THE EFFECT OF THE COMBINATION OF LOCKING SCREWS AND NON-LOCKING
SCREWS ON THE TORSIONAL PROPERTIES OF A LOCKING PLATE CONSTRUCT

A Thesis

Presented To

The Faculty of Graduate Studies

of

The University of Guelph

by

SEBASTIAN S. N. GORDON

In partial fulfilment of requirements

for the degree of

Doctor of Veterinary Science

January, 2010

© Sebastian S. N. Gordon, 2010

Library and Archives
Canada

Published Heritage
Branch

395 Wellington Street
Ottawa ON K1A 0N4
Canada

Bibliothèque et
Archives Canada

Direction du
Patrimoine de l'édition

395, rue Wellington
Ottawa ON K1A 0N4
Canada

Your file Votre référence
ISBN: 978-0-494-64509-3
Our file Notre référence
ISBN: 978-0-494-64509-3

NOTICE:

The author has granted a non-exclusive license allowing Library and Archives Canada to reproduce, publish, archive, preserve, conserve, communicate to the public by telecommunication or on the Internet, loan, distribute and sell theses worldwide, for commercial or non-commercial purposes, in microform, paper, electronic and/or any other formats.

The author retains copyright ownership and moral rights in this thesis. Neither the thesis nor substantial extracts from it may be printed or otherwise reproduced without the author's permission.

AVIS:

L'auteur a accordé une licence non exclusive permettant à la Bibliothèque et Archives Canada de reproduire, publier, archiver, sauvegarder, conserver, transmettre au public par télécommunication ou par l'Internet, prêter, distribuer et vendre des thèses partout dans le monde, à des fins commerciales ou autres, sur support microforme, papier, électronique et/ou autres formats.

L'auteur conserve la propriété du droit d'auteur et des droits moraux qui protègent cette thèse. Ni la thèse ni des extraits substantiels de celle-ci ne doivent être imprimés ou autrement reproduits sans son autorisation.

In compliance with the Canadian Privacy Act some supporting forms may have been removed from this thesis.

While these forms may be included in the document page count, their removal does not represent any loss of content from the thesis.

Conformément à la loi canadienne sur la protection de la vie privée, quelques formulaires secondaires ont été enlevés de cette thèse.

Bien que ces formulaires aient inclus dans la pagination, il n'y aura aucun contenu manquant.

■ ♦ ■
Canada

ABSTRACT

THE EFFECT OF THE COMBINATION OF LOCKING SCREWS AND NON-LOCKING SCREWS ON THE TORSIONAL PROPERTIES OF A LOCKING PLATE CONSTRUCT

Dr. Sebastian Stefan Naudé Gordon
University of Guelph 2009

Advisor:
Dr. N.M.M. Moens

This thesis investigated the effects of replacing one non-locking screw with a locking screw on the torsional properties of a bone-plate construct. Sixty cadaveric canine femurs were divided into 3 groups. In the first group, the plate was affixed using three non-locking screws, in the second group only locking screws were used while a combination of one locking and two non-locking screws were used in the third group. These constructs were then subjected to torsion to failure using a specifically crafted jig on a servohydraulic materials testing machine. Torque, angle of torsion, and work were all calculated at the maximum failure point, as well as at 5 degrees of plastic deformation (offset failure point), which was thought to be more representative of clinical failure. At the maximum failure point, the locking and non-locking groups had a significantly different torque (51.83 Nm vs. 39.17 Nm, respectively), angle (47° vs. 31°), and work (1616.6 Nm-degree vs. 867.8 Nm-degree). The combination group was intermediate to the two other groups and had a significantly lower work compared to the locking group (1121.2 Nm-degree vs. 1616.6 Nm-degree), and a significantly higher torque compared to the non-locking group (47.43 Nm vs. 39.17 Nm). At 5 degrees of plastic deformation, the locking group required significantly higher torque and work than the non-locking group (45.14 Nm vs. 34.78 Nm and 626.37 Nm-degree vs. 461.68 Nm-degree,

respectively). The combination group required a significantly higher torque compared to the non-locking group (40.6 Nm vs. 34.78 Nm). None of the slopes of the curves for the different constructs had a significantly different slope, meaning that the stiffness of all the constructs was similar. This study suggests that a construct composed of all locking screws will fail at a greater angle, at a greater torque value, and sustain greater work to failure in torsion compared to a construct composed of all non-locking screws. The addition of a single locking screw to an otherwise all non-locking construct will increase the torque at the offset failure point and may be of clinical value in constructs subjected to high torsional loads.

Acknowledgements

Many thanks are due to my DVSc advisory committee: Drs. Noël Moens, John Runciman, David Holmberg, and Sarah Boston for their continued support, advice, and encouragement throughout this research project. Special thanks are due to Dr. Moens for the additional hours spent fine-tuning the scientific manuscript. I would also like to thank Dr. Ron McLaughlin (Mississippi State University – external examiner) for their participation in the defense and oral examination process. Mary Leunissen was invaluable for technical support in the materials testing laboratory. Thanks to Gabrielle Monteith for her assistance with the statistical analysis. Finally I would like to thank the Ontario Veterinary College Pet Trust Foundation and Synthes for their generous financial support of this research project.

Declaration of Work Performed

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

Statistical analysis was performed by Gabrielle Monteith, Department of Clinical Studies, Ontario Veterinary College, University of Guelph, Guelph, Ontario.

Table of Contents

	Page
Chapter 1 – Literature Review	1
1.1 – Bone Healing	1
1.2 – Introduction to Conventional Plating and Biological Plate Osteosynthesis	3
1.2.1 – Evolution of Internal Fixators	6
1.3 – Mechanics of Conventional versus Locking Plates	10
1.3.1 – Conventional Plates	10
1.3.2 – Locking Plates	10
1.4 – Biology of Conventional versus Locking Plates	11
1.4.1 – Blood Supply to Fractures	11
1.4.2 – Infection	13
1.4.3 – Implant Stability	14
1.5 – Biomechanical Comparison of Locking and Non-Locking Implants	17
1.5.1 – Force and Displacement	18
1.5.2 – Long Bones, Fractures, and Torsion	20
1.5.3 – Bone Plates and Torsion	22
1.5.4 – Mechanical Testing of Locking Plates Compared to Non-Locking Plates	23
1.5.5 – Mechanical Testing of Locking Plates and Non-Locking Plates in Decreased Quality Bones	28
1.5.6 – Biomechanical Testing of Locking Screws versus Conventional Screws in Locking Plates	32
1.5.7 – Biomechanical Testing of a Combination of Locking and Non-Locking Screws in Locking Plates	37
1.6 – Clinical Studies of Locking Plates	39
1.6.1 – Clinical Studies of Locking Plates in Human Medicine	39
1.6.2 – Clinical Reports and Studies of Locking Plates in Veterinary Medicine	45
1.7 – Summary of Literature Review	49
1.8 – References	51
Chapter 2 – Introduction to Scientific Manuscript	59
2.1 – Materials and Methods	60
2.1.1 – Discussion about Materials and Methods	67
2.2 – References	71

Chapter 3 – The Effect of the Combination of Locking Screws and Non-Locking Screws on the Torsional Properties of a Locking Plate Construct	73
3.1 – Abstract	74
3.2 – Introduction	75
3.3 – Materials and Methods	76
3.4 – Results	80
3.5 – Discussion	83
3.6 – References	91
Chapter 4 – Final Summary	95
Chapter 5 – Data CD	Back Cover

List of Tables

Page

Table 1 – Adverse effects in study on first clinical results of the Locking Compression Plate. 41

Table 2 – Summary of the data, comparing all locking, combination screw, and all non-locking group. 82

List of Figures and Illustrations

	Page
Figure 1 – Demonstrating the undersurface of various plates	7
Figure 2 – Example of LISS for distal femur	8
Figure 3 – An LCP using both conventional (non-locking) screws and locking screws	8
Figure 4 – LCP combination hole without and with screws	9
Figure 5 – Demonstrating toggling of non-locking screw in plate	15
Figure 6 – Comparison of locking and non-locking plates placed under bending loads	16
Figure 7 – Typical load-displacement curve	19
Figure 8 – Diagram of and formula for torque	20
Figure 9 – Demonstration of stress concentration around a hole	22
Figure 10 – Load to failure as a function of bone mineral density	31
Figure 11 – Groups used in Weiss et al, 2008 study on locked versus unlocked plating	32
Figure 12 – Synthes locking tibial plateau leveling osteotomy plate for left limb	34
Figure 13 – Postoperative radiographs of repair of tibial fracture	48
Figure 14 – The three bone-plate construct groups	61
Figure 15 – Testing set-up	62
Figure 16 – Typical calibration curve	63
Figure 17 – Typical output versus force calibration table and curve	63
Figure 18 – Deformation versus torque curve	65
Figure 19 – Trapezoidal method for calculating area under a curve	66
Figure 20 – Testing jig	79
Figure 21 – Example of deformation versus torque curve	79
Figure 22 – Examples of torque/deformation curves comparing locking, combination, and non-locking plates	81

Chapter 1 – Literature Review

1.1 – Bone Healing

Successful fracture fixation depends upon achieving stability by counteracting the various forces acting on the fracture fragments and thus maintaining the strain at the fracture site at a level that will allow the fracture to heal. Depending upon fracture conditions, two different types of healing can be observed at the fracture site: primary or secondary bone healing. Primary bone healing, also known as endosteal healing, occurs without callus formation, while secondary bone healing, also known as endochondral ossification, is characterized by callus formation (1). Strain at the fracture site is one of the main factors that determine the type of bone healing. Strain (ϵ) is the amount of deformation (change in length, ΔL) of a material compared with the original dimension (L) when subjected to a force. This is expressed as the change in dimension per unit of the original dimension.

$$\epsilon = \Delta L/L$$

Strain is a unitless measurement as it is the ratio between two lengths.

Strain at a fracture site is defined as the change in length of the fracture gap divided by the fracture gap length. Primary bone healing may only occur when the strain at the fracture site is less than 2%, while secondary bone healing occurs when the strain is between 2% and 10%. If the strain is greater than 10% bone cannot form (2). The amount of strain at the fracture site is determined by the stability of the fracture and the size of the fracture gap.

Fracture stability can be thought of as the amount of displacement of the fracture surface that occurs when the fracture is loaded (1). If the fracture ends are being

compressed against each other then no displacement will be observed when the fracture is loaded. This is an example of a situation where absolute stability has been achieved. In this type of situation the fracture will heal without radiographic evidence of callus, in a process known as primary healing. Since the fragment edges will never be in perfect apposition on the microscopic level, two types of healing occur under the umbrella of primary healing, contact healing and gap healing. The first, contact healing, occurs in areas where the bone ends contact each other. In contact healing, cutting cones formed of osteoclasts and osteoblasts, will resorb and form bone simultaneously. Gap healing occurs in areas where there is a small gap between the fracture ends, but absolute stability is still achieved due to the zones of contact around these gaps. The gap is generally less than 1 mm. The gap is initially filled with loose connective tissue, before lamellar bone is deposited there by osteoblasts. Cutting cones then remodel this lamellar bone over time (3).

The second type of bone healing is known as indirect or secondary bone healing. This type of bone healing occurs when the implants do not provide enough stability to reduce the strain levels at the fracture site to allow bone tissue to form. In this situation the fracture ends may be seen to resorb. The reason for this can be explained by strain theory, which states that tissue cannot be produced under strain conditions that exceed the elongation at rupture of that given tissue element(1). Since strain is equal to the change in length of the fracture gap divided by the fracture gap length, then for a given amount of instability, resorption will increase the fracture gap and thus reduce the strain.

Concurrently, different types of tissues will be seen in the fracture site, replacing each other and slowly increasing in strength and stiffness. The initial tissues that are

formed are able to withstand high levels of strain without rupturing. As the healing process progresses and the fracture is stabilized by the forming callus, these tissues are progressively replaced by tissues that can withstand lower levels of strain without rupturing.

The initial tissue is the fracture hematoma that forms as a result of the fracture and the subsequent bleeding from the bone and the tissues surrounding the bone. Although there is still debate over the functions of the hematoma, it is believed that it acts as a source of cells to help with the healing process, and also acts as a scaffold for the forming callus. After a few days the hematoma turns into granulation tissue. Fibroblasts, capillary sprouts, and mononuclear cells are found in this tissue. Granulation tissue then matures into connective tissue containing collagen fibers. Fibrocartilage can also be found at this stage of healing. Once the strain is sufficiently reduced these tissues are then mineralized. Remodeling of this mineralized tissue follows, eventually resulting in a healed fracture (3)).

1.2 – Introduction to Conventional Plating and Biological Plate Osteosynthesis

There are numerous approaches to controlling the level of strain at the fracture site including external coaptation, intramedullary pins and cerclage wire, external skeletal fixators, interlocking nails, and bone plates. Plate osteosynthesis refers to the practice of using a bone plate to stabilize a fracture. It was first described in 1886 by Hansmann (4). Since that time there have been numerous adaptations and improvements to the process. In the 1950s the Swiss group, Arbeitsgemeinschaft für Osteosynthesefragen (AO, or Association for the Study of Internal Fixation) was formed and began to scientifically

study fracture healing and to standardize the implants and instruments used to repair fractures (5). The goal of internal fixation is to achieve a stable construct in which bone healing can occur, while still maintaining the function of the surrounding joints and soft tissues and enabling early functional use of the limb (1). The initial tenets of the AO group with regards to internal fixation of fractures included direct fracture exposure with anatomic reduction and rigid fixation of all fragments (2). Anatomic reduction refers to fracture fixation techniques in which the bony column has been perfectly reconstructed. The goal was to achieve bony union by primary bone healing, without callus formation (6).

Conventional plating techniques using plates such as Dynamic Compression Plates (DCPs) or Limited-Contact Dynamic Compression Plates (LC-DCPs) rely on the friction between the plate and the bone to generate stability (7). Achieving sufficient friction between the plate and bone requires extensive pre-contouring of the plate to the bone which can be time consuming and will weaken the plate. The contouring of the plate is also usually not perfect, which leads to incongruence between the bone surface and the plate. This incongruence may cause fragment shifting and fracture gaps to enlarge when the screws are tightened leading to primary loss of reduction (8).

There are a number of other suboptimal conditions created by this type of plating. The precise reduction which this type of plating technique requires usually necessitates an extensive surgical approach. Furthermore, the extent of plate-bone contact negatively affects periosteal blood flow, resulting in bone necrosis and resorption under the plate (9-11). These factors could potentially lead to an increased risk of delayed healing, non-union, infection, hardware failure, and bone re-fracture after plate removal (1,4,12-18).

Recent advances in the understanding of fracture healing have led to the concept of biological plate osteosynthesis in an effort to address these issues. Fixation techniques using this concept were developed as a means to reduce these problems and improve bone healing.

Biological fixation attempts to optimize the bone healing environment by placing less emphasis on absolute mechanical stabilization and more emphasis on preserving the circulation of the fracture site (1). Biological plate osteosynthesis utilizes indirect reduction to minimize the amount of surgical trauma. Restoration of function is still the primary objective in biological plate osteosynthesis. In this type of fixation the plate is anchored to the main proximal and distal fragments and bridges the smaller fractures between these fragments, avoiding the need for precise reduction and absolute stability in fracture repair (4).

Many of the complications due to conventional plating techniques derive from the compressive forces of the plate acting against the bone. These compressive forces can only be avoided when fracture stabilization does not rely on friction. This can be achieved with an angular-stable connection between the screw head and the force carrier (4). When used externally, such a device is known as an external skeletal fixator, and when used internally, it is known as an internal fixator. As opposed to the traditional plates and screws, these fixation systems firmly secure the screw to the plate and thus do not rely upon friction between the bone and plate for stability.

Since the screw effectively locks into the plate, angulation of the screw within the plate hole is not possible and the screw toggling is effectively prevented, even with only one point of purchase in the cortex. Close contact between the device and the bone is no

longer necessary and the need for contouring the plate is minimal. Gutwald et al showed that fracture reduction remains nearly unchanged following application of a locking plate system to the bone compared to a conventional plate (8). Furthermore, periosteal blood flow is better preserved and bone necrosis underneath the plate is minimal (19-22). Internal fixators allow stabilization of the main fracture fragments without direct manipulation of the secondary bone fragments which makes them ideal for comminuted and complex diaphyseal and metaphyseal fractures (10,23).

1.2.1 – Evolution of Internal Fixators

Various examples of internal fixators include the Zespul system (24), the point contact fixator (PC-Fix) (25), the less invasive stabilization system (LISS), and the locking compression plate (LCP). The PC-Fix was one of the first internal fixators used and achieves angular stability because the cone angle of the screw hole is smaller than the angle of the screw head. This centers the screw in the hole and creates a high amount of friction between the plate and screw, effectively locking the screw into the plate. This type of locking mechanism is known as a Morse taper. The underside of the PC-Fix is undercut such that the plate contacts the bone at points (See Figure 1) (25).

Figure 1. Demonstrating the undersurface of various plates. Top = conventional DCP, middle = LC-DCP, bottom = Point-contact fixator (PC-Fix) (1)

However, during clinical handling it was found that the connection between the screw head and the screw hole became nearly “cold-forged”, sometimes making it very difficult to remove screws (4). These issues led to the development of devices with a thread connection between the screw head and the screw hole, resulting in extreme angle-stable devices that did not require any contact between the force carrier and the bone. Two such systems are the less invasive stabilization system (LISS) and the locking compression plate (LCP).

