Biomechanical Evaluation of Locking Compression Plate/Rod Constructs and Limited Contact-Dynamic Compression Plate/Rod Constructs in a Gap Model

Wilburn Maret Maxwell

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BIOMECHANICAL EVALUATION OF LOCKING COMPRESSION PLATE/ROD CONSTRUCTS AND LIMITED CONTACT-DYNAMIC COMPRESSION PLATE/ROD CONSTRUCTS IN A GAP MODEL

By

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BIOMECHANICAL EVALUATION OF LOCKING COMPRESSION PLATE/ROD CONSTRUCTS AND LIMITED CONTACT-DYNAMIC COMPRESSION PLATE/ROD CONSTRUCTS IN A GAP MODEL

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Fractures occur commonly in veterinary medicine and are usually the result of vehicular trauma. Traditionally, comminuted mid-diaphyseal fractures are repaired with an interlocking nail, dynamic compression plate, or combination of a bone plate with an intramedullary pin. In recent years, the locking compression plate has gained popularity in human orthopedics due to its biomechanical characteristics and stability in osteoporotic or periprosthetic bone. In addition, the plate may be applied in a percutaneous manner, thereby allowing for biological osteosynthesis. This study evaluates the combination of a locking compression plate and an intramedullary pin in a Delrin rod gap model. Modalities tested include axial load to failure, torsion, and cyclic loading.
DEDICATION

This manuscript is dedicated to my wife, Christy, who has supported me wholeheartedly in all endeavors.
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The author expresses sincere gratitude to Ron McLaughlin for his guidance during the residency and throughout the master’s degree program. In addition, sincere appreciation is extended to Steve Elder for his expertise in all aspects of biomechanical testing and interpretation.
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CHAPTER I
INTRODUCTION

Comminuted femoral fractures occur frequently in veterinary medicine and are surgically repaired using a variety of fixation methods. Historically, these fractures have been repaired with a dynamic compression or a limited contact dynamic compression bone plate and screws. Traditional plating techniques require reconstruction of the fracture fragments and application of a neutralization plate, or application of plate in buttress fashion to bridge the fracture site. Fracture fixation can be achieved through reconstructing the fracture fragment into the original cylindrical shape or by bridging the gap with anatomical reduction. In general, fracture reconstruction is invasive and may compromise blood supply and healing; while plates placed in buttress/bridging fashion are subject to high bending loads which may result in implant failure. Recent studies have found that stabilizing comminuted fractures with a combination of a bone plate and an intramedullary pin (Plate/Rod fixation) can aid in axial alignment of the fracture, aid in maintaining and/or regaining original bone length, and significantly reduce plate strain, thus decreasing the likelihood of plate failure.\textsuperscript{1,2}

Recently, human and veterinary surgeons have worked to reduce intraoperative trauma, preserve blood supply, and maintain soft tissue attachments during fracture repair
by using minimally invasive techniques to apply orthopedic implants. These minimally invasive techniques have been found to be associated with lower infection rates, reduce intraoperative blood loss, and shorten surgical and anesthesia times. With “biological fixation” in mind, locking compression plates (LCP) have become popular in human orthopedics for the repair of a variety of injuries, including comminuted fractures. Not only can LCP be applied using a minimally invasive technique (percutaneous placement), they also have a biomechanical advantage (fixed angle internal fixation) over standard bone plates, require little if any contouring, and may be applied using monocortical screws. These characteristics would be advantageous when using LCP in plate/rod fashion and may permit the application of plate/rod fixation using a minimally invasive approach. Furthermore, they can be used with bicortical and/or monocortical screws. At this time there is considerable debate about the number of cortices required for LCP constructs. Current recommendations in the human and veterinary literature vary between two and five cortices per main fragment.

There are currently few studies evaluating LCP in veterinary medicine. Initial studies demonstrate similar structural stiffness (in four-point bending) between the LCP and limited contact dynamic compression plates (LC-DCP) constructs. However, there is only one report evaluating the use of LCP in plate/rod constructs and it differs vastly from this study. The purpose of this study is to compare bicortical LCP/Rod fixation with LC-DCP/Rod fixation using a gap model. Delrin® tubes were tested in a 20 mm gap model after application of an intramedullary pin (30% of the internal diameter) and either an 8-hole, 3.5 mm LCP or an 8-hole, 3.5 mm LC-DCP. Eccentric axial load to failure, cyclic failure, and torsion to failure were assessed. The result of this study
provides valuable information with regard to use of LCP plate/rod constructs in veterinary patients and provides new information about LCP rod biomechanics.
References


CHAPTER II
FRACTURE STABILITY

The principles used in fracture management were initially set forth by the AO group in the 1950’s. These principles include anatomic reduction, anatomic fixation, rigid stability, preservation of blood supply, and the early return to function. Prior to the 1950’s fractures were treated primarily with plaster casts or by traction, both of which can lead to prolonged healing times.\(^1\) In addition, problems associated with joint immobilization can also occur. The overall goal of the AO principles is to achieve early anatomic bony union. Although this goal is often achieved, complications such as delayed unions, non-unions, infection, and refracture can infrequently occur. In recent years there has been increased effort in the human and veterinary orthopedic community to reduce complications through more biologically friendly techniques.\(^2\)\(^-\)\(^4\) These techniques, such as minimally invasive percutaneous osteosynthesis (MIPO) and biological osteosynthesis, are directed at maintaining tissue vascularity and rely largely on the understanding of gap strain.\(^4\)\(^-\)\(^7\)

