Biomechanical Comparison of the 1.5mm Locking Compression Plate
with the 1.5 and 2.0mm Mini-Cutable Plates
&
Effect of Bone-Plate Distance on the Biomechanical Properties of the
1.5mm Locking Plate

by
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ABSTRACT

Biomechanical Comparison of the 1.5mm Locking Compression Plate with the 1.5 and 2.0mm Mini-Cuttable Plates & Effect of Bone-Plate Distance on the Biomechanical Properties of the 1.5mm Locking Plate

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This thesis documents a research study investigating the appropriateness of the 1.5mm locking compression plate (LCP) for repair of radial and ulnar fractures in miniature breed dogs. A retrospective study will review the current use and results of the mini-cuttable plates in miniature breeds while a biomechanical study compares the torsional and compression properties of the 1.5mm LCP to the 1.5mm mini-cuttable plate, 1.5mm mini-cuttable plate stacked and 2.0mm mini-cuttable plate. This thesis also documents effects of bone-plate distance on the biomechanical properties of the 1.5mm LCP in compression and torsion.

A fracture gap model was created using a bone surrogate with a 1mm fracture gap and 6-hole plates. For the first phase, 16 constructs were made for each of the plates. Eight each were tested in compression and torsion.

For the second phase, 32 constructs were made with the 1.5mm LCP but 16 constructs had a 0.5mm bone plate off-set and another 16 constructs had a 1mm bone plate off-set. The constructs were tested in compression and torsion.

In compression testing, the 1.5mm LCP was similar in stiffness to the 2.0mm mini-cuttable, 1.5mm mini-cuttable and the 1.5mm mini-cuttable stacked plates. Conversely, the 1.5mm LCP was the weakest construct out of the four constructs tested for maximum load. Minimal displacement occurred during testing with the 1.5mm LCP compared to the other plates.
In torsional testing, the 1.5mm LCP was equivalent to the stiffness of the 1.5mm mini-cuttable plate and the 1.5mm LCP was equivalent to the 1.5mm mini-cuttable and the 1.5mm mini-cuttable stacked for yield torque.

All 1.5mm LCP constructs with 0.5mm or 1.0mm bone plate off-set resulted in significantly weaker constructs in compression and torsion and also showed evidence of early failure.

This research study showed that the 1.5mm LCP was similar biomechanically to the 1.5mm mini-cuttable plate in compression and torsional testing. The use of the 1.5mm LCP can be considered as an option for radial fracture repair in dogs weighing less than 2.2kg where a 1.5mm mini-cuttable plate would otherwise be used. Bone plate off-set of either 0.5mm or 1.0mm for the 1.5mm LCP cannot be recommended.
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Soli Deo Gloria
Declaration of Work Performed

I declare that with the exception of the items below, all work reported in this thesis was performed by me.

Statistical analysis was performed by Gabrielle Monteith, Department of Clinical Studies, Ontario Veterinary College, University of Guelph, Guelph, Ontario
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Chapter 1 – Literature Review

1.1 – Bone Healing

In order to heal, fractures require a suitable environment in which stability is achieved by counteracting the various forces acting on the fracture fragments. Based on the fracture type and stability, either direct healing or indirect healing can be observed.

Direct bone healing occurs without the formation of callus and is generally not considered natural healing as it requires absolute stability with less than 2% strain and an interfragmentary gap less than 1mm; conditions that are rarely met unless open reduction and rigid stabilization have been performed. Under absolute stability, two types of direct healing can be observed based on the size of the gap between bone fragments: contact healing and gap healing. In the presence of absolute stability, contact healing will occur when there is intimate contact between the bone segments. Direct Haversian remodeling will occur following the formation of cutting cones at the ends of the osteons closest to the fracture ends. No lamellar interstitial zone forms with contact healing.

Gap healing is also considered a form of direct bone healing and requires absolute stability but occurs when the fragments are not in direct contact and a small gap (less than 1mm) exists. In gap healing, bony union and Haversian remodeling, are two separate steps. Shortly after fracture and granulation tissue formation within the fracture gap, lamellar bone is deposited oriented transversely to the long axis of the bone. Several weeks later, Haversian remodeling occurs with resorption of lamellar bone and replacement of it with longitudinally directed osteons.

Indirect bond healing, or secondary bone healing, is the most frequent form of bone healing. It occurs through sequential steps of tissue differentiation with healing resulting in callus formation and the development of progressively stiffer tissue types until, ultimately, bone is formed. Three
main phases can be recognized in indirect bone healing: the inflammatory phase, the reparative phase, and the remodeling phase. The inflammatory phase starts immediately after fracture and is characterized by hematoma formation, influx of inflammatory cells and tissue factor release.\textsuperscript{1} The reparative phase begins shortly thereafter with differentiation of pluripotent mesenchymal cells into fibroblasts, chondroblasts, and osteoblasts.\textsuperscript{1} These, in turn, produce fibrous tissue, cartilage, and woven bone, respectively.\textsuperscript{1} The remodeling phase is the slowest phase and may take many years.\textsuperscript{2} It involves the resorption of the bone by osteoclasts and deposition of osteoblasts during Haversian remodeling. Remodeling is regulated by Wolff’s law, which states that bone will respond, or adapt, to the load that is placed on it.\textsuperscript{3,4} The mechanism controlling the progression of the reparative process and progression of the callus is not well understood, however, it has been well established that bone cells and callus cells are highly sensitive to pressure and deformation and “strain” is generally considered the main driving factor determining the progression of healing.\textsuperscript{3}

### 1.1.1 – Vascularization and blood supply to the healing bone

Previous studies have demonstrated the normal direction of blood flow through the diaphyseal cortex of a long bone is centrifugal, meaning, that in normal times, blood travels from medulla to periosteum.\textsuperscript{5} There are three functional vascular entities that comprise the normal circulation of long bones and they are as follows: the afferents (arteries & arterioles), the efferent (veins & venules), and the intermediate vascular system (capillary-sized vessels in rigid bony canals connecting afferent and efferent system).\textsuperscript{5} The afferent vascular system is composed of the principal nutrient artery and the distal metaphyseal arteries.\textsuperscript{5} These two together form the medullary arterial supply, which supplies the entire diaphyseal cortex (except at fascial
attachments) and bone marrow. In areas of muscle and fascial attachments to the bone, the outer one-third of cortical bone is supplied by extraosseous vessels while the remainder of the cortex is supplied by the intraosseous vasculature originating from the nutrient artery.

The intermediate vascular system is the connection between the afferent and efferent vascular systems. It consists of a network of capillary-sized vessels encased in rigid bony canals within the bone cortex. The intermediate system connects to a rich network of canaliculi and provides oxygen, nutrients and cellular waste removal to the bone cells. Canaliculi are the channels by which molecules are transported to the bone in the intermediate vascular system. These canaliculi also contain tissue fluid that is extravascular, extracellular and allows for passive free diffusion of molecules.

The efferent vascular system allows drainage of blood from the bone. There are no valves in this system and various veins drain various associated arteries: metaphyseal arteries are drained by metaphyseal veins, periosteal arteries are drained by periosteal veins, and medullary arteries are drained by accompanying veins at the periosteal surface of the bone.

Following fracture, the normal blood supply to the bone is interrupted. Ischemia and inflammation develop quickly and initiate the healing process. In most cases, the normal vasculature is not sufficient to respond to the injury so the soft tissues surrounding the bone provide the needed blood supply through the formation of a new vascular network termed the “extraosseous blood supply to the healing bone”. This blood supply will provide blood to devascularized bone fragment and assist with early periosteal callus formation. The extraosseous blood supply is only present following injury and will slowly dissipate as the normal blood supply of the bone is re-established and the bone regains the ability to support itself.
been adequately stabilized, the medullary arterioles and capillaries can reform within a week of fracture stabilization.\(^5\)

Disrupting the extraosseous blood supply or preventing the restoration of the medullary blood supply has been linked to delayed union and nonunion.

1.1.2 – Biomechanical considerations in fracture healing: Influence of strain on callus formation

The response of bone or callus to mechanical stimulus has been recognized but has not been fully elucidated. Bone cells and callus cells are highly sensitive and respond to mechanical deformation but the exact mechanism is complex and likely multifactorial.

Piezoelectric stimulation occurs when the bone is loaded and it was originally believed that piezoelectric stimulation was the main factor triggering bone adaptation. Although piezoelectric stimulation does occur, its role remains unclear and recent research indicate that changes in hydrostatic pressure in the bone canaliculi as well as a direct response of the osteocytes to mechanical deformation caused by bone loading are likely the triggers initiating bone remodeling.\(^4\) The concept of bone being able to sense deformation and pressures led Perren in 1979 to propose a unified strain theory to explain the different forms and the progression of bone healing.

In this theory, different tissues have different strain tolerance and a tissue will not form or survive if the strain it is subjected to exceeds tissue tolerance. Granulation tissue can tolerate up to 100% strain. Fibrocartilage can tolerate strain up to 10% while bone can only tolerate strain
less than 2%.\textsuperscript{8} Progression of healing is, in part, dictated by the strain at the fracture site and only tissue that can tolerate the strain present will form. In turn the development of this tissue (callus) progressively stabilizes the fracture, which in turn decreases instability and reduces strain. The reduction in strain allows for the formation of a tissue that is more sensitive to strain but also stiffer. The process continues until the strain is reduced to less than 2% at which time bone can form.\textsuperscript{9}

![Figure 1.1: Strain and how it relates to fracture healing.](image)

In fractures, strain is defined as the change in fracture gap length divided by the original length of the fracture gap.\textsuperscript{1} (Adapted from V. Klika: Biomechanics of Musculoskeletal Injury, Biomechanics in Applications, 2011)

In direct bone healing, absolute stability creates conditions of less than 2% strain and bone can and will form immediately through osteonal remodeling in contact healing or through deposition of lamellar bone following the development of granulation tissue in gap healing. In fractures that have not been rigidly stabilized, the strain is almost always greater than 2% and bone will not form directly. A decrease in strain is therefore required before boney union can proceed. At the bone level, there are two mechanisms by which the body works to decrease
strain. First, fracture resorption occurs at the fracture fragment ends. This involves osteoclastic removal of cellularly dead bone. The removal of the bone widens the fracture gap, therefore reducing the strain. The second mechanism used to decrease strain is by periosteal callus formation. Deposition of callus at the periphery of the bone stabilizes the bone fragments. Although callus also forms intramedullary and intracortically, it is the periosteal callus that contributes most to bone stability as the stiffness is increased exponentially by the radial increase of the callus. As the stability increases, strain decreases and will allow the deposition and/or transformation of the granulation tissue into fibrous tissue, fibrocartilage and ultimately bone.  

1.2 - Fracture Fixation with bone plates and bone plate evolution

Bone plates were first developed around 1895 through the realization of the need to immobilize fracture fragments while allowing use of the limb and motion of the joints. Unfortunately, early plates were met with significant complications that limited their wide acceptance. The rudimentary design and fabrication led to significant issues of stability, corrosion, and re-fracture following plate removal, causing them to be all but abandoned until the 1940’s when stronger bone plate constructs were invented and the understanding for the need for compression of fracture ends was discovered. After this point, the development and application of bone plates took off and bone plates are now ubiquitous. During the past 50 years, bone plates have undergone a significant evolution to keep up with our constantly evolving understanding of fracture healing. From the initial concept of the AO (Arbeitsgemeinschaft fur Osteosynthesesfragen) of absolute stability and perfect reconstruction to the more contemporary concepts of relative
stability and preservation of the blood supply, the plates have evolved from the dynamic compression plate (DCP) to the development of the limited contact plates (PC-Fix, LC-DCP) to ultimately evolve into the concept of locking plates (LCP) that we currently know.\textsuperscript{3,11}

**1.2.1 - Evolution of bone plate design: From conventional to locking plates**

Prior to locking plates, most fractures were repaired with internal fixation using standard plating techniques and following the early concepts of fracture fixation outlined by the AO. Absolute stability and precise anatomical reconstruction were the goals and often resulted in extensive soft tissue dissection and damage to the vascularization.\textsuperscript{3} Although bone healing was often reliably achieved, delayed healing and non-unions were still frequently observed. In fracture fixation, damage to the local blood supply occurs with the surgical approach to the fracture, manipulation of the fracture, reaming of the endosteum, stripping of the periosteum, and continues with application of the bone plate which prevents inflow or outflow of periosteal blood flow.\textsuperscript{3} Reversible signs of bone resorption and necrosis were quickly observed under the plate and were believed to be responsible for some delayed and non-unions.\textsuperscript{12} Stress protection through Wolff’s law was initially believed to be responsible for the temporary osteoporosis under the plate. It was thought that the plate was too stiff and therefore shielded the bone from stress and causing resorption. Further research in the problem disproved this theory and instead showed that disruption of the cortical blood supply under the plate was the cause of the necrosis and subsequent resorption of the bone.\textsuperscript{10} Over the years, the importance of blood supply for fracture healing has become more recognized and further research led to the development of bone plates and techniques that better respect bone
vascularization. This led to the development of the limited contact dynamic plate (LC-DCP™), the point contact fixator (PC-Fix™) and finally to the development of the current locking plates. Although the modern locking plate is a recent creation, the first locking plate was in fact developed over 100 years ago in 1886 by Carl Hansman as a monocortical fixator, which was subsequently modified by Paul Reinhold in France in 1931. These locking plates did not initially gain much popularity and were abandoned until 1974 when the idea re-surfaced with the creation of new locking plate systems: The Litos system, followed by the Zespol system in 1982. By 1995, the Surfix system, created by Patrick Surer, emerged and went through multiple iterations with the collaboration of the AO. In 1995, the point contact fixator (PC-Fix) was approved by the Food and Drug Administration and released for clinical use. The PC-Fix was designed to enhance fracture healing by minimizing contact with the bone and preserving the local blood supply. (See figure 1.2– green areas show the extent of the plate / bone contact for the three different plates)

Unicortical screws were used to act as pegs, which connected the splint to the bone and eliminated the need for friction between the plate and the underlying bone for stability. The Less Invasive Stabilization System (LISS) followed this, in 2001 and was another type of locking internal fixator
with locked full-length metaphyseal screws.\textsuperscript{3,14} Both the PC-Fix and the LISS only allow for screw placement perpendicular to the plate.\textsuperscript{15} This can be a challenge because sometimes screws need to be angled to prevent intra-articular screw placement. After the success of the LISS, work was performed with Professor Michael Wagner of the Wilhelminen Hospital in Vienna to develop a plate with a “combination hole” that could accommodate\textsuperscript{16} either locking or compression screws, with the option for angled standard screws if needed.\textsuperscript{13,15,17}

![Figure 1.3: LCP combination hole with locking screw (blue) and nonlocking screw (gold) on the right and combination hole without screws on left. (Adapted from Frigg R: Development of the locking compression plate. Injury 34: S-B6-S-B10, 2003.)](image)

To date, multiple locking plate systems exist; although many still only allow the screws to be placed at fixed angles relative to the plate, newer advancements have led to the development of angle variable locking systems, allowing a limited angulation of the screw while still locking into the plate. Among others, the most notable locking systems in Veterinary Medicine are the Kyon ALPS\textsuperscript{™}, Trauma Vet Fixin\textsuperscript{™}, Synthes LCPT\textsuperscript{™}, and the Securos PAX\textsuperscript{™}.

