stability, compared with stability associated with unicortical application, wherein screws may toggle about their thin cortical purchase. Finally, precise anatomic contouring of conventional plates is required for proper spatial limb alignment and to prevent displacement of the fracture when the screws are tightened. This is not always feasible when MIPO techniques are used. The ideal plating system would allow percutaneous application without requiring strict anatomic contouring of the bone plate and provide angular stability of screws, even when concurrent use of an IMR necessitates unicortical screw purchase.

The recently introduced LCP uses special screws that lock (at a fixed angle) to the plate. In contrast to conventional plate fixation, the rigid link between the LCP and screwhead obviates the need for the bone plate to be compressed to the bone, minimizes disruption of periosteum and soft tissues, and prevents primary loss of fragment alignment caused by inexact plate contouring. The construct functions as an internal fixator, for which precise anatomic contouring of the plate is not mandatory to maintain fracture alignment and stability. Additionally, the angular stability of the locked screws allows superior fixation, compared with conventional plate constructs, when unicortical fixation is required. These attributes are ideal for bridging osteosynthesis and MIPO techniques, and LCPs have been successfully used in human medicine since their clinical release in 2000. In a study of experimentally induced gap fractures in canine femora, anatomic contouring LCP fixation was associated with greater gap stiffness (in lateral-medial bending) than anatomically contoured conventional fixation. However, in that study, the common clinical practice of combination fixation with an IMR and unicortical screw fixation was not simulated, and anatomic contouring of plates is not always feasible for newer MIPO strategies.

Although there are many potential merits to the use of LCPs in orthopedic procedures in dogs, the authors’ knowledge, there are no published reports of the mechanical behavior of unicortical LCP constructs applied with or without an adjunctive IMR in dogs’ bones. The purpose of the study reported here was to compare the mechanical behaviors of a semicontoured LCP-rod construct and an anatomicomatically contoured LC–DCP-rod construct applied to experimentally induced gap fractures in canine femora. The mechanical behaviors were examined under the conditions of axial loading, prior to and during progressive cyclic loading. We hypothesized that, compared with anatomically contoured LC–DCP-rod constructs, semicontoured LCP-rod constructs would impart equivalent construct stiffness and fracture gap stability but would be more prone to catastrophic failure of the entire lateral cortical wall when loaded to failure.

Materials and Methods

Construct preparation—Sixteen femora were obtained from 8 large-breed dogs (weight, 25 to 36 kg) that were euthanized (via an IV overdose of pentobarbital for reasons unrelated to the study) in accordance with Colorado State University Institutional Animal Care and Use Committee policy. Femora were stripped of soft tissues, evaluated for gross bony abnormalities, double-wrapped in gauze soaked in physiologic saline (0.9% NaCl) solution, placed in double-sealed plastic bags, and frozen at −20°C until hardware application and biomechanical testing were performed. Dual-energy x-ray absorptiometry scans were performed on each femur. The total mean BMD for each femur was determined from 2 craniocaudal and 2 lateromedial views.

One femur from each cadaver was randomly assigned to receive an 11-hole, 3.5-mm, anatomically contoured, LC–DCP with an IMR (LC–DCP-rod construct group). The other femur from each cadaver was assigned to receive an 11-hole, 3.5-mm, semicontoured LCP with an IMR (LCP-rod construct group). Each plate was contoured to the lateral surface of its assigned femur. Screw holes in the plates were numerically designated as 1 through 11 in the proximal to distal direction. In each construct, unicortical screws were positioned in holes 2, 3, 4, 8, 9, and 10; LCPs were applied over a temporary 3-mm spacer while locking screws were inserted to ensure a uniform plate-to-bone gap (for the purposes of the study; this was considered semicontouring of the plate). Following generation of a 39-mm mid-diaphyseal osteotomy beneath plate holes 5, 6, and 7, an IMR that measured 40% of the mid-diaphyseal diameter of the femur was implanted normograde from the intertrochanteric fossa and seated in the distal metaphysis. In the LC–DCP-rod construct group, conventional cortical screws were angled around the IMR in holes 1 and 11 to achieve bicortical purchase as is commonly performed in clinical settings. Unicortical locking screws were inserted in identical positions in the LCP-rod construct group (Figure 1). Radiographic views of each femur were obtained to confirm uniformity of hardware application and the absence of bony abnormalities.

