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AN EXPERIMENTAL TEST FIXTURE AND TEST METHODOLOGY FOR
EVALUATING FRACTURE FIXATION PLATES

by

David S. Backman. B.A.Sc

A thesis submitted to
the Faculty of Graduate Studies and Research
in partial fulfillment of
the requirements for the degree of
Master of Engineering

Ottawa-Carleton Institute for Mechanical and Aerospace Engineering

Carleton University

Ottawa, Ontario

December 2000

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EVALUATING FRACTURE FIXATION PLATES**

Submitted by

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In partial fulfillment of the requirements for
The degree of Master of Engineering

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December 2000

Abstract

From an engineering perspective the testing of orthopaedic implants is often quite challenging since pure mechanical testing does not take into account the interaction between the implant and bone. Any testing method that could provide more information about this interaction would be useful in evaluating new plate designs. In order to fulfil this need two different experimental test fixtures were designed, one to simulate bone during the early phase of fracture healing and the other to simulate bone during the late phase. The test fixtures were designed to allow us to compare the mechanical properties of different plates and to measure how the plates affected the mechanical environment of the underlying bone. Three different types of fracture plates were tested in three separate loading modes: torsion, compression and four-point bending loads. Our test results compared well to those of other researchers and demonstrated that our method provides accurate results. Our test method showed that in order to optimise fracture healing, fracture plates should provide between 0.25-0.28 mm of displacement at the fracture site. New fracture plate designs can now be tested using our test fixtures and test methodology and the data from these tests could provide some preliminary indication as to the ability of a specific plate design to influence fracture healing.

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List of Symbols

Symbol	Description
ACP	Axially Compressible Plate
DCP	Dynamic Compression Plate
F	F-test statistic
LC-DCP	Low-Contact Dynamic Compression Plate
LDPE	Low Density Polyethylene
LVDT	Linear Variable Differential Transformer
P	Probability of Rejecting the Null Hypothesis
PFRP	Polar Flexural Rigidity Profile
PMMA	Polymethylmethacrylate
SI	Stiffness Index

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1 INTRODUCTION

1.1 Problem Statement and Thesis Summary

The use of fracture fixation plates is widely accepted in orthopaedic surgery for the treatment of bone fractures^{15: 41: 47: 62: 63: 77: 79}. Fracture plates are typically rigid metallic implants used in conjunction with specially designed screws, to hold broken bones in the proper position to allow for fracture healing. Although a wide variety of fracture fixation plate designs are currently used, they all share the same goal, which is to improve the mechanical environment surrounding the bone, in order to optimize fracture healing. During the initial phase of fracture healing, fracture plates play an important role in controlling the amount of axial movement or micromotion in bone. During later stages of fracture healing a fracture plate can have the effect of shielding the underlying bone from normal, physiologic stresses. Both of these factors can have a significant effect on the final strength of the healed bone. Currently mandated test methods such as ISO 9585 focus solely on the mechanical properties of the fracture plate with no regard for how the fracture plate modifies the mechanical environment of the surrounding bone.

In designing a new test method for fracture plates one of the goals was to be able to measure how the addition of a rigid fracture fixation plate would change the mechanical properties of the underlying bone during both the early and late stages of

fracture healing. In order to accomplish this two different aluminium bone models were built. Each bone model was instrumented using either strain gages or other displacement sensors. Various fracture plate designs were then fixed to the aluminium bone models using standard 3.5-mm surgical screws. One aluminium model simulated a bone immediately after fracture (GAP model) during the early phase of fracture healing while the other bone model simulated a healed bone (WHOLE model) during the later phase of fracture healing.

In order to prove the accuracy of the mechanical testing system, four different canine fracture plates that had already undergone in-vivo and in-vitro testing were retested, along with a dynamic compression plate (DCP) and a low contact dynamic compression plate (LC-DCP). Both the DCP and the LC-DCP are considered to be the current standard in fracture fixation. The four canine plates that were tested were specially designed such that they differed only in terms of their axial stiffness (1.64×10^6 N/mm to 13.91×10^6 N/mm). The previous mechanical testing performed by Yeadon⁷⁸ only measured the axial and bending stiffnesses of the canine fracture plates. By mounting these plates on the aluminium test bones and subjecting them to the new testing methodology it was possible to compare the results to those of the previous testing and gain additional understanding of how these plates changed the mechanical environment for fracture healing.

The second goal of the project was to measure the amount of axial micromotion allowed by the canine ACP and by the DCP and LC-DCP plates. Small motions in the

axial direction (micromotion) have become increasingly recognised as one of the most powerful stimulators of fracture healing^{5: 16: 27: 35: 39: 41: 60: 80}. One of the fundamental hypotheses in the work of Yeadon⁷⁸ was that plates that allow for enhanced axial micromotion should increase fracture healing. Although the in-vivo results from Yeadon suggest that plates with intermediate stiffnesses provide the greatest increase in fracture healing, no research was done to actually quantify how much micromotion was occurring.

Chapter one provides some of the background information needed to put this research into the proper context. The chapter begins with a review of basic bone anatomy and structure, as well as a brief review of the mechanical properties of bone. Next, is a more in-depth discussion of how fractures heal, with specific reference to both the union and remodelling phase of fracture healing and the role that axial micromotion and stress shielding play. With this groundwork in place the next part of the chapter provides an overview regarding the engineering aspects of fracture plate design. The final part of the chapter provides the results of a literature search centred on the various mechanical test methodologies that have been used by other researchers to study the performance of fracture plate designs.

Chapter two provides an overview of how the aluminium test fixtures were designed. A problem statement is given as well as a design overview and the engineering constraints within which our design had to operate. Next is a brief review of the material selection process followed by a discussion of the test fixtures used to apply compression.

bending and torsional loads to the test fixture. The chapter closes with a brief discussion of the type of instrumentation that was employed to measure displacement during the various loading modes.

Chapter three focuses on an important verification test performed on the canine axially compressible plates. The purposes behind the testing as well as the results are discussed in detail.

Chapter four contains an in-depth discussion regarding the experimental design, the biomechanical testing and the results. A detailed explanation is given as to why the various loading modes are used as well as which variables are of interest. Chapter four also provides details regarding the experimental testing procedures as well as summaries of all the results and relevant statistical data.

Chapter five provides a discussion and analysis of the results, putting them into an understandable context.

Chapter six provides conclusions regarding our research project as well as recommendations for further testing and suggested improvements that could be made to both the experimental method and to the test fixtures.

1.2 Bone as an Engineering Material

There are over 200 bones in the human body, with a wide variety of shapes and sizes⁴⁸. One of the major functions of bone is to provide structural support and anchoring to such soft tissues as muscle, tendons and ligaments. Bone is also used to provide protection to various parts of the body – most importantly to the brain, which is surrounded by a hard thin layer of bone. On a more specialised level, bone provides protection for specialised cells such as marrow cells that are housed in the centre of most long bones. Fracture fixation using rigid internal plates is generally limited to the long bones in the human body and this will be the focus of our discussion on gross anatomy of bone.²⁹

1.2.1 Basic Long Bone Structure

The long bones in the human body all share some basic structural features (Figure 1.1). The central region of a long bone is termed the diaphysis. This is typically the most obviously tubular part of the bone. The diaphyseal shaft or diaphysis is typically composed of hard, dense bone. The hollow space within the diaphysis is called the medullary canal and it is normally filled with bone marrow.

The diaphyseal shaft is surrounded by a fibrous membrane called the periosteum, which plays an important role in fracture healing since it contains dormant cells that become active in the event of a fracture. The endosteum lines the interior of the bone and its structure and function are similar to that of the periosteum.²⁹

Towards either end of a long bone, the bone tends to flare outwards and the internal structure takes on an increasingly porous structure, this region is called the

epiphysial region and it serves to transfer joint and body loads to the diaphyseal region of the bone. (Figure 1.1).

Figure 1.1: Typical structure of a long bone ¹⁹

1.2.2 Bone Matrix Composition

Calcium phosphate makes up approximately two-thirds the weight of bone and it plays an important role in providing strength to the bone matrix since it interacts with calcium hydroxide to form crystals of hydroxyapatite. Hydroxyapatite crystals are very hard, but are relatively inflexible and are quite brittle. The other main constituents of bone are collagen fibres, which make up the remaining one-third of the weight. Collagen fibres are strong in tension, however, unlike hydroxyapatite crystals, collagen fibres alone cannot tolerate any type of compressive load. In bone the collagen fibres provide a framework for the formation of hydroxyapatite crystals. The result is directly analogous to a typical composite material where hydroxyapatite crystals serve as the brittle, high strength reinforcing material and collagen fibres serve as the ductile matrix. As a result

bone has properties superior to either hydroxyapatite or collagen alone. Bone is more ductile than hydroxyapatite, thereby allowing it to absorb more energy before failure. and it is also more rigid than collagen, allowing greater load carrying capacity and greater stiffness ^{44: 63: 66}.