1.3 – Principles of conventional and locking plates

1.3.1 – Conventional Plates
Conventional plates work by compression of the plate to the bone, through the tensile force, originating from the screw, which results in friction in the plate-bone contact zone.\(^1\) By compressing the plate onto the bone, the produced friction, resists tangential loads of approximately 1,000N.\(^3\) The frictional force generated is a product of the normal force (force that presses the plate to the bone) and the frictional coefficient that exists between the bone and the plate.\(^18\) A large area of contact is required between the bone and the plate to generate enough friction in order to create stability. Conventional screws, because they are loaded in tension, need to resist pullout and therefore, have a large, asymmetrical screw thread with small screw core diameter.\(^3,13\) The screw with the greatest torque (tightest screw) bears the greatest load because it contributes the greatest amount of force to the plate and the weakest link in the conventional plating is the screw-bone interface.\(^1,18\)

When conventional plates are loaded axially in compression or tension, the force is converted to shear stress at the bone-plate interface.\(^18\) If this stress exceeds the frictional forces, loosening and screw toggling will occur over time causing failure of the construct.\(^19\) The screw heads are not locked into the plate so bicortical purchase is needed to prevent loosening of the construct.\(^17\)

1.3.2 – Locking Plates

Locking plates do not rely on the friction generated by the plate-screw-bone interface. Locking screws are considered a “fixed-angle” construct, and act as one with the plate, which increases strength of construct.\(^1\) Axial loading or bending is converted to compressive stress at the bone-screw interface because of the fixed-angle construct.\(^1\) Bending loads are applied more evenly
with less stress concentration at individual screw holes and the weakest part of the locking plate construct is at the screw-plate interface.\(^1\) The strength of the construct is equal to the sum of all bone-screw interfaces and if a locking plate fails, it will fail in an “all or nothing” pattern with complete screw pullout.\(^{18,20}\) Screws are no longer subjected to tension forces as the plate pulls off the bone but are subjected to shear and bending forces. The design of the screws has been improved to better oppose these forces when compared to conventional screws.\(^{13}\) Unlike conventional screws, locking screws have a larger core diameter and finer, symmetrical threads that allow them to advance into the bone but provide equal resistance to pullout and advancement.\(^{13}\) The difference in core diameter between the two screw systems, explains the 3-fold increase in resistance in bending and the 2-fold increase in shear of the locking screw compared to the conventional screw.\(^{13}\)

Additional reported benefits of locking compression plates include: stability of small fragments, improved fixation in weak and osteoporotic bone, the ability to use monocortical screws, the need for fewer screws compared with conventional osteosynthesis, the decreased need for bone grafting as well as the improved ability to perform percutaneous osteosynthesis.\(^{21}\)

### 1.4 – Biomechanical Comparison of Locking and Non-Locking Implants

To understand the comparisons of locking and non-locking implants, one must first grasp how they compare in regards to bending, torsion, cyclic testing and fatigue, effect of plate-bone distance, and the advantage of locking implants in low quality bone. The comparison of the biomechanical properties of the different implants is difficult and often leads to conflicting results. These differences are often due to the varying experimental condition between publications but
also because different implants are often compared, each with their own shape, size and material properties.

1.4.1 – Comparison in bending

Because the almost similar physical dimension of the LCP and the LC-DCP, it would be expected that the 2 plates would behave similarly when tested in isolation or when the screw holding is not the limiting factor. Relatively similar properties were demonstrated by Aguila who compared the 3.5mm LCP and the LC-DCP in vitro using canine femurs. The properties of the LCP were similar to the LC-DCP in all bending directions except for a slight increase in stiffness of the LCP in medio-lateral bending. Based on this information, it was concluded that the use of the LCP was a good alternative to the LC-DCP in cases of unstable femoral fracture in the dog. Blake also found similarities between the 2 plates in bending.

Biomechanical properties of a 5-hole 4.5mm narrow locking compression plate were compared to a 5-hole 4.5mm narrow dynamic compression plate in a single-cycle, 3-point bending system using cadaveric adult equine forelimbs. The locking compression plate had significantly greater yield load, failure load and stiffness to failure compared to the dynamic compression plate; failure point and displacement at yield was not significantly different between groups. Under their test conditions, the locking compression plate was significantly more stable than the dynamic compression plate biomechanically. The reason for the difference is complex and may include reasons such as the larger diameter of the screws, the slightly different physical properties of the plate and position of the screws within the plate as well as a difference in the contouring of the plate as it is applied to the bone. Superiority of the 4.5mm LCP over to the 4.5mm LC-DCP was also demonstrated on equine metacarpal bones.
In humans, biomechanical advantages of the locking plates over the non-locking constructs were also demonstrated in a model of comminuted distal radial fractures. The plates; 3.5mm locking T-plates and 3.5mm conventional T-plates were applied dorsally to distal human cadaveric radii with a simulated dorsally comminuted fracture. Sixteen pairs of radii were tested in compressive load to failure: each pair had one radii in the locking group and one in the conventional group. The locking plates were 33% stiffer and had a 91% increase in load at failure compared with the standard T-plates. Based on these results, it was recommended to strongly consider, or even prefer the use of locking T-plates instead of standard non-locking plates, in the treatment of dorsally comminuted distal radius fractures using dorsal plating.

In contrast with these studies, a recent study published in 2014 compared the conventional compression plates and locking compression plates using cantilever bending in a small number of canine cadaveric ilial fracture models. No significant difference was observed between the constructs in mean stiffness, mean yield load, mode of failure, displacement at failure, or ultimate load at failure. Several studies also compare different types of locking plates from different manufacturers. Although informative, because of the significant variation in the size and design of these implants, these studies provide little insight on the benefit of the locking system themselves.

A study compared the LCP to three other locking and non-locking bone plates (LC-DCP, Broad LC-DCP and SOP) in single cycle four-point bending. The LCP and the LC-DCP were statistically similar in structural stiffness and were both weaker than the SOP and the broad LC-DCP with the conclusion that use of a LC-DCP could be replaced with a LCP if applicable. This finding was not surprising as the LCP and the LC-DCP have almost the same profile while both the
Broad LC-DCP and string of pearl plate have both a higher area moment of inertia. It is interesting to note that when the SOP is tested in bending as “a construct” it is no longer superior to the LCP or the LC-DCP. This is likely due to the difference in the diameter of the screws used as the SOP uses regular cortex screws with a small core diameter while the LCP use dedicated locking screws with a larger core diameter.15

The effect of bending direction on the locking plate, String-of-Pearls and the conventional plate, limited contact dynamic compression plate was investigated using a tibial fracture gap model.26 Specimens were tested in both mediolateral and craniocaudal bending for 10 cycles. The SOP plates were significantly stiffer than LC-DCP plates in mediolateral bending; in craniocaudal bending, SOP constructs were significantly less stiff than LC-DCP constructs.26 These differences were attributed to the circular cross sectional area of the SOP compared to the rectangular shape of the LC-DCP and overall, SOP constructs had a more homogenous bending behavior in orthogonal loading directions compared to the LC-DCP.26

1.4.2 – Comparison in torsion

Similarly to bending tests, similar plates tested under the same conditions are expected to behave in very similar fashion. Cabassu compared several plates in torsion and found no difference in stiffness and yield load between the LC-DCP and the LCP.27 This finding contrasts with several studies showing a significant increase in strength and stiffness when locking screws are used. Comparing the 4.5mm locking compression plate to the 4.5mm Limited-Contact Dynamic Compression plate in torsion single cycle to failure with an equine metacarpal fracture model, Sod
et al. determined that the LCP had significantly greater mean yield load and mean composite rigidity compared to the 4.5mm LC-DCP.\textsuperscript{28}

The effect of the screws on the torsional properties of the locking plate constructs have also been investigated by using the same plate but varying the type and position of the screws: the torsional properties of the 3.5 LCP secured to canine femora were assessed using different screw configurations. The plates were secured to the bones using all non-locking screws, all locking screws or a combination of locking and non-locking screws. The results showed that constructs with all locking screws were stronger than constructs using non-locking screws. Hybrid systems, using a combination of locking and non-locking screws had intermediate strength, however, the addition of only one locking screw to an otherwise non-locking construct increased the torsional strength by 17%.\textsuperscript{29}

Another study also compared the effect of different screws on the torsional properties of plates applied to human radii replicas. Composite radius sawbones were divided into four groups based on the number of locking vs. non-locking screws and bicortical vs. monocortical purchase: group 1: all unlocked bicortical, group 2: locked unicortical, group 3: unlock hybrid (monocortical locked screws with one bicortical unlocked screw distal to the fracture site, on either end), and group 4: locked hybrid (all locking screws, monocortical near fracture and one bicortical unlocked screw distal to the fracture site, on either end). These constructs were tested in compression and torsion. When compared in torsion, the weakest construct was the locked unicortical construct with the other three constructs showing similar torsional stiffness with an approximately 51% increase with addition of bicortical screws, locked or unlocked.\textsuperscript{30} Demianiuk also confirmed the torsional weakness of unicortical screws in SOP constructs. Constructs made with monocortical screws were the most compliant, while the addition of even a single bicortical locking screw significantly
increased the stiffness of the constructs. Based on their findings, the authors recommend the use of at least one bicortical locking screw in each of the major fragments to decrease torsional compliance of the repair.\textsuperscript{31} The importance of bicortical screws in torsion, particularly when the bone cortices are thin, has also been highlighted by Gautier.\textsuperscript{32} This is of particular relevance in companion animals as the cortical thickness of the long bones are thin compared to human bones, therefore reducing the working length of the screw and providing little resistance to torsional forces.

1.4.3 – Cyclic testing and fatigue

A study investigated various screw configurations for treatment of comminuted diaphyseal fractures using the 3.5mm Synthes locking compression plate.\textsuperscript{32} The Synthes 3.5mm locking compression plate was tested in cyclic bending (cantilever and compression/bending with constructs being divided into three groups: (1) bicortical locked, (2) unicortical locked, and (3) bicortical unlocked.\textsuperscript{33} The plate with bicortical locked screws withstood significantly more cycles to failure than the other constructs and significantly less displacement occurred after axial loading with bicortical locked screws.\textsuperscript{33} The constructs with monocortical screws fared poorly and the authors concluded that the use of monocortical screws for comminuted diaphyseal fractures could not be recommended.\textsuperscript{33}

An in-vitro biomechanical study comparing 4.5mm locking compression plate fixation to limited-contact dynamic compression plate fixation was performed on osteotomized equine third metacarpal bones.\textsuperscript{28} Four-point bending cyclic fatigue testing was performed and the 4.5mm LCP was superior to the 4.5mm LC-DCP with a significantly greater number of cycles to failure.\textsuperscript{28}
Bending of screw heads and screw breakage at the shaft/head junction occurred with both the LCP and the LC-DCP screws.\textsuperscript{28}

The number of screws used in fracture repair may also influence the cyclic testing properties of the constructs. In human medicine, the current recommendation for placement of locking constructs in non-osteoporotic comminuted fractures is at least three to four monocortical screws or two bicortical screws per fragment.\textsuperscript{32,34,35} A recent study comparing the mechanical behavior of locking constructs with two or three bicortical locking screws per fragment was performed under cyclic torsional testing. The plates used were 10-hole 3.5mm locking compression plates with a fixed 1mm distance away from the bone surrogate.\textsuperscript{35} The results showed approximately 20\% reduction in stiffness and 25\% reduction in fatigue life when only two bicortical screws were used per fragment compared to three screws.\textsuperscript{35} this suggests that if torsional loads are expected additional screws may be needed to ensure long-term stability.

A smaller locking compression plate, 2.4mm, was compared to the 2.4mm limited contact dynamic compression plate in eccentric cyclic loading using a comminuted femoral fracture model in canines.\textsuperscript{36} Three bicortical locking screws were placed in each fragment using a 10-hole plate. Overall, the mean stiffness of LCP constructs was 24\% lower than the mean stiffness for LC-DCP plates and the reason for this could be the longer working length of the LCP plates due to lack of compression of the plate against the bone.\textsuperscript{36} Because of the longer working length and decreased stiffness, if healing is delayed, fatigue failure may occur.

