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Ex vivo cyclic mechanical behaviour of 2.4 mm locking plates compared with 2.4 mm limited contact plates in a cadaveric diaphyseal gap model

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Keywords

Cyclic mechanical testing, locking plate, limited contact plate, gap model, femur, dogs

Summary

Objectives: To compare the mechanical properties of locking compression plate (LCP) and limited contact dynamic compression plate (LC-DCP) constructs in an experimental model of comminuted fracture of the canine femur during eccentric cyclic loading.

Methods: A 20 mm mid-diaphyseal gap was created in eighteen canine femora. A 10-hole, 2.4 mm stainless steel plate (LCP or LC-DCP) was applied with three bicortical screws in each bone fragment. Eccentric cyclic loadings were applied at 10 Hertz for 610,000 cycles. Quasistatic loading / unloading cycles were applied at 0 and 10,000 cycles, and then every 50,000 cycles. Structural stiffness was calculated as the slope of the linear portion of the load-displacement curves during quasistatic loading / unloading cycles.

Results: No bone failure or screw loosening occurred. Two of the nine LCP constructs failed by plate breakage during fatigue testing, whereas no gross failure occurred with the LC-DCP constructs. The mean first stiffness of the LCP constructs over the course of testing was 24.0% lower than that of constructs stabilized by LC-DCP. Construct stiffness increased in some specimens during testing, presumably due to changes in bone-plate contact. The first stiffness of LC-DCP constructs decreased by 19.4% and that of locked constructs by 34.3% during the cycling period. A biphasic stiffness profile was observed: the second stiffness was significantly greater than the first stiffness in both groups, which allowed progressive stabilization at elevated load levels.

Clinical significance: Because LCP are not compressed to the bone, they may have a longer working length across a fracture, and thus be less stiff. However, this may cause them to be more susceptible to fatigue failure if healing is delayed.

Introduction

Comminuted diaphyseal fractures are frequently encountered in veterinary medicine and present challenges for orthopaedic surgeons. Femoral fractures represent 45% of long-bone fractures and diaphyseal fractures occur in 56% of cases (1, 2). The reference treatment for long-bone fractures in small animals is internal fixation and, for many years, the best option has been to use Dynamic Compression Plates[®]. Many other implants have since been developed to limit the potential complications and facilitate treatment of comminuted fractures. The Limited Contact-Dynamic Compression Plate[®] (LC-DCP) was designed to minimize contact between the plate and bone while allowing axial dynamic compression. Because the plate has an even area moment of inertia, stress concentration at the holes is reduced. One of the latest implants to be developed is the Locking Compression Plate[®] (LCP) in which the combination hole accepts both standard cortical screws and locking screws.

The LCP may provide a structural advantage over the LC-DCP because of the additional stability resulting from the creation of a fixed-angle rigid construct. The bone vascular supply is protected due to the absence of bone-plate compression. Precise anatomic contouring of the LCP is not required so it is easier to use in biological osteosynthesis. These features could make the LCP well suited to the stabilization of comminuted diaphyseal fractures (3-7).

Comparisons of the biomechanical characteristics of different implants, and

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Table 1 Number of constructs for which a second or a third stiffness appeared and the time of their appearance.

	Number of constructs		Time of appearance (range of number of cycles)	
	2 nd stiffness	3 rd stiffness	2 nd stiffness	3 rd stiffness
LCP	8 out of 9	3 out of 8	10,000 – 410,000	160,000 – 460,000
LC-DCP	6 out of 9	2 out of 6	10,000 – 310,000	560,000

LCP: Locking compression plate; LC-DCP: Limited contact dynamic compression plate.

more particularly of LCP and LC-DCP, have been reported in the literature (3, 4, 6, 8-15). Many studies with a wide range of protocols have involved bone or synthetic bone under cyclic conditions (3-5, 9, 16-27). To our knowledge, only a few reports have described axial compression cyclic testing with constant load under fatigue conditions (4, 11, 22, 26, 28-31). However only two of these studies compared LCP and LC-DCP, but none specifically evaluated 2.4 mm LCP and LC-DCP (4, 11).

The purpose of this study was to compare the mechanical properties of LCP and LC-DCP constructs in an experimental model of comminuted fracture of canine femora during eccentric cyclic loading.

Materials and methods

Bone specimens

Nine pairs of femora were harvested from adult canine cadavers of the same breed and wrapped in gauze soaked in isotonic saline solution (NaCl 0.9%) before freezing (-20°C). All dogs had been euthanatized for reasons unrelated to orthopaedic disease. All femora were first radiographed to ensure bone maturity and the absence of any bone disease. For testing, the femora were thawed at room temperature (20–22°C) and were kept moist by being wrapped in saline-soaked gauze which was dipped in saline throughout the experiment.

Implants

Ten-hole 2.4 mm stainless steel LCP^a with 2.4 mm self tapping locking screws^a and

10-hole 2.4 mm stainless steel LC-DCP^a with 2.4 mm self-tapping cortical bone screws were used for this study.

Construct assembly

The femora were divided into two groups. Right femora were used for LCP testing, and left femora were used for LC-DCP testing (►Figure 1). For each pair of cadaveric femora, the plates (LCP or LC-DCP) were contoured and applied to the lateral aspect of the bone using bone forceps. Each plate was applied with three bicortical screws in each bone fragment by the same author (AA). Self-tapping 2.4 mm locking screws were used in the LCP group and self-tapping 2.4 mm standard screws were used in the LC-DCP group. All screws were tightened to 0.8 Nm using a torque-limiting screwdriver^a. The proximal screw was inserted in each plate at the level of the minor trochanter. After application of the bone plate, a transverse, mid-shaft femoral 20 mm osteotomy was performed using an oscillating bone saw. Care was taken to ensure that the saw blade had no contact with the plate. The gap between the two fragments was measured. The plates were applied in buttress fashion, leaving four empty holes, two of them being over the fracture gap completely. A new set of implants was used for each construct, and none of the implants were reused for mechanical testing.