Mechanical testing—Specimens were thawed at room temperature (approx 25°C) for 24 hours prior to biomechanical testing. The greater trochanter was excised from each construct to facilitate the phys-
ologic axial load application. Compression loading was applied directly to the femoral head by use of a custom-designed apparatus that included an ultra-high–molecular weight polyethylene cup (diameter slightly larger than that of the femoral head) and a servohydraulic material testing machine (Figure 2). Three wood screws were placed in the femoral condyle of each construct to promote increased rotational stability, and the femur was potted by use of polymethyl methacrylate in a custom-designed aluminum box. In all tested constructs, the femoral axis was aligned parallel to the axis of loading in both the frontal and sagittal anatomic planes.

Construct stiffness and ostectomy gap subsidence were determined for each construct prior to cyclic axial loading by use of a load control protocol, which consisted of a single loading ramp at 40 N/s to a maximum of 20% of live body weight. Thereafter, periods of cyclic loading at 2 Hz were performed for 6,000 cycles at loads of 20%, 40%, and 60% of live body weight (total, 18,000 cycles) to simulate progressively increased weight bearing in the early postoperative period. Following each loading interval, nondestructive construct stiffness and ostectomy gap subsidence were determined by use of the aforementioned load control protocol. Data were collected at 60 Hz. Three matched pairs of constructs representing the limbs assigned to LCP-rod and LC–DCP-rod construct groups from 3 dogs underwent an additional testing protocol (45,000 cycles at 60% of body weight [total, 63,000 cycles/construct]) to simulate a 3- to 6-week convalescence period and to elucidate the effect, if any, of extended exposure to simulated loading on the structural behavior of the construct.

Construct stiffness was defined as the slope of the linear region of the uniaxial force-versus-displacement curve. Translation across and rotation about the ostectomy gap were determined for each construct at each interval of nondestructive uniaxial testing by use of a motion analysis system. Specifically, 2 marker triads were rigidly attached to each femur specimen; 1 was located 10 mm proximal to the proximal border of the ostectomy, and the other was located 10 mm distal to the distal border of the ostectomy (Figure 2). Assessment of ostectomy subsidence in the plane of loading at each loading interval was accomplished by use of single markers placed immediately adjacent to the proximal and distal ostectomy borders on the cranial surface of each construct. Three cameras tracked the 2 marker triads and 2 single markers, and positional data were captured at 60 Hz. Data captured by the servohydraulic material testing machine load cell were synchronized with the marker data to obtain force-versus-local rotation and force-versus-displacement curves.

Following completion of either 18,000- or 63,000-cycle loading, each construct was loaded to failure by use of a displacement control protocol at a...
displacement rate of 5 mm/min. Failure was defined as the point of major yield in the load-versus-displacement curve, and the mode of failure for each construct was documented. By use of a software program, yield energy was determined by integrating tertiary regression curves fitted to each force-versus-displacement curve and incorporating the appropriate upper and lower displacement limits.

Statistical analysis—A Student paired \( t \) test was used to detect differences in total BMD between left and right femora for the cranio-caudal and mediolateral views. ANOVA was then used to determine whether there was a significant difference in total BMD among cadavers. A 2-way repeated-measures ANOVA was used to detect differences between the femoral fixation methods (LCP-rod and LC–DCP-rod constructs) with regard to construct stiffness, local gap subsidence, and in-plane (frontal) rotation about the ostectomy after completion of the 18,000- or 63,000-cycle loading procedures. A paired \( t \) test was used to determine the effect of femoral fixation method on ramp-to-failure variables. The significance level for all statistical analyses was set at a value of \( P < 0.05 \). All statistical tests were performed by use of statistical software.

**Results**

No gross abnormalities were detected in any of the 16 femora, although mild variations in femur length and shape were evident. There was no significant \( (P > 0.05) \) difference in BMD between left and right femora. A significant \( (P < 0.05) \) variation in femur BMD among cadavers was detected in the cranio-caudal view. Mean ± SEM BMDs were \( 0.67 \pm 0.06 \) g/cm\(^2\) and \( 0.75 \pm 0.06 \) g/cm\(^2\) for the cranio-caudal and lateromedial views, respectively. Radiography confirmed relative uniformity in plate, screw, and IMR placement, and no bony abnormalities were detected.