The LISS™ is designed for use in the repair of fractures of the distal femur and proximal tibia (Figure 2). It is composed of anatomically pre-shaped plates that are inserted percutaneously and secured to the bone with screws placed with the assistance of an adaptable insertion guide (4). An angle-stable connection is achieved between the screw and the plate by means of an outer thread of the screw head that screws into an inner thread of the plate hole. Given that these threads need to match, the orientation of the screws cannot be changed.

Figure 2. Example of LISS for distal femur. (a) Lateral view of LISS on distal femur. (b) LISS for distal femur with a mounted insertion guide that, along with the trocar assembly, also serves as an aiming instrument (4).

While internal fixators such as the PC-Fix and LISS offer the many advantages of biological osteosynthesis and the possibilities of minimally invasive osteosynthesis, they also have disadvantages. The screws used with internal fixators such as the PC-Fix must be inserted perpendicular to the plate. Also the only way an internal fixator such as the PC-Fix can exert compression on a fracture site is through the use of a tensioning device (26). The ability to alter screw angle would be useful in several situations, such as for the fixation of a butterfly fragment with lag screws (Figure 3), to assist with fracture reduction or to avoid penetration of a joint.

Figure 3. An LCP plate using both conventional screws (non-locking) and locking screws (From: Wagner M: General Principles for the clinical use of the LCP. Injury 34:S-B31-S-B42, 2003.)

Additionally, it may be necessary to angle screws if the plate is not perfectly aligned with the long axis of the bone and the use of a locked screw at its fixed angle may result in inadequate cortical purchase (19).

In order to address the issues of compression and screw angle a new type of plate was developed: the locking compression plate (LCP). The LCP was developed as a combination between conventional osteosynthesis plates (DCP and LC-DCP) and internal fixators (PC-FIX, LISS). The LCP can be used in conventional technique, bridging technique (internal fixator principle), or a combination technique, using both locking and traditional non-locking screws (21).

The developers of the LCP aimed to maintain the mechanical properties of conventional bone plates so that LCPs would handle similarly to conventional plates and so that the LCP could be used either as a compression plate or as an internal fixator (20). Thus, the LCP combination hole was developed (Figure 4). This hole has a dynamic compression type slope on one end and locking threads on the other end. Studies have shown that the axial push-out strength and angular stability of a locking screw in a combination hole are still very high even with reduced captivation of the screw head (approximately 200° compared to 360° in an isolated locking hole) (26,27).

Figure 4. LCP combination hole without (left) and with (right) screws.
(Adapted from Frigg R: Development of the locking compression plate. Injury 34:S-B6-S-B10, 2003.)

1.3 – Mechanics of Conventional versus Locking Plates

1.3.1 – Conventional Plates

Traditional (non-locking) and locking constructs rely upon different principles for stability. Non-locking plates are reliant upon the friction generated between the plate and the bone for stability of the construct. This friction is created by the compression of the plate to the bone which is generated by the torque of the screws (28). The frictional force generated is equal to the product of force that presses the plate to the bone (the normal force) and the coefficient of friction between the plate and bone (29). Normal force or stress is a force that acts perpendicular to the face of an object. The normal force of a conventional bone-plate construct is equal to the sum of the torques on each screw (30) which is related to the resistance to shear stress of the bone trapped within the threads of the screw (2). Shear stress is stress that is applied tangential or parallel to the face of an object (29). Since non-locking plates are unable to prevent the screw from orienting in the direction of the applied force to the plate, the strength of the connection between the bone and the screws is only due to the shear stress between the screws and the bones (2). The weakest component of the plate-screw-bone construct is the screw-bone interface. The strength of this interface is equal to the stress resistance of the bone multiplied by the contact area between the screw and the bone (2).

1.3.2 – Locking Plates

Unlike traditional plates, locking plates are not reliant upon friction for construct stability. Locking screws result in an angle-stable device, in which the plate prevents the screws from orienting in the direction of the applied force to the plate, thus converting the

shear stress at the screw-bone interface to a compressive stress, which bone is better able to resist (2,28). In order for a locking construct to fail, all screws must cutout of the bone simultaneously, the bone must fracture, or the screws or plate must fail (31). The strength of a locking construct is equal to the sum of all the bone-screw interfaces (2).

1.4 – Biology of Conventional versus Locking Plates

There are numerous theoretical differences between the conventional plate osteosynthesis and biological plate osteosynthesis in the biology of fracture healing. These differences involve the interconnected areas of blood supply to the fracture site, implant stability and infections.

1.4.1 – Blood Supply to Fractures

In the case of a fracture, the blood supply to bone is not only damaged by the trauma that caused the fracture, but also by the surgery required to repair the fracture and the implants that are used to stabilize it (10). Due to the need for close contact between the bone and the plate in conventional plating, conventional bone plates can cause an area of decreased vascularity and bone necrosis directly under the plate. The bone must then go through a process of internal remodeling to re-vascularize these areas (4). This internal remodeling leads to early temporary porosity of the bone until new bone is laid down (1).

Gautier et al placed conventional plates on sheep tibia and demonstrated large areas of non-perfused cortex and bone necrosis under the plates four weeks post implantation, and that the area of non-perfused cortex corresponded with the area of

plate-bone contact. At ten weeks post implantation, resorption cavities contributing to a temporary porosity could be seen histologically (11).

Since the degree of disruption of the bone vascularity seemed to correlate with the area of plate-bone contact, a decrease in the amount of plate-bone contact should also reduce vascular disturbances. Initially this was done by changing the shape of the conventional DCP, leading to the development of the limited-contact dynamic compression plate (LC-DCP). In this plate, areas on the underside of the plate were undercut, resulting in a reduced cortical contact between the bone and the plate while also creating an implant with even stiffness. While the LC-DCP reduces the contact surface by more than 50% compared to the conventional DCP it is still reliant upon compression between the plate and the bone to generate friction and achieve stability (4).

Zhao et al demonstrated the effects of reducing the amount of contact between the bone and the stabilization device by comparing the effects on the cortical perfusion of a goat tibia plated with a point-contact dynamic compression plate (PC-DCP) versus a dynamic compression plate (DCP). This study demonstrated outer cortical bone ischemia under the DCP starting at the day after surgery and continuing until 12 weeks postoperatively. In the tibia that was repaired with the PC-DCP, cortical bone ischemia only occurred in the periphery of the screw holes and at the contact sites of the fixator on the bone the day after surgery and resolved by two week postoperatively, leading the authors to conclude that the PC-DCP was less detrimental to the local bone blood circulation than the DCP (32). Despite a significant improvement offered by the "Limited Contact" plates, a certain amount of contact between the bone and plate is still required in order to generate sufficient friction and stability. This can be avoided by

using angle stable devices that create a stable angle between the implant going into the bone and the load bearing or sharing device.

Implants that are contoured exactly against the bone could also have a higher incidence of re-fracture. Because of the decreased bone vascularity under the plate, the fracture area directly underneath the plate may show decreased or delayed healing due to the damaged blood supply. If the plate is removed, this area of unhealed cortex could create a notch effect and could act as a stress riser, causing the bone to re-fracture (1).

1.4.2 – Infection

The reduced vascularity and increased porosity to the bone underlying conventional plates does not usually have any clinical consequences since the plate protects the temporarily weakened bone. However, if the bone is exposed to further compromise, such as an infection, the various pores resulting from the decreased vasculature and infection could coalesce and lead to the formation of a sequestrum (1). By diminishing the area of contact between the plate and bone, and thus theoretically improving the vascularity and decreasing the porosity of the bone underlying the plate, the likelihood of infection should also be decreased. This was demonstrated in two studies in which either a titanium DCP or a point contact fixator (PC-Fix) was attached to a rabbit tibia and then injected with *Staphylococcus aureus*. The rabbits with the PC-Fix had a significantly lower infection rate than the ones with the DCP (33,34). However, despite the experimental evidence, a large prospective clinical study on the use of PC-Fixators for fracture fixation resulted in an overall infection rate of 1.1% (13/1,172) (34);

similar to infection rates previously reported following fracture repair with conventional plates (35-37).

1.4.3 – Implant Stability

Conventional plates allow the screw to be inserted at a variety of angles within each plate hole. Although the ability to select screw angle is convenient for the surgeon, it is also one of the major drawbacks of conventional plating. Each screw can freely pivot in its hole and therefore must rely on two points of purchase (both cortices) to maintain its stability. Any loss of cortical purchase in either cortex, such as that due to bone resorption or micro-fractures, will result in the loss of one of the two purchase points. This will allow the screw to toggle in the plate hole (also known as the “windshield wiper effect”) causing the loss of stability (2).

If a plate-bone construct using a conventional plate is loaded axially, the forces are resisted by the frictional force between the plate and bone. The conventional screws can toggle within the plate if the axial forces exceed the frictional forces. The plate will then shift and pull on the screw head, the screw will remain in place in the trans-cortex, but toggle in the cis-cortex. The screw will radially compact the bone in the cis-cortex as it toggles. This bone compaction will continue to occur until the remaining bone rim is once again strong enough to resist further compression (Figure 5) (38).

Figure 5. When axial load exceeds the friction force between the plate and the bone, the plate will shift the head of the screw. The screw will remain in place in the trans-cortex, but toggle in the cis-cortex. The screw will radially compact the bone in the cis-cortex (area of high stress near cortex) until this rim of bone in the cis-cortex is strong enough to resist further compression (2).

If the forces acting upon the construct are greater than the coefficient of friction generated between the bone and the plate, the plate may start shifting, causing the screws to pull out of the bone. As the screw pulls out, toggling will occur and the construct loses stability. The ability of non-locking constructs to withstand externally applied stresses without failure should theoretically be equal to the resistance to shear stress of the bone trapped within the threads of the screw since non-locking plates are unable to prevent the screw from orienting in the direction of the applied force to the plate (2). The same is true of bone-plate constructs loaded in bending. As long as the screws can resist pull-out, then the bending forces are converted into shear stresses between the plate and bone (Figure 6).

Figure 6. Comparison of locking (B) and non-locking (C) plates placed under bending loads (2).

The difference in failure mechanisms of conventional versus locking plates has been observed in several studies. In a study comparing conventional versus locking plates for the fixation of intra-articular human cadaveric osteopenic calcaneus fractures the locking plates failed by bending of the plate with the screws maintaining their fixed plate-screw angle. On the other hand the conventional plates remained intact while the screws toggled in their screw holes (39).

Another study compared non-locking and locking plates tested cyclically in axial compression and torsion and then to failure in these modes. It was noted that failure for the compression plate resulted from angulation of the screw in the screw hole, while

failure for the locking system was the result of bending of the plate or breakage of the distal screw (40).

Similar results were obtained in a study comparing locking and non-locking plates in a synthetic, osteoporotic humerus model tested in cyclic torsion. It was noted that for all the specimens in the non-locking group, the construct became unstable due to pullout and gross toggle of the screws, while this toggling did not occur in any of the specimens in the locking or hybrid group (41).

Although, with angle stable constructs it is not essential for the plate to contact the bone, the distance between the bone and the locking plate still needs to be somewhat controlled. Ahmad et al, 2007, demonstrated that increasing the distance between the plate and the bone from 2 mm to 5 mm lead to a significantly lower load to failure when tested in compression. Increasing the distance also resulted in increased rotational deformity for a given torque, however, it was not indicated whether this difference was statistically significant. During cyclic testing, significantly increased displacement was observed for any given load as the distance between the bone and plate increased to 5 mm and it had a significantly lower load to failure compared to the other groups (42). These results can be explained by the fact that, as the distance between the plate and bone increase, the working length of the screw increases and produces a greater lever arm effect during biomechanical testing.

1.5 – Biomechanical Comparison of Locking and Non-Locking Implants

One of the purported benefits of angle-stable devices is their increased stability compared to conventional devices. Numerous studies have compared these two types of

devices. LCPs have been shown to perform similarly to or better than conventional plates when tested biomechanically (22,31,43-45). In order to better understand these studies, one should first understand the concepts of force and displacement.

1.5.1 – Force and Displacement

Force and displacement are used to calculate the strength and stiffness of an object. Force is a vector quantity, with magnitude and direction, which will cause an object to change its velocity, size, or shape. Force is measured in Newtons, which is defined as the force required to accelerate a one kilogram (kg) mass at a rate of one meter (m) per second (s) per second (kg/m/s^2). When a force or a load is applied to an object it can cause that object to displace. In materials testing, the forces are measured by a load cell and the movement of the load crosshead of the load application system coincides with the displacement of the material being tested. The relationship between the amount of force required to achieve a certain amount of displacement is described on a force-displacement graph (Figure 7).

Figure 7. Typical load-displacement curve: The slope of the linear portion is equal to the stiffness of the material being tested. The elastic portion of the curve is the linear portion and the object will not undergo permanent deformation in this portion of curve. The object will undergo permanent deformation in the plastic portion of the curve. The yield load is the point at which the object transitions between elastic and plastic deformation. The area under the curve is equal to the work or energy required to achieve that degree of displacement.

A number of properties about the object being tested can be obtained from the load-deformation curve. Initially there is a preloading or engaging portion of the curve where slack in the testing system is being taken up. This is followed by the linear portion of the curve in which the displacement increases relatively linearly to increasing loads (Hooke's Law). The slope of this portion of the curve is equivalent to the stiffness or rigidity of the object. This portion of the curve is also known as the elastic portion of the curve. Within this portion of the graph, the loaded object will return to its original configuration after the load is removed. The yield point is the point at which the curve transitions from

elastic to plastic deformation. After this point the sample being tested will not return to its original configuration after the load is removed. In this portion of the curve, smaller load increases produce greater deformation compared to similar load increases in the elastic portion of the curve. Catastrophic failure occurs at the fracture, or failure point. The area under the curve is equal to the energy absorbed by the sample at a specific load or deformation. All of these parameters are known as the structural properties of the sample being tested. The parameters obtained from load-displacement curves are going to be affected by the geometry and quantity of the sample being tested (46).

1.5.2 – Long Bones, Fractures, and Torsion

Bones are normally subject to compressive, tensile, bending, and torsional forces in daily life. Torsion can be defined as a force applied to an object that causes it to rotate or twist about its long axis (29). This force is known as torque and is equal to the applied force multiplied by the lever arm (47) (Figure 8).

Figure 8. Diagram of and formula for torque: Torque is the amount of force applied to an object causing it to rotate about a pivot point. It is equal to the product of the lever arm and the perpendicular component of the force.

As a bone is twisted, it is subjected to shear strains and stresses in the horizontal and vertical planes. It is also subjected to tensile stresses that are maximal at a 45° angle to the axis of torsion and fractures will often occur at this angle (29). These forces are proportional to the distance from the center of rotation while the object's ability to resist torsion is for long bones, approximately the radius raised to the fourth power (29). Numerous factors influence the strength of bone in torsion, including size of the defect, cortical thickness, cortical circumference, bone quality, and cross-sectional geometry (48,49).

Long bones are commonly loaded in torsion, both in normal activity and trauma (50). When long bone fractures are repaired, the surgeon must account for and counteract these forces to allow the bone to heal. The surgeon must also consider the impact of his repair method on the torsional properties of the bone as screw holes have been shown to significantly affect the torsional properties of bone (50). An experiment studied the effects of screw holes on the torsional strength of cadaveric sheep tibia. Bones with monocortical screw holes were 21.6% weaker in torsion than intact bones. This difference amounted to 31.4% with bicortical holes. Each of these differences was statistically significant (51). This experiment confirmed that the screw holes act as points of stress concentration in the bone. As the hole is drilled into the bone, it decreases the cross sectional area of the bone at that location, resulting in stress concentration (Figure 9) (29).

Figure 9. Demonstration of stress concentration around a hole: The lines on the object represent lines of force. Since the total load applied to an object must be the same for any cross section any change in the cross sectional area will concentrate these force lines and lead to an area of stress concentration as demonstrated by the concentration of the density of the lines of force (29).

1.5.3 – Bone Plates and Torsion

Several factors affect the behavior of bone plates in torsion. The length of the plate, number of screws, position and type of screws and the distance between the plate and the bone all affect torsional rigidity. For fractures that have been plated, it appears as though the number of screws is more important than the plate length for torsional stability (52). End bicortical screws (the screws furthest from the fracture) also seem to significantly increase the torsional strength of bone-plate constructs compared to end monocortical screws (53).

Since the screws lock into the plate in LCP constructs only monocortical screws are needed theoretically. However, it has been shown that bicortical locked screws are significantly stronger than monocortical locked screws in all modes of loading (31). Specifically, in torsional loading the working length of the screw is of critical importance, thus bicortical screws are generally recommended (54).

Stoffel showed that for a twelve-hole LCP, axial stiffness and torsional rigidity were affected by the distance of the first screw to the fracture site (the working distance) and that torsional rigidity increased with an increasing number of screws up to four screws per fragment (55). It was also found that the stiffness under axial load decreased for the LCP when additional screws were placed closer to the plate ends (55). The gap between an LCP and the bone also affects the torsional rigidity and axial stiffness in an inverse relationship (31).

1.5.4 – Mechanical Testing of Locking Plates Compared to Non-Locking Plates

There are very few studies comparing the use of locking and non-locking screws in LCPs. However, since LCPs were designed to have the same mechanical properties as LC-DCPs, then it should theoretically be possible to use studies comparing LCPs secured with all locking screws to LC-DCPs secured with all non-locking screws as surrogates for studies comparing LCPs secured with locking screws to LCPs secured with non-locking screws.