1.2.3 Types of Bone

The two main types of bone in the human body are compact and spongy bone. Compact bone also known as cortical bone, forms the external layer of all bones in the human body and it makes up the bulk of the diaphyses of long bones. Compact bone is hard and dense and it provides protection and support to sensitive areas of the body such as the brain as well as helping long bones resist the stresses applied to them ^{44: 66}. The basic functional unit of mature compact bone is the osteon or haversian system. The haversian system is composed of collagen/hydroxyapatite fibrils that are formed into sheets (lamellae) with preferred directions. The orientation of the fibres defines directions of maximum and minimum strength ⁶³. The lamellae of the haversian system or osteon are arranged into circular concentric structures elements several millimetres long and 0.2-0.3 millimetre (0.008-0.012 inch) in diameter (Figure 1.2). Osteons exhibit a spiral course oriented along the axis of the bone ^{44: 66}. An osteon has maximum strength along its long axis and a lower strength perpendicular to its circumference ⁶³.

The second major type of bone in the human body is referred to as spongy or cancellous bone. The matrix in cancellous bone forms a loose network of rigid beams called trabeculae. The structure of trabecular bone is similar to that of porous engineering foams. In fact many of the mathematical techniques used to quantify the

behaviour of these foams have been used to predict the material properties of trabecular bone²⁹. Unlike, engineering foams however, trabecular bone is dynamic and can change its trabecular orientation in response to the applied load. While compact bone shares the same microscopic structure as trabecular bone, its density, and thus its overall strength is greatly increased. Spongy bone is typically located in areas where bones are not heavily stressed or where the stresses are multi-directional. The epiphyseal region of the femur is an excellent example of a location in which the trabeculae are oriented along the stress lines that develop in the femur when under compression.

Figure 1.2: Microstructure of an osteon¹⁹

1.2.4 Cells in bone

A discussion of bone microstructure would be incomplete without a brief description of the four characteristic cell types that can be found within the structure of bone. These four cell types are: osteoblasts, osteocytes, osteoclasts and pluripotential mesenchymal cells.

Osteoblasts are responsible for the production and deposition of new bone, a process that is called osteogenesis. These cells synthesise and release the proteins and other organic components that constitute the bone matrix.

Osteocytes are the most abundant type of cell in the bone matrix. They are mature bone cells that have become embedded within the bone matrix. These types of cells are responsible for the assuring the homeostasis of bone by recycling the calcium salts in the matrix around them.

Osteoclasts on the other hand are large multinucleated cells that resorb bone by direct chemical and enzymatic attack. Osteoclasts are always removing bone matrix and osteoblasts are always adding to the bone matrix. This balance is crucial to bone strength since an increase in the activity of one type of cell with a subsequent decrease in the activity of the other can lead to either a weakening or a strengthening of the bone matrix.

Before examining the factors that affect the strength of a whole bone, it is useful to describe the basic mechanical properties of cortical bone. Cortical bone is a heterogeneous, anisotropic, non-linear viscoelastic material ⁶³. Unlike typical engineering materials the properties of bone are not constant, but instead they vary significantly depending on the age, sex and pathology of the individual donor. For modelling purposes bone is often assumed to be an anisotropic linear elastic material.

This assumption holds for as long as the strains and strain rate are in the physiologic range⁶³. A typical stress-strain curve for cortical bone under pseudostatic tension is shown in figure 1.3. It is interesting to compare the results to figure 1.4, which shows the stress-strain curves for bone at various strain rates. There is a significant increase in the modulus of bone as the loading rate increases.

Pluripotential mesenchymal cells, also called osteoprogenitor cells are stem cells that can divide to produce daughter cells that differentiate into osteoblasts or osteoclasts. These cells reside in the tissue between the trabeculae and in the periosteum and give rise to either osteoblasts or osteoclasts given the appropriate stimuli, such as surface damage or fracture^{44: 66}

1.2.5 Mechanical Properties of Bone

Compact bone specimens have been found to have tensile strength in the range of 70 -140 MPa and compressive strengths in the range of 140 - 210 MPa. Bone is strongest along its long axis, roughly parallel to both the collagen fibre axis and to the long axis of the mineral crystals. Engineering estimates put the compressive modulus of elasticity of bone between 11 - 13 GPa and the tensile modulus of elasticity between 17 -20 GPa⁶³. Bone exhibits linear elastic behaviour when subjected to loads of between 30 - 40 percent of its breaking strength. Above this point viscoelastic creep begins to occur (Figure 1.3). It should be noted that the modulus of elasticity in bone is strain rate dependent, with bone increasing in modulus as the strain rate increases, an effect which is clearly shown in figure 1.4. This type of behaviour is a prime indicator that there is an element of viscous flow during deformation²⁸.

Figure 1.3: Typical stress-strain curve for cortical bone ¹⁸

Figure 1.4: Stress-strain curves for cortical bone at different loading rates ¹⁸

1.2.6 Fracture Mechanisms of Long Bones

Bone, as most materials, is susceptible to fracture. In general the three principal stress planes (maximum tensile, compressive and shear stress) dictate the fracture plane and can help predict when and how the material will fail. Cortical bone is generally weaker in tension and in shear, particularly along the longitudinal plane. This anisotropic behaviour plays a large role in influencing bone failure under external loads ⁶³. Most long bone fractures are subjected to a combination of applied loads that produce a complex fracture pattern. By examining some common loading modes in isolation, it is possible to understand some of the more frequent mechanisms of long bone fractures.

When bone is subjected to a pure bending load, the convex side of the bone is loaded in tension and the concave side is loaded in compression. Since bone is weaker in tension than in compression, the first cracks will form on the convex (tension) side of the bone. As the crack propagates, the neutral axis shifts towards the compressive side because the bone no longer has any structural resistance to bending. Thus when examining bone that has failed under a bending load, the crack is most often on the tensile side of the bone, with failure occurring in tension with a transverse fracture plane^{13: 28}

Since the weight bearing long bones are most often subjected to compressive, weight bearing loads, it is of great interest to investigate the combined effects of both bending and axial compression. When bending, which produces tension on the convex side of the bone, is superimposed on axial compression, a combination of two fracture

processes occurs. As was mentioned previously, failure by bending produces a transverse crack on the tensile side of the bone. The compressive force on the bone results in an oblique fracture. When the bone is subject to both bending and compression, the protruding oblique surface impacts the other surface. The result is a characteristic “butterfly segment”, which occurs on the concave side of the bone²⁸.

When a bone is subjected to a purely compressive load the failure mechanism tends to be oblique cracking of the osteons. Fractures produced by compressive loading are usually found in the vertebrae. Compressive loading to failure in a joint can sometime be produced by an abnormally strong contraction of the surrounding muscles¹³.

Another important failure mode in bone is torsion. When a bone is subjected to a purely torsional load the result is typically a spiral fracture with the fracture line oriented at about 45° to the axis about which the torque is applied. From a purely mechanical point of view this failure of the bone is due to tensile forces that have been induced in the bone as a result of the applied torque. Since bone is stronger in compression, tensile failure occurs perpendicular to the diagonal in tension. The crack will be perpendicular to the diagonal in tension, so it will spiral at 45° to the long axis, in the direction opposite to the applied torque.

When loaded in pure torsion, the material property of most importance is the polar moment of inertia (J). The polar moment of inertia for circular shafts is defined as,

$$J = \frac{\pi \cdot d^4}{32}$$

Equation 1.1

where d is diameter of the shaft. The polar moment of inertia (J) serves as a mathematical way of describing the distribution of material around the cross-section of the bone. The region of a long bone with the smallest polar of inertia is subjected to the largest deformation under load and will fail first. From the angle of twist equation for a circular shaft it is clear that decreasing the polar moment of inertia. will increase the angle of twist for a given torque.

$$\theta = \frac{T \cdot L}{J \cdot G}$$

Equation 1.2

In this equation T represents torque, L represents the length of bone, J represents the polar moment of inertia, G represents the shear modulus of the material and θ represents the angle of twist. When the polar moment of inertia decreases the amount of angular deformation increases. This increase in angular deformation causes areas of the bone with low polar moments of inertia to fail first¹³. When examining typical cross sections along tibia, it becomes clear why the portion of the shaft closest to the foot, with its smaller cross-section, is a predominant location for spiral fractures²⁸.

1.3 Fracture Healing

One of the unique features of bone is its ability to restore its original tissue structure after injury. Unlike muscle or skin, bone is capable of healing without the

intervention of fibrous (scar) tissue ⁵⁵. In addition, the quality of the healed bone can be significantly affected by the mechanical environment it is subjected to; from an engineering point of view this makes bone stand alone as a very unique material. This also makes it vital that engineers doing any research or design work in the field of fracture fixation have a firm understanding of the biological, as well as the biomechanical processes that occur during fracture fixation.

Fracture healing is one of the most widely studied subjects in orthopaedic research. Many researchers have studied and put forward theories that try and explain the biological and/or the mechanical factors that drive the fracture healing process ^{9: 13: 21: 45: 53}. The process by which a fractured bone regenerates or repairs itself is extremely dynamic and can be strongly influenced by the way a fracture is treated. It is important to note that fractures will most often heal regardless of the method of treatment, whether it be a fracture plate or a plaster cast that is used to treat the break. Thus the main focus behind much of fracture research is the drive to determine the mechanical environment that will optimise fracture healing. The fracture treatment modality that is used has a significant impact on the type of bone tissue that forms and on the mechanical characteristics of this regenerated bone tissue. For the purposes of this discussion the focus will be on the relationship between the mechanical and biological aspects of fracture healing. The two most general phases of the fracture healing process are termed the union phase and the remodelling phase. Both the biology and the mechanics of the healing fracture will be examined for each of these stages.