Overall, there was no significant difference in construct stiffness between the construct groups after the short-term 18,000-cycle testing procedure (Table 1). Likewise, for the paired constructs that underwent additional cyclic loading (45,000 cycles), there was no significant \( (P > 0.05) \) difference in construct stiffness between construct groups. Furthermore, there was no evidence of implant failure in any of the 6 constructs that underwent the long-term 63,000-cycle testing. Mean pre-cycling construct stiffness in the LCP-rod and LC–DCP-rod construct groups was \( 737.11 \pm 67.81 \) N/mm and \( 775.57 \pm 72.23 \) N/mm, respectively. Temporally, the stiffness of each construct increased from the precycling condition to the first postcycling construct stiffness determination (ie, after \( 0 \) to \( 6,000 \) cycles at 20% of body weight), although this finding was not significant for either construct group \( (P = 0.627 \text{ and } P = 0.684 \) for the LCP-rod and LC–DCP-rod construct groups, respectively). With continued testing at the remaining intervals \( 6,001 \text{ to } 12,000 \) cycles at 40% of body weight \( [8 \text{ femora/construct group}] \), \( 12,001 \text{ to } 18,000 \) cycles at 60% of body weight \( [8 \text{ femora/construct group}] \), and \( 18,001 \text{ to } 63,000 \) cycles at 60% of body weight \( [3 \text{ femora/construct group}] \), construct stiffness for both the LCP-rod and LC–DCP-rod constructs remained statistically equivalent.

No difference \( (P > 0.05) \) in ostectomy gap subsidence at any of the progressively increased loads over time was detected between the construct groups (Table 1). In general, ostectomy gap subsidence increased with progressive increases in axial load for both construct groups. In the LCP-rod construct group, ostectomy gap subsidence at 40% of body weight was significantly \( (P = 0.006) \) increased, compared with findings at 20% of body weight; likewise, ostectomy gap subsidence at 60% of body weight was significantly \( (P = 0.011) \) increased, compared with findings at 40% of body weight. In the LC–DCP-rod construct group, ostectomy gap subsidence at 60% of body weight was significantly \( (P = 0.045) \) greater than the value at 20% of body weight. No change in ostectomy gap subsidence was identified for either group when constructs were tested after 18,000 cycles at 60% of body weight and again after 63,000 cycles at 60% of body weight \( (P = 0.824 \text{ and } P = 0.940 \) for the LCP-rod and LC–DCP-rod construct groups, respectively). Rotation about the 39-mm ostectomy was most pronounced in the plane of construct loading (ie, the frontal plane). Between the construct groups,

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<table>
<thead>
<tr>
<th>Testing period</th>
<th>LCP-rod</th>
<th>LC–DCP-rod</th>
<th>( P ) value*</th>
<th>LCP-rod</th>
<th>LC–DCP-rod</th>
<th>( P ) value*</th>
</tr>
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<tbody>
<tr>
<td>0 to 6,000 cycles at 20% of body weight</td>
<td>861.08 ± 83.60</td>
<td>874.71 ± 81.30</td>
<td>0.881</td>
<td>30.919 ± 3.99</td>
<td>38.33 ± 4.61</td>
<td>0.75</td>
</tr>
<tr>
<td>6,001 to 12,000 cycles at 40% of body weight</td>
<td>773.78 ± 97.00</td>
<td>908.38 ± 97.80</td>
<td>0.322</td>
<td>69.27 ± 13.80†</td>
<td>57.23 ± 7.46</td>
<td>0.35</td>
</tr>
<tr>
<td>12,001 to 18,000 cycles at 60% of body weight</td>
<td>794.31 ± 110.50</td>
<td>899.31 ± 103.80</td>
<td>0.439</td>
<td>104.43 ± 22.69†</td>
<td>71.14 ± 8.16†</td>
<td>0.30</td>
</tr>
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</table>

*Applies to comparison of construct group values for a given variable within a testing period. For this variable, value is significantly \( (P < 0.05) \) increased, compared with the value for the testing period of 0 to 6,000 cycles at 20% of body weight. For this variable, value is significantly \( (P < 0.05) \) increased, compared with the value for the testing period of 6,001 to 12,000 cycles at 40% of body weight.
significant differences in rotation about the ostectomy followed a pattern similar to that determined for ostectomy gap subsidence.