Two predominant methods exists for the open reduction and internal fixation of mid diaphyseal fractures. The first technique involves open exposure and anatomic reconstruction of the fragments. This method allow for load sharing of the bone and restores the original anatomic shape. Unfortunately, anatomic reconstruction requires
significant soft tissue disruption at the fracture site and is associated with increased operative time. The second technique utilizes indirect reduction of the fracture ends and focuses on joint alignment and maintaining limb length. The overall goal in this technique is to align the fracture fragments and, through minimal exposure, keep the fracture site undisturbed thereby maintaining local blood supply. Although the same result is generally obtained with both techniques, varying degrees of strain at the fracture site govern the type of bone healing that occurs.

The strain at a fracture site is defined as the change in fracture gap divided by the original fracture gap and is largely determined by the stability.\textsuperscript{1,8} According to the aforementioned definition, strain at the fracture site can be reduced by either increasing the gap at the fracture site, or by decreasing the amount of motion at the site. In fractures with absolute stability (rigid fixation) there is less than 2\% strain. Under these conditions primary, or endosteal, bone healing can occur. Both compression plating and neutralization plating can help maintain gap strains less than 2\%. However, higher strains can be easily reached with minimal motion if the fracture gap is close to zero. In order to reduce the fracture strain plates are placed on the tension side of the bone and plates applied in compression. Fractures repaired with this technique using rigid fixation usually are associated with an early return to mobility. Early mobility is beneficial for multiple reasons including the ability to maintain range of motion of surrounding and/or involved joints, prevent muscle contracture, minimize the amount of disuse atrophy, and aid in the maintenance of overall joint health through articular cartilage nourishment.

Relative stability is a concept that bone can heal through endochondral ossification at gap strains between two and ten percent. Techniques that employ this
methodology include splints, casts, locking plates placed in a minimally invasive fashion, and external fixators. The rationale is that the original fracture hematoma remains intact and thereby facilitates healing by providing the necessary blood supply and healing factors. The goal is to produce an environment where as the fracture heals the strain is reduced further to a level that is conducive to new bone formation.

Strain can be affected by numerous factors that directly challenge the stability of the fracture site. Two broad categories of forces are encountered in the small animal patient: physiological and non-physiological forces. Physiological forces refer to those derived from muscle contraction, weight bearing and other typical movements such as running or jumping. In canine patients, weight bearing on each individual limb causes a ground reaction force equal to 20-30% of the body weight. However, as acceleration increases during running or jumping, the force can be increased five to tenfold. Nevertheless, physiological forces rarely exceed the ultimate strength of bone resulting in a fracture. On the other hand, supraphysiological force refers to abnormally high forces seen during vehicular trauma, gunshot, or other blunt or penetrating types of trauma. These forces can be transmitted directly to the bone and often exceed the ultimate load of failure resulting in fracture and possible devitalization of surrounding tissues.

Physiological forces can be further divided into axial compression, axial tension, bending, and torsion. These forces, alone or in combination, can exert a wide variety of internal stresses and strains to the bone and must be adequately neutralized during fracture fixation. The summation of forces is also influenced by the lever arm (moment arm), the shape of the bone, and the muscular contraction.¹ Most long bones (radius, tibia, humerus, femur) are primarily subjected to a bending moment. This is due in part to
the natural curvature of the bone, but can also be greatly influenced by the eccentric loading (humerus, femur). In vivo studies demonstrate that about 90% of the internal stress applied to a bone is from bending.\(^8\) Since bending is the most prominent force, methods to aid in the implant strength have been investigated. For example, the addition of intramedullary pin to a bone plate can reduce the stress applied to the implant by twofold and increase the fatigue life by tenfold.\(^9,10\)

Fracture forces can be neutralized using a variety of implants. Devices such as the interlocking nail, locking compression plate, external fixator, limited contact dynamic compression plate counteract all applied forces. This is essential because failure to counteract all forces applied to a bone can result in a delayed union or nonunion. In addition, failure to minimize the applied force can result in plastic deformation and permanent structural change in the bone and/or the implant.

Although the degree of strain encountered and the type of implant are of the utmost importance, additional factors must be considered. Factors such as the healing capacity of the patient, age of the patient, location of the fracture, concurrent disease, type of injury (high versus low velocity), infection, compliance of the owner to adhere to post-operative instructions, and size of the patient all play a part. All these items are carefully weighed in small animal orthopedics in order to make an educated decision about the best option for surgical approach, as well as the best method of fixation.
References


CHAPTER III
FRACTURE HEALING

Fracture healing is a complex process that is dictated by the degree of vascularity and, as discussed in the prior chapter, the overall stability at the fracture site. It is divided into three overlapping phases. The three phases consist of an inflammatory phase, repair phase and a remodeling phase.\textsuperscript{1,2} The length and significance of each phase is affected by the amount of interfragmentary motion and the type of injury sustained.