1.3.1 The Union Phase of Fracture Healing

The union phase of fracture healing is characterised by a bridging process by which the body seeks to join the fractured fragments of bone. There are two separate processes that can result in the fracture being bridged: primary healing and secondary healing. These two types of fracture healing are heavily dependent on the type of fracture fixation that is used.

1.3.1.1 Primary Fracture Healing

Primary healing, or primary cortical healing involves a direct attempt by the cortex to re-establish continuity once it has become interrupted. Primary fracture healing will only occur when two conditions are met. The first is that the fractured bone must be restored to its original anatomical configuration, the second is that there must be a substantial decrease in interfragmentary strain. Both of these conditions are typically met through the use of rigid internal fracture fixation plates²¹. If the ends of the fractured bone are in close enough proximity then osteoclasts on one side of the fracture tunnel into the opposing side of the fracture and re-establish new haversian systems by providing pathways for the penetration by blood vessels. A wide range of bone precursor cells, such as mesenchymal cells accompany these blood vessels. These distinctive tunnels in bone can be easily seen with the aid of a low power microscope and are referred to as cutting cones²¹. The descriptor “primary” derives from the fact that bone healing is completed without the need for any intermediate or precursor cells to become involved⁴⁵. With primary fracture healing there is no periosteal tissue or callus to bridge the fracture

site. In addition, the ongoing work by researchers such as Uhthoff et al ⁶⁸ has shown that primary fracture healing can result in delayed healing, possible refracture and cortical thinning.

1.3.1.2 Secondary Fracture Healing

Secondary fracture healing is so called because initially an intermediate fibrocartilage forms between the bone fragments before being remodelled into new bone. Macroscopically, secondary fracture healing is characterised by the formation of both periosteal and endosteal callus ²⁹. Callus tissue composed of bony and cartilaginous material that forms a connecting bridge across a bone fracture during repair ^{44: 66}. Callus formation plays an important role in providing mechanical stability to the fracture. Relative to a bone that heals with no fracture callus or only a callus tissue internally in the medullary canal, a periosteal callus can increase the bending stiffness of a fractured bone by a factor of two and the strength by a factor of 5/3. Since the periosteal callus is on the exterior surface of the bone, it effectively increases the diameter of the bone surrounding the fracture site and thus it increases the polar moment of inertia of the bone ⁶³. The most potent factor influencing secondary fracture healing is the cyclical transmission of axial micromotion that arises as a result of load sharing between the fracture plate and the bone ^{27: 74: 75}. This transmission of axial micromotion is one of the major factors that can be controlled through the effective design of the fracture fixation device in question.

1.3.2 Axial Micromotion

The transmission of axial micromotion is also one of the more controversial aspects in the design of a fracture plate. The controversy begins with the definition of the term micromotion and whether it differs from the concept of engineering strain. Part of the problem would appear to stem from clinicians mistakenly assuming that micromotion and fracture strain can be used interchangeably. An excellent example of this occurs in a paper by Kenwright et al ³⁷ where the concept of fracture site strain and micromovement are used interchangeably. In order to clarify the concepts involved it is important to understand what is meant by fracture site strain. Fracture site strain is typically measured in units of strain (change in length/gage length) and it normalises the amount of displacement occurring at the fracture site. The gage length in this case is typically defined as the fracture gap width, which is the distance between the primary bone fragments. Herein lies the problem. from a purely engineering point of view the gage length is often sharply defined and it usually remains unchanged during the course of an experiment. In the human body, initial gage length does not in fact remain constant since the body is actively seeking to re-establish a connection between the bone fragments. As a result the gage length, defined here as the fracture gap distance is constantly changing. The variable that does remain constant, or at the very least that can be more easily measured and controlled, is the amount of movement occurring between the bone fragments. The judicious use of a fracture fixation device can have an enormous impact on the amount of relative motion between the bone fragments. The term micromotion is typically applied to the union phase of fracture healing when two (or more) fracture fragments exist. Depending on the stage of union there will be varying degrees of

periosteal and endosteal callus tissue binding the fragments. Of primary interest to researchers is what degree of micromotion is necessary to optimize the environment for fracture healing. For the purpose of this research micromotion and not fracture site strain will be measured. In this instance micromotion can be measured directly with the aid of an extensometer mounted at the fracture gap. Converting this value to strain would be as simple as dividing the results from the extensometer by the gage length, which in this case is an arbitrary fracture gap distance of three millimetres.

1.3.3 The Remodelling Phase of Fracture Healing

Once the fracture site has been adequately bridged, the newly formed temporary bone from the callus begins the remodelling process that changes it into harder and stronger cortical bone. The remodelling stage of fracture healing occurs over a long period of time as excess tissue is removed and transformed into cortical bone. Bone remodelling continues until the bone has returned to its original shape and size and has altered its shape to optimize its load bearing capacity. As with the union phase of fracture healing, the mechanical environment of the bone will have a significant effect on the rate and on the amount of remodelling that takes place ⁴⁵. The process by which bone functionally adapts, or remodels itself in response to its mechanical environment is commonly referred to as Wolff's law.

1.3.4 Stress Shielding

Wolff was one of the first researchers to examine functional remodelling in bone. Wolff put forward the hypothesis that bone is deposited in areas of high stress and is reabsorbed in areas of low stress effectively optimizing the strength to weight ratio of bone. Wolff's law has important implications in fracture healing, especially where rigid internal fixation devices are involved ²⁸. It should be noted that although functional adaptation as described by Wolff refers to bone adapting to areas of high and low stress, it is actually the strain within the bone that regulates the remodelling process. Although stress and strain are related through Hooke's law, strain is a primary, measurable physical quantity, while stress is a mathematically derived quantity ²⁸. Carter et al have refined Wolff's early work and put forward a hypothesis that bone remodelling is related to the strain rate history of bone. According to Carter, strain rate history is a function of the strain range, mean strain magnitude and the number of loading cycles that a particular bone is subjected to. When these values remain in the normal range, then the rate of bone resorption equals the rate of bone deposition. If the strain rate history changes as a result of immobilization or severe loading then this will trigger either a net loss or gain in bone volume ²⁸. A typical stainless steel fracture plate with a modulus of elasticity approximately one order of magnitude greater than that of cortical bone ⁶⁵ can result in a "stress shielding" effect when it is rigidly affixed to bone for the purpose of repairing a fracture. The long term implication of this is that during the remodelling phase of fracture healing the bone directly beneath the fracture plate is not being subjected to its pre-fracture, normal strain values as a result of the stiffer fracture plate. The end result is bone resorption directly beneath the plate ⁶⁵. A paper by Carter et al. was able to

demonstrate in-vivo how sensitive the remodelling process is to small changes in strain. Using canine femurs fixed with steel plates, they were able to show how a difference in applied stress of 1 MPa, representing less than 1% of the ultimate strength of bone, resulted in observable remodelling differences in bone.¹⁰

1.4 An Introduction to Rigid Internal Fracture Fixation Plates

1.4.1 The History of Internal Fracture Fixation

Before the beginning of the 19th century, fracture treatment was more art than science. The treatment standard at that time was internally manipulating the fracture bone so that the fragments were in close proximity, something most often performed by self-proclaimed “bone-setters”. Bone setting was an empirical art and no attempt was made by its practitioners to understand why it worked or how to improve it. Fracture healing was gauged by the presence of a palpable periosteal callus and no undue motion in the affected bone. Internal fracture fixation did not become widespread until the introduction of antiseptic and aseptic techniques reduced the risk of wound infection. The first reported use of a metallic fracture fixation plate was by Hansmann in 1886⁴³. Hansmann’s plate and screws (Figure 1.5) were designed to allow plate removal without re-opening the incision. British surgeon William Lane introduced a metal fixation plate in 1907 (Figure 1.5) and used it widely in his practice. Lane was confident enough with this technique to include it in the second edition of his book *The Operative Treatment of Fractures*. The next major advance in fracture plate design was by William Sherman, who used his connections to the steel industry to produce a stronger, more biocompatible

plate (Figure 1.5) made out of vanadium⁵⁷. Up until this point the plates in use did nothing to aid in aligning or "reducing" the fracture fragments. The first people to address this issue were Townsend and Gilfillian who developed a plate with slotted as opposed to oval holes such as the ones used in the Shermann plate⁶⁷. The plate was applied across the fracture with the screws loosely tightened into the bone. The fracture could then be reduced and the screws would then secure the plate firmly to the bone. Robert Danis⁸ improved on the Sherman plate and produced a plate design called the coapteurs. The coapteur plate was the first plate that effectively combined rigid fixation with axial compression. Danis' plate uses an axially oriented screw located within the plate itself in order to provide compression at the fracture site. In 1956 Bagby developed a compression plate combined oval holes with an integral ramp to produce a compression effect without the use of additional plate hardware. The basic premise behind the Bagby plate has been refined into the dynamic compression plate that is in use today. It is interesting to note that the studies done with the Bagby plate (Figure 1.6) showed that the enhanced healing associated with the induced compression was due primarily to the more secure fixation and close apposition of the fragments rather than the compression itself⁶

Hansmann (1886)

FIG. 1-A

Lambotte (1909)

FIG. 1-B

Lane (1914)

FIG. 1-C

Sherman (1912)

FIG. 1-D

Figure 1.5: Early fracture plate designs ⁶

Figure 1.6: Bagby compression plate ⁶

1.4.2 Mechanism of Action of Compression Plates (DCP and LC-DCP)

Since part of the experimental investigation will involve testing of the dynamic compression plate (DCP) it is important to have a rudimentary understanding of how this plate works. The cortical screws that are used with the DCP have spherical heads that fit into the oval holes in the DCP. The oval holes in the DCP are inclined such that when the cortical screw is tightened, the screw slides down the inclined surface of the hole causing the bone fragment to which the screw is attached to displace along the long axis of the plate. Figure 1.7 illustrates this process graphically. This can be used to produce direct contact (apposition) and compression of the fracture components. ^{52: 63}

Figure 1.7: Mechanism of action behind the dynamic compression plate ⁵²

The axial force that produces compression of the bone fragments depends on the slope of the hole, the coefficient of friction between the plate and the bone, the coefficient of friction between the screw head and the plate, and the amount of torque applied to the screw. It should be noted that the friction developed between the plate and bone can have an effect on the mechanics of a fracture plate. If the bone/plate frictional force is large, then there is less loading of the screws which enhances their fatigue life. On the other hand, a smaller frictional force allows for easier movement between the plate and bone which results in greater compression at the fracture site. This plays an important role in providing mechanical stability to the fracture site but it may result in larger bending stresses being supported by the plate. From a practical point of view, the bone screws used with fracture plates should be as tight as possible, since this protects them from fatigue as well as increasing the compressive force applied to the bone fragments.