DeTora and Kraus tested the differences in bending strength between the Locking Compression Plate (LCP), the Limited Contact-Dynamic Compression Plate (LC-DCP), 3.5 mm Broad LC-DCP, and String of Pearls (SOP). Each implant was tested

independent of any bone in cyclical four point bending. Although significant differences were observed between these implants, they noted that the bending strength and stiffness of the LCP were similar to the LC-DCP (56). This finding was confirmed in Aguila et al who compared 8-hole 3.5 mm LCP and LC-DCP constructs secured to canine femurs with either six locking or six non-locking screws, respectively, under bending and torsion. It was found again that LCPs were, for the most part, biomechanically similar to LC-DCPs. Although the bending stiffness of the different plates in this study was shown to be similar, the LCP constructs had significantly higher gap stiffness in lateral-medial bending. Although no significant difference in the change in the structural stiffness were observed between the two types of plates throughout cyclical torsional testing, the LCP constructs failed at significantly higher torsion than the LC-DCP constructs. Despite these differences, the authors concluded that the LCP was biomechanically similar to the LC-DCP in canine diaphyseal bone (22). These two studies show that although the LCPs and LC-DCPs tested by themselves have similar biomechanical properties, some difference may be detected if those plates are tested as bone-plate constructs. These differences are therefore the result of the differences in the screws used to secure these two systems. Thus, one could theorize that if a LCP secured with all locking screws was compared to a LCP secured with all non-locking (conventional) screws, biomechanical differences similar to Aguila's study would have been found.

Ahmad et al compared the effect of the distance between the plate and the bone on the biomechanical properties of the constructs. In their experiment, DCPs were compared to LCPs applied flush to the bone, at 2 mm and at 5 mm from the bone using a synthetic humeral model. The constructs were tested in cyclical compression,

compression to failure, cyclical torque and torque to failure. Although the effects of moving the plate further off the bone have been discussed earlier, the authors found that when the LCP was either flush or at 2 mm off the bone their biomechanical properties were comparable to the properties of the DCP (42).

Although these studies support the fact that the LCP have relatively similar mechanical properties as the LC-DCPs, other studies have shown that, in some situations, LCPs may have superior yield strength than other types of plates. Florin et al compared 4.5 mm broad DCPs, 5.5 mm broad LC-DCPs, 4.5/5.0 mm broad LCPs, and 4.5/5.0 mm clamp-rod fixators for the repair of a 20° oblique long-bone equine fracture model using CanevasitTM tubes. The tubes were double plated in an orthogonal configuration and tested in four-point bending to failure. Although no significant differences in stiffness were found between the groups at low load. They found that the LCP constructs were significantly stiffer than the other constructs during the last 150 Nm before the point of maximum moment before failure. They also found that while the ultimate strength of the LCP group was significantly lower than the other groups, the yield strength of the LCP was significantly higher than the other groups. Since the yield strength represents the transition between the elastic and plastic portions of the load-deformation curve, it also represents a theoretically more clinically useful point of failure than the ultimate strength (57).

Several other comparisons between locking plates and various other types of plates are available in the literature. However, inferences as to the superiority of the concept of locking systems over the non-locking systems is generally not possible as the implants used to represent each group have different mechanical properties.

Lehmann et al compared locking versus non-locking systems by simulating complete disco-ligamentous instability at the cervical vertebral 4 – 5 level, and stabilizing the vertebrae anteriorly with either locking or non-locking plates and monocortical screws. The constructs were then tested in flexion, bending, extension, and rotation by applying “physiologically” realistic loads for a total of three cycles for each testing phase. No significant difference was found between the average range of motion between the locking and non-locking groups. The authors concluded that the locking plates did not provide any significant advantages over the non-locking plates (58). This finding should be interpreted with caution however, as their testing only consisted of three cycles of physiologic loads, had they loaded to failure, or cyclically loaded for more cycles they may have found significant differences between the two plating systems. Furthermore, the two plates used for comparison were of different sizes, as were the screws. The authors of this study did not mention how these plates compare biomechanically to each other independent of mode of attachment.

Similarly, Chiodo et al did not find any significant difference between a locking and a conventional 2-mm mandibular plate used to repair a mandibular fracture with monocortical screws, loaded in four-point bending in a bovine rib model (45). On the other hand, Boswell et al found that locking plates were significantly stiffer (33%) and had a significantly higher load at failure (91%) than the standard T-plates when comparing locking and compression T-plate constructs applied dorsally to distal radial opening wedge osteotomies on cadaveric human radii tested in compression to failure. All of the plates failed by bending through an empty hole at the osteotomy site. No screw loosening, breaking, or cutting out of the bone was noted. The authors suggest that the

mode of failure seen in their study was due to the fact that they were testing to failure in a single event and did not cycle the constructs (59).

Weinstein et al examined the torsional resistance of three-part proximal human cadaveric humeral fractures repaired using locking plates versus angled blade plates. They found that the mean initial torsional stiffness of the locking plates was significantly higher than the blade plates. The blade plates had greater deformation than the locking plate during cyclic loading (60).

Stoffel et al performed a study comparing the biomechanical stability of bone-plate constructs using a conventional plate, a locking system, and a combination plate in synthetic femurs with a segmental defect and in cadaveric human femurs with a distal intra-articular fracture. Locking and non-locking screws were used in the combination plate. The constructs were tested cyclically in axial compression and torsion and then tested to failure in these modes.

The results were fairly similar for the two fracture models. The stiffness under axial compression was not significantly different, but the locking group had a significantly lower stiffness in torsion compared to the other methods of fixation. However, the locking and combination groups had significantly less plastic deformation under axial load compared to the conventional plate. For the cadaveric specimens, under torsion, the conventional and combination plates showed significantly less plastic deformation compared to the locking system. Also the combination plate withstood a significantly higher load to torsion in the shaft fracture model than all the other constructs. The load to failure in axial compression for the conventional plate in the cadaveric specimens was significantly lower than the combination plate. Some of the

differences between the different constructs seen in this study could be explained by the material differences between the plates being tested, stiffer steel for the conventional plates versus titanium for the locking plate and the different shapes of the plates (40).

1.5.5 –Mechanical Testing of Locking Plates and Non-Locking Plates in Decreased Quality Bones

Lill et al compared five different methods of fixation of cadaveric human proximal humeral fractures, including: a humeral T-plate, cross-screws, an un-reamed proximal humerus nail with spiral blade, a Synclaw™ Proximal Humerus Nail and a locking compression plate. They determined the stiffness of the implants in 0.5 mm axial compression, 4° of torsion, and varus bending by loading the constructs eccentrically with an axial compressive displacement of 4 mm. Finally, the constructs were cyclically tested under varus bending. During the cyclic loading they recorded the maximum load applied during the first cycle (the initial load), the load level at 300 cycles, and the load reduction between the 1st and 1000th cycles. They found that the locking plate group had a low-intermediate stiffness compared with the other implants in static loading. In the cyclic testing, the locking plate had a lower initial load compared to some of the other plates, but had a significantly lower load reduction compared to the other implants. The load reduction was equal to the percentage difference between the load at the first cycle and the load at the last cycle and was an indicator of implant loosening or migration throughout the cycles. The authors hypothesized that this lower load reduction was due to the fact that the locking plate's lower initial stiffness reduced the peak stresses at the implant-bone interface leading to less loosening of the implants, making them more

suited for use in osteoporotic fractures than stiffer implants (61). This study compared different fixation devices, all of different biomechanical properties, so while it is useful for comparing these specific devices, it is difficult to make useful conclusions regarding locking devices compared to non-locking devices from this study.

In another study, blade plates were compared to locking plates for the fixation of proximal humeral fractures in both cadaveric and synthetic human humeri. Blade plates are shaped with a right angle, with one end having a chisel-shape to allow it to be driven into the bone and the other end having screw holes to secure it to the bone. The cadaveric constructs were tested in cyclic bending and torsion, whereas the synthetic humeri were only tested in cyclic torsion. For the cadaveric samples, significant differences in favour of the locking plates were observed for angular displacement in torsion. In bending, less median vertical displacement was observed in the locking constructs compared to the blade plates, although this difference was not statistically significant ($p = 0.079$). Interestingly, for the synthetic constructs no significant difference was found between the locking and blade plate constructs ($p = 0.1$). There were only 0.2 degrees of variation for the median amplitude of rotation for each of the synthetic constructs, indicating that the constructs maintained stability throughout cyclic torsional testing. Significant differences were observed between the cadaveric and synthetic blade plate groups. In this experiment, synthetic bones were used to simulate young healthy bone. Since no significant difference was seen between the synthetic and cadaveric locking constructs, while significant difference were observed with the blade plate, the authors concluded that locking constructs would likely provide similar fixation in bones regardless of the bone quality (62). It must be noted, however, that the group

numbers for each of their tests was fairly small, with only 6 humeri being used in each group in each failure modality, and that the bone quality of the synthetic humeri was not directly compared to that of the cadaveric specimens.

A number of studies have tested locking plates in osteoporotic bone, osteoporotic bone models or in bone with varying bone mineral density (BMD). Snow et al furthered the research into using locking plates in decreased quality bone by comparing LCPs and DCPs for the repair of a fracture in a synthetic osteoporotic bone model. All of the specimens in the DCP group failed initial static testing in axial compression, while all specimens in the LCP group survived fatigue testing without evidence of gross loosening, leading the authors to conclude that the LCP was mechanically superior to the DCP when used as a bridging plate in an osteoporotic bone model and tested in axial compression (63).

Kim et al compared conventional plates to locking plates for repair of distal fibula fractures in osteoporotic bone. No significant difference was found between the stiffness of the two constructs. However, they did find that the torque to failure of the conventional plate was dependent on bone mineral density (BMD) while for the locking plate it was independent of the BMD. This should be interpreted with some caution however, as the correlation between BMD and torque to failure was overall weak ($r^2 = 0.67$) (Figure 10).

Figure 10. Load to failure as a function of bone mineral density (64).

However, if only the bones with a BMD below 0.5 g/cm^3 are considered it can be seen that the locking plates consistently failed at a higher load than the conventional plates. The authors suggest that a biomechanical advantage of locking plates could be demonstrated if only specimens with a BMD of less than 0.5 g/cm^3 were examined (64).

Another study compared conventional versus locking plates for fixation of intra-articular human cadaveric osteopenic calcaneus fractures. Once the fractures were repaired, the specimens were cyclically loaded through the tibia and then loaded to failure. The mean irreversible deformation was significantly less and the load to failure was significantly higher for the locking plates than for the conventional plates. However, the ultimate deformation and work to failure were not significantly different between the

two plates. They also found that a low BMD was significantly correlated with an increased irreversible deformation in the conventional plate group, but not in the locking plate group, indicating that irreversible deformation of a locking plate construct is independent of the BMD (39). This supports the theory that locking plate constructs maintain the same strength and stiffness regardless of the BMD.

1.5.6 – Biomechanical testing of Locking Screws vs. Conventional Screws in Locking Plates

Numerous papers have examined the biomechanical effects of using locking versus non-locking screws in a locking plate. Weiss et al examined locked versus unlocked plating with respect to plate length in a cadaveric ulna fracture model. Ten-hole LCPs were tested with either four locking or four non-locking screws in a near-far configuration (Figure 11).

Figure 11. Groups used in Weiss et al, 2008 study on locking versus unlocked plating with respect to plate length. A. 10-hole plate with four unlocked screws in near-far configuration. B. 10-hole plate with four locked screws in near-far configuration (65).

No significant differences were observed between the two constructs for the bending stiffness, torsional stiffness, or yield strength (65).

Stoffel et al performed a study comparing the biomechanical stability of bone-plate constructs using a conventional plate, a locking system, and a combination plate in synthetic femurs with a segmental defect and in cadaveric femurs with an intra-articular

distal femur fracture. Locking and non-locking screws were used in the combination plate. The constructs were tested cyclically in axial compression and torsion then tested to failure in these modes.

The combination plate was tested as a conventional, combination, and locking plate in the synthetic specimens. It was found that the stiffness in axial compression between the three methods was not significantly different. However, the stiffness in torsion when the combination plate was used as an internal fixator with locking screws was significantly less than the other two methods. There was significantly more plastic deformation under axial compression of the construct when the plate was used as a conventional plate compared to the other two groups. However, no significant differences were seen in plastic deformation between the different groups under torsional loading. Interestingly, the torsional load to failure was significantly higher for the combination group compared to the other two groups, however, no explanation was advanced for these differences (40).

The effect of using non-locking screws in a locking plate were also tested by Leitner et al on a tibial plateau fragment stabilized by a locking tibial plateau leveling osteotomy (TPLO) plate in canine cadaveric tibia. The TPLO plate was a 6-hole, 3.5 mm plate that is anatomically pre-contoured for medial aspect of the proximal tibia (Figure 12).

Figure 12. Synthes locking tibial plateau leveling osteotomy plate for left limb: The three proximal holes are round and threaded and can accept either locking or non-locking screws. The shaft of the plate includes two dynamic compression holes and one combination hole that can be used either in compression or locking fashion.

The three proximal holes are round and threaded and can accept either locking or non-locking screws. The shaft of the plate contains two dynamic compression holes and a central combination hole that accepts either locking or non-locking screws. In the locking group, three locking screws were used in the proximal holes whereas in the non-locking group, non-locking screws were used instead. Non-locking screws were used in holes in the shaft of the plate in all groups. The constructs were tested in axial compression and fragment motion was tracked during plate placement and after cyclic compression. Significantly greater rotation of the fragment about the sawing axis as well as significantly more displacement of the tibial plateau fragment towards the plate was observed for the non-locking group during application of the plate than for the locking group (1.6 +/- 0.6 mm for the non-locking group versus 0.1 +/- 0.8 mm for the locking group). No significant differences in fragment motion were however detected between the groups after the cyclic loading and there were no significant difference between groups in cycles to failure or construct stiffness. The authors believed that the translation of the fragment towards the plate in the non-locking group (1.6 mm mean) would have negligible effects on joint morbidity and that the translation likely occurred because the plate was not contoured perfectly to the shape of the bone (66). One reason that no

significant movement of the tibial plateau fragment was found after cyclic loading between the locking and non-locking groups in this study could be explained by the fact that in both groups, the osteotomy site was compressed, thus the bone was sharing a major portion of the axial loading.

Amato et al compared monocortical locking screws to bicortical non-locking screws in a 2.4 mm UniLOCK™ Reconstruction Plate for fixation of an acetabular osteotomy in canine cadaveric hemipelves. No significant differences were found between osteotomies stabilized with locking monocortical screw fixation or standard bicortical screw fixation with respect to joint congruity, displacement of the fracture gap after cyclic loading, construct stiffness, or ultimate load to failure (67). It should be noted that in this study, the locking screws were monocortical while the non-locking screws were bicortical. The use of bicortical locking screws could have potentially created a difference. Also both locking and non-locking plates were perfectly contoured to the bone, which would not be mandatory if locking screws were used.

The effect of using mono- versus bicortical locking screws was tested by Fulkerson et al. Eight-hole LCPs on a synthetic bone model with a segmental defect were tested in cyclic axial loading and three-point bending. Three consecutive screws on each side, either all monocortical locking, all bicortical locking, or all bicortical non-locking were tested and showed that significantly more displacement of the fragments occurred with bicortical non-locking screws than with bicortical locking screws. Also, bicortical locking constructs withstood significantly more cycles to failure than all other constructs. A significant difference between the number of cycles to failure was

observed between the bicortical non-locking group and the monocortical locking group (31).

Gardner et al. used a synthetic, osteoporotic humerus bone model to test LCP constructs of all locking screws, all non-locking screws, and a hybrid combination of locking and non-locking screws in cyclic torsion testing. The non-locking constructs were significantly less stiff than both the locking and hybrid constructs initially and they retained significantly less stiffness compared with the initial stiffness at each cyclic interval (41).

O'Toole performed a similar study comparing the use of locking or non-locking screws in LCPs for the reconstruction of a comminuted mid-shaft humeral fracture using both synthetic and cadaveric bone. The constructs were tested in torsion, four-point bending and axial compression within the elastic behavior limits of the construct. The constructs in the cadaveric group were then tested under cyclic axial loading for a total of 90,000 cycles. All constructs were then loaded in axial compression to failure. Unlike Gardner, O'Toole found no significant differences between the stiffness of the two groups for both the synthetic and cadaveric bones. The non-locking group in the synthetic bone failed at a significantly higher force than the locking group. For the cadaveric bone constructs, no differences were seen between the groups after cyclic loading or in ultimate failure force (68).

Although Gardner and O'Toole used similar constructs, Gardner found that locking constructs were significantly stiffer than non-locking constructs while O'Toole did not find any significant differences between the stiffness of these two groups. A possible explanation for the differences found between the two studies may result from

the fact that Gardner overdrilled the screw holes to simulate osteoporotic bone, while O'Toole did not. Gardner supported his model for osteoporosis by claiming that it had been used in previous studies (69,70) and that it had been validated by a pilot study in their laboratory. O'Toole argued that this model had not been adequately scientifically validated and that by drilling a hole 0.3 mm less than the diameter of the screw resulted in a hole that is 31% larger than the recommended screw hole size for the non-locking screws, but only 9% larger than the size recommended for the locking screws. This discrepancy could potentially put the non-locking screws at a mechanical disadvantage and could explain the difference between the locking and non-locking groups found in Gardner's study (41,68).

1.5.7 – Biomechanical Testing of a Combination of Locking and Non-Locking Screws in Locking Plates

Only a few studies have examined the effects of using both locking and non-locking screws in the same plate. Gardner et al. used a synthetic, osteoporotic humerus bone model to test constructs of all locking screws, all non-locking screws, and a hybrid combination of locking and non-locking screws in cyclic torsion testing. The hybrid combination consisted of using a non-locking screw in the hole nearest the osteotomy site and two locking screws distally in each fragment. They found that the locking and hybrid constructs were not significantly different from each other. The non-locking constructs were significantly less stiff than both the locking and hybrid constructs initially and they retained significantly less stiffness compared with the initial stiffness at each cyclic interval (41).

