MECHANICAL EVALUATION OF PLATE WORKING LENGTH IN A CANINE FEMORAL MODEL

By

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A THESIS PRESENTED TO THE GRADUATE SCHOOL OF THE UNIVERSITY OF FLORIDA IN PARTIAL FULFILLMENT OF THE REQUIREMENTS FOR THE DEGREE OF MASTER OF SCIENCE

UNIVERSITY OF FLORIDA

2012
To my family.
ACKNOWLEDGMENTS

First, I would like to thank all the people that have contributed to the work described in this thesis.

Dr. Antonio Pozzi as my advisor and mentor has been an inalienable part of this project. His perpetual energy and enthusiasm in clinics and research had motivated me through this research journey. He gave me the confidence and all the support, enabled me to get through every difficulty in front of me.

I would like to acknowledge my thesis committee members. The valuable advice from Dr. Daniel Lewis, always pointed out the key points. Without his support I couldn't have achieved a milestone. I gratefully thank Dr. Bryan Conard for providing me major support, both in knowledge and in techniques, of engineering science. Dr. MaryBeth Horodyski's help in statistics was really precious in this thesis.

Additionally, I appreciate Dan Barousse and Debby Sundstrom for their important experience sharing and all the assistance in every little detail in this study. Last but not the least, this thesis would be incomplete without mentioning my parents: Shereen Lu and Pan-Hwa Chao. Their unconditional support could only be cherished by providing them a good research work. My big brother, Josh Chao has been another powerful support during this whole time. Being my second family here, James Wang and Ting-Bing Wu has not only guided me through many difficulties, but also replaced them with fun and joy.
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<td>ANOVA</td>
<td>Analysis of Variance</td>
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<td>AO-ASIF</td>
<td>Arbeitsgemeinschaft für Osteosynthesefragen – Association for the Study of Internal Fixation</td>
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<td>DCP</td>
<td>Dynamic Compression Plate</td>
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<td>LCP</td>
<td>Locking Compression Plate</td>
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<tr>
<td>MTS</td>
<td>Material Testing Machine</td>
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<td>SD</td>
<td>Standard Deviation</td>
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MECHANICAL EVALUATION OF PLATE WORKING LENGTH IN A CANINE FEMORAL MODEL

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August 2012

Chair: Antonio Pozzi
Major: Veterinary Medical Sciences

To compare the bending stiffness, gap motion, cyclic fatigue life and load to failure of plated dog femora stabilized using either a short or long plate working length technique. Contralateral femora were randomly assigned to one of two stabilization techniques: femora plated with 12-hole 2.4 mm LCP using 2 screws per fracture segment resulting in a long working length or femora plated with the same LCP using 5 screws per fracture segment resulting in a short working length. Constructs stiffness and yield load in load to failure test were compared between techniques using a paired-t test. Bending stiffness and gap motion during cyclic testing were compared between techniques using a 2 x 9 or 2 x 11 (technique x cycle) within subjects repeated measures ANOVA; a p<0.05 was considered significant.

Construct stiffness did not differ significantly between techniques. Implant failure did not occur in any of the plated femora during cycling. Mean ± SD of gap motion of long (0.2 ± 1 mm) and short plate working length constructs (0.1± 0.6 mm) were not significantly different during cyclic test. Mean ± SD yield load in short plate working length (654.5 ± 149.0 N) technique were significantly higher than in long plate working length (452.9 ± 63.4 N) technique. Plate working length had no effect on stiffness, gap
motion and resistance to fatigue in our testing model. Although short plate working length tends to increase the resistance to failure; the yield loads for both stabilization techniques were within physiologic range.
CHAPTER 1
INTRODUCTION

Over the past two decades, there has been a paradigm shift regarding the approach to internal fixation of long bone fractures with bone plates. The prerequisite for open anatomic reduction and rigid stabilization has given way to less invasive application of more flexible constructs with bridging plates \(^1-3\). The use of locking technology allows the plate to function as an internal fixator \(^4-9\). Surgeons can choose among a variety of implants to allow a more or less flexible bone-plate construct \(^8,10-29\). Furthermore, the design and type of plate can play a primary role in fracture reduction \(^12,26,29-32\). Whereas plates with a compression or neutralization function require precise reconstruction and rigid stabilization, plates applied in a bridging fashion circumvent the need for anatomical reduction \(^33\). Bridging plates are often applied utilizing indirect reduction techniques, which mitigate the degree of iatrogenic trauma while preserving fracture vascularity \(^1,2,4,32,34-37\). Understanding the basic biomechanical principles of surgical stabilization of fractures is essential for developing an appropriate preoperative plan as well as making prudent intra-operative decisions.

The objective of this chapter is to provide basic biomechanical knowledge essential to understand the complex interaction between the mechanics and biology of fracture healing. It is clearly understood that careful soft-tissue handling is very important in preserving blood supply to the injured bone. However, the type of healing and the outcome can be influenced by several mechanical factors, which depend on the interaction between bone and implant \(^31,38-41\). The main objective for utilizing less invasive fracture stabilization techniques is to optimize the healing potential by achieving a symbiotic balance between the biologic and the mechanical factors of fracture fixation.
Thus the surgeon should understand the mechanical principles of fracture fixation and be able to choose the best type of fixation for each specific fracture.
CHAPTER 2
BACKGROUND

Basic Mechanics of Materials

**Force, Deformation, Stress and Strain**

The strength of a material depends on its ability to resist failure from an applied stress. Stress is the force acting on an area, and can be compressive, tensile or shear stress. The unit for stress is Newtons per square millimeter (N/mm$^2$). When stress is applied to an object, deformation may occur. Thus, the term deformation is used to describe the change in shape or size of an object due to an applied force. Depending on the size, shape, material of the object, and the force applied, various type of deformation may occur. Elastic deformation is reversible: an object may deform when subjected to an applied load, but the object returns to its original shape once the load is released. Plastic deformation in contrast is irreversible and the object does not return to its original shape once the applied load is released. Another type of deformation, unique to ductile metals, is metal fatigue. This phenomenon describes the progressive formation of cracks which develop in a material subjected to numerous cycles of elastic deformation. The different phases of deformation of an object are readily illustrated by plotting a load-deformation curve (Figure 2-1), which shows the relationship between stress (force applied) and strain (deformation) of an object. The elastic range of the curve ends when the object reaches its yield strength and begins to undergo permanent plastic deformation. If continued load is applied, material failure may occur in the form of a fracture, or the object may just continue to undergo further plastic deformation. When applying these concepts to fracture fixation, a bone plate should function within the elastic region, and should not be subjected to loads which exceed the plate’s yield.
strength. Therefore, yield strength is a very useful parameter for comparing the mechanical properties of different plates; however, this information is most meaningful when the applied load in vivo is known.

The term strain is used in order to give a more standardized and quantified description of material deformation resulting from an applied stress. Strain is defined as the ratio between the measured changes in length during loading to the original length. Strain refers to a change in shape of a specified segment which undergoes either elongation or shortening. Strain is a unitless ratio (length over length), but is commonly reported in units of microstrain, so that a strain of 0.01 (1%) would be 10,000 microstrain. Interfragmentary strain is a term used to describe the mechanical environment within a fracture gap subjected to axial loading. Interfragmentary strain is defined as the relative change in the fracture gap divided by the original width of the fracture gap.