The low contact dynamic compression plate (LC-DCP) relies on the same principle as the DCP except that additional material has been removed on the undersurface of the plate in an attempt to decrease the bone-plate contact area and thus to improve blood flow to the fractured bone. In addition, the LC-DCP is typically manufactured out of titanium in an attempt to reduce the axial stiffness of the plate as compared to a stainless steel plate. Although no study to date has confirmed that these changes result in improved fracture healing,^{32: 33} both these plates are considered to be the current standard in fracture fixation.

1.4.3 Benefits of Rigid Internal Fracture Fixation

One of the key principles in achieving the rigid internal fixation of fractures has been to generate compression at the fracture site. It has been demonstrated that significantly greater bending stiffness is achieved when compression is applied across the fracture line²⁶. Achieving compression across the fracture site is also important in protecting the plate from failure. For the three primary loads that a long bone must bear - axial compression, bending and torsion - compression of the fracture components is a crucial mechanism in ensuring that forces are transmitted through the bone as opposed to through the fracture plate. A compressive load acting on a bone with a large fracture gap produces a significant bending moment in the plate. This large bending moment is transmitted to the screws and the end result may be fatigue fracture of the plate, pullout of the screws, or a combination of both failure mechanisms. Compression also aids in resisting torsion applied to the bone about its long axis, by creating interdigitation and frictional contact of the fracture ends²⁶. From a clinical perspective, the main advantage

of compression fixation was the early mobilisation of neighbouring joints. Although not currently recognised as a benefit it is important to reiterate that the healing of fractures treated by rigid compression occurred through primary union, without cartilage formation and without the development of a periosteal callus, results in poor physiological healing ^{34: 52: 58: 59}.

1.4.4 Problems with Rigid Internal Fracture Fixation

The growth and differentiation of pluripotential tissue into bone forming cells is greatly influenced by mechanical factors ⁴⁵. In fact, elimination of interfragmentary micromotion as achieved by rigid plate's results in the absence of periosteal callus formation. The most potent factor for callus formation is cyclical transmission of axial loads ^{27: 74: 75}. The disadvantages of rigid fixation include: delayed healing, possible refracture after plate removal ^{7: 15: 20: 30: 38: 39: 54: 56}, stress shielding of the plated segment ^{49: 70-73}, suppression of cyclical micromotion at the fracture site and the impossibility to follow the progress of fracture healing radiologically due to the absence of periosteal callus.

A large number of researchers ^{4: 5: 8: 15: 17: 31: 40} have been able to show that the use of less rigid plates can improve the fracture healing process. In most instances the researchers did not assess the degree to which the plate stiffness was decreased ^{46: 50: 61: 62: 69: 79: 80}. It is very difficult to compare the varied methods used to reduce fracture plate stiffness since some researchers focussed on using lower modulus materials, while others focussed on changing the cross-sectional area of the plates in order to modify the plates'

moment of inertia. Still other researchers resorted to combining both structural and material methods. Moreover, the majority of researchers, especially those that focussed on material methods tended to reduce rigidity in all principal directions (axial, bending and torsion) as opposed to just focussing on reducing axial rigidity.

The experimental investigations of Woo et al ^{74: 75} and of Goodship and Kenwright ²⁷, showed conclusively that cyclical motion in the axial plane was a powerful stimulator of postfracture bone remodelling. Sun et al ⁶⁰ and Panagiotopoulos ⁵¹ built on this concept and developed a “sliding” plate that allows for motion in the axial direction while maintaining a high torsional and flexural rigidity. These plates consist of two parts sliding against each other at the fracture site. Their potential disadvantage is a locking of the sliding mechanism under bending and torsion. One of the most successful designs to capitalise on the research of Goodship and Kenwright ²⁷ has been the axially compressible plating system designed and tested by Foux et al ^{23: 24}. Foux et al tested their fracture plate systems in 72 canines and was able to show a definite advantage over DCP plates.

1.5 Literature Review - Testing of Fracture Fixation Plates

Various approaches have been taken to determine the mechanical properties of fracture fixation plates. These approaches can be roughly grouped into three different categories depending on the type of testing that was done. The first group is composed of researchers who focussed exclusively on determining the mechanical properties of the fracture plate itself, without considering how the fracture plate would interact with bone.

The second group is composed of researchers who used bone or bone substitutes in order to determine the mechanical properties of the fracture plate/bone system. The third group of researchers focussed on performing both in-vivo and in-vitro testing on their fracture plates in order to determine whether a correlation existed between fracture healing and the mechanical properties of the fracture plate.

1.5.1 Mechanical Testing of Fracture Fixation Plates

In general when researchers have resorted to testing only the fracture plate, they are generally doing so to meet International Standards Organisation (ISO) standards or to prove that their design meets some specific strength criteria ^{2: 60}. The data obtained from this type of testing is limited as it gives no information about the effect the fracture plate will have on the mechanical environment of the bone. Sun et al ⁶⁰ used a four-point test protocol in accordance with ISO 9585 ¹ in order to verify the performance of their new axially mobile plate (AMP) design. Their test method allowed them to determine the equivalent bending stiffness and the bending strength of their plate design. Although this information is useful in terms of satisfying the strength criteria mandated by governmental agencies it did not prove that their plate design worked in-vivo. This would have been especially helpful in the case of Sun et al ⁶⁰ as their plate was specifically designed to improve fracture healing by allowing for a degree of axial micromotion. Research using a similar design methodology was conducted by Abel² on a minimum-contact fracture (MCP) plate that was designed to improve cortical blood flow when compared to DCP or LC-DCP plates. They, subjected their plate to a four-point bending test in order to meet the requirements of ISO 9585 ¹. Although they were able to

make some comparisons regarding the strength and stiffness of their plate design in comparison to DCP and LC-DCP plate designs, they were not able to prove anything about the in-vivo effectiveness of their design.

1.5.2 Mechanical Testing of Fracture Plate/Artificial Bone Constructs

A more effective method for determining fracture plate performance is to mount the fracture plate on an instrumented bone or bone substitute and subject the plate/bone construct to various loading situations. By properly instrumenting the bone and/or the plate it is possible to learn much more about the way a specific fracture plate modifies the mechanical environment of the bone. Several researchers^{39, 41, 58, 77} have used this approach in their fracture plate research. Woo et al⁷⁷ used this approach in a study designed to provide information regarding the appropriate design parameters for fracture plate systems. Woo fabricated a magnesium tube, 20 cm long with a 1.50 cm outer diameter and a 1.20 cm inner diameter that served as an artificial bone onto which various types of fracture plates were affixed. The magnesium tube was drilled and tapped to accept standard 3.5-mm cortical screws and was then instrumented with strain gages. Two strain gages were located on the bone itself, one on the side of the magnesium tube parallel to the tubes long axis and one directly opposite the fracture plate. A third strain gage was bonded to the top surface of each fracture plate. One of three different fracture plate designs were rigidly affixed to the magnesium tube using cortical screws and the tube was then loaded in both axial compression and four-point bending. The fracture plates being used included a very rigid stainless steel plate that served as a control and two experimental plates, one made out of titanium and one tubular

plate made out of stainless steel. Both of the experimental plates had axial and bending stiffnesses less than half those of the control plate. Since Woo did not instrument the anterior surface of the magnesium tube, below the fracture plate, a finite element analysis (FEA) of the magnesium tube/plate system was also performed in order to investigate strain fields in that region. The results of Woo's experiment showed that the rigid control plate resulted in a significant degree of stress shielding when compared to both the tubular steel and the titanium plate. In conclusion Woo suggested that in order to enhance the environment for fracture healing a fracture plate should have moderately high bending and torsional stiffnesses but low axial stiffness⁷⁷.