The inflammatory phase of fracture healing begins at the onset of the trauma and extends anywhere from 3-4 days to 4 weeks.\textsuperscript{2-4} When the fracture initially occurs there is disruption of the medullary vessels and the periosteum. A fibrin rich blood clot forms to minimize hemorrhage but osteocyte death and canalicular disruption has already occurred. Lysosomal release by the necrotic osteocytes induces resorption of those cells and incites, and maintains, the inflammatory response.\textsuperscript{3} There is considerable debate about the function of the hematoma at the fracture site. It was formerly believed that this provided a scaffold for cells to utilize and migrate across. In contrast, there is also significant evidence that the primary function is to set the stage for the repair phase of healing by stimulating vessel in growth and bone formation.\textsuperscript{3}

The repair phase is the second phase and begins 2-3 days after injury. Typically it coincides with remodeling of the hematoma by fibrin deposition.\textsuperscript{2} There is also a slight
increase in strength as granulation tissue forms and can withstand a force up to 0.1 Nm/mm$^2$ and resist elongation up to 17%. Osteoprogenitor cells migrate to the site from the endothelium, periosteum (cambrium) and medullary cavity. These cells are responsible for angiogenesis. Angiogenesis refers to the ingrowth of new vessels into a hypoxic area. These new vessels are responsible for providing the extraosseous blood supply to the fracture and provide nutrients to the callus and detached bony fragments. The extraosseous blood supply is a transient response to a bony insult and reaches a maximum at about 10 days. From that time period the degree of extraosseous blood supply diminishes. Although this extraosseous blood supply is transient, it is an essential component of early bone healing. It is also the focus of biological osteosynthesis, which strives to align fragments with minimal fracture site manipulation.

As the callus bridges the fracture gap it is initially comprised of chondroblasts that lay down cartilage. Over time the cartilage is replaced by bone through a process known as endochondral ossification. This process is largely dependent of stability as a strain of less than 2% is required for bone formation, but is also controlled by local oxygen tension. Once the callus begins to mature, there is a shift in blood supply as the medullary activity resumes the primary responsibility. As the oxygen tension decreases the cartilage is replaced by bone. At the end of the repair phase bony union is complete.

Although the fracture has achieved bony union it is still vastly different anatomically from the original bone. During the remodeling phase there is reorganization of the Haversian canal system by osteoclasts as well as the transformation from mineralized cartilage to lamellar bone. A study conducted in human beings using radioisotopes has demonstrated that remodeling can occur for up to 9 years after the
initial trauma and represents 70% of the total fracture healing time.\textsuperscript{3,9} During this period the bone responds to stresses placed on it by resorbing excessive bone in low stress areas and depositing bone in the high stress areas. This phenomenon, referred to as Wolff’s law, is a delicate balance between osteoclastic and osteoblastic activity that is controlled by piezoelectricity. Recent research has further defined the remodeling process at a cellular level by describing the three dimensional characteristics of trabecular bone after stresses are applied.\textsuperscript{10}

Fracture healing has further been categorized in direct and indirect bone healing. In direct healing there is rigid stability, fragment compression, and complete apposition. It bypasses the intermediate steps of bone resorption at the ends and tissue differentiation and progresses directly. Although steps in the bone healing process are bypassed, direct healing does not necessarily progress more rapidly.\textsuperscript{11} The primary repair method utilized for direct bony union is the dynamic compression plate and the limited contact dynamic compression plate. Gap healing is a further subdivision of direct union and refers to gaps less than 800 µm to 1 mm with rigid stability (strain less than 2%).\textsuperscript{2}

Although it was once believed that direct bone healing is the desired result, research has demonstrated that this is an excessively slow process.\textsuperscript{12} Indirect bone healing refers to healing that is greater than a 1 mm gap with relative stability. Indirect bone healing undergoes the sequential steps of tissue differentiation, resorption of the fracture surfaces to reduce strain and healing of the fracture via callus formation.\textsuperscript{11} This type of healing, when combined with minimally invasive approaches, has proven to be associated with soft tissue preservation around the fracture, and is associated with early fracture healing and low infection rates. Techniques that utilize this form of healing
include the locking compression plate, external fixator, interlocking nail, dynamic compression plates used in buttress fashion and flexible fracture fixation.13-16
References


CHAPTER IV
BIOMECHANICS

The goal of fracture treatment is to achieve anatomic bony union in the shortest time possible. In order to do so surgeons rely on a number of fixation systems such as dynamic compression plate and screws, external fixator, interlocking nail and locking compression plates. Each of these systems allows varying degrees of stability at the fracture site and possess unique biomechanical properties. The purpose of this section is to describe the plate design and biomechanical function of the dynamic compression plate (DCP), limited contact dynamic compression plate (LCDCP) and the locking compression plate (LCP).

Dynamic compression plates can be applied in neutralization, compression, and buttress fashion and are designed to create absolute stability with a gap strain of less than 2%. When used correctly, and applied to tension side of the bone, they resist axial compression, torsion and bending loads. The DCP plate can be utilized in a number of fractures, but are especially useful for articular fractures where exuberant callus is not desired. In veterinary medicine the DCP and LCDCP are commonly used for mid-diaphyseal long bone fractures.