Stiffness

A material’s or a structure’s stiffness defines its ability to resist deformation resulting from an applied force. For so-called linear elastic materials (such as bone), the elastic region of the load-displacement curve is linear, because the deformation is directly proportional to the applied load (Figure 2-2). The slope of the linear portion of the load-displacement curve is known as Young’s modulus, or the modulus of elasticity and is a measure of the stiffness of the material. Elasticity is a characteristic of a material or object to return to its previous shape after an applied load is released. Plasticity in contrast to elasticity describes residual deformation of a material or object as a result of loading and is an unrecoverable status. More elastic materials can usually sustain considerable plastic deformation, while more brittle materials will fracture at relatively low
loads. As an example, the load-deformation curves of the same object with 3 different material such as bone, glass and metal would look significantly different from each other (Figure 2-2) The more brittle rigid body such as bone undergo minimal plastic deformation before reaching the failure point, while metal has a linear elastic portion of the curve, indicating linearly elastic behavior. The long plastic region of the metal indicates that this material deforms extensively before failure.

Another concept that helps elucidate the biomechanical characteristics of orthopedic implants such as plates, screws, intramedullary pins and interlocking nails is the area moment of inertia \(^46\). The bending stiffness of an object (such as an orthopedic implant) depends on the bending moment (the force applied to the plate), which is the product of the elastic modulus of the object material and the area moment of inertia of the cross-section of the object (Figure 2-3). The area moment of inertia describes the capacity of the cross-sectional profile of an object to resist bending. The greater the area moment of inertia, the less a structure will deflect (higher bending stiffness) when subjected to a bending load. Area moment of inertia is dependent on an object’s cross-sectional geometry (Figure 2-3). The further the object’s mass is located from the neutral axis, the larger the moment of inertia. For this reason area moment of inertia is always considered with respect to a reference axis, in the X or Y direction, which is usually located at the center of an object’s cross section. The area moment of inertia of an object having a rectangular cross-sectional profile, such as a plate, can be derived by the equation \(bh^3/12\) where \(b\) is the base and \(h\) is the height. The base dimension is oriented parallel to the axis of the moment of inertia and height is defined as the dimension parallel to the direction of the applied load. Thus the location a plate is
positioned on a bone and the plate’s orientation to applied bending loads can have a profound effect on a construct’s bending stiffness (Figure 2-3). This effect becomes even more important when fractures are not anatomically reconstructed and plates are applied in bridging fashion. Understanding the area of moment of inertia is important when comparing the mechanical properties of implants of different shape and size but similar metal, such as in case of an interlocking nail, a bone plate and a plate-rod construct (Figure 2-4). As shown in the calculation, the interlocking nail has the largest area of moment of inertia because of its large diameter. It should also be noted that the area of moment of inertia of an intramedullary pin occupying 40% of the medullary canal has a significant contribution to the total area of moment of inertia of a plate-rod construct (Figure 2-4), justifying its recommendation as bridging implants for comminuted fractures. Another approach to fixation of a fracture with a gap is to increase the size of the plate. As shown in the calculation (Figure 2-4), a 3.5 mm broad LCP has an area of moment of inertia 3 times larger than a 3.5 mm LCP and almost twice larger than a 3.5 mm LCP combined with an intramedullary rod.

Although the stiffness of a plate is an important predictor of the implant’s behavior under applied load, the mechanical properties of the combined plate-bone construct are more relevant to predict the type of fracture healing. For this reason, it is important to distinguish between implant stiffness, structural stiffness of the construct and stiffness across the fracture gap. The construct stiffness is determined by numerous variables, including the plate’s composition and geometry, the distance between plate and bone surface, plate length, type of screws and the plate working length. The plate working length is defined as the distance between the proximal
and distal screws closest to the fracture. The gap stiffness is derived from the load-displacement curve describing the mechanical behavior of the tissue forming at the fracture gap. Interfragmentary strain is defined as the relative displacement of the fracture gap ends divided by the initial fracture gap width. For this reason the initial fracture gap is an important factor determining the stiffness of the construct. The relationship between gap strain and fracture healing has been extensively studied and will be discussed in the next section.

**Fatigue Failure**

Mechanical failure of plates can be broadly divided into three categories: plastic, brittle and fatigue failure. Plastic failure describes failure of an implant to maintain its original shape, resulting in a loss of reduction and alignment and clinical failure. Brittle failure, an unusual course of implant failure, results from a defect in design or metallurgy. Fatigue failure occurs as a result of repetitive loading at an intensity considerably below the normal yield strength of the implant. Cyclic loading can lead to the formation of microscopic cracks that can propagate until these cracks reach a critical size which cause sudden failure of the implant. Although the propagation of the microcracks can take a considerable amount of time, there is typically very little, if any, warning preceding ultimate failure. Crack formation is commonly initiated at a “stress concentrator” or a “stress riser” such as a scratch on the plate or at a location where there is a change in the plate’s cross sectional geometry, such as a screw hole. The stress that is focused in these areas can be relatively higher than the average stress of the whole construct. Therefore, local material failure can occur at one of these stress concentrators and eventually propagate through the implant. The number of cycles required to cause fatigue failure decreases as the magnitude of the stress increases. Fatigue failure is a
genuine concern following fracture stabilization because of the high number of repetitive loads that implants are subject to during the post-operative convalescent period. Therefore, every surgeon must be cognizant that they have entered the proverbial race between fatigue failure of the implant and healing of the fracture, when repairing any fracture.

Cyclic testing is useful to detect the performance of an implant in resisting fatigue failure. In general, a predetermined stress or strain limit will gradually be reached while applying a cyclic bending moment or tensile/compressive force on a construct. The principle of this type of testing is to determine the total number of loading cycles that a particular construct can withstand before failing. A construct’s fatigue behavior can be described in an S/N curve; in which the stress to failure, S, is plotted against the number of cycles to failure, N (Figure 2-5) ⁴². A construct’s failure point is termed “allowable stress”. Typically, a construct subjected to a small applied-stress can withstand a large numbers of cycles and vice versa. However, the number of cycles to failure at a constant stress level can be affected by many factors such as the material and the geometry of the construct or the surrounding environment of the implant and the bone.