Roberts et al examined the biomechanical effects of using various fixations on a composite radius fracture model. They compared four constructs: three non-locking bicortical screws on each side, three locking monocortical screws on each side, two monocortical locking screws near the fracture with one bicortical non-locking distant from the fracture on each side, or two monocortical locking screws near the fracture and one bicortical locking screw distant from the fracture on each side. During torsional testing, the monocortical locking group was significantly weaker than the other three groups and the replacement of the end screw of a locking monocortical construct with a bicortical screw significantly increased the construct stiffness. Both of the hybrid constructs provided equivalent stability in torsion compared to the non-locking bicortical group, while the locked hybrid group was significantly stiffer than the other three groups in the anteroposterior bending. The authors concluded that the major factor in determining torsional stability in this experiment was likely the increased working length of the bicortical screws, rather than using locking or non-locking screws (71).

Sod et al compared LCP-combination fixation versus LC-DCP fixation of osteotomized equine third metacarpal bones. For the LCP fixation, a combination method was used with four 5.0 mm locking screws, two 5.5 mm cortical screws, and two 4.5 mm cortical screws. They used four 5.5 mm and four 4.5 mm cortical screws to secure the LC-DCP. No rationale was given for using the combination method for the LCP. They found that for 4-point bending, single cycle to failure, the mean yield load, yield bending moment, composite rigidity, failure load and failure bending moment of the LCP-combination fixation were significantly greater than those of the LC-DCP fixation. Under torsional testing the mean yield load, mean failure load and mean composite

rigidity for the LCP-combination fixation were significantly greater than the LC-DCP fixation. The authors note that the 4.5 mm LCP has the same dimensions as the 4.5 mm LC-DCP, so if just the plates were mechanically tested they should not be significantly different. Thus, the differences found in this study were due to the different fixation methods (72).

1.6 – Clinical Studies of Locking Plates

There have been a number of studies examining the clinical use of locking systems. However, very few papers prospectively compare locking to non-locking systems. Clinically, the superiority of one system over another does not appear evident. It must be noted that complication rates in fracture fixation are relatively low; therefore, the detection of a statistically significant difference between the two fixation methods may require thousands of cases. Papers that address the use of internal fixators clinically are reviewed in the following section.

1.6.1 – Clinical Studies of Locking Plates in Human Medicine

Drobtz and Kutscha-Lissberg report the clinical experience after two years using a new locking system for the treatment of distal radial fractures in humans. Thirty-nine out of the 50 fractures had good to excellent outcomes, however, loss of reduction was observed during follow-up radiographs of 21 fractures. Seventeen of 21 only had a maximum loss of correction of 3 mm or 10° of palmar slope while in the remaining four fractures, the loss of correction was more severe and due to significant loss of angular stability caused by implant failure due to severe osteoporosis in two and severe

comminution in two. Two cases of infections and a number of patients with post operative rupture of flexor tendons were also reported, for an overall complication rate of 32%. Notwithstanding these complications the authors still suggest that the locking plate system could be used for the fixation of distal radial fractures, however, modifications to the plate or removal of the plate may be required to prevent the rupture of the flexor tendons (73).

Another study on the use of a locking plate on the palmar surface for the treatment of unstable dorsal dislocated radius fractures in humans reported good results. In the 19 patients reported, all fractures were radiographically united at 6 months and no hardware failure, or screw loosening was noted. A mild loss (less than 2°) of palmar tilt and radial inclination was observed, which was not noted to affect patients clinically. No other complications were reported (74) .

Koshimune et al compared locking and non-locking palmar plating for unstable distal radial fractures. There were 22 people in the locking plate group and 31 people in the non-locking plate group. There were no significant differences between the groups pre-operatively. In this study, all the fractures healed following either method of fixation and the only significant difference found was a higher range of motion in flexion for patients treated with locking plates compared to the non-locking group (59° vs 49°) (75). No explanation is given for this difference. It may have been a Type I error and even if it were true, it would likely not be that clinically significant.

Sommer et al examined the use of LCPs for treatment of fractures in humans in a multicenter study. Follow-up was available for 151 fractures. Eighty-six percent of the fractures available for follow-up demonstrated uneventful and complete healing, twelve

months after fixation. Twenty-seven adverse events were documented in 19 patients (Table 1).

Adverse Event	No. of cases
Delayed or non-union	5
Loosening of implant and loss of reduction	4
Breakage of implant	4
Refracture	4
Infection	2
Postoperative nerve palsy	2
Implant-related pain	2
Osteonecrosis	2
Loss of reduction	1
Loosening of the implant	1
Total	27

Table 1. Adverse effects in a study on the first clinical results of the Locking Compression Plate (76)

The authors contended that many of these adverse effects arose due to technical errors of application, such as not pre-drilling through the screw-in drill-sleeve, or incorrect choice of plate. If the screw-in drill-sleeve is not used, it could result in the screw not being oriented in the screw hole within the tolerance limits (86-94°) and cross-threading. Notwithstanding these complications, the authors concluded that the LCP system was a valuable addition in the armament of fracture fixation (76).

The point contact fixator was compared with the limited contact dynamic compression plate in a prospective, randomized trial for forearm fractures in humans. There were 45 cases treated with the PC-Fix and 47 cases treated with the LC-DCP. All fractures healed with no significant differences between the groups for operative time, time to union, callus formation, range of motion, or pain. One deep infection and one re-fracture occurred in each group. Both re-fractures occurred through screw tracks after the

plate had been removed. One of the purported benefits of the PC-Fix device is that the use of monocortical screws reduces the risk of re-fracture upon removal of the implant; however this study demonstrates that in spite of the use of monocortical screws, fracture through a screw hole can still occur. Screw pullout was also observed in one fracture treated with a PC-Fix. Two issues with this study are that the fracture subtypes were not matched and the patients were not matched to account for the possible effects of factors such as bone mineral density. A larger, multicenter trial involving a large number of patients would be needed to conduct such a study. Despite these limitations, the authors surmise that the two implants appear to be equally effective for treatment of forearm fractures (77). This study compares conventional osteosynthesis with the theory of biological plate osteosynthesis. And while the PC-Fix is a different fixation device than the LCP, they are based on the same principles of biological osteosynthesis, so the results of this study may be inferred to apply to LCPs as well.

A study examined the results of 29 proximal humeral fractures in 28 human patients that were treated with a Locking Proximal Humerus Plate. A deep infection occurred in one patient. One plate broke at the level of the fracture four weeks postoperatively. Varus mal-alignment was observed in three patients, with seven screws losing their hold on the plate and five screws cutting through the humeral head. The screws likely lost their hold on the plate because they were not inserted at the correct angle and thus were cross threaded. The authors theorized that the screws cut through the humeral head due to the rigidity of the implant and the fact that the bone was osteoporotic. The authors noted the importance of correct positioning of the plate prior to screw placement to avoid the risk of cross-threading the screws. Overall though, the

authors believe that the clinical results are comparable with similar studies using different implants (78).

One of the potential benefits of locking systems, is the ability to use them in a minimally invasive manner. Hazarika et al report 20 cases of distal tibial fractures in humans that were managed with placement of a LCP in a minimally invasive manner with a minimum follow-up of 12 months. Minimally invasive plate osteosynthesis is a technique of fracture fixation that uses the smallest possible approach to the bone in an attempt to protect the fracture site and the surrounding soft tissues as much as possible. Out of the 20 cases, 11 fractures healed within 6 months, 4 within 12 months, 3 after 12 months, and 2 were considered non-unions. The two non-unions occurred in open tibial fractures. There were two cases of delayed distal wound breakdown at 24 and 40 weeks postoperatively. The explanation the authors gave for this was that it was due to the prominence of the plate in the region of the distal tibia and the minimal soft tissue coverage in this area. Similar problems have been noted with 4.5 mm DCPs used in the same minimally invasive manner for distal tibial fractures (79). One case had a wound infection at 15 weeks postoperatively. There was also one case of implant failure due to screw breakage at 32 weeks postoperatively although callus had been visible on the radiographs from 14 weeks. The authors conclude that minimally invasive locking plate osteosynthesis (MILPO) may be a valuable tool in the management of fractures, especially fractures of the distal tibia, however, more research is necessary to evaluate the technique in the long term (80).

Another study retrospectively examined the records of 32 patients who had undergone minimally invasive locking plate osteosynthesis of distal and diaphyseal tibial

fractures in humans. Twenty-four patients were healed at 6 months, 27 patients at 9 months, and 29 patients at 15 months post-operatively. One case of plate bending occurred 5 months post-operatively in an 18-year old patient. The patient resumed normal activity 3 months after fracture, and the authors believe that this excessive weight bearing caused the plate to bend. Two cases of pseudoarthrosis were reported. All 29 patients present at the end of the study reported local disturbance over the medial malleolus due to the high profile of the plate. The authors conclude that MILPO is a viable alternative to other fixation methods for distal and diaphyseal tibial fractures. It appears to avoid soft tissue complications associated with open reduction and internal fixation methods and misalignment (81).

Locking plates have also been used in spinal stabilization. Thalgott et al report on their four year experience with a thoracolumbar locking plate. Twenty-five patients were followed for an average of 38 months. The diseases treated included fractures, tumors, and disc herniations. A total of five broken screws occurred in three patients. Three of the broken screws were of a first generation design which had a stress riser in the screw. The other two broken screws occurred in an elderly patient with osteopenic bone. The authors believed that the screws backed out due to the osteopenic bone and then broke. None of the patients that had screw failure required further surgeries. All of the bone grafts were either fused or heavily consolidated with no pseudoarthrosis or bone graft resorption noted. The authors concluded that the locking device provided good fixation of the anterior spinal column (82).

1.6.2 – Clinical Reports and Studies of Locking Plates in Veterinary Medicine

Locking plates have also been used to stabilize the spinal column in veterinary medicine. Voss et al report the use of the ComPact UniLock™ System for ventral stabilization of the cervical spine in 12 dogs and one cat. The diagnoses included disc-associated lesions, fractures, and luxations. Clinical outcome was good to excellent in 11 out of the 13 patients. Screw pullout occurred in one dog with caudal cervical spondylomyelopathy and four dogs showed some degree of collapse of the intervertebral space. No overt implant failure was noted in these latter dogs, however some of the screws seemed to have bent slightly, and some radiolucency indicative of instability could be observed around some of the screws. Screw breakage or pullout was also detected in a dog with an oblique fracture of the cranial body of second cervical vertebra. However, the fracture had healed and the clinical outcome was good. The authors found the ComPact Unilock system to be useful for the ventral stabilization of a variety of cervical diseases. Their purported benefits of the system included more stability compared to non-locking plates, and the ability to use monocortical screws. One disadvantage of the system is that the screws must be inserted perpendicularly to the plate, which reduces freedom for placing the screws, and plate position must be checked carefully prior to placing the first screw (83). This is especially important when placing a plate to stabilize the spine, as misdirected screws could potentially penetrate the spinal canal and cause nerve damage.

Another type of locking device known as the string of pearls™ (SOP) plates was used for vertebral stabilization in two dogs with triple thoracolumbar disc protrusions that had undergone hemilaminectomies. Follow-up radiographs on one of the dogs showed

that some of the screws were broken. However, spinal pain resolved and neurological function improved in both dogs (84).

Levine et al retrospectively examined the clinical use of LCPs in 31 cases for the stabilization of fractures or arthrodesis in horses. Subjectively, the authors felt that there was no difference in the complication rate of LCPs versus conventional plates. Their complications included incisional infections (32%), implant loosening/screw breakage (22%), and implant infections (19%). At follow-up, 25 of the horses were sound for the intended purpose, one horse was lame, and five were euthanized due to complications associated with their original injuries. The authors concluded that LCPs can be used successfully in the management of a variety of equine fractures and arthrodeses and that their use would provide a better outcome due to their better stability compared to conventional plating (85).

Another equine study retrospectively examined the use of LCPs for fetlock arthrodesis in six horses. No complications were associated with the implant application, although two of the horses were euthanized due to proximal interphalangeal joint luxation, which is a known complication of fetlock arthrodesis. The other four horses were sound for breeding purposes the following year. The authors note that the LCP and screws were easier to use compared with standard LC-DCP and non-locking screws and that surgical time was reduced since the screws were self-tapping (86).

A 3.5 mm Locking Round-Hole Reconstruction Plate was used to provide temporary stabilization of a shoulder joint in a Golden Retriever that had suffered a traumatic shoulder luxation. The plate was left in place for 25 days and then removed.

No complications were noted with its use and the dog was considered to have a good clinical outcome (87).

Locking plates have been used in two case reports of mandibular reconstruction in dogs. The plates were used to span large mandibular defects (7 cm and 5 cm). Intra-oral exposure of a plate occurred in one of the cases, but resolved upon removal of the implant. Both dogs healed well (88,89).

An LCP was used for the stabilization of radius/ulna and tibia/fibula fractures in a Bernese Mountain Dog. The tibia/fibula plate bent two days postoperatively, no screw loosening was noted at that time (Figure 13).

The plate on the tibia was fairly far off the bone, and, as was shown in a previous study, this could have caused a significant reduction in load to failure (42). Revision was performed by placing an intramedullary pin and another locking plate. At 53 days after surgery radiographs revealed that the pin had migrated proximally, but complete bridging and good callus formation of both the radius/ulna and the tibia/fibula occurred. No other complications were noted. The migrating pin could have been indicative of interfragmentary motion, however this was not sufficient to prevent fracture healing as evidenced by the complete bridging (90).

Figure 13. Postoperative radiographs of repair of tibial fracture: a) Original repaired with a 12 hole 3.5 mm LCP. b) 2 days post-operatively, c) 7.5 weeks post-operatively, revealing complete bridging and callus formation. The intramedullary pin has migrated proximally (90).

From these clinical studies, it can be gathered that, although clinical success is consistently achieved with locking plate systems, there is no clear evidence of the biological benefits of locking plates compared to conventional plating. It must however be noted that comparison of the two fixation methods is difficult since very few papers examined both in the same study. Most authors recommend the use of internal fixators such as LCPs and some commented that they were easier to use.

Overall, fractures treated with internal fixators seem to do as well as fractures treated with conventional plates, with most fractures having good to excellent outcomes. Furthermore, both systems appear to have relatively similar complication rates. Of all the complications, it appears that soft tissue problems over the plate were reported more

often with internal fixators than with conventional plates. This problem may be due to the fact that unless the plate is in close contact with the bone, internal fixators may have a higher profile compared to conventional plates, which by function of their fixation method must be directly against the bone.

1.7 – Summary of Literature Review

Internal fixators evolved based on the principle of biological osteosynthesis. Biological osteosynthesis attempts to treat fractures by optimizing the bone healing environment by preserving the circulation of the fracture site. Whenever new devices are introduced, it is always important to ensure that they will not be worse or weaker than the current implements. In various studies it was shown that internal fixators are at least as strong as their conventional counterparts.

One of the problems with internal fixators is that they usually do not allow for angling of the screw within the screw hole and there is also no simple way to compress fractures when using internal fixators. Locking compression plates were developed to address these issues. These plates have been demonstrated to be as strong as their non-locking counterparts. Additional studies have examined the effect of using a combination of locking and non-locking screws in these plates and have found that the hybrid constructs are usually stronger than the non-locking constructs. However, there are still very few guidelines to the use of hybrid internal fixators. Questions such as how many locking screws need to be used and whether their position relative to the fracture matters remain unanswered.

Finally, a clinical picture of the use of internal fixators was obtained, which showed that internal fixators do not appear to worsen the prognosis of fractures, although they do not have any clear benefits compared to conventional plates.

References

- (1) Perren SM. Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. *J Bone Joint Surg Br* 2002; 84(8): 1093-1110.
- (2) Egol KA, Kubiak EN, Fulkerson E, Kummer FJ, Koval KJ. Biomechanics of locked plates and screws. *J Orthop Trauma* 2004; 18(8): 488-493.
- (3) Rahn BA. Bone healing: histological and physiological concepts. In: Sumner-Smith G, editor. *Bone in Clinical Orthopaedics: A Study in Comparative Osteology*. Second ed. New York: Thieme; 2002. p. 287-326.
- (4) Schutz M, Sudkamp NP. Revolution in plate osteosynthesis: new internal fixator systems. *J Orthop Sci* 2003; 8(2): 252-258.
- (5) AO Foundation History. Available at: <http://www.aofoundation.org/>.
- (6) Leunig M, Hertel R, Siebenrock KA, Ballmer FT, Mast JW, Ganz R. The evolution of indirect reduction techniques for the treatment of fractures. *Clin Orthop Relat Res* 2000; 375: 7-14.
- (7) Rüedi TP, Murphy WM. *AO principles of fracture management*. Stuttgart ; New York; Davos Platz, Switzerland: Thieme; AO Pub.; 2000.
- (8) Gutwald R, Alpert B, Schmelzeisen R. Principle and stability of locking plates. *Keio J Med* 2003; 52(1): 21-24.
- (9) Luthi U, Rahn BA, Perren SM. Area of contact between osteosynthesis plate and bone in internal fixation (author's transl). *Aktuelle Traumatol* 1980; 10(3): 131-136.
- (10) Gautier E, Perren SM, Ganz R. (iii) Principles of internal fixation. *Current Orthopaedics* 1992; 6(4): 220-232.
- (11) Gautier E, Rahn BA, Perren SM. Vascular remodelling. *Injury* 1995; 26(Supplement 2): B11-B19.
- (12) Bostman OM. Rotational refracture of the shaft of the adult tibia. *Injury* 1983; 15(2): 93-98.
- (13) Claes L, Heitemeyer U, Krischak G, Braun H, Hierholzer G. Fixation technique influences osteogenesis of comminuted fractures. *Clin Orthop Relat Res* 1999; 365: 221-229.