Conventional plates loaded in axial compression or tension convert the applied force to shear force at the plate bone interface. The frictional force generated at the plate
bone interface is the major counter force in these situations. The frictional force is located between the bone plate and screws and depends on precise contouring and adequate screw purchase. The torque applied to each screw is the major determinant of frictional force and the screw with the most torque is the greatest contributor to load bearing. In human femurs the torque generated by the application of a 3.5 mm screw is 3-5 Nm.\textsuperscript{2,3} Osteoporotic bone, or bone with a high degree comminution, limits the ability to tighten the screw to 3-5 Nm. Recent modifications to the dynamic compression plate has given way to the limited contact dynamic compression plate (LCDCP). The LCDCP has a scalloped underside that decreases the plate bone contact by 50\%.\textsuperscript{1,4} This reduction in cross sectional bone contact increases the amount or perfusion under the plate and reduces the amount of bone resorption around the plate and screws. Less bone resorption allows for better maintenance of the thread bone interface and helps to maintain screw torque. Most studies involving osteoporotic bone have been conducted in the human model. These studies show that osteoporotic bone allows for about 3 Nm of force to be applied to the screw before the shear resistance of the bone is reached and the screw threads lose purchase in the bone.\textsuperscript{3} At 3 Nm the force that can be countered is greatly reduced and motion has been reported at the plate bone interface at loads as small as 500 N. Motion at the plate bone interface can, as previously discussed, produce gap strains in excess of 10\% and result in nonunions.

In order to increase the frictional force at the plate bone interface, and the amount of axial force that can be resisted, the torque or the contact surface must be increased. Alterations to the surgical technique have been investigated and include the use of cancellous screws and the application of polymethylmethacralate (PMMA) to increase
the contact area between the screw and the bone.\textsuperscript{1} Once there is motion between the plate and bone the strength of fixation is reduced at the most proximal or most distal screw (depending on where the load is initiated). A study by Cordey \textit{et al.} demonstrates that the largest force that can be withstood once motion is encountered between the plate and the bone is 1200 N of compression.\textsuperscript{5}

The quality of bone near the fracture site is not only important for holding the screw, it is also essential for preventing motion during the shear strains generated by axial loading. In the DCP and LCDCP the screw is not locked to the plate thereby allowing angulation of the screws when drilling of up to 40°.\textsuperscript{6} Although this allows for greater flexibility when aiming screws, it can lead to bone failure if the axial load exceeds the strength of the cortical bone in the \textit{cis} cortex. When failure of the implant does occur, the bone fractures leading to high gap strains and nonunion.

Bending loads, in contrast to axial loads, applied to fractures repaired with the DCP and LCDCP are countered by the bending stiffness of the plate. This applies when there is no load sharing between the plate and the bone and exists primarily with gaps greater than 1mm. If the plate is applied on the tension surface of the bone the greatest stress is placed, in theory, on the screws furthest from the fracture gap. The opposite applies to plates placed on the bending surface of the bone.

Locking compression plates (LCP) rely on different biomechanical characteristics than the DCP and LCDCP. For this reason they offer several advantages when compared to conventional plating systems such as: better purchase in osteopenic bone, less bone necrosis under the plate, utilization of secondary bone healing, and less refracture after implant removal due to stress shielding.
LCP’s have threads in the plate that allows for each screw to lock into the plate creating a single beam construct. Single beam constructs function as fixed angle implants since all components are locked together. Simply stated, the LCP functions as an internal external fixator. It functions to convert shear forces to compressive forces and since the strength of fixation is the sum of all screw plate interfaces, it is four times stronger than other plating methods. In contrast, the non-locked plates are only capable of functioning in this manner when they all screws are adequately tightened to greater than 3 Nm and there is zero motion at the fracture site. Given these criteria, these plates rarely function in this manner in the clinical setting.

Although locked plating systems function similar to external fixators, they have a distinct biomechanical advantage in that they can be placed closer to the bone. This allows for screw lengths that are 10-15 times shorter than external fixation. This shorter screw increases the rigidity of the frame and, when the plate is appropriately sized, optimizes the strain at the gap for secondary bone healing.

Another advantage of the LCP is that contouring of the plate is not needed in most situations and the plate bone contact is not required. The benefits of locking plate position are twofold. First, since there is no contact between the plate and bone, there is no bone necrosis from plate pressure. This allows for preservation of the periosteal blood supply and, as discussed in previous chapters, the extraosseous blood supply is important in the early fracture healing period. Although mostly theoretical, placement of the plate away from the bone also could be associated with an increased resistance to infection. The second major advantage is that percutaneous plate placement can be utilized. In the clinical setting the LCP can be placed via a few 2 cm incisions, which leaves the fracture
hematoma intact and is associated with lower patient morbidity. This method of plate placement is currently referred to as minimally invasive percutaneous osteosynthesis and is utilized by systems such as the less invasive stabilization system (LISS).

Regardless of the type of implant there are application guidelines that should be followed to minimize complications such as delayed union, nonunion, screw loosening, fracture next to the implant and implant failure. For the DCP and LCDCP the recommendations are more clear-cut and state that five cortices should be obtained on either side of the fracture, the plate should be precisely contoured to fit the bone, and screws should be tightened to 3-5 Nm. In addition it is recommended that the plate be appropriately sized for the patient and span the length of the bone, when possible, to distribute force evenly.