Mechanical testing

The distal part of each femur was potted in polyurethane casting resin^b with at least 10 mm between the distal end of the plate and

the resin. This resin was firmly attached to the testing machine base^c. The load was applied on the femoral head through a cup attached to the actuator (►Figure 2).

Each specimen was subjected to cyclical testing under load control. The test began with four quasistatic loading / unloading cycles between 26 and 260 N. Cyclic compression loadings were then applied from 26 to 260N at 10 Hertz for 610,000 cycles. Quasistatic loading / unloading cycles (26 – 260N) were applied at 0 and 10,000 cycles, and then every 50,000 cycles at a loading rate of 26 N/min. The predetermined intervals of 50,000 cycles were automated and the mechanical loading was continuous for 610,000 cycles without any idle period between cyclic and static loading. The maximal load applied was chosen based on the estimated yield point determined on preliminary tests.

The structural stiffness was calculated as the slope of the linear portion of the load-displacement curves during quasistatic loading / unloading cycles. A relative stiffness was calculated as the percent of the difference between the stiffness at each time interval and the initial stiffness, divided by the initial stiffness. When two or three slopes appeared on the load displacement curve during the cycling tests, the inflection points of the curves were mathematically determined.

Data analysis

Statistical comparisons were done by two-way ANOVA, Yates' chi-square tests, and z tests. The results are reported as mean ± standard deviation. Statistical significance was defined as p < 0.05.

Results

The length of the osteotomized gap was 20.1 ± 0.9 mm for the LCP constructs and 20.7 ± 0.8 mm for the LC-DCP constructs.

No bone failure or screw loosening occurred in any of the models tested. Two out of the nine LCP plates broke at the level of the proximal osteotomy site, between

^a Synthes, Etupes, France

^b Rencast FC 52: Gaches chimie, Toulouse, France

^c Electropuls 1000: Instron, High Wycombe, UK

410,000 and 460,000 cycles for the first construct and between 460,000 and 510,000 cycles for the second one. Both failures occurred at the level of the locking part of the third or fourth hybrid plate hole. The seven other constructs survived fatigue testing with no evidence of gross failure. None of the LC-DCP constructs exhibited any gross failure during fatigue testing.

In many cases, the load deformation curves revealed an evolution from constant stiffness over the elastic loading range to a biphasic or triphasic profile (► Figure 3, ► Table 1). A first, second and third stiffness value was therefore calculated. The differences between the two bone-plate constructs for the appearance of a biphasic or triphasic profile were not significant.

The mean first stiffness of the LCP constructs (210.78 ± 64.08 N/m) was significantly lower than that of the LC-DCP constructs (261.43 ± 74.67 N/m) ($p < 0.05$). The mean second stiffness of the LCP constructs (319.36 ± 60.20 N/m) was also significantly lower than the LC-DCP constructs (435.86 ± 159.77 N/m) ($p < 0.05$) (► Table 2).

For the bone-plate constructs that only had a monophasic profile, the stiffness for the LC-DCP construct (297.80 ± 78.37 N/m) was not significantly different from the second stiffness of the LCP construct (319.36 ± 60.20 N/m) ($p = 0.10$). The difference in first stiffness, between the LC-DCP group (302.12 ± 86.63 N/m) and the LCP group (253.78 ± 52.00 N/m) at the beginning of the fatigue test was not significant. At the end of the fatigue tests, the mean first stiffness decreased to 243.46 ± 76.80 N/m for the LC-DCP group and 166.66 ± 39.51 N/m for the LCP group (► Figure 4).

Over the cycling period, the initial first stiffness decreased by 19.4% for the LC-DCP constructs and by 34.3% for the locked constructs (► Figure 5). This decline stabilized at about 310,000 cycles for the LC-DCP constructs but continued to decrease for the LCP constructs. The change in the first stiffness during cyclic testing differed significantly between the two constructs ($p < 0.05$).

The second stiffness was significantly greater than the first stiffness for both groups. The second stiffness in the LCP

Table 2 First and second stiffness (mean \pm SD) over the cycling period, at the first appearance and at the end of cycling for the locking compression plate and limited contact dynamic compression plate constructs.

	LCP	LC-DCP
First stiffness over cycling period (N/m)	210.78 ± 64.08	$261.43 \pm 74.67^*$
First stiffness of the first cycle (N/m)	253.78 ± 52.00	302.12 ± 86.63
First stiffness of the last cycle (N/m)	166.66 ± 39.51	$243.46 \pm 76.80^*$
Second stiffness over cycling period (N/m)	319.36 ± 60.20	$433.86 \pm 159.77^*$
Second stiffness at the first appearance (N/m)	217.15 ± 76.08	597.55 ± 256.74
Second stiffness of the last cycle (N/m)	338.33 ± 54.60	471.89 ± 164.58
Load at appearance of second stiffness (N)	99.65 ± 28.81	86.47 ± 0.41
Displacement at appearance of second stiffness (mm)	0.64 ± 0.26	$0.53 \pm 0.19^*$

*Differences between the two groups are significant ($p < 0.05$).

group increased from 217.15 ± 76.08 N/m to 338.33 ± 54.60 N/m during cyclic testing. The second stiffness in the LC-DCP group initially decreased from 597.57 ± 256.74 N/m to 405.96 ± 165.78 N/m at

310,000 cycles and then increased to 471.89 ± 164.58 N/m up to 610,000 cycles (► Figure 6). The change in the second stiffness during cyclic testing differed significantly between the two plate constructs ($p < 0.05$).