**Applied Biomechanics**

**Biomechanics of Fracture Healing**

Numerous studies have shown that the mechanical conditions affecting the fracture site, principally the stability afforded by the fixation and the width of the fracture gap, influence callus formation during the healing process ²⁸,⁴⁴,⁶⁰⁻⁷⁰. The process of bone healing is dependent on numerous interactions between biologic and mechanical factors. The type of injury, the location and configuration of the fracture, the magnitude of load acting on the fracture as well as systemic factors, all play a role in the type and efficiency
of bone healing\textsuperscript{41,60,71}. Two principle concerns are whether there is adequate blood supply as well as effective stability to allow a fracture to obtain union. If the local circulation is adequate to support fracture healing, the pattern of bone healing is then dependent on the surrounding biomechanical environment\textsuperscript{41,44,60,65,68-71}. Several mechano-regulation theories of skeletal tissue differentiation have been developed that predict many aspects of bone healing under various mechanical conditions\textsuperscript{59,66-70,72}. The theory proposed by Perren is based on the interfragmentary strain present in the fracture gap\textsuperscript{44,59}. This theory suggests that the type of tissue formed in a healing fracture gap is dependent on the strain environment within the gap. The tissues that are stressed beyond their ultimate strain could not form in the gap. If interfragmentary strain exceeds 100%, non-union will occur. Gap strains between 10 and 100%, allow for fibrous tissue formation. Strains between 2 and 10% allow for cartilage formation and subsequent endochondral ossification. Strains of less than 2% allow for direct bone formation and primary fracture healing. Perren proposed that once any tissue formed, it would progressively stiffen the fracture gap. In turn, the tissue formed in the gap would lead to lower strains, which would allow formation of the next stiffer tissue and the cycle would repeat until bone formed within the gap. An alternate theory relating mechanical stimulus to fracture healing was proposed by Carter and Blenman which predicts that tissue differentiation within the fracture gap is dependent on the magnitude and the type of local stress, including hydrostatic pressure and octahedral shear stress\textsuperscript{68,70,73}. This theory purports that the vascular supply to the tissues at the fracture site is the primary factor determining tissue differentiation. With a good circulation in tissues, Carter and Blenman proposed that fibrocartilage will form if high hydrostatic compressive stresses are
present. In an analysis of fracture healing, Carter and Blenman correlated compressive hydrostatic stress with cartilage formation (chondrogenesis) whereas low hydrostatic stress corresponded to bone formation (osteogenesis) \(^{68-70}\). However, the relationship between the ossification pattern and the loading history was described only qualitatively and not quantitatively. More recently, Claes and Heigele have proposed and tested the quantitative tissue differentiation theory, which relates the local tissue formation to the local stress and strain in a fracture gap \(^{66}\). The results regarding the global strain and hydrostatic pressure fields correlate with the principal results of Carter and Blenman. In contrast to Carter and Blenman's work \(^{68,70,73}\) the quantitative tissue differentiation theory is based on the assumption that new bone formation only occurs on existing bony surfaces and under defined ranges of strain and hydrostatic pressure. The tissue differentiation hypothesis predicts intramembranous bone formation will proceed once interfragmentary strain become less than 5% while endochondral ossification can occur at interfragmentary strains approximating 15% in a diaphyseal fracture, which obtain union by secondary bone healing \(^{45,66}\). Another recent theory on mechanobiology of fracture healing proposed a model dependent on two biophysical stimuli: tissue shear strain and interstitial fluid flow \(^{67}\). The rationale for this approach is that fluid flow increases the biomechanical stress and deformation on the cells above that generated by the strain of the collagenous material \(^{67}\).

Although Perren's theory on interfragmentary strain is important to understand the concept of tissue mechanobiology at the fracture gap, several studies have demonstrated that gap strain higher than 2% is tolerated and that the strain patterns within a fracture gap are heterogenous \(^{74,75}\). It is well accepted that interfragmentary
movement is the most important biomechanical factor in fracture healing, but it is not known which is the optimal range for callus formation and bone healing.

**Bone Healing under Conditions of Absolute and Relative Stability**

The term stability is defined as the load-dependent displacement of the fracture surfaces. Stability in osteosynthesis covers a spectrum from minimal to absolute. Absolute stability is only present when there is no displacement of the stabilized fracture segments under loading (Figure 6). Absolute stability is achieved by 1) applying a compressive pre-load that exceeds the traction force acting at the segments, and by 2) counteracting the shear forces acting on the fracture surfaces with friction. The elimination of relative motion between the bone segments results from the application of interfragmentary compression and requires anatomic reduction. Placement of a lag screw is an excellent example of a fixation that can provide absolute stability (Figure 2-6). In vivo experiments have shown that a lag screw can produce high compressive forces (>2,500 N) across a fracture. While absolute stability was originally thought to be necessary for successful management of most fractures, current thinking suggests that absolute stability is only obligatory when stabilizing articular fractures and only when interfragmentary compression can be achieved without inducing excessive iatrogenic damage to blood supply and surrounding soft tissues. Limiting soft tissue trauma is an essential tenant of any fracture repair even when performing a direct open reduction, efforts should be made to minimize iatrogenic trauma to the regional soft tissues and the periosteum.

Fractures stabilized under conditions of absolute stability will heal by primary or direct fracture healing, if anatomically reduced. Since there is no motion at the fracture site, there will be negligible callus formation. The fracture heals through the formation of
osteonal cutting cones and Haversion remodeling of the compressed cortical bone \(^{78,79}\). Direct bone healing can be further subdivided into two types based on the width of the fracture gap. Contact healing occurs when the ends of the bone segments are in direct contact, the gap between the two bone segments is less than 0.01 mm and when interfragmentary strain is less than 2\% \(^{78}\). If the fracture gap is larger but does not exceed 1 mm, and an interfragmentary strain again is less than 2\%, gap healing will occur, in which intramembraneous bone will be formed directly in the fracture gap \(^{44}\). In both types, a process called Haversian remodeling begins with osteoclastic resorption that result in resorption cavities formed by groups of osteoclasts also called a cutting cone \(^{79}\). Bone resorption is followed by osteoblast activity. The osteoblasts line the resorption cavities and produce layers of new bone. The resorption cavity is filled in with new bone to form a new osteone. Gap healing results from the development of lamellar and cortical bone forming from granulation tissue in small gaps \(^{78,79}\). Intramembranous bone formation occurs during direct bone healing; the surrounding environment can impose up to 5\% strain as long as it allows the differentiation of mesenchymal cells into osteoblasts.

Relative stability is a condition where an acceptable amount of interfragmentary displacement compatible with fracture healing is present (Figure 2-7) \(^{80}\). Relative stability involves placement of implants that provide somewhat flexible fixation which allow an acceptable degree of fracture segments displacement. Fixation modalities that can be employed to provide relative stability include plates, interlocking nails and external fixators applied in bridging fashion to span a bone defect \(^{71,81}\).
Relative stability provides a mechanical environment which promotes indirect or secondary bone healing\textsuperscript{65,80}. Indirect bone healing is very similar to the embryological bone development and occurs via both endochondral and intramembranous ossification\textsuperscript{65,80,82}. The healing process by formation of callus can be divided in 4 stages: inflammation, soft callus, hard callus and remodeling\textsuperscript{82,83}. Mineralized cartilaginous callus develops at the ends of the fracture segments (gap callus), along the medullary canal (medullary callus) and on the outer cortex (periosteal callus)\textsuperscript{82,83}. The majority of the vascular circulation to the callus is derived from the surrounding soft tissues\textsuperscript{84}. Therefore, surgical techniques that preserve the soft tissue envelope adjacent to the fracture are advantageous and promote fracture healing.

The indications for using techniques that achieve absolute or relative stability differ according to fracture location, fracture configuration, soft tissue conditions and vascularity of the bone. Simple transverse, spiral or oblique fractures that can be readily anatomically reconstructed, are good candidates for anatomic reconstruction and compression or neutralization plating. More complex comminuted fractures should be treated with bridging fixation. Articular fractures should be anatomically reduced and stabilized with fixation that generates interfragmentary compression, such as lag screws\textsuperscript{71}. It is always important to consider whether it is possible to implement anatomical reconstruction when choosing the type of fixation. For example, fractures that may initially appear as simple, reconstructable fractures may instead have fragments that are too small for anatomic reconstruction. In these cases an open or closed indirect reduction technique, and a bridging stabilization technique may be indicated. Because
the success of the technique depends on the precision of the reduction, critical preoperative planning should always be performed.