- (14) Finsen V, Benum P. Refracture of the hip rare after removal of fixation device. *Acta Orthop Scand* 1986; 57(5): 434-435.
- (15) Hidaka S, Gustilo RB. Refracture of bones of the forearm after plate removal. *J Bone Joint Surg Am* 1984; 66(8): 1241-1243.
- (16) Kessler SB, Deiler S, Schiffli-Deiler M, Uhthoff HK, Schweiberer L. Refractures: a consequence of impaired local bone viability. *Arch Orthop Trauma Surg* 1992; 111(2): 96-101.
- (17) Mulier T, Seligson D, Sioen W, van den Bergh J, Reynaert P. Operative treatment of humeral shaft fractures. *Acta Orthop Belg* 1997; 63(3): 170-177.
- (18) Riemer BL, Butterfield SL, Burke CJ, 3rd, Mathews D. Immediate plate fixation of highly comminuted femoral diaphyseal fractures in blunt polytrauma patients. *Orthopedics* 1992; 15(8): 907-916.
- (19) Gautier E, Sommer C. Guidelines for the clinical application of the LCP. *Injury* 2003; 34 Suppl 2: B63-76.
- (20) Frigg R. Development of the Locking Compression Plate. *Injury* 2003; 34 Suppl 2: B6-10.
- (21) Wagner M. General principles for the clinical use of the LCP. *Injury* 2003; 34 Suppl 2: B31-42.
- (22) Aguila AZ, Manos JM, Orlansky AS, Todhunter RJ, Trotter EJ, Van der Meulen MC. In vitro biomechanical comparison of limited contact dynamic compression plate and locking compression plate. *Vet Comp Orthop Traumatol* 2005; 18(4): 220-226.
- (23) Burstein AH, Wright TM. Fundamentals of orthopaedic biomechanics. Baltimore: Williams & Wilkins; 1994.
- (24) Ramotowski W, Granowski R. Zespol. An original method of stable osteosynthesis. *Clin Orthop Relat Res* 1991; 272: 67-75.
- (25) Tepic S, Perren SM. The biomechanics of the PC-Fix internal fixator. *Injury* 1995; 26(Supplement 2): B5-B10.
- (26) Frigg R. Locking Compression Plate (LCP). An osteosynthesis plate based on the Dynamic Compression Plate and the Point Contact Fixator (PC-Fix). *Injury* 2001; 32 Suppl 2: 63-66.
- (27) Kaab MJ, Frenk A, Schmeling A, Schaser K, Schutz M, Haas NP. Locked internal fixator: sensitivity of screw/plate stability to the correct insertion angle of the screw. *J Orthop Trauma* 2004; 18(8): 483-487.

- (28) Dickson KF, Munz JW. Locked Plating: Biomechanics and Biology. *Techniques in Orthopaedics* 2007; 22(4): E1-E6.
- (29) Radin EL. *Practical biomechanics for the orthopedic surgeon*. 2nd ed ed. New York: Churchill Livingstone; 1992.
- (30) Miller DL, Goswami T. A review of locking compression plate biomechanics and their advantages as internal fixators in fracture healing. *Clin Biomech* 2007; 22(10): 1049-1062.
- (31) Fulkerson E, Egol KA, Kubiak EN, Liporace F, Kummer FJ, Koval KJ. Fixation of diaphyseal fractures with a segmental defect: a biomechanical comparison of locked and conventional plating techniques. *J Trauma* 2006; 60(4): 830-835.
- (32) Zhao YF, Li QH, Gu ZC, Wang AM. The biomechanics of point contact-dynamic compression plate and its effects on bone perfusion. *Chin J Traumatol* 2006; 9(3): 161-167.
- (33) Arens S, Eijer H, Schlegel U, Printzen G, Perren SM, Hansis M. Influence of the design for fixation implants on local infection: experimental study of dynamic compression plates versus point contact fixators in rabbits. *J Orthop Trauma* 1999; 13(7): 470-476.
- (34) Eijer H, Hauke C, Arens S, Printzen G, Schlegel U, Perren SM. PC-Fix and local infection resistance--influence of implant design on postoperative infection development, clinical and experimental results. *Injury* 2001; 32 Suppl 2: B38-43.
- (35) Garcia S, Lozano ML, Gatell JM, Soriano E, Ramon R, Sanmiguel JG. Prophylaxis against infection. Single-dose cefonicid compared with multiple-dose cefamandole. *J Bone Joint Surg Am* 1991; 73(7): 1044-1048.
- (36) Thanni LO, Aigoro NO. Surgical site infection complicating internal fixation of fractures: incidence and risk factors. *J Natl Med Assoc* 2004; 96(8): 1070-1072.
- (37) Whittem TL, Johnson AL, Smith CW, Schaeffer DJ, Coolman BR, Averill SM, et al. Effect of perioperative prophylactic antimicrobial treatment in dogs undergoing elective orthopedic surgery. *J Am Vet Med Assoc* 1999; 215(2): 212-216.
- (38) Gardner MJ, Helfet DL, Lorich DG. Has locked plating completely replaced conventional plating? *Am J Orthop* 2004; 33(9): 439-446.
- (39) Stoffel K, Booth G, Rohrl SM, Kuster M. A comparison of conventional versus locking plates in intraarticular calcaneus fractures: a biomechanical study in human cadavers. *Clin Biomech* 2007; 22(1): 100-105.

- (40) Stoffel K, Lorenz KU, Kuster MS. Biomechanical considerations in plate osteosynthesis: the effect of plate-to-bone compression with and without angular screw stability. *J Orthop Trauma* 2007; 21(6): 362-368.
- (41) Gardner MJ, Griffith MH, Demetrakopoulos D, Brophy RH, Grose A, Helfet DL, et al. Hybrid locked plating of osteoporotic fractures of the humerus. *J Bone Joint Surg Am* 2006; 88(9): 1962-1967.
- (42) Ahmad M, Nanda R, Bajwa AS, Candal-Couto J, Green S, Hui AC. Biomechanical testing of the locking compression plate: when does the distance between bone and implant significantly reduce construct stability? *Injury* 2007; 38(3): 358-364.
- (43) Gardner MJ, Brophy RH, Campbell D, Mahajan A, Wright TM, Helfet DL, et al. The mechanical behavior of locking compression plates compared with dynamic compression plates in a cadaver radius model. *J Orthop Trauma* 2005; 19(9): 597-603.
- (44) Korner J, Diederichs G, Arzendorf M, Lill H, Josten C, Schneider E, et al. A biomechanical evaluation of methods of distal humerus fracture fixation using locking compression plates versus conventional reconstruction plates. *J Orthop Trauma* 2004; 18(5): 286-293.
- (45) Chiodo TA, Ziccardi VB, Janal M, Sabitini C. Failure strength of 2.0 locking versus 2.0 conventional syntheses mandibular plates: a laboratory model. *J Oral Maxillofac Surg* 2006; 64(10): 1475-1479.
- (46) Carter DR, Spengler DM. Biomechanics of fracture. In: Sumner-Smith G, Fackelman GE, editors. *Bone in Clinical Orthopedics*. 2nd ed. New York: AO Publishing; 2002.
- (47) Cutnell JD, Johnson KW. *Physics*. 6th ed. Hoboken, NJ: John Wiley; 2004.
- (48) Hipp JA, Edgerton BC, An KN, Hayes WC. Structural consequences of transcortical holes in long bones loaded in torsion. *J Biomech* 1990; 23(12): 1261-1268.
- (49) Burstein AH, Currey J, Frankel VH, Heiple KG, Luneth P, Vessely JC. Bone strength. The effect of screw holes. *J Bone Joint Surg Am* 1972; 54(6): 1143-1156.
- (50) Brooks DB, Burstein AH, Frankel VH. The biomechanics of torsional fractures. The stress concentration effect of a drill hole. *J Bone Joint Surg Am* 1970; 52(3): 507-514.
- (51) Remiger A, Miclau T, Lindsey RW. The torsional strength of bones with residual screw holes from plates with unicortical and bicortical purchase. *Clin Biomech* 1997; 12(1): 71-73.
- (52) Tornkvist H, Hearn TC, Schatzker J. The strength of plate fixation in relation to the number and spacing of bone screws. *J Orthop Trauma* 1996; 10(3): 204-208.

- (53) ElMaraghy AW, ElMaraghy MW, Nousiainen M, Richards RR, Schemitsch EH. Influence of the number of cortices on the stiffness of plate fixation of diaphyseal fractures. *J Orthop Trauma* 2001; 15(3): 186-191.
- (54) Kubiak EN, Fulkerson E, Strauss E, Egol KA. The evolution of locked plates. *J Bone Joint Surg Am* 2006; 88 Suppl 4: 189-200.
- (55) Stoffel K, Dieter U, Stachowiak G, Gächter A, Kuster MS. Biomechanical testing of the LCP--how can stability in locked internal fixators be controlled? *Injury* 2003; 34 Suppl 2: B11-9.
- (56) DeTora M, Kraus K. Mechanical testing of 3.5 mm locking and non-locking bone plates. *Vet Comp Orthop Traumatol* 2008; 21(4): 318-322.
- (57) Florin M, Arzendorf M, Linke B, Auer JA. Assessment of stiffness and strength of 4 different implants available for equine fracture treatment: a study on a 20 degrees oblique long-bone fracture model using a bone substitute. *Vet Surg* 2005; 34(3): 231-238.
- (58) Lehmann W, Briem D, Blauth M, Schmidt U. Biomechanical comparison of anterior cervical spine locked and unlocked plate-fixation systems. *Eur Spine J* 2005; 14(3): 243-249.
- (59) Boswell S, McIff TE, Trease CA, Toby EB. Mechanical characteristics of locking and compression plate constructs applied dorsally to distal radius fractures. *J Hand Surg [Am]* 2007; 32(5): 623-629.
- (60) Weinstein DM, Bratton DR, Ciccone WJ, 2nd, Elias JJ. Locking plates improve torsional resistance in the stabilization of three-part proximal humeral fractures. *J Shoulder Elbow Surg* 2006; 15(2): 239-243.
- (61) Lill H, Hepp P, Korner J, Kassi JP, Verheyden AP, Josten C, et al. Proximal humeral fractures: how stiff should an implant be? A comparative mechanical study with new implants in human specimens. *Arch Orthop Trauma Surg* 2003; 123(2-3): 74-81.
- (62) Siffri PC, Peindl RD, Coley ER, Norton J, Connor PM, Kellam JF. Biomechanical Analysis of Blade Plate Versus Locking Plate Fixation for a Proximal Humerus Fracture: Comparison Using Cadaveric and Synthetic Humeri. *J Orthop Trauma* 2006; 20(8): 547-554.
- (63) Snow M, Thompson G, Turner PG. A mechanical comparison of the locking compression plate (LCP) and the low contact-dynamic compression plate (DCP) in an osteoporotic bone model. *J Orthop Trauma* 2008; 22(2): 121-125.

- (64) Kim T, Ayturk UM, Haskell A, Miclau T, Puttlitz CM. Fixation of osteoporotic distal fibula fractures: A biomechanical comparison of locking versus conventional plates. *J Foot Ankle Surg* 2007; 46(1): 2-6.
- (65) Weiss DB, Kaar SG, Frankenburg EP, Karunakar MA. Locked versus unlocked plating with respect to plate length in an ulna fracture model. *Bull NYU Hosp Jt Dis* 2008; 66(1): 5-8.
- (66) Leitner M, Pearce SG, Windolf M, Schwieger K, Zeiter S, Schawalder P, et al. Comparison of locking and conventional screws for maintenance of tibial plateau positioning and biomechanical stability after locking tibial plateau leveling osteotomy plate fixation. *Vet Surg* 2008; 37(4): 357-365.
- (67) Amato NS, Richards A, Knight TA, Spector D, Boudrieau RJ, Belkoff S. Ex vivo biomechanical comparison of the 2.4 mm uniLOCK reconstruction plate using 2.4 mm locking versus standard screws for fixation of acetabular osteotomy in dogs. *Vet Surg* 2008; 37(8): 741-748.
- (68) O'Toole RV, Andersen RC, Vesnovsky O, Alexander M, Topoleski LD, Nascone JW, et al. Are locking screws advantageous with plate fixation of humeral shaft fractures? A biomechanical analysis of synthetic and cadaveric bone. *J Orthop Trauma* 2008; 22(10): 709-715.
- (69) Jazrawi LM, Bai B, Simon JA, Kummer FJ, Birdzell LT, Koval KJ. A biomechanical comparison of Schuhl nuts or cement augmented screws for plating of humeral fractures. *Clin Orthop Relat Res* 2000; 377: 235-240.
- (70) Spivak JM, Neuwirth MG, Labiak JJ, Kummer FJ, Ricci JL. Hydroxyapatite enhancement of posterior spinal instrumentation fixation. *Spine* 1994; 19(8): 955-964.
- (71) Roberts JW, Grindel SI, Rebholz B, Wang M. Biomechanical evaluation of locking plate radial shaft fixation: unicortical locking fixation versus mixed bicortical and unicortical fixation in a sawbone model. *J Hand Surg [Am]* 2007; 32(7): 971-975.
- (72) Sod GA, Mitchell CF, Hubert JD, Martin GS, Gill MS. In vitro biomechanical comparison of locking compression plate fixation and limited-contact dynamic compression plate fixation of osteotomized equine third metacarpal bones. *Vet Surg* 2008; 37(3): 283-288.
- (73) Drobetz H, Kutscha-Lissberg E. Osteosynthesis of distal radial fractures with a volar locking screw plate system. *Int Orthop* 2003; 27(1): 1-6.
- (74) Arora R, Lutz M, Fritz D, Zimmermann R, Oberladstatter J, Gabl M. Palmar locking plate for treatment of unstable dorsal dislocated distal radius fractures. *Arch Orthop Trauma Surg* 2005; 125(6): 399-404.

- (75) Koshimune M, Kamano M, Takamatsu K, Ohashi H. A randomized comparison of locking and non-locking palmar plating for unstable Colles' fractures in the elderly. *J Hand Surg [Br]* 2005; 30(5): 499-503.
- (76) Sommer C, Gautier E, Muller M, Helfet DL, Wagner M. First clinical results of the Locking Compression Plate (LCP). *Injury* 2003; 34 Suppl 2: B43-54.
- (77) Leung F, Chow SP. A prospective, randomized trial comparing the limited contact dynamic compression plate with the point contact fixator for forearm fractures. *J Bone Joint Surg Am* 2003; 85-A(12): 2343-2348.
- (78) Fankhauser F, Boldin C, Schippinger G, Haunschmid C, Szyszkowitz R. A new locking plate for unstable fractures of the proximal humerus. *Clin Orthop Relat Res* 2005; (430):176-81.
- (79) Borg T, Larsson S, Lindsjo U. Percutaneous plating of distal tibial fractures. Preliminary results in 21 patients. *Injury* 2004; 35(6): 608-614.
- (80) Hazarika S, Chakravarthy J, Cooper J. Minimally invasive locking plate osteosynthesis for fractures of the distal tibia--results in 20 patients. *Injury* 2006; 37(9): 877-887.
- (81) Hasenboehler E, Rikli D, Babst R. Locking compression plate with minimally invasive plate osteosynthesis in diaphyseal and distal tibial fracture: a retrospective study of 32 patients. *Injury* 2007; 38(3): 365-370.
- (82) Thalgot JS, Kabins MB, Timlin M, Fritts K, Giuffre JM. Four year experience with the AO Anterior Thoracolumbar Locking Plate. *Spinal Cord* 1997; 35(5): 286-291.
- (83) Voss K, Steffen F, Montavon PM. Use of the ComPact UniLock System for ventral stabilization procedures of the cervical spine: a retrospective study. *Vet Comp Orthop Traumatol* 2006; 19(1): 21-28.
- (84) McKee WM, Downes CJ. Vertebral stabilisation and selective decompression for the management of triple thoracolumbar disc protrusions. *J Small Anim Pract* 2008; 49(10): 536-539.
- (85) Levine DG, Richardson DW. Clinical use of the locking compression plate (LCP) in horses: a retrospective study of 31 cases (2004-2006). *Equine Vet J* 2007; 39(5): 401-406.
- (86) Carpenter RS, Galuppo LD, Simpson EL, Dowd JP. Clinical evaluation of the locking compression plate for fetlock arthrodesis in six thoroughbred racehorses. *Vet Surg* 2008; 37(3): 263-268.

(87) Post C, Guerrero T, Voss K, Montavon PM. Temporary transarticular stabilization with a locking plate for medial shoulder luxation in a dog. *Vet Comp Orthop Traumatol* 2008; 21(2): 166-170.

(88) Boudrieau RJ, Mitchell SL, Seeherman H. Mandibular reconstruction of a partial hemimandibulectomy in a dog with severe malocclusion. *Vet Surg* 2004; 33(2): 119-130.

(89) Spector DI, Keating JH, Boudrieau RJ. Immediate mandibular reconstruction of a 5 cm defect using rhBMP-2 after partial mandibulectomy in a dog. *Vet Surg* 2007; 36(8): 752-759.