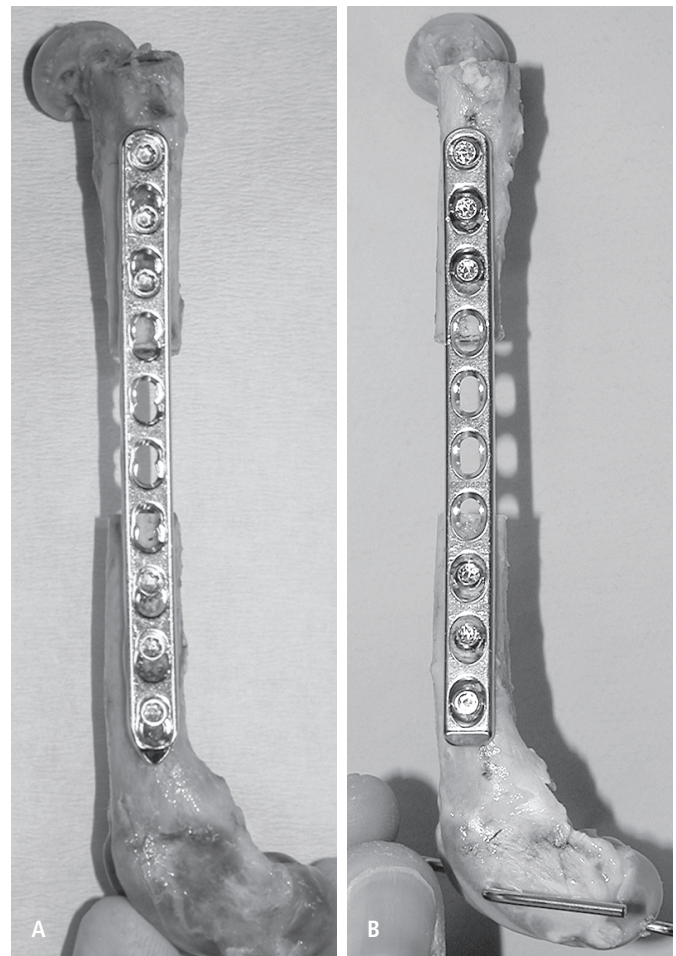


Figure 1 Photographs showing the locking compression plate (A) and limited contact dynamic compression plate (B) constructs.

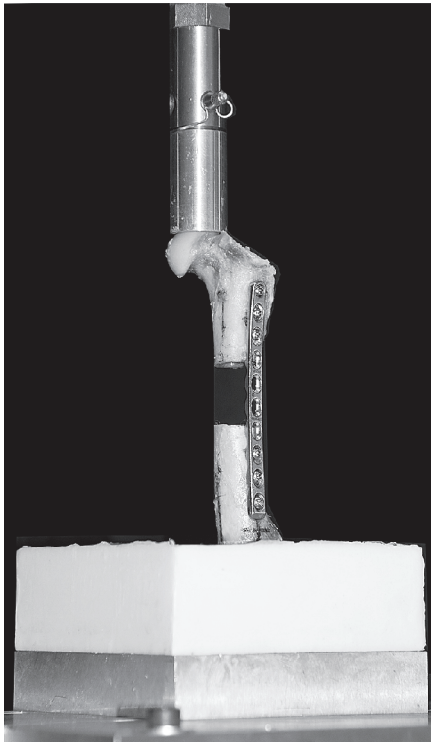


Figure 2 Mechanical test setup for the eccentric loading protocol. The distal femur was secured within the mold which was fixed to the base of the testing machine. The load was applied on the femoral head through a cup attached to the actuator.

There were too few values for the third stiffness (► Table 1) to be taken into account and we decided not to consider these values as being representative of a mechanical characteristic.

The loads that produced the initial changes in stiffness decreased over time. The first time that two slopes were observed, the load was 86.47 ± 0.41 N for the LC-DCP group and 99.65 ± 28.81 N for the LCP group. Both decreased subsequently to 72.76 ± 34.88 N for the LCP group and to 74.73 ± 39.67 N for the LC-DCP group at 310,000 cycles and then increased slightly to 80.31 ± 30.30 N for the LCP group and to 87.21 ± 37.93 N for the LC-

DCP constructs (► Figure 7). These differences were not significant.

The mean displacement observed at the appearance of the second stiffness was significantly higher for the LCP constructs (0.64 ± 0.26 mm) than for the LC-DCP constructs (0.53 ± 0.19 mm) ($p < 0.05$). The maximal displacement was also signifi-