When loaded to failure after completing the 18,000- or 63,000-cycle procedure, constructs in the LCP-rod and LC–DCP-rod construct groups failed at mean ± SEM yield loads of 1,493.83 ± 200.12 N and 1,276.05 ± 156.11 N, respectively; these values were not significantly (P = 0.330) different. Displacement at yield and yield energy were also not different between the construct groups (Table 2). Mode of failure of the 2 constructs was similar. The primary failure event for the most (7/8) constructs in each construct group occurred at the level of the screw hole immediately distal to the ostectomy border (hole 8); failure was either screw pullout or fracture of the lateral cortical wall (initiated at the screw hole). Progressive loading beyond the primary yield event in these constructs resulted in 1 of 2 secondary modes of failure: lateral cortical wall failure in the distal femoral segment or plastic deformation of the plate, IMR, or both. The primary failure event in the remaining LCP-rod construct was plate deformation at the level of the proximal border of the ostectomy. The primary failure event in the remaining LC–DCP-rod construct was fracture at the level of the distal metaphyseal bicortical screw (screw 11), which, with continued testing, propagated through the bone and resulted in collapse of the femur's medial cortical wall.

Discussion

Use of LCPs has inherent biological and surgical application advantages over conventional plating used in biological bridging osteosynthesis and MIPO strategies. On the basis of the results of the present study, there were no significant mechanical differences in construct stiffness and ostectomy subsidence for simulated nonreconstructable femoral diaphyseal fractures stabilized by use of semiconoured LCP-rod constructs or anatomically contoured LC–DCP-rod constructs.

As with any ex vivo biomechanical testing of orthopedic constructs, the conditions of patient loading of the limb cannot be exactly reproduced. In our study, a quasi-physiologic combination of axial compression, a bending moment, and weak torque moment were generated by application of axial load directly to each eccentrically positioned femoral head. The number of cycles in an individual limb of a dog that is taken for a walk for 5 to 10 minutes 4 times daily is estimated to be 1,500 to 3,000 cycles/d.18 Peak vertical forces of 40% to 50% of body weight are achieved in the hind limbs of clinically normal dogs during walking.20,21 Thus, the study protocol of progressive, cyclic nondestructive axial loading (at 20%, 40%, and 60% of body weight) simulated progressively increasing postoperative limb loading during a 6- to 12-day period (18,000 cycles) and during a 21- to 42-day period (63,000 cycles). Progressive fracture zone consolidation during convalescence eventually reduces the load borne by the implant portion of the construct. The fact that none of either type of construct failed during the long-term, progressive-load nondestructive testing procedure suggests that both constructs should remain intact in dogs during a typical postoperative healing period provided appropriate activity restrictions are enforced. In clinical settings, the client and patient are not always compliant with strict activity restriction. Peak vertical forces of 76% to 107% of body weight are achieved in the hind limbs of clinically normal Labradors and Greyhounds running at a trot.22 This would be equivalent to 223 to 315 N in a typical 30-kg large-breed dog. Given that the LC–DCP-rod and LCP-rod constructs failed at mean yield loads of approximately 1,200 and 1,500 N, respectively, in the present study, it is still unlikely that either construct would fail catastrophically in the early postoperative period as a result of an occasional lapse in activity restriction.

There was no significant difference in the construct stiffness between the construct groups after application of the short-term 18,000-cycle or long-term 63,000-cycle loading. This finding was mirrored by the ostectomy subsidence data, which also revealed no significant differences between construct groups, although incremental increases in subsidence occurred with increasing body weight in both groups. Ostectomy gap subsidence is an indirect measure of interfragmentary strain and is inversely related to gap stiffness. The optimal amount of strain ranges between the minimum required for callus induction and maximum that allows bony bridging.12,21 According to this interfragmentary strain theory, an elastic flexible fixation (eg, biological bridge plating) is compatible with indirect healing, provided that very small unstable gaps of high strain are avoided.22 Although not significantly different, the LC–DCP-rod construct consistently had higher stiffness and lower ostectomy gap subsidence than the LCP-rod construct. This may suggest that the LCP-rod construct may promote earlier callus formation; however, in vivo comparison of the 2 constructs is needed.