1.4.4 – Effect of plate-bone distance

One advantage of the locking plates is the fact that the plate does not need to be in intimate contact with the bone to provide stability. The plate can be placed away from the periosteum,
reducing the impact of the plate on the vascularization of the bone. There are two factors influencing fixation stability with the locking compression plates: plate length and distance of the plate to the bone.\textsuperscript{37}

A human study by Haug et al. investigated the effect of plate contouring and offset (distance between the plate and the bone) in surrogate human mandibles. Locking and no-locking plates were contoured to leave an offset of 0, 1 or 2 mm between the plate and the bone. Loading of the mandibles was performed and displacement, stiffness and yield load were measured and compared. The different offsets had an effect on the non-locking plate systems but had no effect on the locking plate systems.\textsuperscript{20}

Similarly, Ahmad et al. conducted research to determine the acceptable distance of the bone plate to the bone in surrogate long bones. A 4.5mm DCP plate was compared to an LCP with 0, 2 or 5mm offset between the plate and the bone. All constructs were tested in axial compression and torsion. The constructs with a 5mm offset were significantly weaker than the other groups. There were no significant differences between the groups with 0 or 2mm offset, leading them to conclude that a 2mm offset or less was acceptable and did not significantly affect biomechanical performance of the constructs.\textsuperscript{38} Because there were no models with either 3mm or 4mm of offset from the bone, it is unknown if these would be acceptable or biomechanically similar to the 2mm offset from the bone, however, it is reasonable to assume that the biomechanical properties deteriorate as the plate is further away from the bone. This was confirmed in a study testing the Vet-Lox\textsuperscript{TM} plate tested in compression/bending. In this study, a 4 mm bone-plate stand-off distance significantly reduced stiffness and yield load by 63% and 69% respectively.\textsuperscript{39} Although other distances were not compared, the authors recommend to appropriately contour the plate to the bone to minimize bone-plate offset distance and decrease risk of plate bending.\textsuperscript{39}
The screw type (monocortical versus bicortical) also influences the stability of offset plate. A study evaluated the effect of the screw type and bone-plate distance using 3.5mm Synthes locking compression plates with 1mm and 2mm offsets. The constructs were tested in cyclic cantilever bending and axial loading. Constructs made with bicortical locking screws withstood significantly more cycles to failure than the monocortical constructs or the non-locking constructs. In cantilever bending, both bicortical locking constructs resisted over 30,000 cycles without failure, regardless of the plate offset. The constructs made with monocortical screws sustained significantly less cycles than the bicortical but were also strongly affected by the offset distance. (Figure 1.4)"
1.4.5 – Advantages in low quality bone

Humans with osteoporotic bone are at increased risk for fractures and increased risk for complications following repair due to the lack of bone density. It can be quite difficult to obtain adequate fixation in poorly mineralized bone and because conventional plating techniques rely on compression between the cortical bone and the plate, the screws must have adequate bone purchase.\(^{40}\) Bicortical screw purchase with conventional plates will usually provide enough stability but the problem arises when bicortical screw purchase is not possible, particularly in the case of periarticular fractures.\(^ {40}\) A cadaveric human tibial fracture model performed in 2007 compared stiffness between locking and conventional plates and concluded no statistically significant difference between the two types of plates although locking constructs may be of benefit in the most severe of osteoporosis cases.\(^ {40}\) A second biomechanical cadaveric study, again, looking at distal fibular fractures, revealed higher torque to failure, maximal torque and higher angle at failure of locking plates compared to conventional plates in osteoporotic bone.\(^ {41}\) Improved fixation strength in an osteoporotic bone model using locking contoured plates with screws of different length concluding locking systems are appropriate in osteoporotic bone with poor mechanical capacity.\(^ {42}\) Clinically, a retrospective study evaluated 46 consecutive patients from 2004-2010 with a mean age of 80 years old that underwent treatment of a distal humeral fracture with placement of a locking compression plate.\(^ {43}\) This study concluded that locking plates provided strong fixation of osteoporotic bone and lead to good clinical and radiological outcome with functional recovery similar to previous case series.\(^ {43}\)

Not all studies, however, support the use of locking plates in osteoporotic bone. Lo, in 2013 tested locking versus non-locking plate constructs in an osteoporotic fibular defect model and although the non-locking constructs were the least stable among the constructs tested, no statistically
significant difference was found between the constructs. The authors, in this model, could not support the use of locking plates for osteoporotic specimens.\textsuperscript{44}

1.5 – Biological comparison of Conventional versus Locking Plates

In addition to the biomechanical advantage of the locking plates, there are a number of biological differences between conventional and locking plates and they will be discussed in the following section. Some of these differences include preservation of blood supply to fractures, rate of infection, speed of healing or delayed/nonunion and stress protection. Some of these advantages have been clearly demonstrated while others remain mostly theoretical and controversy still exists as to the extent of these benefits.

1.5.1 – Preservation of the blood supply to fractures

In contrast to locking plates that allow for space between the bone and implant, conventional plating has been shown to result in significant periosteal and cortical vascular damage. Disruption of the local blood supply from periosteal stripping, with conventional plates, occurred frequently and this can contribute to bone necrosis and increased the risk for infection, delayed healing, ischemia, bone necrosis and the possibility of re-fracture.\textsuperscript{3,38}

In contrast, locking plates do not require intimate contact with the bone to provide stability and can even be safely placed at a distance of up to 2mm away from the bone without decreasing its stability. This allows for a better preservation of the periosteal blood supply as the plate is not in direct contact with the bone.\textsuperscript{38}
1.5.2 – Infection rates

As the locking plates allow preservation of the blood supply to the bone and reduce bone necrosis underneath the plate, it is expected that they would offer better protection against infection than conventional plates. The infection rate for fractures using locking compression plates is 1.5%, as presented in a human journal after treatment of 169 fractures.\(^2\) Infection is thought to occur from necrosis induced by the implant and dead space effect and less from the foreign material itself.\(^3\) Conventional plates cause a large amount of necrosis underneath the plate due to the compression of the bone plate onto the bone whereas locking plates do not.\(^4\) If the necrotic bone becomes exposed to an infection, the already weakened bone with increased porosity and reduced vascularity, could result in formation of a sequestrum.\(^5\) Because of the better preservation of the periosteum and decreased plate footprint on the bone, the use of locking plates should in theory decrease the infections rates. Very few studies actually show an advantage of the locking plates compared to the traditional plates when it comes to infection. In a retrospective study on the infection rates in mandibular fractures, Kirkpatrick concluded that although the locking plates facilitated the treatment of complicated fractures, they did not eliminate complications and infections were still observed.\(^6\) In an experimental infection model in rabbits, the PC-Fix plate was significantly more resistant to infection than the similarly sized DCP plate.\(^7\) The results were attributed to the decreased contact of the PC-Fix with the bone compared to the DCP. The PC-Fix plate was however made of titanium in contrast to the stainless steel DCP. It is therefore not possible to determine if the benefit was indeed provided by the decrease in bone contact of resulted from the difference in material between those 2 implants. It has been shown that titanium implants
are more biocompatible than stainless steel implants and are therefore less prone to infection.\textsuperscript{13,47}

Nevertheless, the overall difference in infection rate using a conventional plate (full contact to bone – DCP in steel) to a locking plate (point contact -PC-Fix in titanium) in this rabbit model was a ratio of 1:450.\textsuperscript{3}

Along with this, there are numerous reports of mal-union and multi-drug resistant infections which resulted in fracture union and resolution of infection following the replacement of the conventional plates with titanium locking plates.\textsuperscript{13}

In veterinary medicine, only a couple of publications highlight the potential decrease in infection rates between the 2 plates: Solano retrospectively compared the postoperative infection rate of conventional non-locking TPLO plate and screw fixation systems to locking TPLO plate and screw fixation systems following tibial plateau leveling osteotomies in 208 dogs, all \textgreater{} 50kg. There was a statically significant decrease in postoperative infection rate in dogs with the locking plate compared to the non-locking plates.\textsuperscript{48} One proposed reason for the decreased infection rate is the fact that stable fractures will have a reduced susceptibility to infection compared to less stable fractures.\textsuperscript{48}

An equine study performing minimally invasive proximal interphalangeal joint arthrodesis with locking compression plates reported a low postoperative infection rate with only one limb out of twelve (16.7\%) requiring additional treatment.\textsuperscript{49} This is much lower than the previously reported infection rate of 35\% with open reduction and internal fixation.\textsuperscript{49} The conclusion was in part, that closed reduction with locking compression plates would result in decreased infection rate although the effect of the minimally invasive treatment of the fracture cannot be differentiated from the effect of the locking implant itself on the infection rates.
1.5.3 – Nonunion and delayed union

Previous research concluded that reduced periosteal vascularity correlated with an increased incidence of delayed union and nonunion. This information confirms the need for adequate blood supply for appropriate fracture healing to occur.\textsuperscript{7} Locking compression plates can be placed in a bridging plate fashion and recent research showed shorter time to fracture healing using these plates on radial fractures in dogs.\textsuperscript{50} One of the benefits of locking compression plates is the preservation of the periosteal blood supply that may allow better healing of fractures and a decreased incidence of delayed union and nonunion.

In a prospective multicentre study, involving 169 fractures, evaluating locking plates in humans, 17 out of 18 cases of primary delayed or non-union went on to heal uneventfully when treated with a locking plate.\textsuperscript{21}

Results from the treatment of nineteen cases of unstable distal radial fractures with locking T-plates were reported.\textsuperscript{51} The mean age of the patients was 61 years old and all fractures were treated with open reduction and no placement of bone grafts, bone substitutes or additional plates.\textsuperscript{51} All fractures were united within six months with no cases of hardware failure or tendon rupture.\textsuperscript{51}

Tibial fractures are associated with a high rate of non-union in human medicine. Percutaneous plating of distal tibial fractures using a limited contact dynamic compression plate was reported in 21 patients. All fractures were closed and seventeen fractures healed within six months but there were two delayed unions and two non-unions, one of which healed after a second surgery to place a bone graft.\textsuperscript{52} A similar study, also looking at tibial fractures, was conducted looking at minimally invasive techniques for treatment of distal tibial fractures using locking plates. The retrospective study included 20 patients, eight had open fractures and 13 patients
received temporary fracture stabilization with an external fixator (six with open fractures) prior to placement of a locking plate.\textsuperscript{53} Out of the eight open fractures, two resulted in aseptic nonunion one year postoperatively but both had been very high-energy injuries.\textsuperscript{53} Out of the closed fracture group, all 12 fractures were healed by 12 months postoperatively. Based on these results, it was concluded that percutaneous plating with locking plates was a good technique for certain types of closed distal tibial fractures and further research was needed to evaluate this surgical technique in the long term.

1.5.4 – Stress protection

Stress shielding, or stress protection, relates to the mechanical effect of fixation of a bone plate on cortical bone with the thought that the bone is overly protected from stress by the bone plate itself and, therefore, the bone remodeling results in a weaker bone due to the lesser stresses placed on it during healing.\textsuperscript{54} An area of temporary osteoporosis was routinely observed underneath bone plates. This phenomenon was believed to be caused by stress protection and resulted from the stiffness of the implant.\textsuperscript{55} Although the osteoporosis was temporary, it was believed that it was responsible for delayed union and increased the risk of re-fracture following implant removal.\textsuperscript{2,56} Further investigation revealed that the area of osteoporosis was in response to devascularization of the cortex and that the extent of the necrosis was related to the footprint of the implant on the bone as opposed to the stiffness of the implant.\textsuperscript{3} Since then, the theory that stress protection was the cause of the early temporary osteoporosis was disproven in several studies based on the composite beam theory and claimed to be a myth but was more likely the result of vascular
disruption of the underlying bone.\textsuperscript{56,57} Similarly, in small breed dogs, the relative stiffness of the plate and the bone were compared. In a cadaveric study comparing the relative stiffness of intact and plated radii in small and large breed dogs, no significant difference was found between large-breed and small-breed radii stabilized with plates and the conclusion was that overly stiff bone-plate constructs was unlikely to be the cause of complications observed with radial fractures in small-breed dogs.\textsuperscript{54}

Although stress protection has been discounted as a cause for the temporary osteoporosis under the plate, stress protection has been incriminated in the delay of callus formation on the cis-cortex of fractures treated with locking plates. Asymmetrical callus formation has been observed in long bone fractures and may be responsible for delayed healing.\textsuperscript{58} It is believed that the superior stability provided by the locking plates does not stimulate callus formation at the level of the cis cortex. As the trans-cortex is located further from the plate, the level of strain is increased and callus forms normally. An increase in callus formation was also seen when titanium plates with lower stiffness were used.\textsuperscript{58} To compensate for this problem, current research focuses on the development of locking plates or screws that allow uniform strain distribution between the different cortices.\textsuperscript{59,60}

1.6 – Clinical results of Locking Plates

The following is a compilation of the clinical use of locking plates, both in human and veterinary medicine. Few, if any studies, prospectively evaluate locking versus non-locking plates and the overall complication rate of fracture repair is low so conclusions regarding the superiority of locking plates is lacking.
1.6.1 – Clinical Studies of Locking Plates in Human Medicine

Locking plates have been used in human spinal surgery. The AO anterior thoracolumbar locking plate (ATLP) has been reported in 25 patients for stabilization of the spine anywhere between T10-L5. Reasons for stabilization included, fractures, metastatic tumors, disc herniation, failed laminectomy, etc.\(^6\) The ATLP is a titanium locking plate with four locking screws that provides rigid fixation and compression across a bone graft in the spine.\(^6\) Out of 25 cases, there were five broken screws and no broken plates with two misplaced screws for postoperative complications. It was concluded that the ATLP system was a safe, low profile MRI/CT compatible implant for spinal stabilization.\(^6\)

A comparison of limited contact dynamic compression plates with the point contact fixator plates for forearm fractures was performed and involved 125 fractures total. The two implants were equally effective for the treatment of diaphyseal forearm fractures, which went against the hypothesis that the PC-Fix would result in better bone-healing and decreased complications than the LC-DCP.\(^6\)

A paper entitled, First clinical results of the Locking Compression Plate (LCP) was published in 2003 out of Switzerland and reported a prospective multicenter study where 144 patients with 169 fractures were treated.\(^2\) Adverse events are summarized below: (Figure 1.5)
Many of the complications were cited to have occurred in the early phase and were due to technical errors intraoperatively. It was also during this time period that it was discovered that 2.5Nm torque for insertion of 3.5mm screws resulted in cold welding and the torque limiter was reduced to 1.5Nm which solved the issue. In 2005, a randomized comparison of locking and non-locking plates for treatment of Colles’ fractures (distal radial fracture) in elderly people was performed and found no significant difference between groups for radiographic assessment and long term outcome.

Distal radial fractures are a common fracture in humans and a study looking at 50 fractures in 49 people using a volar locking screw plate system was successful although there was a 12% rupture of the flexor pollicis longus tendon at 10 months postoperatively. It is thought that the dorsal approach to the distal radius with plate fixation causes irritation of extensor tendons and can be caused by ischemia and trauma or due to direct contact with the hardware devices and the tendon. To address this issue, palmar locking plates were developed and a separate study investigating the use of 2.4mm palmar locking plates for treatment of unstable dorsal dislocated fractures...
distal radius fractures was performed. They evaluated 19 patients; it was noted the approach to
the palmar aspect of the radius is technically less demanding and there were no reports of flexor or
extensor tendon alterations with good radiological and functional results.

Minimally invasive plate osteosynthesis (MIPO) involves small skin incisions with
tunneling of the bone plate extraperiostally to minimize soft tissue injury and preserve local blood
supply. Two retrospective studies evaluated the locking compression plate placed minimally
invasively for tibial fractures. A total of 52 patients were evaluated between these two studies.
Many of these fractures were open fractures and had placement of external fixators prior to
placement of a locking compression plate. MIPO was concluded to be better for soft tissue
preservation although prolonged healing was observed when bridging plate techniques were used
to treat simple fractures and this intuitively makes sense based the analysis of strain at the fracture
level. Two closed fracture cases experienced distal wound breakdown and this was thought to
occur from the plate prominence in this region.

Another study prospectively evaluated 136 patients who were treated for complex articular
distal radial fractures using two different locking compression plates. Follow-up for two years was
present and the two plate designs did not influence the final overall outcome of fracture fixation.