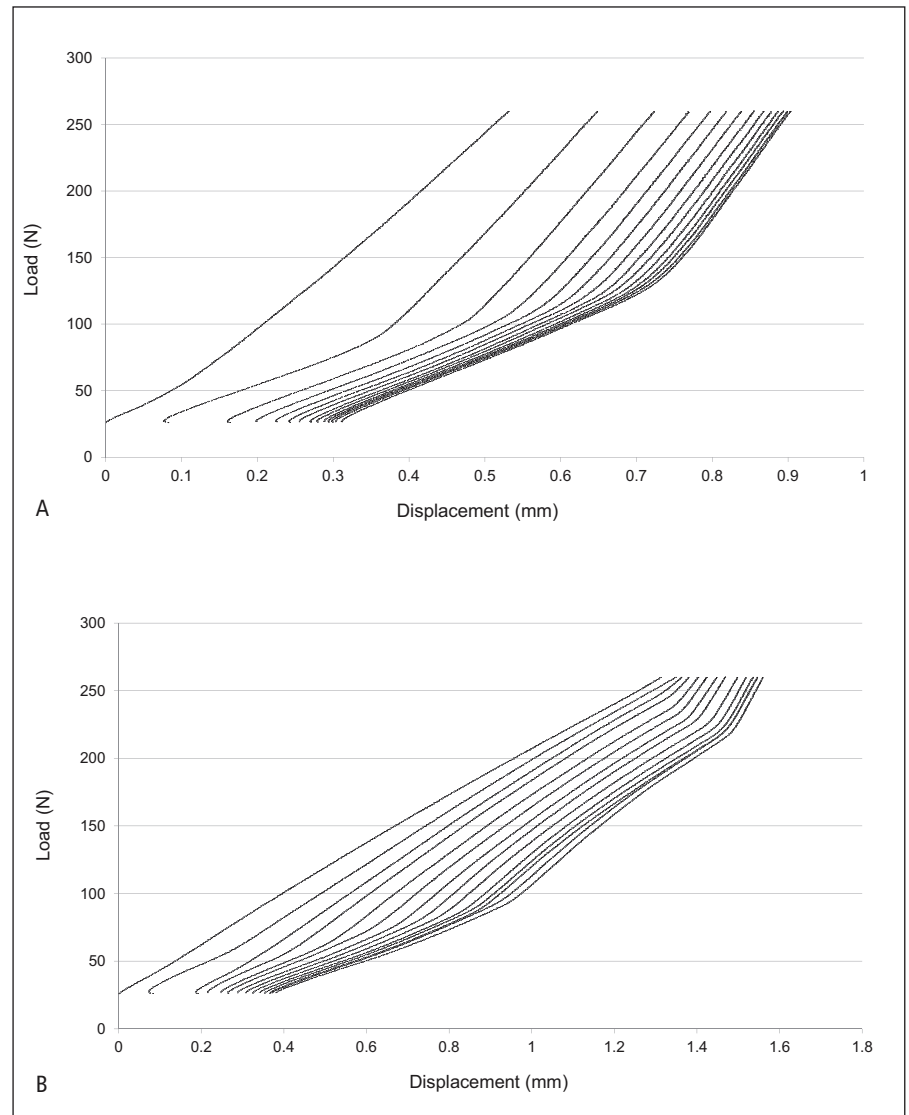


Figure 3 Load deformation curves over 610,000 cycles for the two constructs. **A)** Limited contact dynamic compression plate construct showing the monophasic and biphasic profile. **B)** Locking compression plate construct showing the monophasic and triphasic profile.

cantly greater for the LCP group (0.91 ± 0.18 mm) compared with the LC-DCP group (0.73 ± 0.23 mm) ($p < 0.05$).

Discussion

The mechanical properties of the LCP and LC-DCP bone-plate constructs used in our study were characterized by a quasi-physiological combination of axial compression, a bending moment, and weak torque moment generated by application of axial load directly to each eccentrically positioned femoral head (32).

As previously described by Aguila et al, cyclic loading was considered to be functionally relevant in contributing to implant failure and the disruption of osteosynthesis (3). Throughout the loading/unloading cycles, and according to previous tests performed on pre-test specimens, the load was selected to keep within the linear-elastic domain. The loading rate was low (26 N/min) to suppress the influence of bone viscoelasticity (33, 34). Even though the cycling rate of 10 Hz was greater than that of walking and trotting in dogs, which is between 1.25 Hz and 2.5 Hz, this higher rate was necessary to achieve 600,000

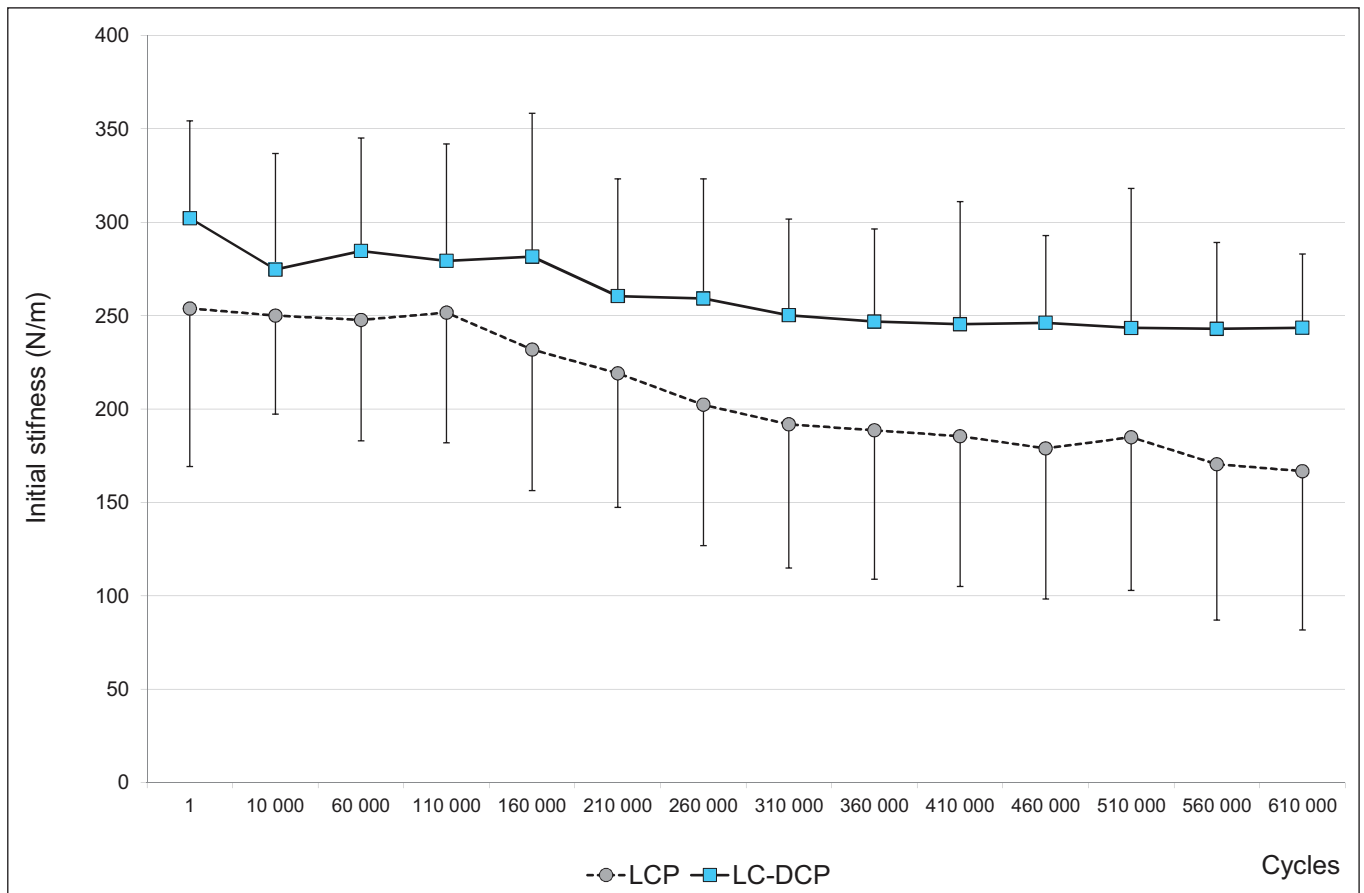


Figure 4 Evolution of the first stiffness for the locking compression plate and limited contact dynamic compression plate groups over the cycling period.