Unfortunately the guidelines for application for the LCP are not as clear. Initial locking plate systems are designed for use with monocortical screws. These systems allow for easier placement of screws with decreased damage to the endosteal blood supply.\(^2\) The limiting factor with monocortical screws is the thickness and quality of the near cortex. If the cortex on the near side is thin, as in cases of osteopenia or metaphyseal bone, the working length of the screw is reduced thereby weakening the implant. Two monocortical screws is the absolute minimum per fracture fragment according to Gautier et al. to keep an implant stable.\(^9\) However, with this low number or cortices failure of the screw due to fatigue fracture, or loosening at to the screw interface can lead to catastrophic construct failure. Furthermore, they recommend that two bicortical locking screws be used to minimize the risk of an unstable interface and be used only in situations
with good bone quality and proper insertion.\textsuperscript{9} Hertel \textit{et al} recommends, based on clinical observation, three cortices be obtained per main fracture fragment.\textsuperscript{10}

Another variable exists in the application of the LCP that is the focus of a recent biomechanical study. The distance of the plate from the bone has been recently investigated as a variable in biomechanical testing. The results of this study by Ahmad \textit{et al.} shows that decreased axial stiffness and torsional rigidity were observed with locking plates place 5mm from the synthetic bone. The authors in that study concluded that plates placed less than 2mm from the bone showed no biomechanical differences when compared to the LDCP.\textsuperscript{10}

Some locking plate systems allow surgeons to blend the two types of plating systems. These constructs run the risk of creating high gap strains due to inadequate contact of the plate with the bone in the area of non-locked screws. In order to avoid the potential problems all non-locking screws should be placed prior to the locking screws. Furthermore, all compression should be generated prior to locking screw placement. Careful attention should also be paid to the position of the plate on the bone since locking screws can only be added perpendicular to the plate and cannot be adjusted during aiming. A commonly performed elective orthopedic procedure in veterinary medicine is the tibial plateau leveling osteotomy. During this procedure the tibial plateau is transected with an oscillating bone saw and repositioned with a bone plate and screws. One of the newer plate designs incorporates locked and non-locking screws into the same plate. A study by Leitner \textit{et al.} demonstrated that the locking screws demonstrated comparable biomechanical stability during cycling and axial load to failure when compared to the
conventional plating group. In addition, the locking constructs had less translation at the osteotomy site than the conventional group.\textsuperscript{11}

Overall, the biomechanics and application principles of the locking and non-locking plating systems have vast differences. Each case should be approached and, depending on a multitude of variables such as bone quality, fracture location and activity level, a plan should be formulated based on the type of desired healing. In general, traditional plating systems should be used in cases where articular or periarticular fracture exists. Locking plate systems are reserved for cases with mid-diaphyseal fractures where anatomic reduction is not necessary for function and in cases with osteopenic or pathologic bone.
References


CHAPTER V

CLINICAL USE OF LOCKING COMPRESSION PLATES

Plate/rod fixation techniques have been used in veterinary medicine for the repair of comminuted, diaphyseal femoral fractures for several years. The construct provides rigid fixation with the addition of an IM pin to aid in alignment and limb lengthening. However, the standard plate/rod technique requires extensive surgical exposure for plate application and thereby disrupts the fracture hematoma and extraosseus blood supply in the immediate post-trauma period. Disruption of the fracture site can be associated with increased healing times as well as a higher incidence of infection.

In contrast, the locking compression plate may be placed percutaneously through small incisions located at the proximal and distal aspect of the fracture, thus reducing disruption of the hematoma. In addition, application of LCP allows insertion of monocortical screws alone, and may require engagement of fewer cortices per fracture segment. Since the IM pin and LCP can be applied using a minimally invasive approach, the use of LCP in plate/rod fashion may further reduce disruption of the fracture site during repair of comminuted diaphyseal fractures. Furthermore the IM pin can aid in the alignment of the limb and help to maintain length during plating. The purpose of this study is to determine the biomechanical characteristics of the LCP/rod construct and determine the feasibility for use in veterinary comminuted femoral fractures. To the
authors knowledge no studies regarding the use of the LCP/rod with this screw configuration in veterinary medicine have been conducted.
CHAPTER VI
METHODOLOGY

Experimental Design

A plate-rod construct was mounted to Delrin rods and biomechanical testing performed. Five constructs from each group, locking compression plate and limited contact dynamic compression plate, underwent axial load to failure, cyclic load to failure, and torsion to failure. Construct stiffness, yield force and yield torque were calculated from the data obtained.

Construct Preparation

Fifteen Delrin® tubes (McMaster-Carr, Atlanta, GA item #8627K59) 3/4” diameter, 12 cm in length were used in each of the two groups. Each rod has a wall thickness of 1/8” with an inside diameter of 0.5”. Each rod was mounted on a drill press and a 9/16” bit used to center the hollow tube for a depth of 0.125”. Next, 9/16” Delrin rod (McMaster-Carr, Atlanta, GA, item #8497K231) was transected using a band saw to create discs that were 0.125” long and 9/16” in diameter. The discs were sanded briefly to remove any roughened edges. The hollow Delrin tube was mounted in a vice and holes drilled at approximately 120° intervals around the circumference (for a total of 3) of the Delrin and threads tapped to match the stainless steel cone point set screws (McMaster-
Carr, Atlanta, GA, item number 90778A178). The Delrin discs were then inserted into the 9/16” predrilled recession in the Delrin tube and the three cone set point screws tightened to ensure adequate contact.