**Factors Affecting Stiffness of the Plate-Bone Construct**

The stiffness of the bone-plate construct is a major determinant of the mechanism and progression of bone healing \(^1,2,8,50,68,69\). There are several parameters in addition to the material properties of the implants that need to be considered when applying a bone plate. Understanding the effect of plate type, size, length, position, screw type and screw placement is important because successful fracture healing depends on appropriate fixation stability \(^53,55,57,58,85-88\). Furthermore a multitude of plate types and concepts have been described and proposed in the last decade, in an attempt to decrease the complications and improve the reliability of bone plating. The development of new implants and techniques have followed a shift in emphasis of AO philosophy from obtaining anatomic reconstruction and absolute stability to obtaining anatomic alignment and appropriate stability utilizing more atraumatic application techniques \(^9,71\). Concurrent with this change in emphasis in internal fixation, newer implant systems such as internal fixators, locking plates or angular stable devices, have been developed to improve bone-plating technique \(^9\). Understanding the mechanical properties of locking plates and conventional plates is important for choosing an appropriate implant system.

**Choosing the type of plate: locking versus non-locking plates**

Gautier and Sommer recently presented prudent guidelines that may improve the individual learning curve of surgeons who are less familiar with locking plates \(^31\). However, it is important to understand the concepts behind these recommendations for successful utilization of the vast choice of plates available \(^10,27\). There are distinct principal biomechanical differences between bridging plates and locked internal fixators.
with regard to load transfer through a fractured bone. In conventional compression plate
constructs or nonlocking bridging plate constructs, fixation stability is limited by the
frictional force generated between the plate and the bone. This force is created by axial
screw forces and the coefficient of friction between the plate and the bone \(^8,89\). If the
force exerted on the bone while the patient is ambulating exceeds the frictional limit,
relative shear displacement will occur between the plate and the bone, causing a loss of
reduction between the bone fragments, or loosening of the screws, or both. Conventional
plates, including dynamic compression plates (DCP) \(^90\) and limited contact dynamic
compression plates (LC-DCP) \(^91\) allow compression of bone fragments utilizing the
dynamic compression holes. In a transverse fracture that has been anatomically
reduced, stability can be further increased by utilizing the plate to generate
interfragmentary compression between the ends of the fracture segments when the
screws are inserted eccentrically at the end of the oval hole (away from the fracture), the
lower hemi-spherical part of the screw head will contact the dynamic compression incline
of the compression hole. This interaction between the screw head and the compression
incline results in sliding of the screw down the incline hole in the plate, producing
compression of the ends of the fracture segments during screw tightening \(^90,91\).

Locking plates differ from non-locking plates because stability is not dependent on the
frictional forces generated at the bone-plate interface. The first plate that functioned as
an internal fixator (Zespol system) was developed in 1970 in Poland \(^92\). Since then, a
number of locking plates have been developed that utilize the concept of angular
stability. These implants consist of a plate and locking head screws which together act
as an internal fixator. Locking the head screw into the hole confers axial and angular
stability of the screw, relative to the plate. Because the stability of the construct does not depend on frictional forces generated between plate and bone, the bone-screw threads are unlikely to strip during insertion. The fixed-angle connection between the screw and the plate clearly affords improved long-term stability. Plate failure by “pull-out” is unlikely because the screws cannot be sequentially loaded or pulled out.

Locking plates have both mechanical and biological advantages. The periosteal blood supply beneath the plate is not compromised because compression between plate and bone does not occur. Preservation of the periosteal vasculature may improve healing and decrease the risk of cortical bone necrosis and infection. Another advantage is that the plate does not need to be perfectly contoured, because the bone is not “pulled towards” the plate while tightening the screw. For this reason, locking plates are often used for minimally invasive plate osteosynthesis (MIPO), which involves closed reduction and percutaneous fixation of the fracture. Several different locking plate systems are available. Some plates may have combination holes that allow placement of a locking screw or a conventional non-locking screw in either a compressive or neutralize fashion or a locking screw.

Several biomechanical studies have compared locking and non-locking plates in dogs. These studies have conflicting results. While some studies demonstrated that locking plate constructs were stiffer than non-locking plate constructs when tested in axial compression, torsion, and bending, others did not find any significant differences between the two. The most consistent finding has been that locking plates perform better than non-locking plates in osteoporotic bone.

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The biomechanical advantages of locking plates may be less evident in normal bone.

**Choosing the length of the plate**

The selection of an appropriate length plate is a very important step in the preoperative plan. Appropriate plate length is dependent on the location and configuration of the fracture as well as the intended functional application of the plate. In bridge plating, longer plates lower the pullout force acting in screws due to an improvement of the working leverage for the screws and better distribution of the bending forces along the plate. The theoretical advantage of using a longer plate without placing screws in the center portion of the plate is supported by several biomechanical studies. Sander et al compared three different plate lengths, six-, eight- or ten-hole 3.5 mm dynamic compression plates fixed on dog cadaveric ulnae, tested in four-point bending to failure. The results revealed that ten-hole plates with four screws (widely spread on the fracture fragment), failed at higher peak loads than six-hole plates with six screws, supporting the recommendation that longer plates with fewer screws provide superior bending strength than shorter plates with a greater number of screws. In another study, Weiss evaluated eight- and ten-hole 3.5 mm locking compression plates used to stabilized human cadaveric ulnas. This study found that 10-hole plates secured with two non-locking screws placed in a near-far configuration on either side of the fracture demonstrated an increased yield strength compared to 8-hole plates with the same number of screws and configuration in four-point bending to failure. Iatrogenic trauma associates with the open application of a long plate can be substantially mitigated by using less invasive application techniques such as MIPO.
Two values have been used to determine the length of the plate to be used. The plate span ratio is a quotient derived by dividing plate length by the segmental length of the fracture. Based on guidelines developed for fracture fixation in man, the plate span should be more than 2 to 3 in comminuted fractures and more than 8 to 10 in simple fractures\textsuperscript{9}. Plate-screw density is the quotient derived by dividing the number of screws inserted by the number of holes in the plate. Empirically, values below 0.4-0.3 when applied in simple fracture and value below 0.5-0.4 when applied in comminuted fracture have been recommended\textsuperscript{9,31}. These guidelines were formulated for the application of plates in human patients, and need to be evaluated in dogs.

**Effect of the position of screw placement in the plate**

In comminuted fractures which have not been reconstructed, stress is distributed over the fracture gap and is dependent on the number and location of screws placed\textsuperscript{9,53} in addition to other factors. The lowest stress in the plate occurs when the screws are as close as practical to the fracture\textsuperscript{53}. This, however, leads to the highest axial stiffness as well as very small interfragmentary movements and strains beneath the plate. It has been recommended to increase the plate working length to reduce axial stiffness of a plate-bone construct\textsuperscript{9,31,110}; however, previous mechanical studies have yielded conflicting results\textsuperscript{53,87,111}. Based on mechanical tests performed in our lab\textsuperscript{34} we suspect that the variability in the results amongst reported studies might be attributed on how the plate is applied to the bone. In constructs which utilize non-locking plates, the contact between the plate and the bone segments causes the bending moment to concentrate between the end of the bone segments regardless the positioning of the screws. Therefore, the functional plate working length does not correspond to the distance between the screws placed closest to the fracture gap, but rather to the length of the
fracture gap. In contrast, the physical offset of a locking plate which is applied without the bone and plate in intimate contact enables a locking plate to bend along the entire segment of the plate between the two most centrally positioned screws.