cycles before bone dehydration (35). Particular attention was paid to the storage of specimens. Freezing has long been the most widely accepted means of bone storage (34, 36). Freezing and thawing the femurs in saline-soaked gauze, and wetting the gauze that was wrapped around the femurs in saline kept the specimen moist and prevented any adverse effects of dehydration throughout the testing period (34).

In the fracture gap model that was used in our study, offset axial loading induced plate bending. As loads are applied to such a configuration, the maximal plate deflection becomes proportional to the working length squared (37). Stoffel et al have shown that the working length has the greatest effect on construct stability (30). Due to differences in plate design and plate hole geometry, the working length of the LCP construct (37 mm) was less than the LC-DCP construct (40 mm) in our study. This theoretical difference in working length of 7.5 % between the two plates was

likely to increase the stiffness of LCP constructs compared with that of the LC-DCP constructs. However, the compression forces between the plate and the femur induced by the tightening of the conventional cortical screws should decrease the working length of LC-DCP constructs, as the plate and bone are in more intimate contact.

When the LCP constructs were cyclically loaded in eccentric compression, the mean initial stiffness over the course of testing was 21.4 % lower than that of LC-DCP constructs. Similar results have been previously described with cyclic loading in bending and in torsion for 4.5 LC-DCP and LCP (19, 38). In contrast, stiffness of the LCP was significantly higher than that of conventional plating configuration when tested in axial compression (19, 38). Structural stiffness did not differ significantly between 3.5 LC-DCP and LCP constructs in a canine femoral fracture gap model subjected to four-point bending tests (3).

Other studies also reported an absence of significant differences in stiffness between locked and non-locked constructs (30, 31, 38).

The influence of slippage between plate and bone or the influence of working length on the structural stiffness of plate-bone constructs depends on the loading mode. Direct comparisons of our results and these other studies are hindered by the variations in loading applications, experimental models, implant types and construct materials.

Over the cycling period, the starting point of the load-displacement curves of both the LCP and LC-DCP constructs shifted to the right as clearly demonstrated in ► Figure 3 A and B. This suggests a permanent deformation of the system that could be attributed to changes in the cup-femoral head interface or changes in the screw-bone or bone-plate interfaces.

In the current study, both plate constructs exhibited a biphasic stiffness profile