The use of locking plates has recently been evaluated for use after juxta-articular
oncological resections and use in pathologic fractures in Pakistan. It was noted that many people in
Pakistan present with large tumors and metastasis is common. The goal with surgery was to
remove the tumor with a narrow margin of normal bone and follow-up with neoadjuvant and
adjuvant chemotherapy. Twenty-five patients were included in this retrospective study – eight
patients had pathologic fractures treated with a locking plate and 17 patients had limb salvage
procedures performed with a locking plate. Although three patients experienced complications
(one each: nonunion, wound infection and periprosthetic fracture), the results were favorable for joint sparing limb salvage surgery and the authors concluded that the locking plates resulted in a good and predictable rate of union and that the locking plates performed well in the presence of poor quality and diseased bone.\textsuperscript{67}

Not all studies are in favor of the locking plates: A retrospective study involving forty-two patients with diaphyseal forearm fractures was reported and compared the use of locking plates to dynamic compression plates. Only one patient in each group experienced delayed union and there was no difference between the two groups regarding time to union, operative time, complications and range of motion.\textsuperscript{68} It was concluded that correct surgical technique was more important than the type of plate used in this type of fracture.\textsuperscript{68} The cost-benefit of locking plates has also been questioned in a study looking at the factors influencing the re-operation rate in distal tibial fractures, treated with medial plates (both locking and non-locking plates). This retrospective study evaluated 93 patients over a 10-year period.\textsuperscript{69} Locking plates were more approximately US $1,000 more expensive than non-locking plates but were not associated with a decreased risk of re-operation.\textsuperscript{69}

\textbf{1.6.2 – Clinical Studies of Locking Plates in Veterinary Medicine}

Locking plates have been used to treat radial and tibial fractures in a six-month old dog; both fractures repaired with 3.5mm LCP.\textsuperscript{70} The radial fracture repaired without issue but the tibial fracture sustained implant failure two days postoperatively.\textsuperscript{70} All locking screws were firmly anchored to the plate at the time of revision surgery. The dog went on to heal without
complications following revision surgery with placement of a locking plate and intramedullary nail.  

Both a retrospective study and case report involving locking plates with vertebral stabilization have been reported. Thirteen consecutive cases were treated for ventral stabilization of the cervical spine using the ComPact UniLock system. Two cases had complications with one case needing revision surgery and one dog died in the postoperative period due to pneumonia and deterioration of neurologic status. The authors concluded that the ComPact UniLock system was a suitable implant for treating cervical instabilities, from various causes, in patients with lesions from C1/C2 to C6/C7. Two dogs suffering from triple adjacent thoracolumbar disc protrusions underwent hemilaminectomy, partial annulectomy and bilateral quadruple vertebral body stabilization using the String-of-Pearls locking plates. Both patients experienced resolution of spinal pain postoperatively and improvement in pelvic limb ataxia and paraparesis although screw breakage was present five months postoperatively.

The use of locking plates for facial reconstruction has also been reported for both a comminuted maxillary fracture in a dog and repair of a mandible after partial mandibulectomy in a dog. Full function and cosmetic appearance was present over a year postoperatively in the dog with comminuted maxillary fractures with no complications and only minor complications were present in the dog with mandibular repair (repeated exposure of implant intraorally) which resolved with implant removal.

Haaland et al. retrospectively examined the clinical use of locking compression plates in appendicular fractures of dogs. Forty-seven cases were evaluated and included 34% simple fractures, 6% wedge fractures, 60% comminuted fractures. The fractured bones involved: 11% humerus, 30% radius and ulna, 34% femur and 25% tibia and fibula. The authors reported good
success with bony union present in 46 out of the 47 fractures by 20 weeks postoperatively. Four major complications were present which required revision surgery but it was stated that all implant failures were due to surgical errors and the authors concluded that locking plates are advantageous when other implants prevented the use of bicortical screws, when the locking plate could be used as a bridging plate and in instances when exact plate contouring was not possible.\textsuperscript{50}

The temporary use of a locking plate has been reported. A 32 kg dog sustained trauma, which resulted in a left medial luxating glenohumeral joint, and a right lateral luxation of the elbow.\textsuperscript{75} Closed reduction of the right elbow was performed and the left shoulder was unstable with closed reduction so surgery was indicated. The joint was repaired and a 3.5mm Synthes Locking Round-Hole Reconstruction Plate placed with four screws in the scapular spine and four screws along the humerus (Figure 8) The implants were removed 25 days postoperatively and the dog was deemed to have a good outcome with grade I/V lameness long-term. The authors concluded temporary use of a locking plate for internal fixation of a medial luxated glenohumeral joint yielded satisfactory functional results.\textsuperscript{75}
Figure 1.6: Temporary Transarticular Stabilization with a Locking Plate (Adapted from C. Post: Temporary transarticular stabilization with a locking plate for medial shoulder instability. Vet Comp Orthop Traumatol 2008; 21: 166-170.)

Locking plates were also combined with corrective osteotomy to address procurvatum deformity of the distal femur sustained after mal-union of distal femoral physeal fractures in two dogs. A caudal opening-wedge osteotomy was performed in one dog and a cranial closing-wedge osteotomy in the other dog and both dogs had placement of String-of-Pearls locking plates. Both dogs had excellent outcomes and it was concluded that this type of locking plate was a viable treatment option for cases of severe procurvatum caused by distal physeal fracture mal-union.

Six cases of fetlock arthrodesis using a 4.5mm 14-16-hole broad locking compression plate were retrospectively reviewed. Four of these cases were breeding sound one year postoperatively and the other two were euthanized within weeks of the surgery due to proximal interphalangeal joint luxation. It was concluded that the locking compression plates are a viable option for fetlock arthrodesis although the cost was three times as much as using a limited contact dynamic compression plate with cortical screws. Surgical time was also decreased when using the locking
compression plates and the LCP screws were technically easier to place compared to the standard cortical screws which required tapping.\(^\text{77}\)

Another equine study retrospectively examined the use of locking compression plates in 31 cases for the treatment of fractures and arthrodesis. Complications included 32% incisional infection, 19% implant infection, 22% implant loosening/screw breakage, 16% contralateral limb laminitis, 3% colic, and 3% diarrhea.\(^\text{78}\) Out of the 31 cases, 27 were discharged from the hospital with long term outcome being, 25 sound for intended purpose, 1 lame, and 5 euthanized from complications related to original injury.\(^\text{78}\) It was the author’s conclusion that LCP can be used with success to treat a variety of fractures and arthrodesis with the major drawback being the increased cost of the implants.\(^\text{78}\)

A Fixin locking plate system was reported in six dogs that underwent stifle arthrodesis. Complications occurred in two cases: one dog fractured his tibia at the level of the distal screw 20 days postoperatively and one case experienced a fissure fracture intraoperatively which was addressed with a second locking plate. Mechanical lameness was present in all cases, which is to be expected, and all healed without further complications leading to the conclusion that stifle arthrodesis can be successfully performed using a Fixin locking plate system.\(^\text{79}\)

From these clinical studies, it is evident that locking compression plates can be used successfully to manage fractures in veterinary medicine.

1.7 – Summary of Literature Review

This literature review illustrates the vital need for protection and preservation of the blood supply to healing bone and the detrimental effect that the bone plate can have on the blood supply. Biological osteosynthesis attempts to preserve this blood supply to the fracture site.
As internal fixation of fractures has evolved, conventional plates were found to be lacking in several ways (damage to underlying bone, poor holding power in osteoporotic bone, etc) and this lead to the development of the locking compression plate. With the advent of the locking compression plate, preservation of blood supply is possible and there is greater holding power in osteoporotic bone. Anatomical contouring of the locking plates is not required, although studies show a small bone-plate gap of less than 2mm should be maintained to avoid losing biomechanical stability. Infection rates and delayed union / non-union were not significantly different between the conventional and locking plates and more research will need to be performed to evaluate if there is benefit of using locking plates for these reasons alone.

1.8 – References:


2.1- Introduction to the retrospective study:

This chapter consists of a retrospective study of radial and ulnar fractures in small breed dogs treated with cuttable plates (including the 1.5mm and 2.0 mm mini cuttable plates) in order to determine the current results and complications associated with those implants. The study will also attempt to determine the appropriate weight range for the use of those plates. The study will provide a background on the results of the treatment of radial fractures in miniature breeds and provide a basis for comparison between the 1.5mm locking compression plate and currently used plates.

Abstract - This retrospective study evaluated complication rates for radius and ulnar fractures in small breed dogs in which 1.5 mm to 2.7 mm cuttable bone plates were used for internal fixation. The medical records of all cases from 2004 to 2011 that were presented to our clinic were reviewed. Inclusion criteria were: dogs with bodyweight < 9 kg, fracture of the radius and ulna with open reduction and internal fixation utilizing a cuttable bone plate. Thirty-four fractures in 31 dogs met the inclusion criteria. Of 25 dogs that were available for follow-up, all achieved union, minor complications occurred in 9 and major complications occurred in 8. External coaptation was responsible for complications in 8 cases and the need for coaptation needs to be investigated. Excluding minor complications, 32% of patients required at least 1 additional surgery or additional hospitalization. All but 2 of the dogs returned to full function. The 1.5 mm straight plate was successfully used in all dogs with a body weight of 0.9 to 2.6 kg.
**Introduction**

Radius and ulna fractures represent the third most common fractures in dogs and account for approximately 17% of all fractures in dogs, with many of these occurring in small breeds \(^1\).

Although various techniques can be used for the treatment of radius and ulna fractures in small breed dogs, bone plating remains one of the most frequent methods of stabilization for these fractures. In a retrospective study of 22 small and miniature dogs with radial and ulna fractures treated with bone plates, 89% successfully returned to function. However, complication rates were high and reported to affect 54% of the cases with 18% considered major complications and 36% minor complications \(^2\).

Bone size and small bone fragments are always significant challenges in miniature breeds and the surgeon must choose the most appropriate plate for each dog. Several types of plate are available for the repair of radial fractures in small and miniature dogs and the results for several of these plate types have already been described in the literature \(^2-4\). Cut-to-length plates provide unique characteristics that make them appealing for the treatment of radial fractures in miniature breeds. They are versatile and economical; they come in multiple small sizes and generally offer a short hole-to-hole distance, allowing the placement of several screws in relatively short bone segments \(^5,6\). On the other hand, their small size and high hole-to-plate ratio make them subjectively flexible and weak. Results associated with the use of these cut-to-length plates have not been reported.

It has been over 15 years since any literature has been published reviewing the complication rate and long-term outcome of radius and ulna fractures repaired by internal fixation in small breed dogs. The purpose of this retrospective study was to determine the current complication rate over
an 8-year period treating small breed dogs with internal fixation for radial and ulnar fractures using cut-to-length plates.

**Materials and methods**

**Inclusion criteria**

The medical records of all cases from 2004 to 2011 in which a 1.5 mm, 2.0 mm or 2.7 mm cuttable bone plate was used were reviewed and totaled 97 cases. The criteria for inclusion of cases in the study were: fracture of the radius and ulna with open reduction and internal fixation utilizing a cuttable bone plate (1.5 mm straight plate™, 2.0 mm straight plate™, 2.0 mm/1.5 mm Cut-To-Length Plate™, formerly known as Veterinary Cuttable Plates (VCP) or 2.7 mm/2.0 mm Cut-To-Length Plate™ (DePuy-Synthes, Paoli, Pennsylvania, USA) in dogs with body weight less than 9 kg.

Data pertaining to breed, gender, age, body weight, clinical history, time from injury to surgery, fracture description, previous repair attempts, duration of surgical repair, plate size and configuration, utilization of cancellous bone graft, postoperative fracture alignment, postoperative management, postoperative complications, lameness outcome, and time from fracture fixation until last follow-up radiographs were recorded. Owner compliance was not recorded.

Complications were classified into major and minor based on criteria proposed by Cook et al. Major complications were defined as complications that required further treatment based on current standards of care (implant failure, surgical intervention or hospitalization for bandage complications). Minor complications were defined as complications not requiring additional
surgical or medical treatment to resolve (long-term lameness, bandage complications not requiring specific treatment or hospitalization)\textsuperscript{7}.

All referring veterinarians were contacted to determine if the patient was still living at the time of the survey. A questionnaire was created and mailed to all owners for long-term follow-up, excluding those clients whose dogs were known to have died. A $10 gift card incentive was offered and mailed to every client who completed and returned the questionnaire.

**Results**

Thirty one dogs were included in the study. Pomeranians constituted 10 of the 31 dogs; other breeds were poodle (\(n=7\)), Yorkshire terrier (\(n=4\)), mixed breed (\(n=3\)), Chihuahua (\(n=3\)), Chinese crested hairless (\(n=2\)), Jack Russell terrier (\(n=1\)) and Italian greyhound (\(n=1\)). The mean (median) age at fracture repair was 14.9 mo (7.0 mo). Bilateral fractures occurred in 2 of 31 dogs and 1 dog was presented for fracture of the contralateral radius 7 mo following the first repair for a total of 34 fractures in 31 dogs. There were 14 females and 17 males. Clinical history for all cases included minimal trauma: falling (\(n=12\)), jumping a short distance (\(n=11\)), unknown trauma (\(n=5\)), playing with other dogs (\(n=2\)) or being stepped on (\(n=1\)).

All fractures consisted of complete fracture of the radius and ulna. Of the 34 fractures that were treated initially, 20 were located in the distal diaphysis of the radius, 10 in the mid-diaphysis, 1 in the proximal diaphysis and the location was not recorded in 3 cases. All recorded fractures were either a short oblique or transverse fracture except for 3 cases: 1 case of bilateral fractures (both comminuted) which went on to experience implant failure and 1 mid-diaphyseal fracture (mild comminution) which was lost to follow-up (Table 1).
The majority of dogs had their fracture(s) treated within 3 d after the injury; 8 fractures were repaired between 4 to 10 days after injury. One dog had been treated unsuccessfully with a splint for 9 wk prior to presentation.

All fractures were repaired at our clinic. A total of 45 surgeries were performed on 31 dogs. Bilateral fractures were present in 3 dogs, 1 of which had bilateral radial fractures 7 mo apart. Implant failure requiring plate replacement occurred in 4 dogs (Table 1). Seven additional surgeries were performed to provide additional bone graft and/or to remove or replace screws during the healing process (n=4) or to remove the plate following complete healing because of perceived ongoing complication associated with the implant (n=3). Excluding the surgeries for bilateral fractures, 5 dogs required 2 surgeries and 3 dogs required 3 surgeries. Among the 4 dogs that suffered catastrophic implant failure, 2 were repaired using a plate other than a cuttable plate (2.0 mm DCP™, DePuy-Synthes). For these 2 cases, the revision surgeries were counted as a complication but excluded from our analysis of plates.