More recent strategies to decrease the stiffness of locking plate-bone constructs include new designs of locking screw that allow to increase the fracture gap micromotion with axial loading \textsuperscript{112-114}. The goal of this novel approach to locking plate is to promote more reliable healing, and prevent late failures observed in several clinical studies in people \textsuperscript{115-117}. The far cortex locking screw has a smooth shaft with threads at the tip that achieve purchase in only the far cortex \textsuperscript{50,112,113,118}. The smooth shaft of this screw decreases the stiffness of the plating construct and allows greater callus compared to standard locked implants \textsuperscript{112}. Another screw design attempting to combine the advantages of locking screws and controlled axial micromotion is the dynamic locking screw \textsuperscript{114}. This dynamic locking screw is composed of an outer sleeve with threads that engage the bone and an inner pin with threads that lock to the plate. By allowing motion between inner pin and the outer sleeve, the dynamic compression screws reduced the axial stiffness by 16\% \textsuperscript{114}. 
A load-deformation curve shows the relationship between stress (force applied) and strain (deformation) of an object. The elastic range (A-B) of the curve ends when the object reaches its yield strength (B) and begins to undergo permanent plastic deformation (B-C). If continued load is applied, material failure may occur in the form of a fracture, or the object may just continue to undergo further plastic deformation.
Figure 2-2. The load-deformation curves of the same object with 3 different materials: bone, glass and metal. More elastic materials can usually sustain considerable plastic deformation, while more brittle materials will fracture at relatively low loads. The more brittle rigid body such as bone undergo minimal plastic deformation before reaching the failure point, while metal has a linear elastic portion of the curve, indicating linearly elastic behavior. The long plastic region of the metal indicates that this material deforms extensively before failure.
Figure 2-3. The area moment of inertia describes the capacity of the cross-sectional profile of an object to resist bending. The greater the area moment of inertia, the less a structure will deflect (higher bending stiffness) when subjected to a bending load. Area moment of inertia is dependent on an object’s cross-sectional geometry.
Figure 2-4. Area moment of inertia is important when comparing the mechanical properties of implants of different shape and size but similar metal, such as in case of an interlocking nail, a bone plate and a plate-rod construct. As shown in the calculation, the interlocking nail has the largest area of moment of inertia because of its large diameter. It should also be noted that the area of moment of inertia of an intramedullary pin occupying 40% of the medullary canal has a significant contribution to the total area of moment of inertia of a plate-rod construct, justifying its recommendation as bridging implants for comminuted fractures. Another approach to fixation of a fracture with a gap is to increase the size of the plate. As shown in the calculation, a 3.5 mm broad LCP has an area of moment of inertia 3 times larger than a 3.5 mm LCP and almost twice larger than a 3.5 mm LCP combined with an intramedullary rod.
Figure 2-5. A construct’s fatigue behavior can be described in an S/N curve; in which the stress to failure, S, is plotted against the number of cycles to failure, N. Typically, a construct subjected to a small applied-stress can withstand a large numbers of cycles and vice versa.
Figure 2-6. Absolute stability is only present when there is no displacement of the stabilized fracture segments under loading. The elimination of relative motion between the bone segments results from the application of interfragmentary compression and requires anatomic reduction.

Figure 2-7. Relative stability is a condition where an acceptable amount of interfragmentary displacement compatible with fracture healing is present.
CHAPTER 3
EXPERIMENT

Effect of Plate Working Length on Plate Stiffness and Cyclic Fatigue Life in a Cadaveric Femoral Gap Model Stabilized with a 12-hole 2.4 mm LCP

There are several factors that can affect the fatigue life of plated fracture construct. In addition to the mechanical properties of the isolated plate, the location, type and complexity of the fracture influence the load acting on the plate. Bridging plates are often utilized in comminuted fractures to span a large segment of the fractured bone which is not reconstructed, having a long segment of the plate unsupported by screws. The distance between the proximal and distal screw in closest proximity to the fracture is defined as the “working length” of the plate. Plate working length has been shown to influence construct stiffness, plate strain and cyclic fatigue properties of the plated constructs. The lack of load sharing between the stabilized bone segments and the implants increases the risk of cyclic fatigue and potentiates early failure of the implant. A mechanical study evaluating the mechanical endurance of human femora stabilized with 14-hole broad 4.5 mm locking compression plates (LCP) found that constructs with load sharing resisted 20 times more cycles than the constructs with an 8 mm segmental diaphyseal gap.

Controversy still exists regarding the effect of plate working length on stiffness and resistance to fatigue failure. Stoffel et al. reported that increasing the plate working length by omitting one screw placed adjacent to the fracture in each of the major fracture segments made a locking plating construct nearly twice as flexible when loaded in axial compression and torsion. In contrast, Field et al. reported that omitting two screws proximal and distal the fracture had no significant effect on either bending or torsional stiffness of a conventional plate construct in a comparable bridge plating configuration.
Studies evaluating the effect of working length on the cyclic fatigue properties of plated constructs have yielded inconsistent results. Stoffel et al. found that the constructs with a shorter plate working length were more resistant to cyclic loading. Maxwell et al. reported similar results when screw placement was evaluated in 3.5 mm dynamic compression plates and limited contact dynamic compression plates applied in a fracture gap model. Recent studies comparing the cyclic fatigue of plates applied with a short and long working length found that the constructs with a longer working length withstood more cycles before failure, although the differences between stabilization techniques were not significant.

The purpose of the present study was to evaluate the effect of plate working length on construct stiffness, gap motion and resistance to cyclic fatigue in dog femora with a simulated fracture gap stabilized using a 12-hole 2.4 mm LCP. Our first hypothesis was that constructs with a longer plate working length would be more flexible and have greater gap motion than constructs with a shorter plate working length. The second hypothesis was that constructs with a longer plate working length would withstand more cycles before succumbing to fatigue failure than constructs with a shorter plate working length.

**Materials and Methods**

**Specimen Preparation**

This study was approved by the University of Florida Institutional Animal Care and Use Committee. Ten pairs of femora were collected from adult dogs (mean ± SD: 17.9 ± 1.9 kg) that were euthanized for reasons unrelated to the study. Once all the soft tissues were removed from the femurs, the bones were radiographed in order to exclude specimens with osseous abnormality. The femora were then wrapped in saline-soaked
towels and stored at -20°C until testing, at which time the bones were thawed to room temperature. One femur from each dog was randomly assigned to the short plate working length stabilization group. The contralateral femur was assigned to the long plate working length stabilization group.