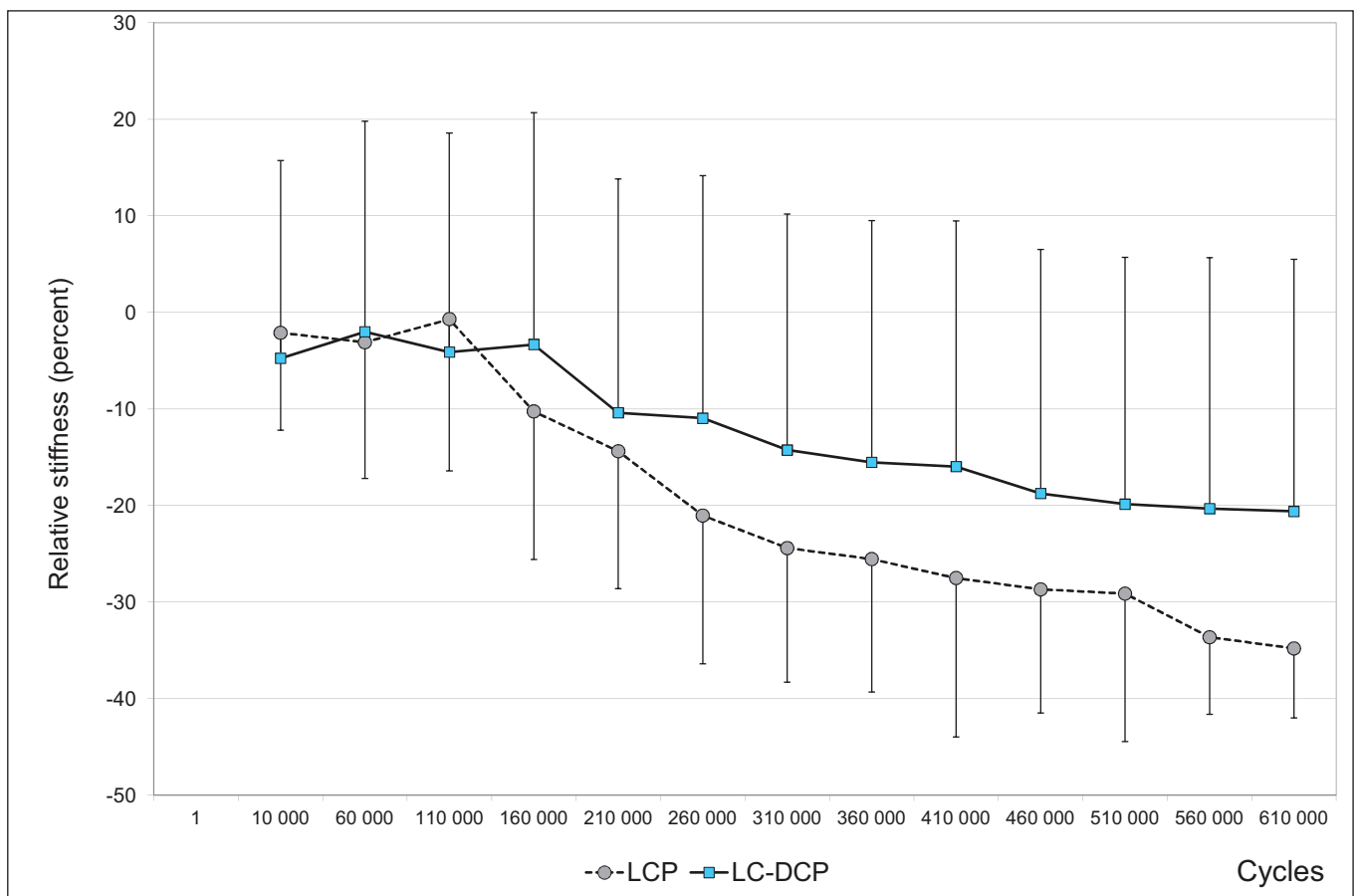


Figure 5 Evolution of the relative stiffness of both constructs over the cycling period. The relative stiffness was calculated as the percent of the difference between the stiffness at each time interval and the first stiffness, divided by the first stiffness.

with a first stiffness and a second stiffness. The second stiffness appeared when the load was around 60% to 70% of the body weight and allowed progressive stabilization at loads that can occur in clinical situations in the dog (32). This finding suggests that both constructs would probably provide acceptable clinical stability for bone healing.

Since the first stiffness of the LC-DCP bone-plate constructs was equivalent to the second stiffness of LCP constructs, the functional working length of the former was probably shorter than that of the latter. For LC-DCP bone-plate constructs, the functional plate working length was shorter than the distance between the two screws placed closest to the osteotomy gap. Under the conditions of our gap model subjected to offset axial compression, loading of the construct led to plate bending. Whichever plate was used, bending was assumed to increase the contact area between

plate and bone immediately adjacent to the femoral osteotomy gap and thereby decrease the working length of the plate (► Figure 8). This led to the appearance of the second stiffness. The difference in stiffness is probably due to the extent of bone contact between the bone and plate for the two bone-plate constructs. The compression forces induced by the two screws placed closest to the osteotomy gap act on the working length but this effect progressively decreases as one moves away from the screw to the osteotomy gap. In contrast, the working length for the LCP group was the distance between the central screws that were not in intimate contact with the femur. This difference in functional working length explains the lower first stiffness of LCP constructs compared with the LC-DCP group.

This ability to bear elevated loads as a result of modification of the bone-plate construct may be important in preventing

fixation failure. Biphasic profiles have already been reported in far cortical locking constructs in which a contact between screw and bone in the near cortex occurred during loading (20). The biphasic stiffness profile can be compared to the nonlinear behaviour of Ilizarov fixators that become progressively stiffer with an increase in load (39).

The decrease in initial stiffness after 610,000 cycles was greater for LCP constructs (34.3%) than for LC-DCP constructs (19.4%). A decrease in stiffness during cyclic loading has been previously demonstrated in experimental studies on human cadaveric humeri, human cadaveric clavicles, canine cadaveric humeri, as well as fiberglass and epoxy composite humeral models (4, 23, 40, 41). The most important finding in the current study was that, after cyclic loading, the stability of the locked screw LCP constructs was significantly lower than that of unlocked LC-DCP con-