An oscillating bone saw was used to transect central aspect of each Delrin rod. The rod was then placed on a custom jig to allow for consistent fracture gap distance of 20 mm between the Delrin rod segments. (fig.6-1) A 5/32” x 9” double trocar point Steinman pin (IMEX, Longview, TX) was inserted normograde into each tube after the gap has been created and advanced until it contacts the distal aspect of the Delrin insert. Excess pin was transected flush with the Delrin at the insertion point using a pin cutter. The pin diameter was 30% of the internal diameter of the Delrin tube.

![Figure 6.1](image)

Figure 6.1  Locking compression plate with intramedullary pin applied to the Delrin rod.

In 15 specimens (LCP/Rod group), an 8-hole, 3.5 mm LCP (Synthes, Paoli, PA) was applied to the Delrin® tube. Current recommendations for application of the LCP were followed and include the use of 2.8 mm drill bit and corresponding locking drill
guide to ensure axial alignment. The plate was allowed to contact the surface of the Delrin but no compression was generated. Self-tapping 3.5 mm locking screws 26mm in length (Synthes, Paoli, PA) were placed with two bicortical screws in the most proximal and distal plate holes. A torque-limiting adapter was utilized to ensure that the screws were tightened to the recommended 1.5 Nm. The remaining plate holes over the fracture site remained unfilled.

In the other 15 specimens (LC-DCP/Rod group), an 8-hole, 3.5 mm LC-DCP (Synthes, Paoli, PA) was applied to the Delrin tube using current AO/ASIF guidelines. Three bicortical screws will be inserted both the proximal and distal aspects of the plate. Screws were hand tightened to simulate the clinical setting by one veterinary surgeon. Two empty plate holes were maintained over the fracture gap in each specimen.

**Mechanical Testing**

Plated Delrin® tubes were mounted in a servohydraulic materials testing machine (Bionix 858 Test System, MTS Systems, Eden Prairie, MN, USA) for mechanical testing. The distal aspect of the tube was placed in a custom jig made to match the rod diameter for placement in the unit. A single point of fixation was used at the distal aspect in the form of bolt for attachment to the MTS. This allowed for translation during weight
Figure 6.2  Assembled LC-DCP mounted within the materials testing machine.
bearing along the loading axis during displacement. The proximal aspect of the jig was eccentrically loaded (offset 3cm) to simulate weight bearing. (see fig 6-2, 6-3, 6-4) This portion of the jig was held to the Delrin tube by four cone point set screws (McMaster-Carr, Atlanta, GA, item number 90778A178). The portion of the proximal jig that contacted the MTS machine (point of loading) was fitted with a ball and socket mechanism to allow loading with slight angulation during construct displacement to simulate the clinical weight bearing of the canine femur.

Figure 6.3  Close-up of the ball and socket apparatus in the materials testing machine.

Each construct was axially compressed by 20 mm at a constant rate of 0.2 mm/s. A scan rate of 98 Hz was used for data acquisition. Each group, LCP/rod and LC-DCP/rod, was statically loaded to failure to compare structural stiffness, offset yield force and maximum yield force. Failure limit was defined as the ultimate load before breakage
Figure 6.4  Close-up of the distal jig apparatus in the materials testing machine.
(acute decrease in load with continued displacement). Stiffness was defined as the slope of the best fit line through the linear portion of the force vs. displacement curve. A displacement of 1.5 mm was utilized to determine the yield force via the offset method.

Cyclic loading was performed in a similar manner. Constructs were cyclically loaded in ramp fashion under load control to lower and upper limits of 10 N and 400 N. Each loading/unloading ramp took one second to complete for a cyclic loading rate of 0.5 Hz. Mean cycles to failure was determined for each construct type. Failure limit was defined as the number of cycles before fatigue failure (acute decrease in load with continued displacement or plate breakage).

Torsional testing was performed by mounting the rod in a custom cable and pulley fixture attachment to a universal testing machine (Instron Model 1011, Instron Corp. Canton, MA, USA). The fixture converts the vertical linear motion of the testing machine’s crosshead to a rotational motion. Applied rotation was calculated as vertical displacement x (360°/pulley circumference). Measured tension in the cable was converted to torque by multiplying the pulley’s radius. Rods were twisted to failure at a rate of one degree per second. Torque versus rotation curves were generated to define failure limit and yield limit between the groups. Failure limit was defined as the ultimate torque before breakage (acute decrease in load with continued rotation). Stiffness was defined as the slope of the best fit line through the linear portion of the torque vs. rotation curve.
References