Simon⁵⁸ and Korvick³⁹ both performed similar experiments using aluminium tubing instead of magnesium. The experiment by Simon⁵⁸ was very similar to that performed by Woo⁷⁷ in that both mechanical and FEA were performed. In Simon's experiment a titanium fracture plate was mounted onto an aluminium bone model. The aluminium bone was instrumented with strain gages bonded underneath the fracture plate and directly opposite the plate. A third strain gage was mounted on the side of the aluminium tube at right angles to its long axis, in order to measure circumferential strains. In order to provide clearance for the anterior strain gage, plastic spacers were placed underneath the fracture plate to lift the plate off the bone. No analysis was done to quantify the effect that this would have on the performance of the bone/plate system. The plate/bone model was then loaded in four-point bending to a maximum applied load of 1864 N. The purpose of Simon's research⁵⁸ was to clarify the complex nature of the stress field in a plated bone. As such they did not draw any conclusions regarding the

design of new fracture plates. Korvick³⁹ also used an aluminium bone model to examine the effect of screw removal on bone strain. For this study Korvick instrumented an aluminium bone model with strain gages directly underneath and directly opposite the fracture plate. Although screw removal is not a clinically viable option, Korvick's results demonstrated that an instrumented aluminium bone model would be an excellent test platform for studying changes in bone strain caused by changing different fracture plate parameters – in their case, screw removal.

Kowalski⁴¹ also performed a similar type of experiment to that of Sun⁶⁰ and Abel² in order to test the mechanical properties of a noncontacting plate (NCP) used for tibial fixation. In the case of Kowalski, the NCP was mounted on a plastic tibia model and then subjected to axial compression, four-point bending and torsion. The plastic tibia was not instrumented with strain gages and only the displacement transducers on the testing machines were used to determine the mechanical properties of the bone/plate construct. Kowalski's experiment was interesting since he was the only one to perform an osteotomy on the test bone, and he was also the only one to test two different configurations for the model bone. Kowalski described the first configuration as “contact osteosynthesis” where both fragments of plastic composite tibia were in direct apposition. The second configuration was described as “bridging osteosynthesis” where a 5-mm gap was left between both fragments. The results of Kowalski's experiment showed that the NCP plate has properties approaching those of a DCP plate in most loading modes in both the contact and bridging osteosynthesis model. The most significant difference occurs with the bridging osteosynthesis configuration when the

NCP plate is loaded in torsion. Perhaps the most interesting results are achieved in studies where in-vitro mechanical testing is correlated to, in-vivo testing.

1.5.3 Mechanical and In-vivo Testing of Fracture Plates

The most informative experimental design is the one where in-vivo testing is combined with mechanical testing. This has allowed researchers to make some correlations between the mechanical properties of various fracture plates and their influence on healing in-vivo. Although this research is significantly more expensive than simple mechanical testing, several researchers have published the results of these types of studies^{3: 7: 23: 24: 42: 64: 65}.

In 1976 both Tonino⁶⁵ and Akeson³ published papers examining the biomechanical and histological effects of rigid versus non-rigid internal fixation plates. Both these experiments involved performing osteotomies on canine femurs and plating the osteotomies with either a standard 316L stainless steel fracture plate or a fracture plate made out of a composite material. In the case of Tonino⁶⁵ a plate composed of polytrifluoromonoethylen (PTFCE) was used and in the case of Akeson³ a graphite fibre methylmethacrylate (GFMM) resin plate was used. In both cases the composite/plastic plate had a modulus of elasticity of at least one order of magnitude lower than that of the metallic plate. In addition, as a result of the composition and design of these composite plates the bending and torsional stiffnesses were reduced as well as the axial stiffness. The results of both studies showed quite conclusively that there was a significant decrease in both bone mass and in bone mechanical properties for

the femurs plated with the metallic versus the composite fracture plates. These early experiments were groundbreaking in some ways since they conclusively showed that rigid internal fixation provided inferior biomechanical and histological results when compared to less rigid fixation methods. What these studies did not provide was any information regarding the ideal rigidity of a fracture plate in order to optimise fracture healing. One of the first people to perform a study examining the effect of fracture plate mechanical properties on fracture healing was Bradley ⁷ in 1979. Bradley's research involved producing seventeen fracture plates, with a range of bending stiffnesses from 1.4 to 16.8 Nm², but all having similar axial stiffnesses. The conclusion Bradley arrived at was that the resultant strain in the healing bone is relatively independent of the bending stiffness of the plate, but is substantially affected by changes in axial strain. This eventually led to the studies by Terjesen, Foux, and Yeadon ^{23, 24, 64} that set about examining the issue of what constituted the ideal axial stiffness for an internal fracture fixation plate.

The study by Terjesen ⁶⁴ involved using four different fracture plates to heal rabbit tibias subjected to experimental osteotomies. The four fracture plates varied in both their bending and their axial stiffnesses. The plates ranged in axial stiffness from 490 kp/mm (4806.9 N/mm) to 26 kp/mm (255.1 N/mm) and they ranged in bending stiffness from 29.4 kp/mm (288.4 N/mm) to 1.5 kp/mm (14.7 N/mm). Although all the rabbit tibias healed, the results showed that the two plates with the intermediate bending and axial stiffness provided for optimal fracture healing ⁶⁴. The only possible criticism of this excellent study was the fact that they used a rabbit model and the fact that the plating

systems differed in both bending and axial stiffness. The study by Foux and Yeadon^{23, 35} was similar to that done by Terjesen⁶⁴ except that a canine model was used along with four plating systems that varied only in terms of their axial stiffness. Both in-vivo and in-vitro testing was conducted on these canine axially compressible plates (ACP).

The decision was made to acquire and re-test the canine plates used by Foux and Yeadon in order to verify the new mechanical testing methodology. Since in-vivo testing was performed with these plates, it was felt that the additional in-vitro testing results might shed some light on which mechanical properties of the canine plates were responsible for optimal fracture healing. As such it is important to understand in greater detail the methodology and results behind Foux and Yeadon's experiment.

1.5.3.1 Canine Axially Compressible Plate (ACP) System

The goal behind Foux and Yeadon's study^{23, 24} was to design a fracture plate that provided adequate bending and torsional stiffness while allowing for an enhanced degree of axial compressibility. To do this they designed a new type of fracture plate system. Their system was designed specifically to fit canine anatomy in order to facilitate an in-vivo study.

Each canine ACP fracture plate consisted of three components: stainless steel plate, polymer inserts, and stainless steel screws. The canine ACP was designed to meet two opposing criteria: provide high flexural and torsional stiffness while at the same time

reducing axial stiffness. The design of the stainless steel plates provided high flexural and torsional stiffness, while the polymer inserts functioned to reduce the axial stiffness.

Three plates manufactured from 316 LVM implant grade stainless steel were used. All three plates had six holes through their thickness. The plates differed only by the shape of these holes. The first plate type had spherical 3.6 mm diameter holes. The second type of plate had elongated holes with a centre to centre distance of 0.75 mm (Figure 1.8). The third type of plate also had elongated holes, but the centre to centre distance of these holes was 1.50 mm. The elongated holes were designed to accept polymer inserts of different sizes. The smaller insert had 0.75-mm thick "cushion" that rested against the cortical screw (Figure 1.9) while the larger size of insert had a 1.5-mm cushion. All three plate types were 56 mm long and 9.5 mm wide. The bottom surface of all plates was contoured to approximate the radius of a beagle femur. All cross-sections without screw holes had an area of 48.2 mm^2 , a flexural rigidity about a tangential centroidal axis of 17.3 Nm^2 , a flexural rigidity about a radial centroidal axis of 177.2 Nm^2 , and a torsional rigidity of 74.6 Nm^2 .

The polymer inserts were manufactured in two different sizes, in order to fit into the different sizes of elongated holes. As illustrated in figure 1.9, these inserts took the form of a long, flat washer with a crescent shaped bumpers protruding from the bottom surface. The inserts were manufactured using either low density polyethylene (LDPE) or polymethylmethacrylate (PMMA). These materials were chosen for their proven

biocompatibility and for their compressibility. The elastic modulus of the LDPE was approximately 0.20 GPa while that of the PMMA was approximately 3.10 GPa.^{24, 78}

Custom flat head cortical bone screws were designed for use with the ACP fracture plate system. The screws were machined from 316 LVM implant grade stainless steel with a thread length of 16 mm. These custom screws were manufactured with cross-slotted flat heads and a 3.30-mm length of unthreaded shank immediately beneath the heads. The unthreaded shank was intended to bear against the bumpers protruding from the inserts and to compress the bumpers slightly when subjected to an axial load. The amount of compression varied depending on the material and the thickness of the cushion portion of the insert. Table 1.1 outlines the four different configurations of the canine ACP fracture plates that were used in the experiment:

<i>Name</i>	<i>Insert Material</i>	<i>Insert Size</i>
System 1: Round	Countersunk holes – no insert	N/a
System 2: PMMA059	Polymethylmethacrylate	1.5 mm
System 3: PMMA030	Polymethylmethacrylate	0.75 mm
System 4: LDPE030	Low Density Polyethylene	0.75 mm

Table 1.1: Various configurations for the canine ACP

Some preliminary mechanical testing was performed on the four canine fracture plate systems in order to determine both the axial and bending stiffness of each plate. A summary of the mechanical testing results is given below in table 1.2⁷⁸:

<i>Plate Type</i>	<i>Axial Stiffness (N/mm)</i>	<i>Bending Stiffness (N/mm)</i>
ROUND	13 910	7 772
PMMA030	8 390	7 482
PMMA059	4 260	7 559
LDPE030	1 640	7 795

Table 1.2: Results of mechanical testing on canine ACP ⁷⁸

The canine ACP plates were tested in 72 dogs that underwent unilateral femoral osteotomies. Eighteen dogs were plated with each of the four plating systems. Groups of dogs were killed at three different time points, 8 weeks, 16 weeks and 24 weeks. The osteotomised femur and the contralateral control were explanted and subjected to a non-destructive four-point bend test ^{23: 24: 78}. Twenty-four separate bending tests were conducted for each of the bones. Bending tests were conducted at 15-degree increments around the circumference of each bone. The results of the four point bending tests were used to produce a derived value called the stiffness index. The stiffness index (SI) was used to classify the quality of bone healing with each of the different plate types.

$$SI = \left[\frac{EI_{os}}{EI_c} \right]_{Minimum}$$

Equation 1.3

Where SI is the stiffness index, EI_{os} is the flexural rigidity of the healing bone and EI_c is the flexural rigidity of the contralateral control bone. The ratio shown in equation 1.3 was calculated at each 15-degree increment around the circumference of the bone.