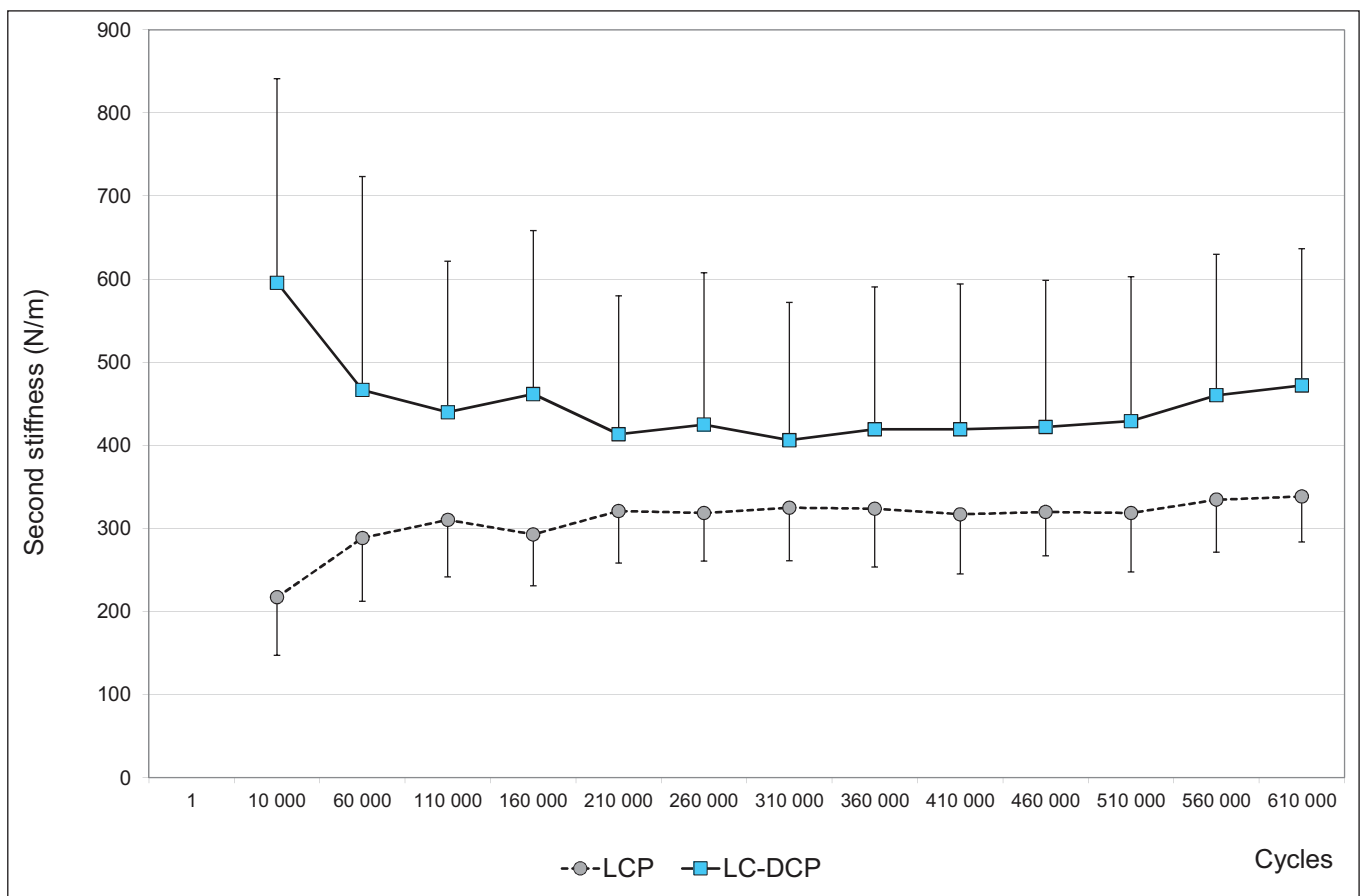


Figure 6 Evolution of second stiffness for the locking compression plate and limited contact dynamic compression plate groups over the cycling period.

structs. This decrease in stiffness during cyclic loading suggests that some degree of construct deformation had occurred, possibly caused by a slippage at some interface within the system, such as the junction of the plate and bone, the screw and bone or the screw and plate.

At present, LCP are frequently used in minimally invasive plate osteosynthesis for the treatment of comminuted diaphyseal long-bone fractures (42, 43).

It has been proposed to treat diaphyseal long-bone fractures with flexible fixation using long plates and with the screws positioned at the plate extremities (7, 44-48). Under these conditions, from a biological standpoint, the greater flexibility of LCP observed in our study could be considered as an advantage compared with LC-DCP. The benefits of flexible fixation in promoting bone healing have been well documented (7, 42, 47, 49, 50). In recent studies, it was found that fractures in dogs

that were treated by minimally invasive plate osteosynthesis, using flexible plates, healed faster than fractures stabilized with conventional plating (46, 51). However, this potential advantage must be balanced with the risk of plate failure as occurred in two LCP constructs in our study.

This *ex vivo* mechanical study has several inherent limitations. The *in vivo* situation is far more complex than that of a cadaveric femur and biomechanical results cannot be directly extrapolated to the clinical setting (30). When performing cyclic testing designated to resemble a clinical environment after fracture fixation, the loading plane should be considered (44). As previously described, our setup used an offset axial loading to simulate loading of a plated femoral fracture (3, 32, 44). This testing methodology had the limitation of being isolated to a single plane, without considering more complex forces such as a combination of bending and torsional

forces. In the diaphyseal region of the femur, however, axial and bending forces predominate and these forces were replicated in our testing protocol (52, 53). Using uniform testing conditions allowed for valid comparisons between treatment groups.

As in other studies, *in vivo* factors, such as callus development, were not included in this model and were not addressed in the current study (28). Thus, our results, which suggest that the biomechanics of locked plate-screw constructs differ from those of traditional compression plates under cyclic loading, may not accurately reflect the difference in biomechanical behaviour between LCP and LC-DCP constructs on comminuted fractures of canine femora *in vivo*.

In the stabilization of fractures of long bones, there is a compromise between flexible fixation, which enhances callus formation and improves the healing pro-