Of the initial fracture fixation (34 fractures in 31 dogs), 21 fractures were stabilized with a 1.5 mm straight plate, 4 fractures with a 2.0 mm straight plate, 5 with a 2.0/1.5 mm VCP, and 4 with the 2.7/2.0 mm VCP. The average (median) weight of the dogs treated with each type of plate were 2.2 (2.2) kg for the 1.5 mm straight plate, 2.4 (2.8) kg for the 2.0 mm straight plate, 2.9 (2.0) kg for the 2.0/1.5 VCP, and 5.4 (4.9) kg for the 2.7/2.0 VCP. All fractures were initially repaired using a single plate. The 2 plate revision surgeries included were revised with a 1.5 mm straight plate (from a 2.0 mm straight plate) on a 0.94-kg dog and a stacked 2.0 mm straight plate (from a single 2.0 mm straight plate) on a 2.8-kg dog.

Either a cancellous or cortico-cancellous bone graft (autograft or allograft) was used in 8 of the initial 34 fractures. Three additional bone grafts were performed on 3 dogs during revision.
surgery (Table 1). Two of the bilateral fractures and the fracture that had been splinted for 9 wk prior to presentation were grafted at the time of initial surgery.

Immediately after surgery for the initial 34 fractures, 23 caudal splints were placed, including both of the bilateral fracture repairs. Caudal splints generally consisted of spoon splint, caudal splint made of fiberglass casting material or a portion of a tongue depressor, at the discretion of the clinician. Splints or bandages were rechecked weekly. Three limbs had a soft padded bandage placed after surgery for an undetermined period of time (Table 1). Eight cases had either no bandage post-surgery or only a soft padded bandage placed for a short period of time post-surgery (1-5 days) (Table 1). Of these 8 cases, 4 were lost to follow-up immediately after surgery. One dog required bandage removal because of skin irritation and was lost to follow-up after 27 d. The other 3 were followed a minimum of 101 d post-surgery and all 3 had healed fractures.

Of the 31 dogs, 6 were lost to follow-up after the initial surgery (Table 1). The remaining 25 dogs had at least 1 set of recheck radiographs with a follow-up range of 27 to 169 d. The mean (median) follow up time was 64 (57) d. Complete fracture healing was recorded for 17 cases with the remaining 8 cases having evidence of progression of bony healing. The number of follow-up visits ranged from 2 to 15 and included 2 patients that were hospitalized for 4 d and 10 d each, to address severe bandage complications and associated wounds. Of the 25 cases with at least 27 days of follow-up, major postoperative complications were recorded in 8 patients and minor complications occurred in 9 patients (Table 1).
**Major complications**

Four dogs suffered a catastrophic plate failure (9 to 58 d following initial surgical repair) (Table 1, dogs 1 to 4). Two (dogs 1 and 2) were in the group repaired with the 1.5 mm mini straight plate and had a splint or padded bandage following surgery. The dogs weighed 3.5 and 5 kg and were the largest dogs in the 1.5 mm plate group. Both were successfully repaired with a 2.0 mm DCP and were censored from the study. The 2 other failures happened with the 2.0 mm straight plate. Dog 3 had a bilateral comminuted fracture with postoperative splints. One of the plates broke and the dog developed pressure sores requiring additional hospitalization. The surgery was revised with a stacked 2.0 mm straight plate and no coaptation. Bone graft was used for both the initial and revision surgeries. The fourth dog (#4) failed after a bone screw was removed because of delayed healing and progressive osteopenia. The dog was a 0.94-kg Chihuahua and the smallest dog in the series. The plate and screws were considered oversized and the surgery was revised successfully with a 1.5 mm straight plate. Bone graft and coaptation were used for both initial and revision surgeries.

Dog 5, which had been treated initially with a splint before a 2.0 mm straight plate was applied developed a delayed union and required 2 additional surgeries, 1 for bone grafting, the other to replace failing screws. Healing was confirmed 151 d following surgery and required 15 hospital visits in total.

Three dogs (dogs 6, 7, 8) healed uneventfully but required the implant to be removed because of progressive osteopenia or cold sensitivity. One of these dogs was a 1-kg dog treated with a 1.5 mm straight plate and no coaptation, one was a 3-kg dog treated with a 2.0/1.5 mm VCP and no coaptation and the third (dog 8) was a 4.6-kg dog treated with a 2.7/2.0 VCP and
coaptation. This dog was also hospitalized for 4 days because of a deep olecranon ulcer. Two of these plates were removed in a single staged procedure and the third plate was removed in 2 stages.

**Minor complications**

Nine dogs suffered only minor complications; some dogs suffered more than 1 complication. Minor complications included: bandage complications that did not require specific treatment other than bandage change or removal (5 cases), premature suture removal by the animal (1 case), carpal swelling (1 case), valgus deformity (1 case), intermittent lameness (3 cases) and 1 non-weight bearing lameness at 8 wk.

Long-term lameness outcome among the 25 cases with follow-up was as follows: full return to function in 15, acceptable function in 5; unacceptable function with continued lameness in 2 dogs, and undocumented lameness status in 3 dogs. One of the dogs with unacceptable function was the dog which had the fracture splinted before treatment and required multiple surgeries due to delayed union (Table 1, dog 5). This dog was also diagnosed with elbow incongruity. The other dog with unacceptable function was non-weight bearing 8 wk following fracture fixation despite bone healing on radiographs (Table 1, dog 14). This dog had mild swelling of the carpal joint, however no further follow up was available. Of the 3 dogs that had undocumented lameness status; 2 had mild decrease in carpal and elbow range of motion.

Twenty nine questionnaires were mailed to owners, excluding 5 whose dogs had died. Six questionnaires were returned. All owners were satisfied with the surgery and only 1 dog had residual lameness after strenuous activity. The overall use of the limb was satisfactory for all owners. Two of the 6 patients had implants removed. The long term complications noted from surgery included a scar from a bandage sore and a little tenderness after running too hard resulting
in favoring of the leg for a brief period with a mild lameness. A third owner reported the leg to be sensitive and become cold easily. The questionnaire answers were factored into the overall complication rate.

**Discussion**

Small breed dogs appear to be predisposed to radial fractures with approximately 85% of radius and ulna fractures occurring in the distal third. Morphological differences in the antebrachium of small breed dogs compared to larger breeds are believed to be responsible for this predisposition. Radius and ulna fractures in small breed dogs have been associated with a high complication rate and a high incidence of delayed and non-unions. One study reported up to 54% of small breed dogs treated with plate osteosynthesis developed postoperative complications. The complication rate is approximately 83% if the radial fractures are treated with cast fixation. The reasons for this high rate of complication likely include the size and shape of the bones, technical difficulties associated with the size of the bone fragments and the paucity of soft tissues surrounding the distal antebrachium.

Our retrospective study on radius and ulnar fractures in small breed dogs treated with cuttable plates showed similar trends to previous studies but a higher complication rate with 8 major and 9 minor complications in 25 cases. The overall complication rate of 68% is higher than the previously reported overall complication rate of 54%. However, unlike previous publications, we adopted a stringent definition of complications as suggested by Cook et al. Furthermore, complications associated with coaptation were also counted as if they resulted in alteration of the
original postoperative plan (such as premature bandage removal), even though they may not have
influenced the final outcome.

In our study, coaptation was used in 23 of the 31 cases following the initial repair of the radius
and ulna. External coaptation accounted for complications in 8 cases and resulted in prolonged
hospitalization in some cases. A high complication rate associated with coaptation (up to 63%) has
also been observed by others following the application of bandages, splints or casts in small
animals\textsuperscript{11-14}. External coaptation has the potential for causing pressure sores, swelling and
dermatitis\textsuperscript{13,14} and in this study, all of these were observed within the minor complication category.
Extensive lesions and a deep olecranon ulcer were observed in the major complication category.

The need for external coaptation following internal stabilization is controversial. The decision
to apply postoperative coaptation was made by the surgeon, based on their evaluation of the
surgical repair and the implant strength. In this study, out of the 21 fractures in which the smallest
1.5 mm straight plate was used, 16 had external coaptation placed post-surgery. Four minor and 2
major complications were at least partially attributed to the bandage in that group. Bandages were
also used with all other plate types and complications occurred in all groups. There are however
too few cases of other plate types and our ability to make comparisons between groups was limited.

One would expect that postoperative coaptation would be used in cases deemed at risk for implant
failure and that coaptation would protect against plate breakage. Surprisingly, all 4 implant failure
cases had postoperative coaptation. Two of those dogs were in the 1.5 mm straight plate group.
Two of those dogs had body weights of 3.5 and 5 kg, well above the average weight of the dogs in
that group (2.2 kg). Although there are no published guidelines for the use of these plates, it is
likely that those dogs were too heavy for the plate. All dogs which had successful fracture repair
with the 1.5 mm straight plate had an average body weight of 1.9 kg (range 0.9 to 2.6 kg) and we
therefore suggest that this plate be used for dogs within that body weight range. The other 2 cases that underwent catastrophic failure were in the 2.0 mm straight plate group and both had bone graft and postoperative coaptation. The failure of 1 of the cases remains unexplained as the dog weight (2.8 kg) was in the range of the other dogs in that group and no predisposing factor could be identified other than the fact that this dog had bilateral comminuted fractures. The last case that suffered implant failure was the smallest dog of the cohort and weighed only 0.94 kg. The fracture was initially stabilized with a 2.0 mm straight plate and 2.0 mm screws. In the face of progressing osteopenia and delayed healing, concerns of overly rigid fixation prompted the removal of 2 of the central screws and precipitated the implant failure and fracture of the bone. Fixation with a smaller plate and smaller screws, a bone graft and coaptation resulted in healing of the fracture.

Eight dogs did not have external coaptation following surgery or had coaptation for 5 days or less. Four of those dogs were immediately lost to follow up. Interestingly, none of the remaining 4 dogs for which follow up was available experienced implant failure, suggesting that postoperative coaptation may not always be required.

Osteopenia was diagnosed in several dogs. The degree of osteopenia was subjectively considered significant in 3 cases, prompting surgical removal of all or some of the implants. Self-limiting osteopenia not requiring treatment was not counted as a complication. Concerns about osteopenia, delayed healing or non-union in small and miniature dogs are often raised following radius and ulna fractures. Although the causes of osteopenia following plate fixation have been extensively debated, it is generally accepted that vascular impairment to the bone cortex plays a larger role than stress protection in the development and progression of osteopenia.

Determination of the degree of osteopenia is often subjective and the need for the removal of the implant could be questioned. In 1 dog, the plate was believed to be oversized and although
stress protection cannot be totally ruled out in this dog, vascular impairment caused by the oversized plate and screws was likely to also be a contributing factor.

In addition to the effect of the implant, vascular insufficiency of the distal radius has also been suggested as a cause of healing impairment in small and miniature breeds. Decreased vascular density of the intraosseous blood supply to the distal diaphyseal-metaphyseal junction of the radius has been demonstrated compared to large breeds. Although vascular density is not necessarily synonymous with blood flow, decreased vascular density may contribute to the decreased prognosis for fracture healing in small breed dogs and an increased frequency of delayed union and nonunion compared to similar fractures in large breed dogs or compared to fractures in other bones. To the best of our knowledge, true vascular insufficiency has not been demonstrated. All fractures for which follow-up was available healed and non-union was not observed in this series. Because of the retrospective nature of the study, accurate time to healing was not available.

The lack of long term follow up, inability to compare risk factors due to small group sizes, inability to compare time to healing with or without external coaptation, and the retrospective nature of the study present significant limitations of this study. Because of institutional policy regarding client communication, we were not allowed to contact owners of pets that had died. Furthermore, the response rate to the long-term survey was low. These factors reduced our ability to collect long-term data on several of the dogs. It is possible that these limitations have biased our results. The direction of the bias is however unknown.

Our study on radial and ulnar fracture fixation using 1.5 mm to 2.7 mm “cut-to-length” plates demonstrated a higher overall rate of complication than previously reported in the literature for other plate types. The high complication rate may have been, at least in part associated with the
strict definition of complications used in this publication. Despite the high complication rate, the results were similar to those reported for different plates in a similar population of dogs. All fractures that we could follow achieved clinical union and we believe “cut-to-length” plates remain a good choice for fracture repair. Bandages and splints were used in the majority of cases and were responsible for a large number of complications. The fact that all catastrophic failures occurred despite the splint raises questions about the efficacy of coaptation or the need for coaptation. Perhaps more emphasis should be placed on strict exercise restriction and external coaptation reserved only for tenuous repairs, however, owner compliance could not be assessed and may have played a role. The retrospective nature of the study and the small number of cases that were treated without coaptation limit the strength of this conclusion and additional studies should be conducted. The 1.5 mm straight plate was successfully used in all dogs with a body weight between 0.9 and 2.6 kg. We suggest that this range is appropriate for use of this plate although plate selection and the decision to apply coaptation will remain subjective until stronger guidelines can be developed.

References


Chapter 3 – Biomechanical Comparison of the 1.5mm Locking Compression Plate and 1.5 and 2.0mm mini-cuttable plates & Effect of Bone-Plate Distance on the Biomechanical Properties of a 1.5mm Locking Compression Plate

3.1 Introduction to Scientific Manuscript

This thesis documents a research study investigating the possible use of the 1.5mm locking compression plate for the repair of radial and ulnar fractures in miniature breed dogs. This chapter compares the torsional and compression properties of the 1.5mm locking plates to frequently used alternative for the repair of these fractures: The 1.5mm mini-cuttable plate, 1.5mm mini-cuttable plate stacked and 2.0mm mini-cuttable plate. This chapter also documents the effect of bone-plate distance on the biomechanical properties of the 1.5mm locking compression plate in compression and torsion.

Radius and ulna fractures represent the third most common fractures in dogs and account for approximately 17% of all fractures in dogs.\(^1\) Small breeds appear to be predisposed to radial fractures with approximately 85% of radius and ulna fractures occurring in the distal third.\(^1,2\) Although multiple methods have been described for the fixation of these fractures, bone plating remains the method of choice for small and miniature breeds. Because of the small size of the bones 1.5mm or 2.0mm bone plates are often used for the repair. Despite good overall success rates, radius and ulna fractures in small breed dogs continue to be associated with a high complication rate. Some studies as well as our own retrospective study reveal a complication rate 54% to 68%.\(^3,4\)

The reasons for this high rate of complication are likely multifactorial and may include the size and shape of the bones, technical difficulties associated with the size of the bone fragments and the paucity of soft tissues surrounding the distal antebrachium.
Concerns about vascularization of the distal radius in small breed dogs have been raised and a decreased vascular density of the intraosseous blood supply to the distal diaphyseal-metaphyseal junction of the radius has been demonstrated compared to large breeds.\(^5\)

The fractures often affect the distal part of the bone leaving only a small fragment available for fixation. In many cases, only 2 screws can be placed in the distal fragment and the use of special “T plates” is necessary to achieve adequate fixation. As the fracture is more frequent in young dogs, the relative softness of the bone and the small fragment size pose a real challenge to the surgeon trying to get adequate screw purchase to achieve fixation.