A 12-hole 2.4 mm LCP\(^1\) was contoured and clamped temporarily to the lateral surface of each femur. Two lines, 10 mm apart were marked on the mid-diaphyseal region of the bone adjacent to the central 2 holes of the plate. The plate was removed, and an oscillatory saw was used to initiate two mid-diaphyseal partial osteotomies in the lateral cortex of the femur at the marked locations. In the femora assigned to the long plate working length stabilization group, the plate was affixed with one 2.4 mm locking and one 2.4 mm cortical screw placed at the ends of the plates, leaving eight empty holes in the middle of the plate (Figure 3-1). The femora assigned to the short plate working length stabilization group were fixed with one 2.4 mm locking and four 2.4 mm cortical screws in each femoral segment, leaving two empty holes in center of the plate (Figure 3-1). All screws had bicortical purchase and were placed in the same order. Cortical screws at each end of the plate were placed first, followed by the locking screw in the second hole and the remaining cortical screws. All screws were placed and hand-tightened by an experienced board-certified surgeon (AP) in accordance with standard AO-ASIF techniques.\(^1\) Prior to testing, the torque of each locking and cortical screw was tightened to 0.8 Nm or 0.44 Nm, respectively.\(^2\) After applying all screws, osteotomies were completed from the medial cortex of the femur resulting in a mid-diaphyseal segmental ostectomy.

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\(^1\) Synthes USA, Paoli, PA

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The distal ends of femora were potted using epoxy resin\(^2\). A goniometer was used to ensure that the specimens were potted perpendicular to the testing mold. Two points located on the cranial and caudal surface of each femur segment adjacent to the fracture gap were first marked then scribed using a 1.5 mm drill bit, so these points could be accurately identified with the tip of the articulating arm of a three-dimensional positioning device\(^3\).

**Mechanical Testing**

All testing was performed using a servohydraulic MTS\(^4\). During testing, each specimen was positioned vertically with the epoxy resin block secured to the machine using a specially designed fixture. A polyoxymethylene cup simulating the shape of the acetabulum was secured to the mechanical test system and used to load the femora. In order to measure gap displacement of the bone segments, three points on each bone fragment were digitized using the Microscribe while the limb was preloaded to 10 N.

A cyclic axial compressive load from 0 N to 100% body weight (mean ± SD: 175.5 ± 18.6 N) was applied at 2 Hz, while displacement data was recorded at 100 Hz throughout testing. Testing was paused at 1000, 2000, 5000, 10000, 20000, 50000, 100000, 120000, and 18000 cycles to obtain gap displacement measurements as each specimen was loaded. If fatigue failure did not occur after 180,000 cycles, the specimen was loaded in axial compression at a rate of 1 mm/seconds until failure was achieved. Displacement data was recorded at 100 Hz while specimens were loaded to failure.

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\(^2\) Bondo Corp, Atlanta, GA

\(^3\) Microscribe 3DX digitizing arm (Immersion Corp., San Jose, CA)

\(^4\) MTS 858 Mini Bionix II, MTS Systems Corp, Eden Prairie MN
Failure was defined as breakage or plastic deformation of the implants or any visually detected loosening of screws in the plate or bone. Video was recorded during testing to identify the mode of failure for each construct.

**Data and Statistical Analysis**

Displacement values were recorded at 1000, 2000, 5000, 10000, 20000, 50000, 100000, 120000, and 180000 cycles. A load-displacement curve was calculated by plotting the load versus displacement data for each specimen for each time point during cyclic testing in a scatter graph using spreadsheet software\(^5\). A best fit trend line, straight line slope, and corresponding \(R^2\) value were determined for each load-displacement curve; using a sum of least squares method for the linear elastic region of the load-displacement curve using the same software. The slope of this line defined the stiffness value and was reported in units of N/mm.

Relative gap displacement of two bone segments in the sagittal and frontal plane was analyzed using classical principles of rigid body mechanics\(^{123}\). Gap displacement of each construct was also calculated to compare whether there was any significant difference between short and long plate working length stabilization techniques. Calculations were performed using a custom-written computer program\(^6\).

A separate load-deformation curve was determined similarly from the initial linear segment of load to failure test for each construct. The elastic limit or yield load was defined as the loading value at which the linear phase of the curve terminated\(^{124}\).

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\(^5\) Microsoft Office Excel 2003, Microsoft Corporation, Redmond, Washington

\(^6\) MATLAB®, MathWorks Corporation, Natick, Massachusetts
values and standard deviations of stiffness, gap displacement and yield load were determined for both the short and long plate working length stabilization techniques.

Stiffness values, cranial/caudal translation, medial/lateral translation of two bone segments and gap displacement measured during cyclic testing were compared between the short and long plate working length stabilization techniques using a 2 x 9 (technique x cycle) within subjects repeated measures ANOVA. When appropriate a post hoc Bonferroni procedure\textsuperscript{7} was used to adjust for multiple pairwise comparisons. The yield load determined during load to failure testing was statistically compared between the short and long plate working length stabilization techniques using a paired t-test. For all statistical analyses, significance was set at p < 0.05.

**Results**

When the constructs were cyclically loaded, there was no significant difference (p=0.435) in stiffness between the short and long plate working length stabilization techniques. There was a trend for the stiffness to increase during cycling in both short and long working length plate stabilization techniques until 50,000 cycles. After reaching 50,000 cycles, the mean stiffness of long working length plate technique decreased while the mean stiffness of short working length plate technique continued to increase, although these trends never became significant between stabilization techniques (Figure 3-2). All constructs completed 180,000 cycles of loading without plate or screw breakage, screw loosening or visible plastic deformation of the implants observed in any of the constructs.

\textsuperscript{7} SPSS Statistics 17.0, SPSS Inc., Chicago, IL
There were no significant differences in the relative displacement of the bone segments between stabilization techniques in the sagittal (p=0.448) or frontal plane (p=0.504). Gap displacement also did not differ significantly within the designated cycles and between stabilization techniques (p=0.116). The ends of two stabilized bone segments did not come into contact in any of the constructs during cyclic compression loading.

During load to failure testing, constructs in the short plate working length stabilization technique had a significantly (p=0.016) higher yield load (mean ± SD: 654.5 ± 149.0 N) than constructs in the long plate working length stabilization technique (mean ± SD: 452.9 ± 63.4 N). The mean ± SD stiffness of the short (223.8 ± 46.7 N/mm) and long (219.4 ± 80.3) plate working length stabilization techniques during load to failure testing was not significantly different. Failure patterns of the short and long plate working length stabilization techniques were distinctively different (Figure 3-3): the proximal femoral segment bent medially during axial loading in both the short and long plate working length constructs. The lateral surface of the proximal femoral segment acted as a fulcrum against the plate and there was no separation of the plate from the lateral cortex of the femur. The plate elevated from the lateral cortex of the distal femoral segment in the long plate working length constructs. The plate remained apposed to the lateral cortex of the femur in the short plate working length constructs.

Review of video obtained during the load to failure testing revealed that eccentric axial compressive load applied to the femoral head leads to the formation of a bending moment, with the center of rotation located at the plate bone interface. Bending moment was defined as the product of applied force (F) amount and the perpendicular distance
(d) from the applied force to the axis of rotation (Figure 3-3). The bending moment increased during the load to failure test as the plate deflected away from the lateral cortex of the femur. This effect was more pronounced in the long plate working length constructs.