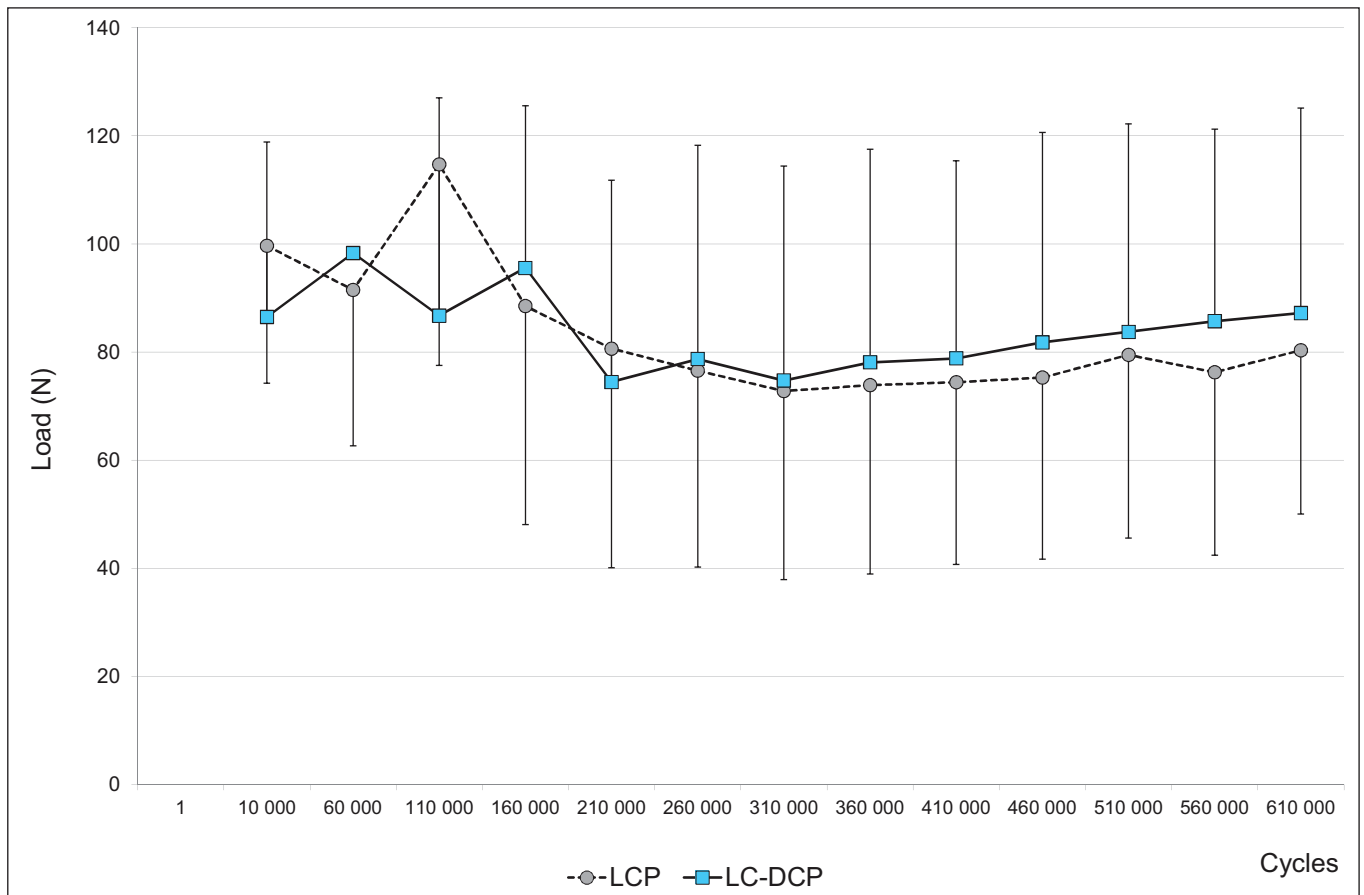


Figure 7 Evolution of loads that induced a second stiffness for the locking compression plate and limited contact dynamic compression plate groups over the cycling period.

cess, and unstable fixation, which leads to non-union or implant failure (30). The main problem, when selecting a plate for osteosynthesis of comminuted fractures, is to determine how the mechanical environment of the fracture and implant failure can be controlled (30). In the current study, two LCP plates broke during fatigue testing whereas no failure occurred with the LC-DCP constructs. It is possible that, in the case of these two LCP constructs, the flexibility was too high and this threshold was crossed. Even though the LCP constructs were less stiff than the LC-DCP constructs in this *ex vivo* study, nothing suggests that the mechanical environment of the fracture would be unfavourable to bone healing.

Interfragmentary motion plays a significant role in fracture healing. The identification of motion properties (translation, rotation, combined translation-rotation) is important since shear and tension-compression strain will condition the tissue

phenotype and healing patterns. It was particularly challenging to obtain real-time measurements of this evolving motion during the fatigue tests. Despite this, our experimental data showed that the effective stiffness of LCP plate-bone constructs was lower than that of LC-DCP plate-bone constructs and that the maximal displacement of the LCP group was higher than that of the LC-DCP group. As the boundary conditions and loading conditions were identical for each tested construct, and since the plate-bone constructs were mounted in series with the caudal and distal intact bone parts, we could conclude that the interfragmentary translation motion due to compression was lower for the LC-DCP plate bone constructs.

Evaluation of the effects on fracture healing would require an *in vivo* study of bilateral osteotomies stabilized with either a LCP or a LC-DCP. However this would raise ethical and technical problems and

involve limitations. Firstly, consideration of the pain associated with a bilateral osteotomy would be mandatory, and the clinical relevance of the surgical technique would need to be discussed. Secondly, bias would be introduced due to different weight bearing on the two limbs during the healing period.

Locking compression plates are widely used for minimally invasive plate osteosynthesis. To our knowledge, no comparative studies of periosteal or bone vascularization with standard and locked plates have been published. A few studies compared minimally invasive plate osteosynthesis and open reduction internal fixation on canine tibia, and radius and ulna fracture repair, and proposed a benefit of the minimally invasive plate osteosynthesis technique (46, 51). One cadaveric study demonstrated preservation of the vasculature at the fracture site using 3.5 mm LCP and non-locking screws with the minimally in-

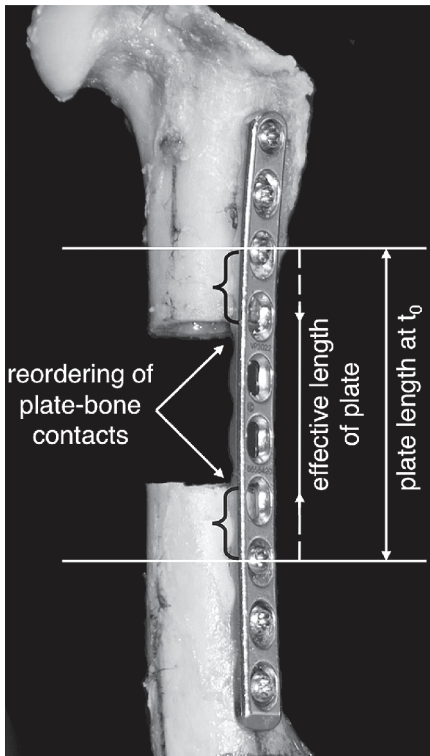


Figure 8 Photograph showing reordering of plate-bone contacts with changes in effective plate length. During loading, plate bending increased the contact area between plate and bone close to the proximal or distal femoral gap and thereby decreased the working length of the plate.

vasive plate osteosynthesis technique as compared to open reduction internal fixation (50). Bone healing was shown to be faster when the minimally invasive plate osteosynthesis technique, as compared with open reduction techniques, was used to treat tibia fractures in the dog (42).

The suggested benefits of locking plate technology in improving the biology of fracture healing and reducing disruption of the soft-tissue envelope and the periosteal vascular supply, as a result of minimally invasive plating have to be balanced with the mechanical behaviour of such plates.

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Conflict of interest

There is no conflict of interest.

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