Because the repairs are often considered tenuous, external coaptation is often applied postoperatively and may lead to many bandage complications.\(^3\),\(^6\)-\(^8\)

Synthes recently released a 1.5mm locking compression plate that could be suitable for use in distal radial fractures in miniature breeds. Because of the potential benefits of locking plates such as preservation of the blood supply, increased purchase in small or poor quality bone fragments etc., this plate could be ideal for treatment of distal radial fractures in small breed dogs but no biomechanical studies have been performed yet to confirm its suitability.

The purpose of this research was to compare the biomechanical properties of the new Synthes 1.5mm locking compression plate to the 1.5mm mini-cuttable plate and 2.0mm mini-cuttable plate in compression and torsion as well as evaluate the biomechanical properties of a 1.5mm locking compression plate in compression and torsion with increasing distances between the plate and the bone surrogate. We hypothesized that the 1.5mm locking compression plate would be equivalent or superior to the 1.5mm mini-cuttable plate and the 2.0mm mini-cuttable plate in compression and torsion. We further hypothesized that increasing the bone-plate distance
of the 1.5mm locking compression plate would not significantly affect its biomechanical properties
and that the results in compression and torsion would be comparable to the results of the 1.5mm
locking plate placed in contact with the bone.

3.2 – Materials and Methods

Determining appropriate simulated bone size:
Lateral radiographs of the radius-ulna of 10 miniature and small breed dogs repaired with a 1.5 or
2.0mm mini-cuttable bone plate were reviewed to determine the average offset distance between
the plate axis and the mechanical axis of the radius. The mechanical axis was determined by
drawing a line between the center of the radio-humeral and radio-carpal joints on the lateral
radiographic view. The perpendicular distance between the mechanical axis and the central portion
of the plate was measured and averaged.

Radial surrogate:
The simulated bone segments were made from acetron rods. The diameter of the acetron rods was
selected so that the distance between the mechanical axis of the rod and the plate would be similar
to the distance previously calculated on radiographs. This was done to replicate loading conditions
of the plate caused by the natural curvature of the radius.
The rods were 80mm in length. The ends of the acetron rods were hollowed out to create a shallow
spherical cavity to accommodate two 10mm diameter steel balls of the compression testing jig.
To accommodate the plate, one side of the cylindrical rod was shaved to create a 4mm wide flat surface. Each rod was cut into 2 halves and the screw holes were drilled and tapped with either a 1.1mm drill bit and 1.5mm tap for all 1.5mm screws, or a 1.5mm drill bit and a 2.0mm tap for all 2.0mm screws. All constructs were secured in a custom made vise for all drilling and tapping to ensure consistency.

**Plates and screws:**

Six different types of plate constructs were tested. These included the 1.5mm mini-cuttable plate, stacked 1.5mm mini-cuttable, 2.0mm mini-cuttable plate and 1.5mm locking compression plate, 1.5mm locking compression plate with 0.5mm space between the plate and the bone surrogate, and 1.5mm locking compression plate with 1mm space between plate and bone surrogate. All plates were 6-hole plates and six screws were used for each construct, the screw size corresponding to the plate size recommendations. All screws were 12mm in length to allow for full purchase of the acetron rod and bone plate, except for the stacked 1.5mm mini-cuttable, which were 14mm in length to accommodate for the thickness of the stacked plates. All screws exited past the acetron rod.

**Plate-construct assembly:**

The bone plates were secured to the acetron rods with three screws on each side, leaving a 1mm fracture gap in the middle. The acetron rods were beveled on the side opposite to the plate to allow bending of the constructs without causing contact of the rods, creating a true open gap model throughout the testing range. The insertion torque applied to each screw was standardized to 10 cNm using a torque-limiting screwdriver for all 1.5mm screws and 15 cNm for all 2.0mm screws.
The torque-limiting screwdriver was certified to be calibrated to +/- 0.06cNm when set to 2.0cNm torque and increased in upper and lower limits to +/- 0.46cNm when set to 15cNm torque. The 1.5mm mini-cuttable, 1.5mm mini-cuttable stacked and 2.0mm mini-cuttable plate constructs had two sets of constructs made and the one set was hand tightened to two finger tight and the other set used the aforementioned torque limiter.

For both constructs with bone-plate offset, either 0.5mm or 1mm metal spacers (stainless steel wires) were placed between the acetron rod and bone plate while locking the screws to provide uniform and consistent spacing. The spacers were removed following screw tightening. Sixteen constructs were made with each type of plate totaling 96 constructs. For each assembly type, 8 samples were tested in compression and 8 in torsion (n=8).

**Figure 3.1: Plate-Bone model constructs**
Top of image: 2.0mm mini compression plate. Middle of image: 1.5mm mini cuttable plate. Bottom of image: 1.5mm locking compression plate. Not pictured: 1.5mm stacked mini cuttable.

**Mechanical testing:**

Compression testing:
Each construct was mounted in a custom designed fixture consisting of 2 steel balls mounted on the servo-hydraulic testing machine (Instron 5965, Norwood, MA) mounted with a 5kN load cell. The specimens were pre-loaded to 5N then compressed at a constant rate of 0.1mm/sec until failure.

Data were collected at 100 Hz frequency using an IEEE interface\textsuperscript{a} with software written for control of displacement and data collection from the mechanical testing frame\textsuperscript{b}

The data was exported into Excel\textsuperscript{c} for processing.

The load deformation curves were graphed and the straight portion of the curve was determined visually. Bending stiffness was calculated by determining the slope of the straight portion of the load-displacement curve by using linear regression analysis for a best fit. Bending strength was determined by finding the intersection of the load deformation curve with a line parallel to the straight portion of the curve but offset by 0.02mm.

\textsuperscript{a} IEEE 488 Connector: Austin, TX, USA
\textsuperscript{b} Bluehill Version 3.25 Testing Software for Mechanical Testing Systems: Instron, Norwood, MA, USA
Figure 3.2: Compression Testing Assembly

Figure 3.3: Diagram illustrating methods for determining bending properties of bone plates
Torsional testing:

Eight constructs of each construct type were tested in torsion to failure. The ends of each construct were secured using set screws in premade aluminum blocks and fit securely into the custom made torsional jig, allowing 3 degrees of freedom (shortening and rotation in 2 planes). The testing jig was connected to the servo-hydraulic testing machine (Instron 5965, Norwood, MA) using a stainless steel cable, resulting in a lever arm of 130mm between the rotational axis and the attachment of the cable. Displacement rate for the testing machine was set at 2.27mm/sec, resulting in a rotational displacement of 1°/s. A preload of 0.25N (0.01625Nm torque) was applied initially and torque was applied until failure.

The data files were smoothed using a moving average ± 18 data points prior to graphing the torque/deformation curves. Torsional stiffness was calculated by determining the best straight line in the initial linear region of the torque/deformation curve by using linear regression analysis for a best fit. A second line parallel to this linear portion was drawn but offset by 1 degree and the intersection between the two lines was used to define the torsional strength.
Figure 3.4: Torsional Testing Unit (Arrow: Cable)

Figure 3.5: Diagram illustrating methods for determining torsional properties of bone plates

Sample collection
All tests were performed on a servo-hydraulic testing machine\textsuperscript{d} with a 5kN load cell. Data were collected using an IEEE interface \textsuperscript{e} with software written for control of displacement and data collection from the mechanical testing frame at 100 Hz intervals (0.01 sec) (Bluehill 3.25)\textsuperscript{f} The data was exported into Excel\textsuperscript{g} for processing.

Statistical analysis

All data was tested for normality using a Shapiro-Wilk test for normality and the examination of the residuals. Seven total outlier observations were examined and excluded. Log transformation was used to improve normality. Kruskal Wallis Anova was used if normality was not achieved, otherwise, an analysis of variance with post hoc Tukey adjusted t-tests were used to detect differences between groups for stiffness, ultimate load at failure, yield load, and displacement. A standard statistical software package was used for statistical testing.\textsuperscript{h} Significance was set at $p \leq 0.05$ for all tests.

3.3 – Results

Offset distance between plate and mechanical axis:

Average offset distance between the mechanical axis of the radii and the axis of the plate measured on radiographs was 4.65mm (range: 3.50mm-6.98mm). Doubling this distance provided the required diameter of the bone surrogate. Acetron rods with the diameter closest to this distance

\begin{itemize}
  \item \textsuperscript{d} Instron 5965, Norwood, MA
  \item \textsuperscript{e} IEEE 488 Connector: Austin, TX, USA
  \item \textsuperscript{f} Bluehill Version 3.25 Testing Software for Mechanical Testing Systems: Instron, Norwood, MA, USA
  \item \textsuperscript{g} Microsoft. (2011). Microsoft Excel [computer software]. Redmond, Washington: Microsoft.
  \item \textsuperscript{h} SAS® 9.3, SAS Institute Inc., Cary, NC, USA
\end{itemize}
were chosen. Acetron rods with a diameter of 3/8” (or 9.5 mm) were selected. After shaving a flat surface of 4mm to accommodate the plate, the distance between the center of the rod and the center of the plate was (4.19mm+1/2 plate thickness).

**Compression testing: (excluding plate offset groups)**
When tested in compression, all specimens failed by plastic deformation of the plates and no implant breakage or bone surrogate breakage occurred during testing.
The stiffness of the different constructs ranged from a mean of 74.7 to 153.4 N/mm. The 2.0mm mini cuttable, the 1.5mm mini cuttable stacked and the 1.5mm LCP constructs had the highest stiffness and were all statistically similar. Of all the constructs, the 1.5mm mini cuttable had statistically the lowest stiffness and had approximately half of the stiffness of the other groups.

Mean yield strength was 89.1N for the 2.0 mini cuttable, 62.3N for the 1.5 mini cuttable stacked, 38.4N for the 1.5 LCP and 44.5N for the 1.5 mini cuttable plates. There was statistical significance between all plate types for load at failure. The 2.0 mini cuttable had the highest load at failure while the 1.5 LCP displayed the lowest load at failure of all the constructs. (Table 3.1)
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<td>2.0mm mini</td>
<td>89.1 d</td>
<td>83.0</td>
<td>95.8</td>
</tr>
<tr>
<td>1.5mm LCP</td>
<td>0.3 b,c</td>
<td>0.2</td>
<td>0.4</td>
</tr>
<tr>
<td>1.5mm mini</td>
<td>0.9 a</td>
<td>0.7</td>
<td>1.0</td>
</tr>
<tr>
<td>1.5mm mini stacked</td>
<td>0.8 b</td>
<td>0.7</td>
<td>0.9</td>
</tr>
<tr>
<td>2.0mm mini</td>
<td>1.1 c</td>
<td>1.0</td>
<td>1.2</td>
</tr>
</tbody>
</table>

Table 3.1: Summary of Plate Bending Data: Different superscript letters within each of the parameters tested indicate statistical differences.
Compression testing: Stiffness

Compression testing: Yield Strength
Figure 3.6: Summary of Compression test Data. Different superscript letters indicate statistical differences. Lower limit: 95% Confidence interval lower limit, Upper Limit: 95% Confidence interval upper limit

**Torsion testing:**

Torsional stiffness ranged from 50.0 to 82.8 Nmm. The 2.0mm mini cuttable was the stiffest construct, followed by the 1.5mm mini cuttable stacked. The 1.5mm LCP and 1.5mm mini cuttable constructs were the most compliant and were statistically similar to each other.

Mean yield torque was similar for the 1.5mm LCP, the 1.5mm mini cuttable and the 1.5mm stacked mini cuttable. The 1.5mm mini cuttable statistically sustained less torque prior to failure than the 1.5mm mini stacked although both plates were statistically similar to the 1.5mm LCP plate. The 2.0mm mini cuttable sustained the highest torque prior to failure and was statistically different from the others. (Table 3.2)

Displacement was not significantly different between any of the constructs.
Table 3.2: Summary of Torsional Testing Data: Different superscript letters within each of the parameters tested indicate statistical differences.
Figure 3.7: Summary of Plate Torsional Data. Different superscript letters indicate statistical differences. Lower limit: 95% Confidence interval lower limit, Upper Limit: 95% Confidence interval upper limit.
Results of testing with plate offset (0.5mm and 1mm from the bone surrogate)

(1.5mm Locking Compression Plates Only):

Compression

The average stiffness obtained during axial compression testing was 139.1 N/mm for the 1.5mm LCP flush (0mm offset) with the surrogate, 88.1 N/mm when the plate was offset from the surrogate by 0.5mm and 87.7 N/mm when the offset was 1mm. Statistical significance was present between the 0mm offset and both the 0.5 and 1mm offsets. There was no statistical significance between the 1.5mm LCP flush and the 1.5mm LCP with 1.0mm distance but it should be noted that early failure of 6 of the 8 constructs was observed in the 1mm offset group and only 2 samples yielded load deformation curves that could be reliably analyzed, making statistical comparison difficult. Only 1 sample from the 0.5mm offset group underwent early failure and of the load-displacement graph was removed from the analysis.