CHAPTER VII
RESULTS

After analysis there were no statistically significant differences in construct stiffness between the LC-DCP rod and the LCP rod in axial load to failure (p=0.95). Mean stiffness of the LCDCP-rod was 92.24 N/mm (Std Dev 20.99N/mm). Mean Stiffness of the LCP-rod was 92.88 N/mm (Std Dev 13.72 N/mm). Furthermore, there was no statistically significant difference in 1.5 mm offset yield force (N) between the LCDCP-rod and LCP rod (p=0.51). Mean 1.5 mm offset yield force of the LCDCP rod was 490.0 N (Std Dev 30.82 N) and 503.0 N (Std Dev 29.7 N) for the LCP rod. (Fig. 7-1, 7-2)

Figure 7.1 Graphical depiction of axial results comparing the LC-DCP rod and the LCP rod.
The maximum force (N) to implant failure during axial loading was also compared between the two constructs. Mean maximum load to failure for the LCDCP-rod group was 570 N (Std Dev 28.8 N). Mean maximum load to failure for the LCP-rod group was 598.8 N (Std Dev 37.54 N). No statistically significant differences were noted between the two groups (p=0.17).

The torsional stiffness of the two constructs was also compared. Mean torsional stiffness of the LCDCP rod construct was 0.445 N-m/Deg (Std Dev 0.123 N-m/Deg). Mean torsional stiffness for the LCP Rod was 0.382 N-m/Deg (Std Dev 0.38 N-m/Deg). Significant differences were noted between the two constructs (p=0.0006) with the LCDCP having a higher mean torsional stiffness. (Fig. 7-3)
Cycles to failure was recorded after a 10-400 N ramp at 0.5 Hz with the LCDCP-Rod failing at a mean of 62,809.4 (Std Dev 29955.3 cycles). In contrast, the LCP-Rod constructs failed at a mean cycles of 95,671.8 (Std Dev 27580.1 cycles). No significant differences were noted within the two groups with regard to cycles to failure (p=0.10). All constructs failed via fatigue fracture at a screw hole over the simulated fracture gap.

Figure 7.3 Summation of data regarding the torsional stiffness between the two plate-rod constructs.
CHAPTER VIII
DISCUSSION

The recent use of minimally invasive percutaneous osteosynthesis techniques in surgery allows for biologically friendly repair utilizing indirect bone healing.\textsuperscript{1-5} The Synthes 3.5 mm locking plate system provides a method to apply, in a minimally invasive manner, a plate to the lateral cortical surface of the femur for the treatment of comminuted mid-shaft fractures. With the inclusion of an intramedullary pin, normal limb length and axial alignment can be more readily achieved. In addition, the bending strength can be greatly improved.\textsuperscript{6,7} The goal of this research project was to compare the biomechanical properties of the LCDCP Rod with the LCP rod under similar testing criteria. No significant differences in eccentric axial load to failure were found. Torsional stiffness of the two constructs was significantly different. A difference in mean cycles to failure was noted between the two construct; however this difference was not statistically significant and has limited clinical applicability due to the variability between specimens.

This study utilized a Delrin rod with a critical gap where no load sharing could occur between the two simulated fracture ends. Delrin rods were used because of their uniform size, stiffness and the relative unavailability of cadaveric femurs. The uniformity of the Delrin rod decreases the risk of settling that can be seen in research studies and allows more direct comparison between implants. According to previous
recommendations the intramedullary Steinman pin was measured to ensure it occupied 30% of the internal diameter of the Delrin. In the clinical setting intramedullary pins are usually placed in the normograde or retrograde fashion and penetrate one end of the metaphyseal bone. In order to mimic this aspect, Delrin discs were precisely fitted within the Delrin and secured with three, cone-point set screws set at 60° intervals. Furthermore, to ensure the disc did not subside during testing the Delrin rods were predrilled with a press to a depth of 0.125” to provide a ledge for the disc. The intramedullary pin was then drilled normograde down through the disc into the center of the medullary canal. The distal aspect of the pin was allowed in all cases to engage the Delrin disc but did not penetrate it entirely.

In order to simulate a worst case scenario the minimum number of cortices was utilized for the both the LCDCP and LCP constructs. According to the rules set forth by AO Principles of Fracture Management a minimum of 5 cortices must be obtained per fracture fragment. To meet this minimum and simulate a clinical setting three bicortical screws were used per fragment for the LCDCP. The LCP plate application guidelines are, unfortunately, less straightforward as recommendation vary between two and four screws per segment with varying number being bicortical in nature. In this study, two screws per fragment were used in the most proximal and distal screw holes which coincides with minimal requirement. The torque limiting adapter was used for the LCP constructs to ensure that all screws were tightened to the recommended 1.5 nM. The LCDCP were all tightened by hand by one veterinary surgeon. This is a potential limiting factor since variation could exist in torque and the LCDCP does rely on friction between the implant and the surface for stability.
No significant differences were noted in axial load to failure between the LCDCP and the LCP construct. Failure of the construct was associated with plastic deformation of the plate and no screw pull-out or screw failure (Delrin fracture or screw unlocking) was noted. It is interesting to note that the maximum force to implant failure was much less than reported by Goh et al as those implants failed at approximately 1500 N after 60,000 cycles. In that study LCDCP rod constructs were compared to LCP rod constructs with 5 monocortical screws used in the LCP group. Several variations do exist that may account for the difference. First, screws were placed in a different arrangement with 5 screws per fracture fragment. Second, an intramedullary pin of 40% of the medullary diameter was utilized which will increase the bending stiffness. It has been reported that the peak vertical force of 76%-107% body weight of a 30 kg Labrador at a trot corresponds to a force of 223 N to 315 N respectively. Although it is difficult to compare an in vitro Delrin model to clinical application, it is fair to suggest that the failure limit of the implants in our study is higher than the expected force encountered in the post-operative period.