(90) Schwandt CS, Montavon PM. Locking compression plate fixation of radial and tibial fractures in a young dog. *Vet Comp Orthop Traumatol* 2005; 18(3): 194-198.

Chapter 2 – Introduction to Scientific Manuscript

Internal fixators are a relatively new concept in fracture fixation. Unlike conventional plates, locking plates counteract the forces acting on fractures by creating an angle-stable construct by having the screws effectively lock into the plate. This technique can have a number of benefits compared to conventional, non-locking plating, such as decreased contact of the plate on the bone, decreased need for contouring of the plate, and stronger plate-bone constructs in weaker bone. However, there are a number of limitations to using a locking construct. Since the screws lock into the plate at a given, fixed angle, it is not possible to angle the screws within the screw hole as it is in conventional plates. In some occasions, fracture configuration may dictate that screws be placed at an angle relative to the plate. This would be the case for the placement of lag screws across oblique fractures, to assist with fracture reduction, or to avoid penetration of a joint surface. Other limitations of the locking plates are the fact that there is no simple way to create compression of the fracture using a locking plate and finally, locking screws are significantly more expensive than non-locking screws. Some of these draw backs can be addressed by replacing some of the locking screws with non-locking screws. The holes in the Synthes locking compression plate (LCP™) are designed to accommodate both types of screws, making it possible to select which type of screw is used. Although the biomechanical properties of locking plates have received great attention during the past few years, very little is known about the effects of combining locking and non-locking screws in the same construct. Additional knowledge in this area

would assist surgeons in choosing the most appropriate fixation method for fracture fixation.

The objective of this study was to determine what effect the addition of a locking screw would have on the biomechanical properties of a non-locking bone-plate construct tested in torsion.

2.1 Materials and Methods

The basic outline of this research project was that an “all locking” screw construct, an all “non-locking” screw construct, and a combination construct of both locking and non-locking screws would be tested in torsion. The first phase of the project was to test all locking and all non-locking constructs. The second phase was to test a construct secured with a combination of one locking and two non-locking screws. The plates were attached to the distal aspect of a canine femur and then mounted in a jig that exerted torsional forces on the bone-plate construct and the constructs were then tested to failure.

Phase I consisted of testing locking constructs and non-locking constructs. In the locking group, the plate was attached to the bone using all locking screws, and in the non-locking group, the plate was secured to the bone using all non-locking screws. Phase II consisted of testing a construct secured using a combination of one locking and two non-locking screws (Figure 14). Cadaveric canine femurs were used for the testing and the bone-plate constructs were loaded to failure.

Figure 14. The three bone-plate construct groups: From top to bottom; all non-locking group, all locking group, and combination group (locking and non-locking screws). The osteotomy is to the right in this diagram as represented by the jagged line on the right

The testing jig was designed such that the constructs would be subjected to torsion. The jig allowed for two degrees of rotational freedom and one degree of translation freedom on one side, and three degrees of free rotation on the other side. Torsion was applied by attaching one end of a wire to the load cell of the servohydraulic materials testing machine (Instron 4204, Canton, MA, USA) and the other end to the end of a lever arm. Another piece of metal in the shape of an arc of a circle was attached to the free end of the lever arm and the wire was fixed to this. The jig was positioned such that the wire was tangential to the arc of the lever arm (Figure 15). This would result in the force acting perpendicular to the lever arm.

Figure 15. Testing set-up: (a) Jig (b) Close-up of mounted bone

For the testing, femurs without gross evidence of orthopaedic disease were harvested from adult dogs between 20 and 25 kg body weight and euthanized for reasons other than the purpose of this study. All soft tissue was removed from the femurs, and the femurs were wrapped in moist sponges, double bagged, labeled and stored at -20°C in a conventional freezer until bone plate application and biomechanical testing.

Before every testing session a machine calibration was performed on the servohydraulic materials testing machine. This was done by sequentially loading and

unloading five kilogram weights on the load cell of the Instron machine and recording the voltage outputs. These data were then graphed (Figure 16).

Figure 16. Typical calibration curve: Each step corresponds to an additional 5 kg being added to the load cell.

The loads used for calibration were converted to units of Newtons. The average output voltage for each of the loads (0, 5, 10, and 15 kg) was calculated and compared to the corresponding applied loads. This was then graphed and linear regression was used to calculate a formula for the resulting line. This formula was then used to convert the output (volts) into force (Newtons) (Figure 17).

Output (v)	Weight (kg)	Force (N)
0	0	0
0.49	5	49.04
0.98	10	98.07
1.48	15	147.1

Figure 17. Typical Output vs. Force calibration table and curve.

Before testing, the femurs were thawed at room temperature in a water bath (while still in the bags) and the distal end was potted in plastic cups using a polyester resin (Bondo, Atlanta, Georgia) to the level of the fabellae. Initial testing consisted of testing all locking screw constructs and all non-locking screw constructs with eight paired femurs. This initial testing demonstrated that in order to detect a difference of 20% in the torque at failure with a power greater than 80%, 20 femurs would be needed in each group. The femurs were thus divided into two groups of 20 femurs each. The level of the osteotomy was marked on the bone using a standard template to ensure repeatability of the cut and to standardize the distance of the cut to the potting material. Depending upon group assignment, one half of a 6-hole 3.5 mm locking compression plate (LCP) (Synthes Ltd, Mississauga, ON, Canada) was attached to each femur using either 3 locking screws or 3 non-locking screws.

The locking and non-locking groups were paired right and left femurs from the same dog, with the side decided randomly by coin toss. Given the straight nature of the canine mid-femur no contouring was needed for any of the plates.

All bone screws were inserted following standard AO principles. All non-locking screws were tightened by hand, as would be done clinically, by a surgeon experienced in orthopaedic surgery. The locking screws were tightened using the recommended torque-limiting device (1.5 Nm torque-limiting attachment, Synthes Ltd, Mississauga, ON, Canada). All screws were bicortical. Once the plate was secured to the bone with three screws, a transverse osteotomy was performed at the previously determined level and the cranial-caudal and medial-lateral diameters of the bone were measured at the osteotomy site using a caliper. Once the osteotomy was performed, the free half of the plate was

bolted to a torsion jig using machine screws. No gaps were present between the end of the jig and the osteotomy, however, compression at the fracture site was not performed. The constructs were tested in torsion to failure at a rate of 0.5°/s using the servohydraulic testing machine (Figure 15). This rate was obtained from a previous publication (1).

The output was converted from mV into force (N), and the time measurements were converted to angular displacement (degrees). The force was converted to torque applied to the sample by multiplying the applied force by the lever arm length of the testing jig using the following formula:

$$T = r \times F$$

T = torque

r = moment arm, distance from the pivot point to the point where the force acts

F = force applied

The data were then graphed on a torque/deformation curve. Torque at failure, angle at failure and torsional stiffness were determined by analysis of the torque/deformation curves (Figure 18).

Figure 18. Deformation vs. Torque Curve: The slope of the linear region of the line is equivalent to the stiffness of the construct. The line at (x+5) is parallel to the stiffness of the curve and represents 5 degrees of plastic deformation.

Stiffness was calculated by creating a best-fit line in the initial linear region of the load displacement curve for each sample. A second line parallel to the linear portion of the torque/deformation curve but offset by 5 degrees was drawn on the graph and the intersection between this line and the torque/deformation curve was used to define clinical failure. This line represented 5° of plastic deformation, which was thought to be more representative of clinical failure than the maximum torque. The work to clinical failure and work to maximal failure were determined by calculating the area under the curve. This was done by using the trapezoidal method (Figure 19).

$$\text{Area total} = \sum_{n=1}^6 A_n$$

Figure 19. Trapezoidal method for calculating area under a curve.

The values for the torque, angle, and work at maximum failure point and the clinical failure point for the locking and non-locking groups were tested for significance. When significant differences between the two groups were found Phase II was performed.

Phase II consisted of testing constructs composed of a combination of one locking and two non-locking screws. The testing was performed as described for Phase I.

Twenty femurs from 20 different dogs were used and the side was once again randomized with a coin toss. For the combination group, the locking screw was placed in the most distal hole to the planned fracture site after the non-locking screws had been placed.

A general linear mixed model accounting for the random effect of dog and the fixed effects of construct, side and their interaction was run for the parameters measured. Residual analysis and a Shapiro Wilk test were used to determine if the data met the assumptions of the ANOVA. If the overall F-test was significant, pairwise comparisons were based on 1 sided Tukey tests. Statistical analyses were performed with standard analysis software. Statistical significance was set at $p < 0.05$ for all tests.

2.1.1 – Discussion about Material and Methods

Torsion was chosen as our testing modality for several reasons. Torsion loads the entire bone evenly along its length, whereas tests such as four point bending will create areas of stress concentration where the jig interacts with the bone. Minor variations of bone geometry, such as length will have little effect upon torsional testing. The torque applied and angular velocity are easily measured and calculated. Finally, torsion is a common force in the cause of fractures seen clinically (2,3).

Initially plaster-of-paris was used as a potting material. However, it was found that it did not dry sufficiently in a reasonable amount of time (less than 6 hours), and was thus inappropriate for use with bones which had the potential to dry out. For accurate biomechanical testing, bones need to be kept moist (4,5). The difficulty in keeping the bones moist while the plaster-of-paris dried led us to find another potting material. A

polyester resin (Bondo, Atlanta, Georgia) was tested as the next potting material and was found to harden quickly enough to be an adequate potting material for this project.

A cadaveric femur model was chosen to most closely approximate the clinical situation. While using a synthetic model would have eliminated interspecimen variability, these models do not have all the same biomechanical and frictional properties as bone. Many of these models have similar biomechanical properties to bone when tested whole (6,7), however, the strength of these surrogate models can become significantly different from bone when subjected to a notch test which is used to determine fracture toughness (8). Several studies have shown that screw holes can significantly weaken bone (9-12) and that the screw hole will influence the behavior of the fracture. Screw holes can cause bone surrogates to have significantly different fracture toughness than real bone which may affect the load that the constructs fail at, and may also cause them to fail in a different pattern compared to real specimens (8). This is further demonstrated in a study in which locking plates were compared to non-locking plates in both cadaveric humeri and synthetic humeri tested in torsion. This study demonstrated that the locking plates had significantly less angular displacement than the non-locking plates in cadaveric specimens, while the two constructs were not significantly different in the synthetic specimens (13).

In a previously mentioned study, locking and non-locking screws were used to secure plates on both cadaveric and synthetic bone. Both groups in the synthetic bone failed at a mean force greater than 4200 N, while in the cadaveric bone they failed at a mean force greater than 3400 N. The authors did not compare this statistically or comment on this. However, this may indicate the drawbacks of using synthetic

specimens to compare fixation methods. The authors also did not examine whether the stiffness of the constructs in the synthetic bone were different from the ones in the cadaveric bone (14).

Freezing and thawing bone is known to have damaging effects upon the internal structure of tissues including bone (15). There are a number of studies with conflicting results as to whether the changes wrought upon the internal structure of bone due to freezing and thawing will significantly affect the mechanical properties of the bone. Strömberg and Dalén demonstrated that frozen bone is significantly weaker (4.6 ± 2.3 %) than fresh bone (16). However, other studies have demonstrated a non-significant increase in strength of bones that have been frozen compared to fresh bones (17). With the uncertainty surrounding the effects that freezing and thawing may have on bones, the measurements from bones that have been frozen should be interpreted with the consideration to the effects that freezing and thawing may have had. However, it should also be noted that in the studies comparing frozen and fresh bones, freezing and thawing only mildly affected the overall percentage strength of the bone.

The screw hole most distal to the osteotomy was chosen for the locking screw position in the combination group because it was hypothesized that this hole would experience the greatest degree of torque during torsional testing since it had the greatest lever arm from the fracture line. Previous studies of conventional, non-locking plates tested in torsion have shown that these constructs will most likely fail at the point of fixation furthest from the fracture (18-20). A locking screw in this hole should therefore have the greatest effect on the torsional strength of the construct. It is not known if the same biomechanical effect would have been obtained if the locking screw had been

placed in a different plate hole, closer to the fracture. Future studies are needed to determine the effects of the locking screw location or the placement of additional locking screws.

It was decided to pair the locking and non-locking combinations by using contralateral femurs from the same dog since these two groups were the first to be tested. It was important to ascertain that there was a significant difference between the all locking group and the all non-locking group before investigating whether a combination of locking and non-locking screws would affect the torsional properties of bone-plate constructs. If no difference between the all locking and all non-locking groups had been found, then one would have been able to assume that combinations of the two screw types would not be significantly different either from each other or from the all locking and all non-locking groups. Locking and non-locking plates were used on contralateral femurs in an effort to minimize interspecimen variability. Strömberg and Dalén demonstrated that there is close agreement in the maximum torque capacity between contralateral bones from dogs (2).

Only one side of the plate was affixed to bone in our study. This was done in an attempt to minimize the variability of having the plate attached to bone on both sides of the osteotomy. Since the results of this study are derived from 5 degrees of plastic deformation from only one side of the plate, it could mean up to 10 degrees of plastic deformation if both sides were attached to bone.

References

- (1) Weinstein DM, Bratton DR, Ciccone WJ, 2nd, Elias JJ. Locking plates improve torsional resistance in the stabilization of three-part proximal humeral fractures. *J Shoulder Elbow Surg* 2006; 15(2): 239-243.
- (2) Stromberg L, Dalen N. Experimental measurement of maximum torque capacity of long bones. *Acta Orthop Scand* 1976; 47(3): 257-263.
- (3) Burstein AH, Frankel VH. A standard test for laboratory animal bone. *J Biomech* 1971; 4(2): 155-158.
- (4) Kim WK, Donalson LM, Herrera P, Woodward CL, Kubena LF, Nisbet DJ, et al. Research note: Effects of different bone preparation methods (fresh, dry, and fat-free dry) on bone parameters and the correlations between bone breaking strength and the other bone parameters. *Poult Sci* 2004; 83(10): 1663-1666.
- (5) Crenshaw TD, Peo ER, Lewis AJ, Moser BD. Bone strength as a trait for assessing mineralization in swine: A critical review of techniques involved. *J Anim Sci* 1981; 53: 827-835.
- (6) Cristofolini L, Viceconti M, Cappello A, Toni A. Mechanical validation of whole bone composite femur models. *J Biomech* 1996; 29(4): 525-535.
- (7) Heiner AD. Structural properties of fourth-generation composite femurs and tibias. *J Biomech* 2008; 41(15): 3282-3284.
- (8) Gibson TW, Moens NM, Runciman RJ, Holmberg DL. Evaluation of a short glass fibre-reinforced tube as a model for cat femur for biomechanical testing of orthopaedic implants. *Vet Comp Orthop Traumatol* 2008; 21(3): 195-201.
- (9) Brooks DB, Burstein AH, Frankel VH. The biomechanics of torsional fractures. The stress concentration effect of a drill hole. *J Bone Joint Surg Am* 1970; 52(3): 507-514.
- (10) Remiger A, Miclau T, Lindsey RW. The torsional strength of bones with residual screw holes from plates with unicortical and bicortical purchase. *Clin Biomech* 1997; 12(1): 71-73.
- (11) Burstein AH, Currey J, Frankel VH, Heiple KG, Luneth P, Vessely JC. Bone strength. The effect of screw holes. *J Bone Joint Surg Am* 1972; 54(6): 1143-1156.
- (12) Malone CB, Heiple KG, Burstein AH. Bone strength: before and after removal of unthreaded and threaded pin and screws. *Clin Orthop Relat Res* 1977; 123: 259-260.

- (13) Siffri PC, Peindl RD, Coley ER, Norton J, Connor PM, Kellam JF. Biomechanical Analysis of Blade Plate Versus Locking Plate Fixation for a Proximal Humerus Fracture: Comparison Using Cadaveric and Synthetic Humeri. *J Orthop Trauma* 2006; 20(8): 547-554.
- (14) O'Toole RV, Andersen RC, Vesnovsky O, Alexander M, Topoleski LD, Nascone JW, et al. Are locking screws advantageous with plate fixation of humeral shaft fractures? A biomechanical analysis of synthetic and cadaveric bone. *J Orthop Trauma* 2008; 22(10): 709-715.
- (15) Mazur P. Cryobiology: the freezing of biological systems. *Science* 1970; 168(934): 939-949.
- (16) Stromberg L, Dalen N. The influence of freezing on the maximum torque capacity of long bones. An experimental study on dogs. *Acta Orthop Scand* 1976; 47(3): 254-256.
- (17) Sedlin ED, Hirsch C. Factors affecting the determination of the physical properties of femoral cortical bone. *Acta Orthop Scand* 1966; 37(1): 29-48.
- (18) Miclau T, Remiger A, Tepic S, Lindsey R, McIff T. A mechanical comparison of the dynamic compression plate, limited contact-dynamic compression plate, and point contact fixator. *J Orthop Trauma* 1995; 9(1): 17-22.
- (19) ElMaraghy AW, ElMaraghy MW, Nousiainen M, Richards RR, Schemitsch EH. Influence of the number of cortices on the stiffness of plate fixation of diaphyseal fractures. *J Orthop Trauma* 2001; 15(3): 186-191.
- (20) Mensch JS, Markolf KL, Roberts SB, Finerman GM. Experimental stabilization of segmental defects in the human femur. A torsional study. *J Bone Joint Surg Am* 1976; 58(2): 185-190.