<table>
<thead>
<tr>
<th>Variable</th>
<th>LCP-rod construct</th>
<th>LC–DCP-rod construct</th>
<th>P-value*</th>
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<tbody>
<tr>
<td>Yield load (N)</td>
<td>1,493.83 ± 200.12</td>
<td>1,276.05 ± 156.11</td>
<td>0.330</td>
</tr>
<tr>
<td>Displacement at failure (mm)</td>
<td>5.46 ± 0.47</td>
<td>5.62 ± 0.87</td>
<td>0.876</td>
</tr>
<tr>
<td>Yield energy (N-mm)</td>
<td>5,462.90 ± 808.38</td>
<td>4,696.28 ± 1,149.65</td>
<td>0.512</td>
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</table>

*Applies to comparison of construct group values for a given variable; a value of P = 0.05 was considered significant.
In the present study, the LCP was semicontoured by use of a 3-mm offset from the bone; this was designed to emulate the clinical practice of semicontouring from the radiographic views of the contralateral limb without introducing a variable plate-to-bone distance within the construct group. The complete lack of plate-to-bone contact represents the worst-case scenario with regard to clinical application. As the distance from the plate to bone increases from 2 to 6 mm, both torsional rigidity and axial stiffness decrease by 10% to 15%.24 In the present study, no difference in the construct stiffness was evident between construct groups; thus, the effects of the 3-mm offset were either countered by the increased mediolateral bending stiffness of the locking plate25 or obscured by addition of the IMR to the constructs. It is frequently clinically possible to place bicortical locking screws in the proximal and distal plate holes of an LCP-rod construct, which may be mechanically advantageous. In our study, we elected to use only unicortical locking screw purchase to again simulate the worst-case scenario during clinical application, wherein the angle-stable locking screws may prevent angulations of screws around the IMR. When the constructs were loaded to failure, yield energy and displacement at yield were not significantly different between groups. In 14 of 16 constructs, the primary failure event was either fracture of the lateral wall or screw pullout at the screw hole distal to the distal border of the osteotomy. Although the mode of failure in clinical settings is likely to differ somewhat from this ex vivo situation, finite element analysis has revealed highest stress concentrations for the screws closest to the fracture gap in applications of the bridge plating technique.24 The closer the screw is positioned toward the fracture gap (ie, the shorter the plate's working length), the stiffer the construct becomes under axial compression, especially in fractures with a large gap.24 Additionally, failures associated with screw pullout or cutout of the implants are less likely to occur if screws with angular stability are used.13 This is explained by the reduction of localized stress concentration at the bone-implant interface. Pullout forces are transferred into compression and shear forces between the screws and adjacent osseous structures.13 Nonetheless, no significant differences in failure characteristics between the LCP-rod and LC–DCP-rod construct groups were evident in our study.

To simulate a type of nonreconstructable comminuted femur fracture that is frequently encountered clinically, we used femora from dog cadavers in each of which an ostectomy was performed and a gap created. Biological variation in dog breed and age could not be controlled because of dependence on cadaver donation. Differences in BMD (craniocaudal view) and anatomic variance in femur length and shape between limbs in matched pairs likely mimic normal clinical variation. Although use of homogenous synthetic bone models would have eliminated biological variation, assessment of the mechanical performance of these fixation systems in canine bones was our primary interest, and a synthetic model may have prevented meaningful interpretation of the study data regarding the clinical application of the LCP-rod construct in dogs. Additionally, the contralateral femur of each dog also acted as an internal control specimen; no significant differences in BMD between left and right femora were detected, as anticipated on the basis of a concept validated by Markel et al.27,28

The use of an IMR to maintain approximate axial alignment in combination with a conventional bone plate in nonreconstructable fractures in dogs has been established.27,28 The similarity in mechanical performance of the LCP-rod construct and that of the clinically validated LC–DCP-rod construct determined in the present study suggests that the LCP-rod construct could be successfully applied to a similar spectrum of fractures. Additionally, technical differences in the use of locking plates in general may simplify application when MIPO strategies are used. Although the plate-rod combination is well suited to complex, nonreconstructable fractures of the femur, the IMR may not be advantageous in other fracture configurations for which more flexible fixation is desired.

Locking plates may lend themselves to minimally invasive fixation techniques that provide for spatial alignment of the fractured bone, sufficiently stable fixation, and undisturbed indirect bone healing. The semicontoured LCP-rod construct is biomechanically similar to the more conventional anatomically contoured LC–DCP-rod construct. Because of inherent biological and surgical application advantages of locking plates, clinical investigations of the LCP-rod construct for biological bridge plating of nonreconstructable fractures of the femoral shaft in dogs are warranted.

References