**Discussion**

Increasing working length by placing a limited number of screws at each end of the plate has been recommended as a strategy to decrease construct stiffness and therefore allow more motion at the fracture gap \(^9,31,48,53,110\); however, previous mechanical studies have yielded conflicting results \(^53,57,87,111\). Based on our results, we cannot support our hypotheses that constructs with fewer screws and a longer plate working length would be more flexible, have greater gap motion and more resistant to cyclic fatigue failure than constructs with more screws and a short plate working length. We suspect that the inconsistencies in the results amongst reported studies might be attributed to the manner in which the plate was applied to the bone. In constructs which utilized non-locking plates \(^58,111\), the contact between the plate and the bone segments causes the bending moment to concentrate between the ends of the bone segments regardless the positioning of the screws. Therefore, the functional plate working length does not correspond to the distance between the screws placed closest to the fracture gap, but rather to the length of the fracture gap. In contrast, the physical offset of a locking plate which is applied without the bone and plate in intimate contact enables a locking plate to bend along the entire segment of the plate between the two most centrally positioned screws \(^53\). If our suppositions are confirmed, the concept of short and long plate working length should only be applied to locking plates and should not be applied to conventional non-locking plates.
The results of our cyclic fatigue testing are not consistent with previous studies which found differences between constructs with short and long plate working lengths. These contrasting results might be ascribed to differences in the fracture models employed and methods of loading. In our study the applied load was based on the dogs' body weight in an attempt to replicate clinical situations. In other studies the upper load threshold was set to induce implant failure within a predetermined number of cycles or was established using displacement control based on plate strain. Establishing significance may have been masked in our study because we did not apply high enough loads as none of our constructs failed during cyclic testing. The magnitude of the load applied in diaphyseal fracture models is more complex than simply applying a load equivalent to mean body weight. In human patients which have been instrumented with distal femoral diaphyseal prosthesis with telemetric strain gauges, load values as high as three times body weight and bending forces up to ten times body weight have been recorded. Other loading protocols such as progressive loading could have been used to shorten the test protocol and may have yielded significant differences between stabilization techniques.

Both the short and long plate working length stabilization constructs sustained 180,000 cycles with the equivalent of full body weight loaded in axial compression without failure, which approximates 2 to 4 months of weight-bearing during normal ambulation. This finding suggests that both constructs would likely provide acceptable clinical stability as most fractures are expected to obtain union within 12 weeks of stabilization. The yield load for both stabilization techniques ranged from 453 to 655 N, which corresponds to 2 to 4 times the hindlimbs peak vertical force measured from a
normal 20 kg dog running at a trot\textsuperscript{15,127}. Although kinetic values measured with gait analysis are only an approximation of the forces tolerated by implants during loading, our results suggest that it is unlikely that either of our constructs would fail catastrophically in the early postoperative convalescent period.

Cadaveric studies have numerous limitations. When performing cyclic testing designated to resemble a clinical environment after fracture fixation, the loading plane should be considered\textsuperscript{128}. We selected an offset axial loading to simulate loading of a plated femoral fracture. Our testing methodology had the limitation of being isolated to a single plane, without considering more complex forces such as a combination of bending and torsional forces. In the diaphyseal region of the femur, however, axial and bending forces predominate and these forces were replicated in our testing protocol\textsuperscript{125}. Another limitation was that we did not test a construct which exclusively utilized locking screws. This omission limits our ability to make conclusions regarding the effect of working length of a locking plate employed in its intended application\textsuperscript{53,87}. A hybrid construct in which both locking and non-locking screws were used was selected because these constructs are commonly used in dogs\textsuperscript{16}. Furthermore, previous studies have shown that placement of a single locking screw in each of the major fracture segments increases construct stability subjected to axial and torsion loading\textsuperscript{16,96,129,130}.

We were unable to establish a significant effect of working length on stiffness and resistance to cyclic fatigue in a fracture gap model plated with 2.4 mm LCP applied in direct cortical contact with the femur. Our results question the advantage of long working length in decreasing stiffness of the bone-plate construct and protecting the portion of unsupported plate at the fracture gap\textsuperscript{6,48,131}. Neither of the constructs tested in our study
failed during cyclic load. Therefore both short and long working lengths constructs would appear to be acceptable for clinical applications. Placing five bicortical screws in each of the major fracture segments may not be necessary. Other studies have reported that fewer, more widely spaced screws increased the bending strength of screw–plate fixation more effectively than increasing the number of screws. Guidelines regarding the optimal number of screws will, however, need to be derived from future mechanical and clinical studies.
Figure 3-1. Constructs tested: long plate working length stabilization technique (left) and short plate working length stabilization technique (right). Each plate had a 2.4 mm locking screw (○) placed in the second hole from each end of the plate. The other screws were 2.4 mm cortical screws (●).

Figure 3-2. Comparison of mean ± SD values of construct axial stiffness (N/mm) loaded to body weight between long plate working length stabilization technique (□) and short plate working length stabilization technique (■). Measurements recorded at 1000, 2000, 5000, 10000, 20000, 50000, 100000, 120000, and 180000 cycles. No significant difference was found between the two stabilization techniques at any of the evaluated cycles.
Figure 3-3.  Photographs of a long and short plate working length construct before (left) and after (right) load to failure testing. The adjacent free body diagrams represent how bending moments were created. Bending moments (Force (F) x Distance (d)) were calculated based on measured scale distances obtained from photographs and the yield load of each construct.
Plate working length is a major determinant of construct stiffness. A guideline of whether to keep plate working length short or long remains controversial. The recommendation of keeping at least two or three empty plate holes at the fracture site were made in order to allow adequate construct flexibility when bridging a fracture. Additionally it was theorized that two or three empty plate holes at the fracture site would decrease plate strain at the fracture gap, therefore decreasing the risk of fatigue failure. However, several studies have demonstrated that a screw should be placed as close to the fracture gap as possible in each major fracture segment, to decrease plate strain and increase the resistance to fatigue failure. Based on our testing results we cannot support the hypotheses that constructs with longer plate working length stabilization technique have lower stiffness, greater gap motion and are more resistant to fatigue failure. No significant effect of working length were found on construct stiffness, gap motion and resistance to cyclic fatigue in a fracture gap model plated with 2.4 mm LCP applied in direct cortical contact with dog femur. Both short and long plate working length stabilization techniques should be able to sustain through post-surgery coalescent period in dogs.

Our results are inconsistent with previous studies in which they found that longer plate working length stabilization technique has less stiffness, higher tensile strain and earlier fatigue failure than short plate working length stabilization technique. Inconsistent findings may results from different fracture model and loading methods. In our study we fixed the plates flush on the femoral cortex which probably limited the effective working length to the portion of the plate bridging the gap. In other words, the
working length may equal the size of fracture gap in a non-locking plate-bone construct. Stoffel et al. applied plates at 2 mm offset from plate to bone, allowing plate bending over the entire distance between two inner most screws on the plate. Therefore the plating working length in a locking plate-bone construct truly corresponds to the distance between the two innermost screws. Stoffel et al. focused on the contribution of the plate to stiffness of the construct. This result results are similar to other studies that found no significant difference of constructs stiffness and fatigue resistance between long and short plate working length stabilization techniques. The mode of failure and plate deformation observed in our study was consistent with Hoffmeier et al.’s study. In long plate working length stabilization constructs, the free plate sections on both sides of the gap bulged away from the bone during bending. Hoffmeier et al. found that the larger the working length, the larger the radius of curvature would be and the evenly the stress would be distributed. We might found the similar result of the stress distribution on the plate because we have used similar femoral fracture model without leaving distance between plate and bone as in Hoffmeier et al.’s study.