Twenty-four different ratios were calculated and the minimum value of this ratio was termed the stiffness index ^{23: 24: 78}. The closer the stiffness index (SI) is to one, the closer the mechanical properties of the operated bone approached those of the control bone.

The stiffness index results for the canine ACP bones are displayed in Table 1.3:

Plate Type	Stiffness Index
ROUND	0.70
PMMA059	0.85
LDPE030	0.72
PMMA030	0.80

Table 1.3: Stiffness index values for the various canine plates

By plotting stiffness index versus healing efficiency (SI) the researchers involved in this study were able to conclude that optimum fracture healing would most likely be obtained with a plate that had axial stiffness properties somewhere between those of the PMMA030 and the PMMA059 plate.

Figure 1.8: Engineering drawing of a canine fracture plate with 0.75-mm holes

Figure 1.9: Typical engineering drawing of the corresponding polymer insert

2 CONCEPTUAL DESIGN

2.1 Problem Statement

In order to be able to measure how a particular fracture plate design changes the mechanical environment of the underlying bone it is necessary determine the changes that occur during both the union and the remodelling phase of fracture healing. During the union phase of fracture healing the variable of the greatest interest is the amount of axial micromotion allowed by the particular plate design. During the remodelling phase of fracture healing the variable of greatest interest is the degree of stress shielding that occurred in the area directly underneath the test plate. Any new test method would have to be able to simulate both these phases of fracture healing.

2.2 Design Overview

The results of the literature review made it very clear that in general, the testing of fracture plates has been limited to either testing the plate in isolation, or to mounting the plate onto an artificial bone and subjecting the plate/bone construct to various loading modes. Although testing the fracture plates on a single artificial bone did not appear to be the final solution, the methodologies that had been used were important since the testing of an intact bone essentially represented a bone in the remodelling stage of fracture healing. With the exception of Kowalski ⁴¹ there was no effort made to

investigate how a fracture plate performs during the union phase of fracture healing.

Kowalski used a composite femur model in one of two configurations to test his fracture plate design. One model had the ends of the osteotomized bone in contact and the other had a 5mm fracture gap. It is interesting to note that Kowalski made no effort to measure axial micromotion or microstrain, two important factors in fracture healing. It was Kowalski's model that eventually led to the proposed design. The use of Kowalski's "contact osteosynthesis" model, where the two osteotomized bone fragments are in contact was rejected. This model did not accurately simulate bone during the remodelling phase of fracture healing and in addition, instrumenting the bone with strain gages would have been impossible.

As a result of the design review it was decided to produce two separate models. identical in all ways except that one would have a defined fracture gap. The model without the fracture gap would represent the bone during the remodelling phase of fracture healing and the model with the fracture gap would represent bone during the union phase of fracture healing.

2.3 Design Criteria

In order to meet the requirements outlined in the problem statement, several design criteria had to be met by the proposed design. Since several different types of fracture plates would be tested with the proposed testing system, ease of manufacture would be an issue. The final testing system should be produced from an easy to obtain and economical material and it should be easily machinable using standard machine

tools. In addition due to the desire to avoid the effect of confounding variables, the design should contain features that lend themselves to precision manufacture. From an instrumentation point of view the material used should be readily compatible with standard strain gages.

2.4 Proposed Design

In order to solve the problem of testing a fracture plate during both the union and the remodelling phase of fracture healing, it was proposed to design two separate test fixtures. The basic test fixture for both tests would be a machined bone substitute with fixed overall dimensions and a specific inner and outer diameter. To test the fracture plate during the union phase of fracture healing, the basic test fixture would be cut into two pieces, thus simulating a simple fracture. The metal test bone would have drilled and tapped holes spaced in such a way that they matched the hole spacing of the particular fracture plate being used. Once the fracture plate has been secured to the bone model with the proper cortical screws, the design of the metal test bones would ensure that there would be a 3-mm gap present at the midpoint of the bone model. The benefit of this is that the metal test bone would be completely unstable until a fracture plate was rigidly attached to it. The fact that there is no mechanical connection between the two bone fragments is the reason that this bone model simulates the conditions inherent with a simple fracture during the very beginning of the union phase of fracture healing.

In order to measure the effect of various fracture plates during the remodelling phase of fracture healing a second metal bone model was created. To reduce

manufacturing costs. the same basic bone model was used except that this model was not cut into two halves. Otherwise, the inner and outer diameters of the two bone models were identical. The only modification made to this metal bone model was the addition of a small, machined landing in the area directly underneath the midpoint of the fracture plate. The purpose behind this small landing was to allow for strain gage placement directly underneath the fracture plate, at a location that corresponded to the location of the fracture site in-vivo. Although the machined landing did result in small stress concentrations at the edges of the cuts, the strain gage was located at a distance more than five times that of the cut radius, reducing the effect of this stress riser. Thus, this method was deemed superior to that used by Simon et al ⁵⁸ where a plastic spacer was used to raise the fracture plate off the bone. Proper fracture plate application is very sensitive to many factors, including bone/plate friction and the addition of a plastic spacer would have changed the mechanism of action of the fracture plate. This would have been especially crucial in the case of the canine plates, where screw insertion torque was limited and the addition of a plastic spacer may have radically changed the effectiveness of the canine plates. Although the argument can be made that true surface strain was not measured, the results that were obtained were consistent from test to test and they allowed for comparisons between different types of fracture plates, without changing the mechanism of action of the plates.

2.5 Material requirements

Several different materials were evaluated and prototyped. From a design standpoint the ideal material would have an elastic modulus similar to that of bone and it

would be compatible with standard resistance wire strain gages. From an economic standpoint the ideal material would be easy to machine, readily obtainable, and be of relatively low cost. The literature search that was conducted showed that a wide variety of materials had been used successfully for this purpose.

2.5.1 Material Selection

2.5.1.1 Cadaveric bone

The first material that was evaluated was human bone. Initially, it appeared that this would be the ideal material for this research. However, there are certain technical difficulties that exist when using human bone. The use of cadaveric material involves certain potential health risks during handling and transporting the material. Secondly, the material properties of cadaveric bone are sensitive to preservation and storage conditions. Lack of moisture during storage or formalin preservation can both strongly change the characteristics of human bone. In addition, there can be a significant amount of variation in surface geometry, bone geometry and material properties between various bone samples increasing the risk of adding confounding factors to the analysis. When the high price of obtaining this cadaveric bone is factored in, using cadaveric bone specimens was not an option.

2.5.1.2 Polycarbonate

The second material that was investigated was polycarbonate tubing. Several different diameters of polycarbonate tubing were available, making this a very cost-effective solution for producing the bone models. From a material perspective the elastic

modulus of polycarbonate (approximately 3.0 GPa) was somewhat lower than that of bone (17-20 GPa). However, polycarbonate had been used successfully by Cheal¹⁴ and thus the decision was made to produce a prototype. Several problems became apparent with the polycarbonate prototype. Due to the nature of the material, stress cracking at the screw holes was soon a problem; in addition, although strain gage application was not overly difficult internal stresses in the material caused spurious strain readings. Although the polycarbonate prototype allowed for the testing of the strain gage setup it was decided to use a different material for the final test bones.

2.5.1.3 Brass

Brass was the next material that was evaluated. From an economic perspective a bone model made out of brass would have been ideal. Brass is relatively inexpensive, easy to obtain and very easy to machine. From an engineering perspective there was some concern regarding the relatively high modulus of brass (approximately 117 GPa) as compared to bone. Whereas polycarbonate had a modulus approximately five times less than bone, brass had a modulus five times greater. Given the ease of machining and the availability of the raw material, a brass prototype was produced. The brass prototype was quite functional, there was no stress cracking at the screw holes and the material was highly compatible with the strain gage adhesives. The initial compression pilot test showed that the strain gage readings were quite stable and very repeatable over time. One problem was the low strain values that occurred as a result of the low axial loads mandated in the experimental protocol. At these low axial load levels (441.45 N) the strain gages were registering approximately 5 microstrain. Although the pilot study

showed that microstrain readings at this level could be recorded reliably, there was some concern about the effect that experimental error and low signal to noise ratios would have on the ability to detect differences in surface strain between various types of fracture plates. Instead of increasing the applied load by at least one order of magnitude, it was decided to select a material with a lower modulus.