The results obtained during axial compression revealed that the average yield load was 38.4N for the 1.5mm LCP flush with the bone surrogate (n=8), 25.2N for the 1.5mm LCP at 0.5mm distance (n=7) and 26.3N for the 1.5mm LCP with 1.0mm distance away from bone surrogate (n=2). The 1.5mm LCP flush against the bone surrogate was statistically stronger than either plate with distance away from bone surrogate. Increasing the distance of the plate to the bone decreased the load required to failure. (Table 3.3)

No statistical difference between any constructs was present for displacement during compression testing. (Figure 3.8)
<table>
<thead>
<tr>
<th>Offset distance (mm)</th>
<th>Mean</th>
<th>95% Confidence Interval</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td></td>
<td>Lower</td>
</tr>
<tr>
<td><strong>Stiffness(N/mm)</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>139.1</td>
<td>125.0</td>
</tr>
<tr>
<td>0.5</td>
<td>88.1</td>
<td>73.9</td>
</tr>
<tr>
<td>1.0</td>
<td>87.7</td>
<td>59.4</td>
</tr>
<tr>
<td><strong>Yield Load(N)</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>38.4</td>
<td>35.9</td>
</tr>
<tr>
<td>0.5</td>
<td>25.2</td>
<td>23.4</td>
</tr>
<tr>
<td>1.0</td>
<td>26.3</td>
<td>23.0</td>
</tr>
<tr>
<td><strong>Displacement (mm)</strong></td>
<td></td>
<td></td>
</tr>
<tr>
<td>0</td>
<td>0.3</td>
<td>0.2</td>
</tr>
<tr>
<td>0.5</td>
<td>0.3</td>
<td>0.2</td>
</tr>
<tr>
<td>1.0</td>
<td>0.3</td>
<td>0.1</td>
</tr>
</tbody>
</table>

Table 3.3: Summary of Plate Bending Data (Offset): Different superscript letters within each of the parameters tested indicate statistical differences. n=8 for the 0mm offset distance, n=7 for the 0.5mm offset and n=2 for the 1mm offset.
Figure 3.8: Load / Displacement Curves – 1.5mm LCP flush (top image), 1.5mm LCP 0.5mm offset (middle image) and 1.5mm LCP 1.0mm offset (bottom image). Progressive distance away from bone results in less consistent graphs.

Failures occurred either from screw head unlocking from the plate or screw bending. Screw breakage or bone surrogate fracture did not occur in any construct. Almost all constructs of the
1.5mm LCP with 1.0mm distance had screw bending and screw head unlocking with plate slippage occurring early during testing.

**Torsion**

During torsion testing the 1.5mm LCP fixed with 0mm offset to the bone surrogate remained the strongest and stiffest construct statistically. Both constructs with an offset of 0.5 or 1mm were similar to each other and statistically weaker and more compliant than the 1.5mm LCP with 0mm offset.

There was no statistical difference between the displacement at yield for any of the constructs.

<table>
<thead>
<tr>
<th>Offset distance (mm)</th>
<th>Mean 95% Confidence Interval</th>
<th>Lower</th>
<th>Upper</th>
</tr>
</thead>
<tbody>
<tr>
<td><strong>Stiffness (Nm/degree)</strong></td>
<td>0 0.05 (^a) 0.04 0.05</td>
<td>0.04 0.05</td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.5 0.03 (^b) 0.02 0.03</td>
<td>0.02 0.03</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1.0 0.03 (^b) 0.02 0.03</td>
<td>0.02 0.03</td>
<td></td>
</tr>
<tr>
<td><strong>Yield Torque (Nm)</strong></td>
<td>0 0.55 (^a) 0.46 0.55</td>
<td>0.46 0.55</td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.5 0.30 (^b) 0.29 0.34</td>
<td>0.29 0.34</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1.0 0.28 (^b) 0.25 0.30</td>
<td>0.25 0.30</td>
<td></td>
</tr>
<tr>
<td><strong>Torsion Displacement (degrees)</strong></td>
<td>0 9.1 (^a) 7.9 10.4</td>
<td>7.9 10.4</td>
<td></td>
</tr>
<tr>
<td></td>
<td>0.5 7.1 (^a) 6.2 8.1</td>
<td>6.2 8.1</td>
<td></td>
</tr>
<tr>
<td></td>
<td>1.0 7.0 (^a) 6.1 7.9</td>
<td>6.1 7.9</td>
<td></td>
</tr>
</tbody>
</table>

Table 3.4: Summary of Plate Torsional Data. \(^a\) Different superscript letters within each of the parameters tested indicate statistical differences. \(n=8\) for all groups.
3.5 – Discussion

Compression testing:

Because of the curvature of the radius in dogs, the mechanical axis of the radius is offset relative to cranial surface of the bone. Axial compression of the radius therefore creates a bending moment on the radius causing tension on the cranial cortex and compression on the caudal cortex.\textsuperscript{9,10} In a fracture situation with loading of the leg, the bone plate would be subjected to a similar bending moment, as its axis would also be offset relative to the mechanical axis of the bone. In order to simulate loading conditions of a bone plate in our experiment, the average offset between the mechanical axis of the radius and the bone plate was measured on postoperative radiographs from 10 clinical cases. The radius of the bone surrogate was chosen to closely match the averaged offset measured on radiographs so that the loading of the plates would more closely simulate in vivo loading conditions, creating both compression and bending on the implant as it is loaded.
The results of this study showed that the 1.5mm LCP was biomechanically similar in stiffness to the larger 2.0mm mini cuttable, 1.5mm mini cuttable stacked and 1.5mm mini cuttable. Stiffness is a measure of the amount of deflection that a load causes in a material and within these study parameters the 1.5mm LCP deformed approximately the same amount as the three other tested plates for the same amount of load applied. Stiffness is important because it controls the amount of strain at the fracture. Improved stiffness prevents the plate from bending during cycling after internal fixation of a fracture. On the other hand, if the implant is too stiff, stress protection may occur and delayed healing may be observed. The bending stiffness of an implant is dictated by the material properties of the implant but also by its moment of inertia. The moment of inertia is a mathematical formula taking into account the size of the implant and the distribution of the material around its axis or plane of bending. For a rectangular structure, the formula is (base (height)^3 / 12). Although both dimensions of the plate are important, the height of the structure in the plane of bending has the most influence on bending stiffness. In this case, the thickness of the plate is likely going to be the determinant factor influencing the bending stiffness. The stacked 1.5mm plates had the greatest thickness and the highest resistance to bending, although not significantly different from the LCP and the 2.0 mini plate.

The working length of the implant will also influence bending stiffness of the constructs. The bending stiffness of an object is proportional to its length and the working length of an implant corresponds to the distance between the screws closest to the fracture. The distance between the center of one screw hole to the next was different for all plates tested. (Table 3.5) The largest distance was for the 1.5mm LCP, then the 2.0mm mini cuttable and smallest for the 1.5mm mini cuttable. The longer the working length, the more flexible the plate, so a decrease in stiffness would be expected. Despite having the longest working length out of the plates tested, the 1.5mm
LCP had a stiffness that was similar to both the 2.0mm mini cuttable and 1.5mm mini cuttable stacked. This is likely due to the relatively minor differences in working length between implants and the more significant effect of the thickness of the plate on the bending stiffness. Large hole-to-hole distances, however, may have a clinical impact when fixing distal radial fractures as the small size of the distal fragment may preclude its use.

Yield strength, is a measure of the maximum load that can be placed on a material before plastic deformation occurs. It is mostly a function of the material properties of the implant and its size. The 1.5mm LCP constructs were the weakest (failed with the lowest load) construct out of the four plates tested. Despite its larger width and thickness, the LCP plate was 14% weaker than the 1.5mm mini plate. It must be noted however that the underside of the LCP plates is undercut to allow periosteal vascularization and to create a more uniform area bending moment of inertia along the length of the implant. These undercuts reduce the cross section of the implant between the screw holes, decreasing the strength of the implant relative to a more traditional plate with no undercuts. Because our fracture gap model left no plate holes open between the fragments, the LCP plate with its undercuts might have been at a disadvantage relative to the 1.5mm cuttable plate which has a solid section between screw holes. A similar effect has been observed when testing the biomechanical properties of the LCP plates against traditional DCP plates. A fracture model with a large gap causing one of the screw hole to be left open was not tested but would have likely produced results more favorable to the 1.5mm LCP.

The 2.0mm mini plate constructs were the strongest constructs tested and exceeded the strength of the stacked cuttable plates constructs. Although the cross sectional area of the stacked 1.5mm plates is larger than the cross section area of the 2.0mm plate it is possible that other factors such as
the size of the screws, the strength of the screws or the frictional forces generated by the screws may have influenced the results. (See Table 3.5). Research on stacked VCP plates has shown that stacked plates increase the strength and stiffness of constructs compared to single plate constructs by a statistically significant amount.\textsuperscript{16}

Minimal displacement occurred during testing with the 1.5mm LCP compared to the other plate constructs. It is expected that the high stiffness and lower strength would result in a shorter displacement at yield compared to plates of similar stiffness but higher yield strength. It is, however likely that the locking of the screws within the plate also contributed to a smaller displacement of the constructs during testing compared to the non-locking plates.

<table>
<thead>
<tr>
<th>Plate</th>
<th>Distance between center of screw hole</th>
<th>Plate Thickness</th>
<th>Plate width</th>
</tr>
</thead>
<tbody>
<tr>
<td>2.0mm mini cuttable</td>
<td>6mm</td>
<td>1.0mm</td>
<td>5.0mm</td>
</tr>
<tr>
<td>1.5mm mini cuttable</td>
<td>5mm</td>
<td>0.8mm</td>
<td>3.8mm</td>
</tr>
<tr>
<td>1.5mm LCP</td>
<td>7mm</td>
<td>1.0mm</td>
<td>4.3mm</td>
</tr>
</tbody>
</table>

Table 3.5: Comparison of plate width and distance between screw holes

**Torsion**

Torsional stiffness and strength are determined by the material properties and the size and geometry of the material being tested. The torsional stiffness of the 1.5mm LCP was equivalent to the stiffness of the 1.5mm mini plate despite the fact that the 1.5mm LCP is larger and thicker than the 1.5mm mini plate. Similarly to the compression testing, the scalloped underside of the plate as well as the beveled edges of the 1.5mm LCP likely played a role in decreasing the polar moment of inertia of the implant, relative to the solid portion of the mini plate. Not surprisingly, the stacked
1.5mm mini plate had a higher torsional stiffness than the single layer plate due to the increased cross section area on the stacked implant. The 2.0 mini plate had the highest stiffness of all the constructs, consistent with the larger size of the implant.

The yield torque was identical for the 1.5mm LCP, 1.5mm mini and the stacked 1.5mm mini plate. The 2.0mm mini plate had the highest torque at yield, however, because of its higher stiffness, angular displacement at yield was not different from the other plate constructs.

In this study, we aimed to provide biomechanical evidence for the use of the 1.5mm LCP in a clinical setting by comparing the 1.5mm LCP to commonly used plates for the repair of radial fractures in miniature breeds. The 1.5mm LCP has similar biomechanical properties to the 1.5mm mini cuttable, both in stiffness and yield torque. It was the weakest plate in compression testing by a small, but statistically significant margin to the 1.5mm mini cuttable plate. Our retrospective study on radial fractures in miniature breeds found that the 1.5mm mini cuttable plate was successful for all dogs with an average weight of 2.2kg. As the properties of the 1.5mm LCP are similar to those of the 1.5mm mini plates, we would anticipate that the plate would be adequate for similar population of dogs. Although the strength of the LCP was 14% less than the strength of the 1.5mm mini plate, this difference may not be clinically relevant. Kinetic studies have shown that dogs exert approximately 106% of their body weight on a single forelimb at the trot. This means a 2kg dog would exert approximately 20.6N force during the trot (2.0kg*106% = 2.1kg; 2.1kg*(9.81Newton/1kg) = 20.6Newton). Even though the 1.5mm LCP was statistically the weakest of the plates tested, yield load at failure was 38.4N, which is almost double the force applied for a 2kg dog at the trot in the forelimb. Fatigue testing of the implants should be performed to further characterize these implants.
The model that we tested did not leave a screw hole open over the fracture site. This may have disadvantaged the LCP relative to the mini plate. It is likely that the properties of the plate would have exceeded those of the mini plate in a model with a screw hole open. Additional research will be needed to confirm this hypothesis.

Although the distance between screws in the LCP could be a disadvantage in fractures with small distal fragment, the benefit of the locking screws over traditional screws may negate this disadvantage.

**Effect of the off-set distance:**

One of the great benefits of locking plates over traditional plates is the fact that the plate does not need to be in contact with the bone to confer stability.\(^{18}\)

This reduces the need for contouring the plate but also preserves the periosteal vasculature and decreases the area of contact between the plate and the bone.\(^{18}\) In a study done by Haug et al, the effect of bone plate contouring and offset (distance between the plate and the bone) in surrogate human mandibles was investigated. Locking and non-locking plates were contoured to leave an offset of 0, 1 or 2 mm between the plate and the bone. Loading of the mandibles was performed and displacement, stiffness and yield load were measured and compared. The different offsets had an effect on the non-locking plate systems but had no effect on the locking plate systems.\(^{19}\) In long bone fracture, the effect of the bone–plate distance was investigated with the 4.5 DCP. The results confirmed that an offset distance between the bone and the plate up to 2mm, did not significantly affect the biomechanical properties of the constructs while larger distance between the bone and the plate significantly weakens the constructs.\(^{20,21}\)
Our results contrast drastically with those of Ahmad and Tomlinson. Both constructs with 0.5 and 1mm offsets were significantly weaker than the constructs made with the plate against the bone surrogate in compression and in torsion. Not only the stiffness and strength of the constructs with 0.5 and 1mm offsets were statistically weaker than the 0mm offset group but they also showed evidence of early failure. In compression testing, failure of the screw locking mechanism was evident in 4 out of 8 constructs in the 0.5mm offset group as opposed to only 2/8 in the 0mm offset group. Six constructs in the 1mm offset group displayed evidence of early failure and only 2 constructs produced usable data.

Similarly, the 0.5mm and 1mm offset group displayed early failure in torsion when compared to the 0mm offset group. Although the 2 offset groups failed at low torque, uncoupling of the screws from the plate was less evident and did not appear to play a major role in the failure mode in torsion testing compared to the compression testing. Bending of the screw was however observed in most samples suggesting that the screws themselves were the weakest link in the constructs. In compression, the uncoupling of the screws from the plate was evident by a clicking sound and a sharp but limited drop in the load/deformation curve. Because the compression resulted also in bending of the implant, it is likely that, as the construct bent, the screw closest to the fracture was pushed out of the plate, resulting in the uncoupling of the head of the screw from the plate. Although similar uncoupling happened in 2 of the 0mm offset specimens (1 screw each), uncoupling was likely prevented by the contact of the bone surrogate with the plate, limiting movements of the screw relative to the plate. In the 0.5mm offset group of the 4 constructs with evidence of uncoupling, three showed uncoupling of at least 2 screws and one showed uncoupling of 4 of the screws.
Uncoupling was not readily observed in torsion. This could be because the screw heads are more resistant to extrusion from the plate in this testing mode. The bending of the screws and the low stiffness and strength observed from the load deformation curves, suggests that the screws themselves were the weakest part of the constructs when the plate is not in contact with the bone. In the 0mm offset group, the screws are embedded into the plate or the bone surrogate, protecting the screws from bending. In the 2 remaining groups, 0.5mm or 1mm of the screws was exposed between the bone surrogate and the plate and subject to bending. As the screws are smaller than the plates, they bent before the plate. Locking screws are subjected to higher bending forces than traditional screws. Because of this, locking screws are generally made with a larger core diameter than traditional cortical screws. In the 1.5mm LCP system, however, the locking screws have the same core and thread diameter as the non-locking screws, making them susceptible to bending. A similar weakness has been identified in locking systems that use regular cortical screws as their mode of attachment to the bone.