We applied LCP in hybrid fixation in which one locking screw was placed next to other cortical screws in each bone segment. The effect of using locking screws in combination with cortical non-locking screw in LCP applications is multifold: using locking screws has been reported to increase axial strength and torsion resistance, as well as increase the remaining torque in nearby cortical screws than all non-locking screws fixation. Even though there is currently no study comparing the effect of different plate working length on hybrid plate fixation, one study suggested that hybrid plate fixation has similar bending strength and higher torsional strength than an
all-locked bridge plating construct in the osteoporotic diaphysis. Therefore, it is unlikely that the hybrid plate fixation we used was the reason led to different results between our study and other all-locked-screw study.

Cyclic loading is the mechanical test that more closely resembles the conditions of fracture-healing in the clinical patient. However, it presents several limitations and challenges. First, the loading plane should simulate the in vivo conditions that the implant would experience in vivo. In our study we selected axial compression loading which cause mainly bending force on the femur, because it is one of the most commonly force that femurs encountered in physiological conditions. The magnitude of load is another major factor that should be considered when designing a cycling test. To simulate in vivo condition of loading, clinically relevant forces should be used. In people, the physiological basis for testing loads is based on telemetrized implants such as interlocking nails or total knee replacements, which directly measure the forces acting at the level of the femoral diaphysis or at the knee joint, respectively. Because we lack of these data for dogs, we chose testing loads based on dog’s body weight in an attempt to replicate clinical situations.

The number of cycles required to test an implant for fracture fixation is controversial. The presence of callus is a major factor in construct stability because it may develops as early as 2 weeks following fracture fixation. Fracture callus protects the implant from failure by providing load sharing, but cannot simulated in a laboratory setting. Our study design is different than other studies in which fatigue tests were designed either the upper load threshold was set to induce implant failure within a predetermined number of cycles or the number of cycles are as high as 1,000,000 cycles. Some other
studies used protocols as control displacement based on plate strain\textsuperscript{87} or applied progressive loading\textsuperscript{118}, aimed at shortening the duration of cyclic fatigue test while also assuring if significant differences between stabilization groups. Because all constructs sustained through 180,000 cyclic weight-bearing loading in our study, the loads and the number of cycles we chose were not high enough for constructs to reach fatigue failure. Therefore, the significance may have been masked in our study.

The material of the plate is another factor affecting fatigue endurance of plate-screw constructs. Hoffmeier et al compared the fatigue life between short, medium and long working length in both stainless steel and grade-2 titanium 4.5 mm distal femoral plates which bridged over a 10 mm fracture gap on a composite human femur\textsuperscript{85}. When plates of different metals were compared, the stainless steel plates sustained higher cyclic load than the grade 2-titanium plates. For the steel plates, no significant correlations were found between fatigue strength and different working length groups. For the titanium plates the fatigue strength was greater in constructs with long plate working length but this correlation was not significant. Therefore, the effect of plate working length on fatigue endurance may be greater in titanium plates than in stainless steel plates.

We did not evaluate other factors that affect plate-screw constructs stability such as the number of screws inserted in each fragment. Studies which evaluated the effect of different screw numbers on stiffness of plate-screw constructs reported that fewer, more widely spaced screws increased the bending strength of screw–plate fixation more effectively than increasing the number of screws\textsuperscript{31,55,131}. Other studies compared the effect of different screw positions on plate strain or fatigue resistance\textsuperscript{85,111}. Therefore, the effect of screw numbers should be minimal compared to the position of screws.
Future studies are necessary to evaluate if it is necessary to place screws in each screw holes in the plate, especially considering the vascular trauma caused by drilling screw holes in cortical bone \textsuperscript{132}.

Stoffel et al made the statement that as long as more than four screws were placed each bone segment, the torsional rigidity would not be affected by the position of the screws. Similar with other study suggested that regardless of the spacing of screws and the plate length, strength in torsion was dependent on the number of screws securing the plate \textsuperscript{53,55}. One study also reported that add a single locking screw to an otherwise non-locking construct would increase the torsion rigidity \textsuperscript{16}. Therefore, if we tested our constructs in torsion, the major variable that causes any difference should be the number of screws rather than plate working length.

Longer plate working length stabilization technique has been shown to result in higher plate strain in both experimental and computational study models \textsuperscript{53,57,119,133}. The authors concluded that screws should be placed as closed to the fracture as possible in order to decrease plate strain. However, one study evaluated the strain at and around the fracture site in a gap model stabilized by 12-hole 3.5 mm LC-DCP and DCP, and reported that the gap strain does not significantly change after screw removal \textsuperscript{111}. The other study used a femoral fracture model and found a reduction of plate strain with increasing plate working length stabilization technique \textsuperscript{87}. This finding is supported by a computational simulation showing that a short plate working length produces stress concentrations in the plate, which leads to fatigue failure \textsuperscript{134}. We did not measure plate strain during test; however, we did not find any significant difference in gap motion during cyclic testing. Based on our results, it is likely that we would not find any difference in gap
strain because gap strain is affected by gap motion. After load to failure test, both short and long plate working length stabilization techniques have similar bending fulcrum at the end of the proximal bone segment, which is likely the area where we would find the greatest plate strain. However, the strain distribution along the entire plate between two plate working length stabilization techniques during cyclic loading test would need further testing to be determined.

Both short and long plate working length stabilization constructs should be acceptable for clinical applications, because no construct failed during cyclic weight-bearing loading. Based on our study design, no conclusion regarding clinical guidelines can be provided and further mechanical and clinical studies are warranted. Future study design should exclude the effect from different number of screws and the application of both locking and non-locking screws in the plate.

When preparing a fracture stabilization plan, the choices will be different based on different intention. If one would like to provide a fixation without traumatizing excessively the surrounded soft tissue, a MIPO technique and a longer plate working length should be acceptable based on our study result. However, if the higher strength of the fixation construct is the goal, we would suggest placing screw as closed to the fracture gap as possible in each of the major fracture segment. Future mechanical and clinical studies are needed to establish guidelines regarding the optimal application of locking compression plates.
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BIOGRAPHICAL SKETCH

Peini Chao grew up in Taipei, Taiwan with her family. She graduated from Taipei First Girl High School in 2003. She studied in the college of Veterinary Medicine in National Taiwan University. After one-year rotational internship training in Animal Hospital of National Taiwan University, she earned her D.V.M. in National Taiwan University in 2009. During the summer in 2007, she completed an externship in Elwood Animal Clinics in California.

Upon graduating in August 2009 with her D.V.M., Peini enrolled the Master of Science program in Department of Small Animal Clinical Science in College of Veterinary Medicine in University of Florida. She works in Comparative Orthopedic Biomechanics laboratory, research strengthens on mechanical evaluation of plate-screw constructs. She earned her M.S. in summer 2012.

After completion of her M.S. program, Peini will be back to Taipei, Taiwan, continuing work in small animal practice.