Chapter 3 - The Effect of the Combination of Locking Screws and Non-Locking Screws on the Torsional Properties of a Locking Plate Construct

S. Gordon¹, N. M. M. Moens¹, J. Runciman², G. Monteith¹

1. Department of Clinical Studies, Ontario Veterinary College, University of Guelph,
Guelph, Ontario, Canada, N1G 2W1

2. School of Engineering, University of Guelph, Guelph, Ontario, Canada, N1G 2W1

Please address correspondence to Dr. S. Gordon, Department of Clinical Studies, Ontario
Veterinary College, University of Guelph, Guelph, Ontario, Canada, N1G 2W1

3.1 – Abstract

Little is known about the torsional properties of bone-plate constructs using a combination of locking and non-locking screws. Sixty cadaveric canine femurs were divided into 3 groups. In the first group, the plate was affixed using three non-locking screws. In the second group only locking screws were used while a combination of one locking and two non-locking screws were used in the third group. All constructs were subjected to torsion until failure. Torque, angle of torsion, and work were all calculated at the maximum failure point, as well as at 5 degrees of plastic deformation, which was thought to be more representative of clinical failure. At maximum failure point, the locking group had significantly higher torque, angle, and work values than the non-locking group. The combination group was intermediate to the two other groups and significantly differed from the non-locking group in torque and from the locking group in work. At 5 degrees of plastic deformation, the locking group required significantly higher torque and work than the non-locking group. The combination group required a significantly higher torque than the non-locking group. This study suggests that a construct composed of all locking screws will fail at a greater torque value, and sustain greater work to failure in torsion compared to a construct composed of all non-locking screws. The addition of a single locking screw to an otherwise non-locking construct will increase the torque at the offset failure point and may be of clinical value in constructs subjected to high torsional loads.

3.2 – Introduction

Successful fracture fixation results when the forces acting on the fracture fragments are counteracted by the mechanical stiffness of the bone-implant construct during loading. Conventional fracture plating relies on the friction between the plate and the bone to generate stability (1). Achieving such friction requires intimate contact between the plate and the bone and therefore extensive pre-contouring of the plate to the bone. This extensive contact can negatively affect periosteal blood flow, resulting in bone necrosis and resorption under the plate (2,3). Conventional dynamic compression plates have smooth, elongated screw holes allowing the placement of screws in a large range of orientations to match the needs of each specific fracture. Although the ability to vary screw angle is convenient for the surgeon, it may also represent a significant drawback of conventional plating. Since each screw can freely pivot in its hole it therefore relies on two points of purchase (both cortices) to maintain its stability. Any loss of cortical purchase in either cortex, due to bone resorption, microfracture etc., will result in the loss of one of the two purchase points and will allow the screw to toggle in the plate hole and the bone fragment to shift under the plate.

Locking plates were developed as a means to reduce these problems and improve bone healing, particularly in poor quality bone (4). In these systems, a thread is cut in the head of the screw and a matching thread is cut into the plate screw hole. When the screw is tightened, the head of the screw engages the thread in the plate, effectively locking the screw to the plate. Because any angulation of the screw within the screw hole is rendered impossible, the construct acts as an internal fixator and does not rely upon friction for stability. Any screw toggling is effectively prevented, even with monocortical screws

with only one point of purchase in the bone. Because this system does not rely on plate to bone friction to achieve stability, the plate does not have to be in direct contact with the bone and perfect contouring is no longer necessary. The space between the plate and the bone may result in an improved periosteal blood flow, causing reduced bone necrosis under the plate and may also decrease infections rates (4-7).

The holes of the locking compression plates (LCP, Synthes Ltd, Mississauga, ON, Canada) are designed such that either a non-locking or locking screw can be used in them. The use of a traditional (non-locking) screw in a locking plate may be indicated when angulation of the screw is necessary, such as for the fixation of a butterfly fragment with lag screws, to assist with fracture reduction, or to avoid penetration of a joint surface. The use of non-locking screws may also be selected in order to decrease the overall cost of the fixation as locking screws are significantly more expensive than regular screws. Although several studies have shown that LCPs perform similarly to conventional plates when tested biomechanically (8-10), only a few studies have evaluated the effect of using a combination of locking and non-locking screws on the biomechanical properties of plate-bone constructs subjected to torsion (11,12). Because locking screws create angle-stable constructs we hypothesized that the addition of a locking screw to a non-locking construct would significantly affect the torsional strength of the bone-plate construct.

3.3 – Materials and Methods

Sixty femurs without gross evidence of orthopaedic disease were harvested from adult dogs between 20 and 25 kg body weight and euthanized for reasons other than the

purpose of this study. All soft tissue was removed from the femurs, and the femurs were wrapped in moist sponges, double bagged, labeled and stored at -20 °C in a conventional freezer until bone plate application and biomechanical testing. Before testing, the femurs were thawed at room temperature in a water bath (while still in the plastic bags) and the distal end was potted in plastic cups using a polyester resin (Bondo, Atlanta, Georgia) to the level of the fabellae.

Initial testing consisted of testing all locking screw constructs and all non-locking screw constructs with eight paired femurs. This initial testing demonstrated that in order to detect a difference of 20% in the torque at failure with a power greater than 80%, 20 femurs would be needed in each group. The femurs were thus divided into three groups of 20 femurs each. The constructs used for the initial testing were included in their respective groups. The level of the osteotomy was marked on the bone using a standard template to ensure repeatability of the cut and to standardize the distance of the cut to the potting material. Depending upon group assignment, one half of a 6-hole 3.5 mm locking compression plate (LCP) (Synthes Ltd, Mississauga, ON, Canada) was attached to each femur using either 3 locking screws, 3 non-locking screws, or a combination of 2 non-locking and one locking screw (n=20). The locking and non-locking groups were paired right and left femurs from the same dog, with the side decided randomly by coin toss. For the combination group, 20 femurs from 20 different dogs were used and the side was once again randomized with a coin toss. Given the straight nature of the canine mid-femur no contouring was needed for any of the plates.

All bone screws were inserted following standard AO principles. All non-locking screws were tightened by hand, as would be done clinically, by a surgeon experienced in

orthopaedic surgery. The locking screws were tightened using the recommended torque-limiting device (1.5 Nm torque-limiting attachment, Synthes Ltd, Mississauga, ON, Canada). All screws were bicortical. For the combination group, the locking screw was placed in the most distal hole to the planned fracture site after the non-locking screws had been placed. Once the plate was secured to the bone with three screws, a transverse osteotomy was performed at the previously determined level and the cranial-caudal and medial-lateral diameters of the bone were measured at the osteotomy site using a caliper. Once the osteotomy was performed, the free half of the plate was bolted to a torsion jig using machine screws. No gaps were present between the end of the jig and the osteotomy, however, compression at the fracture site was not performed. Only one side of the plate was affixed to bone in an attempt to minimize the variability of having the plate attached to bone on both sides of the osteotomy. The torsion jig was designed to allow 6 degrees of freedom during testing. The constructs were tested in torsion to failure at a rate of 0.5°/s using a servohydraulic testing machine (Instron 4204, Canton, MA, USA) (Figure 20).

Torque at failure, angle at failure and torsional stiffness were determined by analysis of the torque/deformation curves (Figure 21).

Figure 20. a) Potting fixture and torsional testing jig seen. The longitudinal axis was visually assessed. (b) Close up of potted femur attached to testing jig.

Figure 21. Example of torque/deformation curve. The line at $x + 5$, is parallel to the linear portion of the torque/deformation curve but offset by 5 degrees. The intersection between this line and the torque/deformation curve was used to define the clinical failure point. The torque and angle at failure at the intersection between the 5° offset line and torque/deformation curve were calculated. The work to clinical failure (the 5° offset failure point) and work to maximal failure were equal to the area under the curve.

Stiffness was calculated by creating a best-fit line in the initial linear region of the load-displacement curve for each sample. A second line parallel to the linear portion of

the torque/deformation curve but offset by 5 degrees was drawn on the graph and the intersection between this line and the torque/deformation curve was used to define clinical failure. This line represented 5° of plastic deformation, which was thought to be more representative of clinical failure than the maximum torque. The torque and angle at failure at the intersection between the 5° offset line and torque/deformation curve were calculated. The work to clinical failure (the 5° offset failure point) and work to maximal failure were determined by calculating the area under the curve.

A general linear mixed model accounting for the random effect of dog and the fixed effects of construct, side and their interaction was run for the parameters measured. Residual analysis and a Shapiro Wilk test were used to determine if the data met the assumptions of the ANOVA. If the overall F-test was significant, pairwise comparisons were based on 1 sided Tukey tests. Statistical analyses were performed with standard analysis software^a. Statistical significance was set at 0.05 for all tests.

3.4 – Results

For all values for the torque, angle, and work at the maximum and offset failure points, the greatest value was always the all locking construct, the smallest value was always the non-locking construct, and the combination construct was always intermediate to the other 2 groups.

The mean +/- standard deviation of the average diameter (average of the medial-lateral and cranio-caudal diameters) of the bones was 15.20 +/- 2.19 mm, 15.05 +/- 1.970 mm, 14.90 +/- 0.904 mm for the locking, non-locking and combination group respectively. There were no statistical differences in bone sizes between groups.

The mean $\pm$ standard deviation of the torsional stiffness for the locking construct was 2.19 ± 0.381 Nm/degree, for the non-locking construct it was 2.03 ± 0.569 Nm/degree, and 2.31 ± 0.305 Nm/degree for the combination construct. The torsional stiffness of the different constructs were not significantly different. Representative curves for each construct are demonstrated in Figure 22.

Figure 22. Examples of torque/deformation curves comparing locking, combination, and non-locking plates.

At the maximum failure point, the locking group had significantly higher torque than the non-locking group (52.6 ± 8.43 Nm vs. 38.7 ± 11.57 Nm, respectively, $p = < 0.001$). Failure angle ($47.8 \pm 18.88^\circ$ vs. $30.8 \pm 13.46^\circ$, $p = 0.001$), and work (1694 ± 886 Nm-degree vs. 842 ± 609 Nm-degree, $p = 0.002$) were also significantly higher in the locking group than in the non-locking group (Table 2).

Construct	Maximum torque (Nm)	Maximum angle (degrees)	Maximum work (Nm-degree)	Offset torque (Nm)	Offset angle (degrees)	Offset work (Nm-degree)
All locking	52.6 +/- 8.43 (48.6, 56.7) ^a	47.8 +/- 18.88 (38.9, 56.6) ^c	1694 +/- 886 (1266, 2120) ^{d, e}	45.1 +/- 7.26 (41.7, 48.5) ^f	25.2 +/- 3.95 (23.3, 27.1) ^{h, i}	625 +/- 149.8 (553, 697) ^j
Combination	47.4 +/- 5.74 (44.1, 50.7) ^b	36.6 +/- 12.36 (30.8, 42.4)	1119 +/- 473 (842, 1340) ^d	40.7 +/- 5.74 (38.0, 43.4) ^g	22.1 +/- 3.28 (20.6, 23.7) ^h	526 +/- 126.0 (467, 585)
All non-locking	38.7 +/- 11.57 (33.3, 44.1) ^{a, b}	30.8 +/- 13.46 (24.5, 37.1) ^c	842 +/- 609 (557, 1127) ^e	34.8 +/- 9.2 (30.5, 39.1) ^{f, g}	20.6 +/- 4.69 (18.32, 22.8) ⁱ	454 +/- 178.6 (370, 537) ^j

Table 2: Summary of the data, comparing all locking, combination screw, and all non-locking group: Results are shown for both the maximum failure point and the 5-degree offset failure point. Data represents the mean +/- standard deviation (95% confidence interval). Data with superscripted letters denote significant differences between groups

(a). $p = <0.0001$; (b). $p = 0.02$; (c). $p = 0.001$; (d). $p = 0.04$; (e). $p = 0.002$; (f). $p = <0.0001$; (g). $p = 0.04$; (h). $p = 0.03$; (i). $p = <0.001$; (j). $p = 0.001$

At the maximum failure point, the combination group was intermediate to the two other groups and had a significantly lower work compared to the locking group (1119 +/- 473 Nm-degree vs. 1694 +/- 886 Nm-degree, $p = 0.04$), but a significantly higher torque compared to the non-locking group (47.4 +/- 7.11 Nm vs. 38.7 +/- 11.57 Nm, $p = 0.02$).

At 5 degrees of plastic deformation (clinical failure), the locking group had a significantly higher torque (45.1 +/- 7.26 Nm vs. 34.8 +/- 9.20 Nm, $p = <0.0001$), work (625 +/- 149.8 Nm-degree vs. 454 +/- 178.5 Nm-degree, $p = 0.001$), and angle (25.2 +/- 3.95° vs. 20.6 +/- 4.69°, $p = < 0.001$) than the non-locking group. The combination group had a significantly higher torque compared to the non-locking group (40.7 +/- 5.74 Nm vs. 34.8 +/- 9.20 Nm, $p = 0.04$) and a lower angle at the offset compared to the all locking construct (22.1 +/- 3.28° vs. 25.2 +/- 3.95°, $p = 0.03$).

3.5 – Discussion

The current study found that an all locking LCP construct was significantly stronger in torsion than an all non-locking LCP construct. The combination construct of locking and non-locking screws was always intermediate to the other two groups for torque, angle, and work.

Bone plates are one of the most frequently used implants for fracture fixation. Several types of bone plates have been designed but traditionally all relied on the friction between the plate and bone generated by the compression of the plate to the bone by the torque of the screws. The frictional force generated is equal to the product of force that presses the plate to the bone (the normal force) and the coefficient of friction between the plate and bone (13). The normal force of a conventional bone-plate construct is equal to the sum of the torques on each screw (14) which is related to the resistance to shear stress of the bone trapped within the threads of the screw (15). Since non-locking plates are unable to prevent the screw from orienting in the direction of the applied force to the plate, the strength of the connection between the bone and the screws is only due to the shear stress between the screws and the bones (15). The weakest component of the plate-screw-bone construct is the screw-bone interface. The strength of this interface is equal to the stress resistance of the bone multiplied by the contact area between the screw and the bone (15). Recently a new type of plate was introduced which does not rely upon the friction between the plate and the bone for stability. These new plates differ from the traditional plates in that the screws that secure the plate to the bone also lock into the plate.

Traditional (non-locking) and locking constructs rely upon different principles for stability. Non-locking plates are reliant upon the friction generated between the plate and the bone for stability of the construct. If the forces acting upon the construct are greater than the coefficient of friction generated between the bone and the plate, the plate may start shifting, causing the screws to pull out of the bone. As the screw pulls out, toggling will occur and the construct loses stability. The strength of fixation of non-locking constructs is equal to the resistance to shear stress of the bone trapped within the threads of the screw since non-locking plates are unable to prevent the screw from orienting in the direction of the applied force to the plate (15). On the other hand, locking plates are not reliant upon friction for construct stability. Locking screws result in an angle stable device, in which the plate prevents the screws from orienting in the direction of the applied force to the plate, thus converting the shear stress at the screw-bone interface to a compressive stress, which bone is better able to resist. In order for a locking construct to fail, all screws must cutout of the bone simultaneously, the bone must fracture, or the plate must fail (16).

There have been several biomechanical studies comparing the properties of locking versus non-locking constructs and these studies often have conflicting results. While some studies demonstrated a significant improvement in the strength of locking construct in axial compression, torsion, and bending when compared to non locking constructs (11,16-23), others did not find any significant differences between the two (8,10,24-30). Reasons for the conflicting results found in the various studies include a variety of locking systems used in a variety of bones, both cadaveric and synthetic, in a variety of species, tested using a variety of methods.

A number of studies have specifically examined the torsional properties of LCPs. Aguila et al compared LCPs using all locking screws to LC-DCPs in cadaveric canine femurs and found that the mean twist to failure of the LC-DCP was significantly lower than for the LCP, however the torsional stiffness of the two constructs was not significantly different (8). This is in contradiction with Sod et al who compared 4.5 mm LCP to 4.5 mm LC-DCP in equine third metacarpal bones. In their model they used a combination of locking and non-locking screws in the LCP and found torsional stiffness was significantly greater for the locking plates (17).

Since non-locking and locking constructs employ different biomechanical principles to impart stability to the construct it is generally recommended that using combinations of locking and non-locking screws in the same construct be avoided. However, there are a number of situations in which using a combination of fixation techniques would be beneficial. If compression at the fracture site is desired, then non-locking screws must be used. Non-locking screws give surgeons the freedom to angle screws to engage bone fragments that locking screws would be unable to engage sufficiently, also non-locking screws are sometimes used to reduce fractures by drawing the bone to the plate. Finally, non-locking screws are significantly less expensive than locking screws, and thus may be selected in order to decrease the cost of the construct. When non-locking and locking screws are used in the same construct it is recommended that the non-locking screws be used first to establish the friction fit between the plate and bone, followed by locking screws to protect the fixation (4,31).

The addition of a locking screw to a non-locking construct should theoretically change the construct into an angle stable construct. Thus, the addition of a single locking

screw should significantly increase the torque at failure and work as demonstrated in this study in which a 17% increase in torque at 5 degrees offset was observed after addition of a locking screw to a non-locking construct. However, the addition of a single locking screw was not equivalent to an all locking construct (Table 2).

There are very few studies examining the biomechanical effects of using a combination of locking and non-locking screws in the same construct. Gardner et al. studied the effects of using a combination of two locking and one non-locking screw in a LCP in a synthetic human humeral osteoporotic model on cyclic torsion testing. That study found no significant differences between the all locking and the combination group, but both of these groups were significantly stiffer than the non-locking group throughout all phases of cyclic testing. After 1000 cycles, the locking and combination constructs had declined to 80.0 +/- 10.2% and 79.2 +/- 9.5% of their initial stiffness, while the non-locking group had declined to 22.3 +/- 12.6% (11).