2.5.1.4 Aluminium

Initially, magnesium appeared to be a likely candidate since it had an elastic modulus of approximately 45 GPa. However, magnesium was ultimately rejected on the basis of its higher cost and availability difficulties. The next choice was one of the aluminium alloys. The 6061-T6 aluminium immediately appeared to be a likely candidate due to its low elastic modulus (approximately 70 GPa), its ease of machinability and its low cost. In addition aluminium bone models had been used with success by other researchers^{39: 58}. As such it was decided to proceed with the creation of an aluminium prototype. The aluminium prototype proved to be very successful. The cost to purchase the material and to machine it was extremely reasonable. Although strain gaging the aluminium bone was not as easy as with the brass or polycarbonate prototypes, clean, reproducible strain signals were obtained.

2.6 Design Constraints

Several different design constraints had to be satisfied in determining the final dimensions for the aluminium bone model. It was decided that the overall shape of the bone should be cylindrical in order to generate a more uniform strain pattern along the

length of the bone. The next step was to determine the design constraints for the aluminium bone model in order to facilitate the design process.

2.6.1 Design of Cylinder Ends

Of the three loading modes that were applied to the aluminium bones, torsional loading provided the most rigid design constraints. For the torsion testing an Instron 8521 bi-axial testing machine was used. The torsional grip interface on the Instron was comprised of two sockets mounted on both the upper and lower crosshead of the Instron. The sockets measured 31.75mm in diameter and were 63.5mm deep. Each socket had 12.7mm hole cut transversely through the socket to allow for the use of a shear pin. The sockets were provided on the Instron as a way of facilitating the testing of cylindrical specimens under 31.75mm in diameter. Thus in the design of the aluminium bones it was decided to oversize both ends of the bone to 31.75mm in diameter and to drill a 12.7mm hole through either end. This allowed the aluminium bone models to be secured to the Instron by passing shear pins through both the upper and lower holes; no additional test fixtures were required.

2.6.2 Design of main shaft

The design of the main shaft of the aluminium bone proved to be more difficult. Several important variables, such as the outer and inner diameter of the cylinder had to be carefully calculated since these values would have an impact on the surface strain. In addition, the tests with the polycarbonate model showed that to place a strain gage

directly underneath the fracture plate. a flat landing would have to be machined in this area to ensure that the fracture plate would not interfere with the strain gage.

2.6.2.1 Engineering calculations

In determining the dimensions of the main shaft of the aluminium bone, there were two issues of concern. The first was the amount of microstrain that would be generated given the applied load and the second was the size of the landing for the strain gage. Although at first these two issues appeared independent of one another, they were in fact inextricably linked together. Changing the inner/outer diameter of the aluminium bone, modified the cross-sectional area and thus the amount of axial deformation (microstrain) that occurred under load. This also changed the amount of available space from which the landing could be machined (dimension "c" on Figure 2.1) into the aluminium bone. In order to be able to solve this problem in an iterative fashion both the magnitude of the applied load and the size of the strain gages were fixed. Thus the design solution now had to allow for a strain gage measuring 8mm x 5mm to fit onto the landing and it had to allow for axial deformation (microstrain) of at least 15 microstrain in order to ensure adequate signal strength.

Figure 2.1: Cross-sectional diagram of aluminium bone with landing

Given the clinically relevant load of 441.45N (45KgF), it was possible to estimate the amount of microstrain that would be generated in the aluminium bone for a given wall thickness. The maximum width of the landing ("c") was now constrained only by the outer and inner diameter of the cylinder and "h", the height of the landing. The height of the landing ("h") had to provide enough room to ensure that the strain gage leads did not contact the fracture plate and cause a short circuit.

The width of the landing could easily be calculated from the equation for the chord of circle:

$$c = 2 \cdot \sqrt{h \cdot (2 \cdot r - h)}$$

Equation 2.1

where c is the chord length, r is the radius of the circle and h is the perpendicular height from the chord to the circle. The key dimension is h, which is the perpendicular distance from the chord to the circle. It was critical that h not be too small as this could result in a short circuit between the strain gage and the fracture plate.

The perpendicular height h, can also be found from equation 2.1:

$$h = r - 1/2 \sqrt{4r^2 - c^2}$$

Equation 2.2

It was decided to pick a value of h and then use that value to determine c. The landing height h was set to a value of one-half the thickness of the aluminium cylinder. This ensured that the landing would be thick enough to machine easily, without worrying about breaking through the tube during the manufacturing process. With the value for h set it was possible to calculate a value for the chord length c for the aluminium tube given its inner and outer diameters.

Obviously some additional constraint was needed in order to limit the number of possible inner and outer tube dimensions. An outer diameter of 2.0 cm was selected

based on the results of previous studies^{39: 58} and based on geometrical compatibility with a wide range of fracture plates. An iterative solution was found where each set of bone dimensions was then used to calculate axial strain values. The final dimensions of the landing ensured that the strain gage was located 1mm away from either edge. The radius of the edge was machined to less than 0.127mm, ensuring that the stress concentration would have minimal impact on the operation of the strain gage. A sample calculation is given in Appendix A. The result of this design analysis is a set of aluminium bones with specific dimensions as detailed in Appendix B.

In order to be able to simulate the union phase of fracture healing, one group of aluminium bones were machined without landings for strain gages and then cut in half. This model was termed the GAP bone model, the uncut version of the aluminium bone was termed the WHOLE bone model.

2.7 Design of test fixtures

Since the bone models were going to be tested in three distinct loading modes, it was necessary to devise testing fixtures that would allow both the GAP and the WHOLE bone models to be tested in all three loading modes: axial compression, four-point bending and torsion. From a design standpoint the easiest test fixture to design and produce was the torsional test fixture.

2.7.1 Torsional Test Fixture

No actual fixture had to be designed in order to be able to test both bone models in torsion as the enlarged portion at either of the bone models were sized to fit directly

into the test fixtures on the Instron 8521 bi-axial testing machine. The transverse holes in the end portions of the shaft allowed a shear pin to securely fix both the upper and lower part of the bone model to the testing machine. Unfortunately, the test fixtures for axial compression and four-point bending were not quite as simple to produce.

2.7.2 Compression Test Fixture

The purpose behind the compression fixture was to secure the aluminium bone models to the moving crosshead and to the base of the MTS machine. It was also important that the design of the compression fixture help ensure that the line of action of the compressive load was consistent between each of the aluminium bone models. Since the load cell is not capable of measuring bending moments any significant offset from the main axis of the load cell would result in inaccuracies in the force measurements. In order to minimise the degree to which the line of action of the force might be offset, an alignment system was designed that would account for any misalignment between the moving crosshead and the base of the MTS machine. It was important that the design of the compression apparatus should also allow each aluminium bone to be placed in a consistent position. For the purposes of this test, it was decided that each aluminium bone would be positioned in such a manner that the flat anterior landing underneath the fracture plate would be parallel with the front of the testing machine. This requirement imposed a rigid constraint on the design of the compression fixture. The implication behind this was that the rotational alignment of the aluminium bone had to be adjusted independently from its fixation to the load cell.

The complete compression fixture was composed of several pieces of equipment that served to either attach the aluminium bone to the MTS machine or to ensure proper alignment of the aluminium bone. In order to attach the aluminium bone to the MTS machine, two separate fixtures were designed. The MTS/Aluminium bone interface fixture, shown in Figure 2.2, was made from stainless steel rod stock and had one end that was externally threaded (3/4-10 UNC) and the other end that was machined down to 12.7-mm (0.5-in) in diameter. Once the MTS/Aluminium bone interface unit was screwed into the aluminium bone, it was held in place with a set-screw.

Figure 2.2: Exploded view of Interface Fixtures

The MTS connector interface was designed to allow the aluminium bone to connect to the MTS machine. This adapter had one end that was externally threaded (1/2-20 UNF) so that it could be threaded directly into the load cell. The other end of this adapter had a 12.7mm diameter hole that was designed to accept the 12.7-mm rod on the MTS/bone interface unit. An assembly drawing of this fixture is shown in figure 2.3. Another set-screw was used to secure the aluminium bone with the adapter into the MTS interface.

The benefit of the design is that by releasing the set-screw on the MTS connector the aluminium bone could have either its rotational position adjusted or it could be removed completely from the MTS adapter.

The next important part of the fixture was designed to allow for the alignment of the aluminium bone models under a compressive load. Figure 2.3 graphically illustrates that underneath the aluminium bone is a ball bearing that rests in a spherically cut baseplate. The radius of the spherical cut in the base is equivalent to the length of the aluminium bone. As a result the line of action of the load is now in line with the long axis of the aluminium bone.

Figure 2.3: Assembly drawing of compression fixture

2.7.3 Bending Test Fixture

The purpose behind the four-point bend fixture was to be able to apply a precise bending moment to each of the aluminium bone models. From a design perspective the four-point bending apparatus had to accommodate the geometry of the bone models and allow for repeatable measurements of both surface strain and deformation to be made. In order to measure midspan deflection, the four-point bend fixture was designed to accept a linear variable differential transformer (LVDT). The aluminium bones were supported in a custom designed cradle on cylindrical supports located 177.8-mm (7-in) away from one another. The load platens were attached to a separate baseplate that could be secured to the load cell. The load platens were separated from one another by a distance of 101.6-mm (4-in) and with the applied load of 45 kgf (441.41 N) they generated a maximum moment of 8.38 Nm. In order to ensure repeatable alignment of the aluminium bones, light score marks were placed at the ends of all the bones.