Torsional stiffness between the constructs was different with the LCDCP being significantly stiffer. Data regarding torsion in locking plate construct is scarce in the veterinary literature. Filipowicz et al reported in a humeral LCP-rod model that the LCP model showed significantly less resistance to torsion when cycled than the LCDCP but it was not evident until after 280 cycles. A study by Gordon et al found that the LCP construct was significantly stronger than the LCDCP in torsion. Another veterinary study by Sod et al also showed that the torsional stiffness of a 4.5 LCP had a higher torsional stiffness than the 4.5 LCDCP in an equine metacarpal. Studies in the human
literature are more readily available but the findings vary as some show increased strength of the LCP in compression, torsion, and bending while others show no significant difference.\textsuperscript{16-27} It is important to note that a variety of types/sizes of LCP plates were used in those studies and that testing materials and testing methods varied greatly. It is likely that the screw position in the two most proximal and distal plate holes within the LCP/Rod constructs decreased the performance in torsion as the screws were further from the simulated fracture gap than the LCDCP/Rod model.

Cycles to failure was performed using a 10 N to 400N ramp. This corresponds to the to greater than 107\% of peak vertical force in a 30 kg dog. Furthermore it has been estimated that a canine during 5-10 minute walks four times per day will have approximately 1500-3000 cycles per day. Extrapolation of that data, and using the lower estimate since patients activity would be restricted during fracture healing, indicates that within a 6-week period there would be approximately 63,000 cycles. Since our failure data during cycling was similar to and greater than this number, and the force applied is designed to emulate an undesirable scenario (dog at a trot), it is unlikely that either construct would fail due to cyclic motion in the immediate recovery period. It is important to note that there was a wide standard deviation and this data should be confirmed with additional testing.

A potential advantage of the LCP when compared to the LCDCP is the decreased plate contact with the bone surface. Since the LCP does not rely on the friction generated between the implant and bone, perfusion under the plate should be improved. In this study the plate was allowed to contact the Delrin rod in both groups. Previous locking plate-rod studies have semicontoured the LCP to a distance of 3mm away from the bone.
In the clinical setting, and when utilizing minimally invasive percutaneous osteosynthesis, it is almost certain that there will be varying degrees of LCP contact with the bone. Clinical case studies are needed to determine the outcome as it is difficult to account for variations in plate distance from the bone during in vitro testing.

The limitations of this study include in the in vitro nature and the relatively low sample size. Delrin rods provided a uniform material for testing but cannot simulate the variation seen in actual bone. Furthermore, there was no way to account to muscular attachments around the bone that apply force in varying directions. Also, one veterinary surgeon tightened all LCDCP constructs but there still could be small variance in screw tightness. Since this directly impacts the stability of the LCDCP constructs it could cause slight variations in the data.

Overall we found no significant differences in construct stiffness between the LCDCP/Rod and LCP/Rod in eccentric axial load to failure. Although torsional stiffness was different the primary load applied to the femur it is our belief that the LCP /Rod with 2 screws per segment would perform similarly to the LCDCP/Rod in the clinical setting. Further studies are needed to evaluate the most advantageous screw position as well as the optimum intramedullary pin size and number of cortices per segment.
References


CHAPTER IX
FUTURE DIRECTION

The increased use of locking plates in veterinary surgery is sure to have an impact of small animal fracture repair in the future. This study, coupled with the work of Goh et al., highlights the idea of percutaneous plate placement and provides biomechanical data comparable to other plate/rod constructs. Although preliminary work has been completed there are many other avenues to be investigated.

Minimally invasive percutaneous osteosynthesis has become the preferred method of fracture treatment, when possible, in human orthopedics. However, additional studies are still needed to determine the optimal number of screws for the locking system as well the location of the screws. Screw configuration with combinations of bicortical and monocortical screw configurations with an intramedullary pin should be investigated in the future in the cadaver model and eventually in the canine patient. Also, variation in the diameter of the intramedullary pin (percent medullary canal fill) can also be altered to aid in the placement of bicortical screws.

In human orthopedics it is currently recommended that 40-50% of the plate holes be filled and the plate should span the entire length of the bone. Other variations with open holes between screws may be beneficial in veterinary medicine and allow for smaller incisions during the fracture approach. The longer plate, with distribution of
strain along the length of the entire bone, should be investigated in the clinical setting to assess healing. Other biomechanical testing is necessary, such as four-point bending, to ensure that adequate fixation is achieved regardless of the screw position.

Overall, the management of fracture has changes dramatically over the last 20 years. Fractures are now repaired through small incision with particular attention paid to the amount of strain and type of bone healing desired. The locking compression plate with an intramedullary pin provides an exciting alternative to traditional fracture repair options. Future research is needed to further define ideal screw and plate configurations.