In our study, the non-locking plates failed at a lower torque and at a lower angle than the combination plates or the locking plates, and the all locking plates failed at a higher torque and angle compared to the other two groups of plates. Since all of the plates tested were the same type (LCP), it was the attachment of the plate to the bone that was being tested and any differences seen would be due to how the plate was secured to the bone. The fact that the same type of plate was used in all three groups also explains why the stiffness of the constructs was similar between groups as long as the torsional force did not exceed the screws' holding power or overcome the frictional forces between the bone and the plate. Similar stiffness in torsion between locking and non-locking constructs using LCPs was also found in both synthetic and cadaveric human humeral

fractures (26) and in cadaveric human ulnar fractures (29). This is in contradiction with Gardner's study which found a significant difference between the stiffness of the locking versus the non-locking constructs. The difference found in Gardner's study may be due to the difference in testing material (synthetic humeri versus cadaveric femurs) but more likely due to the fact that Gardner et al overdrilled the screw holes to simulate reduced screw purchase in osteoporotic bone while our study used normal canine bones and no attempts were made to stimulate osteoporosis (11).

Another study examined the effects of replacing the end screws of a locking unicortical construct with bicortical (locking or non-locking) screws and compared these constructs to a non-locking bicortical construct. During torsional testing, they found that the unicortical locking group was significantly less stiff than the other three groups and that replacing the end screws of a locking unicortical construct with bicortical screws significantly increased the construct stiffness. Both of the hybrid constructs provided equivalent stability in torsion compared to the non-locking bicortical group. The locked hybrid group was significantly stiffer than the other three groups in anteroposterior bending. The authors concluded that the major factor in determining torsional stability was likely the increased working length of the bicortical screws, rather than using locking or non-locking screws (12). This latter study is difficult to compare to our study because of the confounding effect of using unicortical versus bicortical screws.

We used bicortical screws in this study since bicortical screws are most commonly used in small animal surgery. While some studies have shown that locking unicortical screws provide sufficient stability to tolerate loading in excess of physiological loads (32), other studies demonstrated that constructs composed entirely of

locking unicortical screws are significantly weaker in torsion than constructs in which locking or non-locking bicortical screws are used (12,16). This is due to the increased working length or amount of bone purchase present. Thus, clinically it is recommended that bicortical screws be used whenever possible (6,16,33).

The screw hole most distal to the fracture was chosen for the locking screw position in the combination group because we hypothesized that this hole would experience the greatest degree of torque during torsional testing since it had the greatest lever arm from the fracture line. Previous studies of conventional, non-locking plates tested in torsion have shown that these constructs will most likely fail at the point of fixation furthest from the fracture (25,34). A locking screw in this hole should therefore have the greatest effect on the torsional strength of the construct. It is not known if the same protective effect would have been obtained if the locking screw had been placed in a different plate hole, closer to the fracture. Future studies are needed to determine the effects of the locking screw location or the placement of additional locking screws.

The non-locking screws were hand tightened by an experienced orthopaedic surgeon. This was done in an attempt to mimic the clinical setting. It should be noted that different clinicians will tighten screws to different degrees. This variation could potentially alter the results of this study. However, experienced clinicians will generally apply a consistent level of torque (35). The amount of torque applied to the screw will directly affect the amount of compression achieved between the plate and the bone, and thus the friction between the two (14).

One of the major limitations of this study was its *in vitro* nature. The femurs were stripped of all surrounding soft tissues which could contribute to fracture stability *in vivo*.

A simple, transverse fracture model with no bone loss was used, which does not occur commonly in clinical situations. It has been shown that bone resorption occurs around implants when even minute instability exists. This resorption could cause loose implants that would then affect the stability of the construct (36). For this resorption to occur, active blood supply is required and therefore resorption cannot be accurately reproduced *in vitro*, even with cyclic testing. For this reason, we tested the specimens by applying a single load to failure. Static testing is often used as a first step for testing constructs and provides information regarding the levels of force that a construct can stand prior to failure (28,37). The behavior of these constructs in the presence of bone resorption or in poor quality bone warrants further investigation.

We chose to use a cadaveric femur model to most closely approximate the clinical situation. While using a synthetic model would have eliminated interspecimen variability, these models do not have all the same biomechanical and frictional properties as bone, nor do they always have the same screw holding power as cadaveric femora (38). Many of these models have similar biomechanical properties to bone when tested whole (39,40). However, the strength of these surrogate models can become significantly different from bone when subjected to a notch test which is used to determine fracture toughness (38). Several studies have shown that screw holes can significantly weaken bone (41-44) and that the screw hole will influence the behavior of the fracture. Screw holes can cause bone surrogates to have a significantly different fracture toughness than real bone which may affect the load that the constructs fail at, and may also cause them to fail in a different pattern compared to real specimens (38). This is further demonstrated in a study in which locking plates were compared to non-locking plates in both cadaveric

humeri and synthetic humeri tested in torsion. This study demonstrated that the locking plates had significantly less angular displacement than the non-locking plates in cadaveric specimens, while the two constructs were not significantly different in the synthetic specimens (21).

From previous studies and the current study it can be seen that LCPs used in locking fashion are as strong, if not stronger than conventional plates or LCPs used in non-locking fashion. Although the torsional stiffness of the constructs in the initial phase of loading was similar, non-locking constructs failed at lower torque than the locking and combination constructs. Adding a single locking screw to an all non-locking construct increased the torque to the offset failure point by 17%. In the current study, no significant differences were found between the combination and all locking groups for the offset torque and angle. This would suggest that the addition of just one locking screw is able to provide an angle-stable construct and it is unnecessary to replace non-locking screws that have been used to initially reduce the fracture with locking screws to create an all-locking construct. This study also suggests that an all-locking construct be used when the greatest implant strength is needed, and that adding locking screws to a non-locking construct can increase the torsional strength of that construct.

^a SAS Institute Inc. 2004. SAS OnlineDoc® 9.1.3. Cary, NC: SAS Institute Inc.

References

- (1) Rüedi TP, Murphy WM. AO principles of fracture management. Stuttgart ; New York; Davos Platz, Switzerland: Thieme; AO Pub.; 2000.
- (2) Luthi U, Rahn BA, Perren SM. Area of contact between osteosynthesis plate and bone in internal fixation (author's transl). *Aktuelle Traumatol* 1980; 10(3): 131-136.
- (3) Gautier E, Perren SM, Ganz R. (iii) Principles of internal fixation. *Current Orthopaedics* 1992; 6(4): 220-232.
- (4) Wagner M. General principles for the clinical use of the LCP. *Injury* 2003; 34 Suppl 2: B31-42.
- (5) Frigg R. Development of the Locking Compression Plate. *Injury* 2003; 34 Suppl 2: B6-10.
- (6) Gautier E, Sommer C. Guidelines for the clinical application of the LCP. *Injury* 2003; 34 Suppl 2: B63-76.
- (7) Eijer H, Hauke C, Arens S, Printzen G, Schlegel U, Perren SM. PC-Fix and local infection resistance--influence of implant design on postoperative infection development, clinical and experimental results. *Injury* 2001; 32 Suppl 2: B38-43.
- (8) Aguila AZ, Manos JM, Orlansky AS, Todhunter RJ, Trotter EJ, Van der Meulen MC. In vitro biomechanical comparison of limited contact dynamic compression plate and locking compression plate. *Vet Comp Orthop Traumatol* 2005; 18(4): 220-226.
- (9) Gardner MJ, Brophy RH, Campbell D, Mahajan A, Wright TM, Helfet DL, et al. The mechanical behavior of locking compression plates compared with dynamic compression plates in a cadaver radius model. *J Orthop Trauma* 2005; 19(9): 597-603.
- (10) Korner J, Diederichs G, Arzendorf M, Lill H, Josten C, Schneider E, et al. A biomechanical evaluation of methods of distal humerus fracture fixation using locking compression plates versus conventional reconstruction plates. *J Orthop Trauma* 2004; 18(5): 286-293.
- (11) Gardner MJ, Griffith MH, Demetrakopoulos D, Brophy RH, Grose A, Helfet DL, et al. Hybrid locked plating of osteoporotic fractures of the humerus. *J Bone Joint Surg Am* 2006; 88(9): 1962-1967.
- (12) Roberts JW, Grindel SI, Rebholz B, Wang M. Biomechanical evaluation of locking plate radial shaft fixation: unicortical locking fixation versus mixed bicortical and unicortical fixation in a sawbone model. *J Hand Surg [Am]* 2007; 32(7): 971-975.

- (13) Radin EL. Practical biomechanics for the orthopedic surgeon. 2nd ed ed. New York: Churchill Livingstone; 1992.
- (14) Miller DL, Goswami T. A review of locking compression plate biomechanics and their advantages as internal fixators in fracture healing. Clin Biomech (Bristol, Avon) 2007; 22(10): 1049-1062.
- (15) Egol KA, Kubiak EN, Fulkerson E, Kummer FJ, Koval KJ. Biomechanics of locked plates and screws. J Orthop Trauma 2004; 18(8): 488-493.
- (16) Fulkerson E, Egol KA, Kubiak EN, Liporace F, Kummer FJ, Koval KJ. Fixation of diaphyseal fractures with a segmental defect: a biomechanical comparison of locked and conventional plating techniques. J Trauma 2006; 60(4): 830-835.
- (17) Sod GA, Mitchell CF, Hubert JD, Martin GS, Gill MS. In vitro biomechanical comparison of locking compression plate fixation and limited-contact dynamic compression plate fixation of osteotomized equine third metacarpal bones. Vet Surg 2008; 37(3): 283-288.
- (18) Kim T, Ayturk UM, Haskell A, Miclau T, Puttlitz CM. Fixation of osteoporotic distal fibula fractures: A biomechanical comparison of locking versus conventional plates. J Foot Ankle Surg 2007; 46(1): 2-6.
- (19) Florin M, Arzdorf M, Linke B, Auer JA. Assessment of stiffness and strength of 4 different implants available for equine fracture treatment: a study on a 20 degrees oblique long-bone fracture model using a bone substitute. Vet Surg 2005; 34(3): 231-238.
- (20) Snow M, Thompson G, Turner PG. A mechanical comparison of the locking compression plate (LCP) and the low contact-dynamic compression plate (DCP) in an osteoporotic bone model. J Orthop Trauma 2008; 22(2): 121-125.
- (21) Siffri PC, Peindl RD, Coley ER, Norton J, Connor PM, Kellam JF. Biomechanical Analysis of Blade Plate Versus Locking Plate Fixation for a Proximal Humerus Fracture: Comparison Using Cadaveric and Synthetic Humeri. J Orthop Trauma 2006; 20(8): 547-554.
- (22) Boswell S, McIff TE, Trease CA, Toby EB. Mechanical characteristics of locking and compression plate constructs applied dorsally to distal radius fractures. J Hand Surg [Am] 2007; 32(5): 623-629.
- (23) Weinstein DM, Bratton DR, Ciccone WJ, 2nd, Elias JJ. Locking plates improve torsional resistance in the stabilization of three-part proximal humeral fractures. J Shoulder Elbow Surg 2006; 15(2): 239-243.

- (24) Leitner M, Pearce SG, Windolf M, Schwieger K, Zeiter S, Schawalder P, et al. Comparison of locking and conventional screws for maintenance of tibial plateau positioning and biomechanical stability after locking tibial plateau leveling osteotomy plate fixation. *Vet Surg* 2008; 37(4): 357-365.
- (25) Miclau T, Remiger A, Tepic S, Lindsey R, McIff T. A mechanical comparison of the dynamic compression plate, limited contact-dynamic compression plate, and point contact fixator. *J Orthop Trauma* 1995; 9(1): 17-22.
- (26) O'Toole RV, Andersen RC, Vesnovsky O, Alexander M, Topoleski LD, Nascone JW, et al. Are locking screws advantageous with plate fixation of humeral shaft fractures? A biomechanical analysis of synthetic and cadaveric bone. *J Orthop Trauma* 2008; 22(10): 709-715.
- (27) Stoffel K, Lorenz KU, Kuster MS. Biomechanical considerations in plate osteosynthesis: the effect of plate-to-bone compression with and without angular screw stability. *J Orthop Trauma* 2007; 21(6): 362-368.
- (28) DeTora M, Kraus K. Mechanical testing of 3.5 mm locking and non-locking bone plates. *Vet Comp Orthop Traumatol* 2008; 21(4): 318-322.
- (29) Weiss DB, Kaar SG, Frankenburg EP, Karunakar MA. Locked versus unlocked plating with respect to plate length in an ulna fracture model. *Bull NYU Hosp Jt Dis* 2008; 66(1): 5-8.
- (30) Amato NS, Richards A, Knight TA, Spector D, Boudrieau RJ, Belkoff S. Ex vivo biomechanical comparison of the 2.4 mm uniLOCK reconstruction plate using 2.4 mm locking versus standard screws for fixation of acetabular osteotomy in dogs. *Vet Surg* 2008; 37(8): 741-748.
- (31) Gardner MJ, Helfet DL, Lorich DG. Has locked plating completely replaced conventional plating? *Am J Orthop* 2004; 33(9): 439-446.
- (32) Marti A, Fankhauser C, Frenk A, Cordey J, Gasser B. Biomechanical Evaluation of the Less Invasive Stabilization System for the Internal Fixation of Distal Femur Fractures. *J Ortho Trauma* 2001; 15(7): 482-487.
- (33) Kubiak EN, Fulkerson E, Strauss E, Egol KA. The evolution of locked plates. *J Bone Joint Surg Am* 2006; 88 Suppl 4: 189-200.
- (34) ElMaraghy AW, ElMaraghy MW, Nousiainen M, Richards RR, Schemitsch EH. Influence of the number of cortices on the stiffness of plate fixation of diaphyseal fractures. *J Orthop Trauma* 2001; 15(3): 186-191.

- (35) McGuire RA, St John KR, Agnew SG. Analysis of the torque applied to bone screws by trauma surgeons. Comparisons based on years of experience and material of implant construction. *Am J Orthop* 1995; 24(3): 254-256.
- (36) Perren SM. Evolution of the internal fixation of long bone fractures. The scientific basis of biological internal fixation: choosing a new balance between stability and biology. *J Bone Joint Surg Br* 2002; 84(8): 1093-1110.
- (37) An YH, Draughn RA. Mechanical testing of bone and the bone-implant interface. Boca Raton: CRC Press; 2000.
- (38) Gibson TW, Moens NM, Runciman RJ, Holmberg DL. Evaluation of a short glass fibre-reinforced tube as a model for cat femur for biomechanical testing of orthopaedic implants. *Vet Comp Orthop Traumatol* 2008; 21(3): 195-201.
- (39) Cristofolini L, Viceconti M, Cappello A, Toni A. Mechanical validation of whole bone composite femur models. *J Biomech* 1996; 29(4): 525-535.
- (40) Heiner AD. Structural properties of fourth-generation composite femurs and tibias. *J Biomech* 2008; 41(15): 3282-3284.
- (41) Brooks DB, Burstein AH, Frankel VH. The biomechanics of torsional fractures. The stress concentration effect of a drill hole. *J Bone Joint Surg Am* 1970; 52(3): 507-514.
- (42) Remiger A, Miclau T, Lindsey RW. The torsional strength of bones with residula screw holes from plates with unicortical and bicortical purchase. *Clin Biomech* 1997; 12(1): 71-73.
- (43) Burstein AH, Currey J, Frankel VH, Heiple KG, Lunseth P, Vessely JC. Bone strength. The effect of screw holes. *J Bone Joint Surg Am* 1972; 54(6): 1143-1156.
- (44) Malone CB, Heiple KG, Burstein AH. Bone strength: before and after removal of unthreaded and threaded pin and screws. *Clin Orthop Relat Res* 1977; (123): 259-260.

Chapter 4 – Final Summary

In order for fractures to heal adequate blood supply and fracture stabilization must be present. Historically fracture stabilization has aimed at anatomic reconstruction and absolute stability, often with detriment to the soft tissues surrounding the fracture. In addition, the plates that have been used in fracture reconstruction have relied upon friction between the bone and the plate for stability. This close contact of the plate on the bone can cause negative effects to the blood supply to the bone. Biological plate osteosynthesis is a recent theory that has been developed to avoid some of the problems associated with conventional bone plating.

Locking compression plates are an angle stable device, such that the screws lock into the plate. This means that stability is not reliant upon friction between the plate and the bone, thus the plate can be slightly off the bone which would theoretically improve the blood flow to the bone. Locking plates also have a number of other theoretical advantages including, not requiring exact anatomic contouring, the ability to place them through a minimally invasive manner, less disruption of the fracture fragments, and better holding in weak bone.

Numerous studies have compared locking and non-locking devices biomechanically. The locking devices are usually as strong, if not stronger than the non-locking devices. In the few clinical studies on locking devices, they appear to be useful clinically, however no statistically significant benefits compared to conventional plates were found.

Locking compression plates have holes that are designed to accept either locking or non-locking screws. Non-locking screws may be indicated in situations in which the

screw needs to be angled such as for placing a lag screw across a butterfly fragment, to assist with fragment reduction, to avoid a joint, or to gain better purchase in the bone if the plate is off the axis of the bone.

This study investigated the effects of using a combination of locking and non-locking screws on the torsional properties of a bone-plate construct in torsion. It was found that the all locking plate was always statistically stronger than the all non-locking plate. The combination plate was intermediate to the other two groups, however this was not always statistically significant.

From this we can conclude that an all locking construct is going to be the strongest construct, and that adding locking screws to a non-locking construct will make it stronger in torsion.