Regardless of the testing mode or mode of failure, both 0.5mm and 1mm offsets resulted in significantly weak systems and were prone to early failure. For this reason, we cannot recommend leaving a space between the bone and the plate when using the 1.5mm LCP.

3.5 - Limitations of the study:

Similarly to all in vitro studies, there are several limitations to our study:

A bone surrogate model was chosen to allow for uniformity among constructs and it was stronger than bone to allow for full testing of the plates. Bones surrogates are often used because they decrease the inherent variability of biological specimens, but they could also change our results compared to testing with small breed dog bone.
Although we emulated the weight bearing condition of the distal radius in miniature breeds by taking into account the offset distance between the plate axis and the weight bearing axis, in vitro testing conditions can only represent a crude and simplified representation of the in vivo loading conditions of bone. Our model used three screws per fragment but in most clinical cases, the distal radial fracture may only allow for 2 screws and therefore, these results may not apply to the most common fracture seen in small breed dogs. Along with that, our study had a fracture gap model for all testing but in clinical cases, many of these fractures are simple and load shearing can be achieved with fracture fixation.

The torsional testing jig allowed 3 degrees of freedom by allowing shortening and angulation in 2 planes. Ideally a jig allowing for 5 degree of freedom should be so that the specimens are not constraint. This is particularly important if the implant is not perfectly aligned with the torsional axis of the jig. In our case, the bone plate was slightly offset from the torsional axis of the jig by the amount equivalent to the radius of the acetron rods. This means that the plates were not tested in pure torsion.

However, as the conditions were similar to all constructs, we believe that the comparisons between constructs are still relevant, although they may not depict the exact properties of the implants.

A torque-limiting screwdriver was used for all initial testing. This was chosen to standardize the torque of all constructs. Under-tightening screws would not generate the friction required for non-locking plates to achieve stability while over-tightening locking screws could lead to cold welding. A recent study examined the effect of technical errors on accuracy of the torque limiter in locking plate osteosynthesis. They examined the use of torque limiters under hand power and drill power
using low and high velocity. There was a significant difference in measurements between all groups and it was concluded that a higher than expected torque could occur when placing locking screws inserted with improper technique. In the 1.5mm LCP we experienced over-tightening of the locking screws evidenced by the plastic deformation of the screwdriver. There were no recommendations for the amount of torque to be applied for the 1.5mm locking screws so this was decided in the lab testing a variety of torque settings until a subjectively acceptable tightness was achieved: 10cNm for all 1.5mm screws and 15cNm for all 2.0mm screws. Higher torques for the 1.5mm LCP resulted in torsional damage to the screwdriver during preliminary testing. This torque-limiting screwdriver was used for initial testing for both compression and torsion of all six constructs.

Following analysis of the load/displacement curves, which showed inconsistent curves with great variation during compression testing, it was concluded that the torque-limiting screwdriver was likely under tightening the screws and skewing the data. The torque limiter data for the 1.5mm mini cuttable plate, 1.5mm mini cuttable stacked plate and 2.0mm mini cuttable plate was replaced by hand tightening for the compression testing only. No apparent effect was observed on the torsion testing.

<table>
<thead>
<tr>
<th>Group</th>
<th>Type</th>
<th>Stiffness (N/mm)</th>
<th>Yield Load (N)</th>
<th>Displacement (mm)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.5 mini</td>
<td>Hand tighten</td>
<td>71.6</td>
<td>42.8</td>
<td>0.94</td>
</tr>
<tr>
<td>1.5 mini</td>
<td>Torque limiter</td>
<td>32.8</td>
<td>55.0</td>
<td>2.18</td>
</tr>
<tr>
<td>1.5 mini stacked</td>
<td>Hand tighten</td>
<td>152.9</td>
<td>62.4</td>
<td>0.78</td>
</tr>
<tr>
<td>1.5 mini stacked</td>
<td>Torque limiter</td>
<td>62.2</td>
<td>73.9</td>
<td>2.09</td>
</tr>
<tr>
<td>2.0 mini</td>
<td>Hand tighten</td>
<td>138.6</td>
<td>89.2</td>
<td>1.10</td>
</tr>
<tr>
<td>2.0 mini</td>
<td>Torque limiter</td>
<td>239.0</td>
<td>69.7</td>
<td>0.39</td>
</tr>
</tbody>
</table>

Table 3.6: Summary of Hand Tightening vs. Torque Limiter
Legend: 1.5 mini: 1.5mm mini cuttable plate; 1.5 mini stacked: 1.5mm mini cuttable plate stacked; 2.0 mini: 2.0mm mini cuttable plate
An additional set of compression testing was then performed using all new constructs - 1.5mm mini cuttable plates, 1.5mm mini cuttable plates stacked and 2.0mm mini cuttable plates. Eight constructs of each of these were, again, tested as previously described in compression with the exception that they were all hand-tightened. This data was more consistent. The data that was ultimately analyzed included torque-limiter tightening for all torsional testing and all 1.5mm locking compression plates. Hand tightening data was used for the 1.5mm mini cuttable plates, stacked 1.5mm mini cuttable plates and 2.0mm mini cuttable plates. In an ideal world, the entire construct fleet would have been retested with hand tightening in both compression and torsion but this was not possible due to finances. Based on Table 3.6, it is evident that hand tightening gives stiffer constructs than using the torque limiter for the 1.5mm mini cuttable, single and stacked. For the 2.0mm mini cuttable, hand tightening resulted in significantly less stiff constructs. It is unknown why this was the case for the 2.0mm mini cuttable plates. Perhaps using the torque limiter tightened the plates to the bone surrogate better than two finger tightening or perhaps when hand tightening, the acetron rod absorbed most of the torque and more force should have been used.

The acetron rods consisted of a solid cylindrical rod with no distinction between cortices and medullary cavity. This resulted in the screws being fully embedded into the acetron material. Although, the holes were drilled and tapped prior to screw insertion, it is likely that the friction of the screw in the solid rod was higher than what would be encountered in a clinical situation. The excessive friction might have triggered the torque limiter before full tightening of the screw and plate had a chance to occur.
In a clinical setting, with actual bone, 10cNm setting may be sufficient on a torque limiter or potentially, no torque limiter should be used and hand tightening would be better because the risk of cold welding is less than the risk of screw loosening and the subsequent consequences.\textsuperscript{24} The use of solid acetron rods may also have influenced the fitting and subsequent locking of the screw head into the threaded hole of the locking plate. Because of the solid core of the rods, the direction of the screws could not be altered as the screw is inserted and may have prevented some screw head from fully locking into the plate. Following tightening some of the screw heads appeared slightly more prominent than others and could have been the result of incomplete insertion or cross threading. Unlocking of the screw heads was prevalent during the compression testing when a space between the plate and the bone was present but did not appear to be an important failure more in all other testing. Cross threading also occurs in clinical situations and it is therefore uncertain if and how the Acetron influenced the results.
Chapter 4 – Final Summary

4.1 Advantages and disadvantages of the locking plate for radial fractures in dog:

Prior to locking plates, most fractures were repaired with internal fixation using standard plating techniques using conventional plates, which relied on friction of the bone plate to the underlying bone for stability. Absolute stability was the goal along with precise anatomical reconstruction. This approach and type of plating resulted in extensive soft tissue injury and damaged to the underlying blood supply. Over time, it was discovered that the underlying blood supply is vital for fracture healing and the previous approach is being replaced with the intent to provide the best biological conditions for optimum fracture healing.

Locking compression plates are a fixed angle construct that works by screw – plate strength and does not rely on compression of the bone plate onto the bone for stability. Preservation of the blood supply is maintained and this type of plate is used extensively in osteoporotic bone, small bone fragments or weak bone with great success. Multiple studies have confirmed that the plates do not need to be in contact with the bone and that leaving a small space between the bone and the plate was not detrimental to stability.

Radial fractures in small breed dogs are very frequent and continue to have a high complication rate with implant failure and external coaptation complications. Our retrospective study confirmed that although the fractures do heal when properly stabilized, the complication rate approaches 68%. Although many of the complications could be associated with the use of external coaptation a better solution is clearly needed. The use of locking plates has become common practice in veterinary surgery. Locking plates offer multiple advantages over traditional plates and could be beneficial for the treatment of radial fractures in miniature breeds.
In our retrospective study, of the 21 fractures treated with a 1.5mm mini plate 14 achieved union without implant failure, with 4 lost to follow-up, and all the fractures in dogs weighing between 0.9kg and 2.6kg (average 2.2kg) had a successful outcome suggesting that the 1.5mm mini plate is adequate for that weight range.

This experimental study investigated the Synthes 1.5mm locking compression plate and compared it to three cuttable plates commonly used in radial fractures, including the 1.5mm mini plate. It also investigated the effect of bone-plate distance on biomechanical properties of the constructs.

It was concluded that the 1.5mm locking compression plate with biomechanically similar to the 1.5mm mini cuttable plate in compression and torsion and could be used clinically in radial fractures in dogs weighing around 2.2 kg where a single non-stacked 1.5mm mini cuttable plate would otherwise have been used.

Although locking plates do not require intimate contact with the bone, we found that placing the plate away from the bone by as much as half a millimeter resulted in failure of the constructs and cannot be recommended. Whether the contact of the plate and the bone negates some of the benefits of the locking plates on the viability of the periosteum remains to be seen.

The 1.5 LCP presents several advantages over the traditional 1.5mm mini plate. The angle stable constructs allow for improved fixation in weak bone and allow fixation using less screws than for traditional fixation. Locking plates are therefore ideal in those fractures as many of the radial fractures occur in young dogs and involve the distal third of the bone, rendering many of the repairs with traditional plates tenuous. Locking screws could greatly improve the fixation of fractures in which only 2 screws can be inserted into the distal fragment. The current design of the 1.5mm LCP has a larger hole per unit length than the mini cuttable plate and there might be situations in which the bone fragments are too small and preclude the use of the 1.5mm LCP.
Preservation of the blood supply to the bone is a major factor affecting bone healing and is of particular concern in radial fractures in miniature breeds. Although locking plates have shown to better preserve periosteal vascularization and minimize bone necrosis under the plate in vivo studies must be performed to confirm that it is also the case for this plate in radial fractures; particularly in light of the fact that a gap between the plate and the bone cannot be recommended. Because of the improved fixation in small fragments and weak bones, it is possible that external coaptation may not be required following fracture repair. External coaptation was shown to be responsible for significant morbidity during the postoperative period. Unfortunately, our study did not allow us to draw conclusion in this regard and caution must still be applied, as the plate was slightly weaker than the 1.5mm mini plate in our testing conditions. We suspect that the 1.5mm LCP would perform better than the 1.5mm mini plate in a situation involving a comminuted fracture as the LCP has a larger cross section area at the level of the screw hole, however, this hypothesis was not tested and was completely in vitro.

There are disadvantages associated with the use of the locking plates: The distance between the screws is one of them and has already been discussed. Placement of the screws is delicate and slight misalignment may result in cross-threading or incomplete locking of the screw and there is no ability to angle the screws.

The cost of the implant is also a disadvantage of the locking plates as the cost of locking plate and screws tend to be higher than regular plates. A single 6-hole 1.5mm LCP cost approximately $244.00 and the cost of a 20-hole 1.5mm mini cuttable plate cost approximately $158 (approximately $53.00/6-hole). The cost of a single 1.5mm locking screw is $68.00 compared to a single 1.5mm cortical screw at approximately $17.00. The overall cost for a 6-hole construct with
six screws is $652.00 (for 1.5mm LCP) and $155 (6-hole 1.5mm mini cuttable) – this is a 4x increase in cost for the locking construct. These costs are in Canadian dollars.

4.2 – Recommendations & future studies

This research study showed that the 1.5mm locking compression plate was similar biomechanically to the 1.5mm mini cuttable plate in compression and torsional testing. The use of a 1.5mm locking compression plate can be considered as an option for radial fracture repair in dogs weighing less than 2.2kg where a 1.5mm mini cuttable plate would otherwise be used.

A prospective clinical trial evaluating the use of both the 1.5mm locking compression plate and the 1.5mm mini cuttable plate in small breed dogs, less than 2kg, for radial and ulnar fractures is recommended. The use of external coaptation should also be closely documented and all cases followed.
4.3 – References:


Figure 7.1: Load/Displacement curves of eight of the 1.5mm mini cuttable plate constructs tested in bending at a constant rate of 0.1mm/sec until failure
Figure 7.2: Load/Displacement curves of eight of the 1.5mm mini stacked plate constructs tested in bending at a constant rate of 0.1mm/sec until failure
Figure 7.3: Load/Displacement curves of eight 2.0mm mini cuttable constructs tested in bending at a constant rate of 0.1mm/sec until failure
Figure 7.4: Load/Displacement curves of eight of the 1.5mm locking compression plate constructs tested in bending at a constant rate of 0.1mm/sec until failure
Figure 7.5: Load/Displacement curves of eight of the 1.5mm locking compression plate 0.5mm bone-plate distance constructs in bending at a constant rate of 0.1mm/sec until failure
Figure 7.6: Load/Displacement curves of eight of the 1.5mm locking compression plate 1.0mm bone-plate distance constructs in bending at a constant rate of 0.1mm/sec until failure
Figure 7.7: Torque/Deformation curves of eight 1.5mm mini cuttable plate constructs tested in torsion at 2.27mm/sec (constant displacement rate)
Figure 7.8: Torque/Deformation curves of eight 1.5mm mini cuttable stacked plate constructs tested in torsion at 2.27mm/sec (constant displacement rate)
Figure 7.9: Torque/Deformation curves of eight 2.0mm mini cuttable plate constructs tested in torsion at 2.27mm/sec (constant displacement rate)
Figure 7.10: Torque/Deformation curves of eight 1.5mm locking compression plate constructs tested in torsion at 2.27mm/sec (constant displacement rate)
Figure 7.11: Torque/Deformation curves of eight 1.5mm locking compression plate 0.5mm bone-plate distance constructs tested in torsion at 2.27mm/sec (constant displacement rate)
Figure 7.12: Torque/Deformation curves of eight 1.5mm locking compression plate 1.0mm bone-plate distance constructs tested in torsion at 2.27mm/sec (constant displacement rate)