Figure 2.4: Four-point bend assembly

2.7.4 Strain Gage Selection

Strain gages were used to measure surface strain on the WHOLE bone models. Two strain gages were used for each of the WHOLE bone models. one mounted underneath the plate and one located on the side of the bone. perpendicular to the fracture plate, at 45 degrees to the longitudinal axis. Given the wide variety of strain gages available, proper strain gage selection is critical to achieving accurate repeatable strain readings. There are three important points that were taken into consideration in selecting the type of strain gage to use with the aluminium bone models. The selection criteria included: static/dynamic testing, material type and operating temperature. For the test protocol the strain gages would be used to measure surface strain during static loading conditions. at room temperature (22 °C) on a 6061-T6 aluminium surface. There was nothing in the test protocol that suggested the use of anything but a general-purpose constantan grid strain gage. For this research CEA (Constantan Encapsulated Alloy) gages were selected. These strain gages are designed to be flexible, with large copper soldering tabs, cast polyimide backings and encapsulated constantan alloy grids. CEA type strain gages are available in a wide variety of sizes and configurations with a choice of either 120 or 350-ohm resistances. Although 350-ohm gages would have reduced substrate heating, 120-ohm gages had to be used due to the system requirements of the data acquisition system at Queen's University, where the torsion testing was performed. Due to the isolated loading modes used in this test protocol it was determined that using gages with a single longitudinal pattern would provide sufficient information. In addition, the data acquisition package on the MTS Sintech has only one channel available

for receiving strain gage data. Since there was no need for temperature compensation or strain amplification, the strain gages were balanced using a quarter bridge configuration.

3 VERIFICATION TESTING OF THE CANINE AXIALLY COMPRESSIBLE PLATES

3.1 Purpose

One of the problems posed by reusing the canine axially compressible plates for this test protocol was the possibility that the polymer inserts had changed properties as a result of the in-vivo implantation process. All the polymer inserts were inspected macroscopically and any inserts that showed visible signs of damage were discarded. Although the ideal solution would have been to manufacture new inserts the economics of the situation dictated that a thorough mechanical testing would be more cost-effective and just as relevant. As was discussed in chapter one, the canine ACP plates relied on a pair of polymer inserts, to reduce the axial stiffness of the plates. Different sizes and materials of polymer inserts were used to provide a range of stiffnesses. A large number of canine plates (with the corresponding polymer inserts) were retrieved from Yeadon's in-vivo experiment. Unfortunately, no information existed regarding the mechanical characteristics of each of the polymer inserts. Although there were no concerns regarding the stainless steel part of the plate, any changes in material properties of the polymer inserts would significantly affect the mechanical characteristics of the plate system. Thus it was deemed critical to proceed with a preliminary experiment that would allow for the measurement of the stiffness of each canine plate/polymer pair. At the same time this

experiment would allow for a comparison between the results obtained using the same test fixture but a different testing machine and different displacement sensors.

3.2 Experimental Fixture Design

The original canine ACP testing device was composed of two separate fixtures machined from cold rolled steel (Figure 3.1). Each fixture was designed so that the perpendicular distance between each of the fixtures "arms" could accommodate the thickness of a canine ACP and two spacers. The function of the spacers was to position the plates within the fixtures in such a way that the line of action of the applied force coincided with the centroidal axis of the plate. Each fixture had three holes that were spaced according the hole spacing on the canine ACPs'. Six stainless steel pins with a diameter of 3.14-mm were inserted through these holes and through the holes in the ACP plates in order to hold the ACP plates in the test fixtures. The diameter of the smooth pins was chosen so that it would be approximately equal to the inner diameter of the cortical screws that would be used with the canine ACP plate. The polymer inserts normally employed with the canine ACP plate were installed in the opposite orientation to what would be used in-vivo. In the in-vivo situation the polymer inserts are oriented to allow for a small degree of deformation under a compressive load, but no movement under a tensile load. For the in-vitro test a tensile load would be applied and thus it was necessary to reverse the orientation of the polymer inserts in order to simulate what would occur with an actual plated bone/plate assembly subjected to a compressive load⁷⁸. This made the test fixture used by Yeadon unsuitable for the purpose of measuring axial micromotion occurring at the fracture gap.

The canine ACP test fixture was mounted to the fixed bed of an MTS testing machine by securing the bottom of the fixture into a stainless steel cup and using a 12.7mm (0.5in) stainless steel shear pin to secure the canine test fixture to the cup. The top portion of the test fixture was secured to the moving crosshead using a small universal joint and a 4.5mm diameter stainless steel shear pin.

Figure 3.1: Tensile test fixture

Figure 3.2: Cross-sectional view of MTS fixture and shear pin

3.3 Experimental Procedure

Each pair of polymer inserts was catalogued and installed in the corresponding canine ACP plate. Each plate was then held in place in the test fixture using stainless steel pins. The test fixture was then fastened to the upper and lower crossheads of an MTS model Sintech 1G material testing machine. Four different variants of the canine ACP system were tested. Variant number one had polymethylmethacrylate inserts with a thickness of 1.5mm (0.059"). variant number two also had polymethylmethacrylate inserts except with thinner 0.75mm (0.030") inserts. Variant number three had low-density polyethylene inserts with a thickness of 0.75mm (0.030"). The last canine ACP variant had round holes with no polymer insert. For the sake of identification each plate was abbreviated as follows: PMMA059, PMMA030, LDPE030 and ROUND.

Initially all the ACP plates using polymer inserts were subjected to a series of pseudostatic loading cycles as prescribed in the previous testing protocol, in order to precondition the polymer inserts. Twenty preconditioning cycles were applied to each

pair of inserts using a maximum load of 500 N and a crosshead speed of 0.5mm/min.

After the preconditioning cycles were complete, each pair of inserts were tested in tension ten times up to a maximum load of 500 N. Crosshead speed was constant at 0.5mm/min and the data was acquired at 10 Hz. Axial stiffness was calculated using linear regression to determine the line of best fit for the load versus displacement data. The slope of this line gave the axial stiffness of each specimen in N/mm. Displacement was measured from the crosshead position sensor and was thus a measure of the overall displacement of the entire plate/fixture construct. This was one of the major differences in experimental procedures between this experiment and the one performed by Yeadon. In Yeadon's experiment an LVDT was mounted between the two halves of the fixture allowing for the measurement of plate stiffness. In order to compare the results from the two experiments it was necessary to determine the stiffness of the test fixtures that were used and then use those values to calculate the theoretical stiffness of the plates.

3.4 Results

Each insert number represents a pair of inserts that were used together in a specific plate. Table 3.1 details the average result of the ten repeated tension tests for each of the insert pairs. It should be noted that due to damage and loss, the same number of inserts of each type were not available for testing. Since the round plate did not employ any polymer inserts, only one plate was tested for comparison purposes.

Insert #	Plate stiffness (N/mm)			
	PMMA030	PMMA059	LDPE030	ROUND
1	3023.34	2480.26	1333.28	3591.05
2	2987.34	2670.48	1164.15	
3	3008.14	2888.85	1421.76	
4	3011.77	3056.31	1501.71	
5	3042.47	2976.65	1491.32	
6		2089.04		
7		2310.75		
8		2470.65		
9		2146.04		
10		2535.89		
<i>Average</i>	<i>3011.96</i>	<i>2561.59</i>	<i>1382.43</i>	<i>3591.05</i>
<i>Std. Dev</i>	<i>20.21</i>	<i>335.01</i>	<i>139.30</i>	<i>20.21</i>

Table 3.1: Results from the verification testing of the polymer inserts

In order to be able to perform theoretical calculations to determine true plate stiffness, it was necessary to determine the relative stiffnesses of the various fixture components. A theoretical analysis was performed to determine the axial stiffness of the stainless steel shear pin. The results of this analysis showed the stiffness of the shear pin to be approximately 6.1×10^{11} N/mm. A separate experiment was run in order to determine the stiffness of the universal joint. The average stiffness of the universal joint was measured to be 4949.02 ± 13.76 N/mm. The results of this experiment are detailed in Table 3.2:

Test #	Stiffness (N/mm)
1	4939.41
2	4926.78
3	4937.22
4	4937.86
5	4951.40
6	4954.50
7	4953.30
8	4954.17
9	4961.90
10	4973.67
Mean	4949.02
Std. Dev.	13.76

Table 3.2: Results of universal joint stiffness test

An overlay of the all the load versus displacement data has been included in Figure 3.2 below. The line of best fit for the first data set has also been included on this graph.

Figure 3.3: Overlay of all load versus deformation curves - universal joint

In order to confirm these results a cold rolled steel test piece was tested using the fixtures as for the plates. The results of this test series are detailed in Table 3.3:

Test #	Stiffness (N/mm)
1	4946.98
2	5042.16
3	5071.94
4	5121.82
5	5158.57
6	5181.09
7	5194.06
8	5216.66
9	5240.50
10	5256.50
Mean	5143.03
Std. Dev.	97.87

Table 3.3: Results of test bar stiffness test

An overlay of the all the load versus displacement data has also been included in Figure 3.4:

Figure 3.4: Overlay of all load versus deformation curves - test bar

The line of best fit for the first data set has also been included on this graph.

The theoretical stiffness of the steel test bar was calculated to be approximately 1.68×10^{12} N/mm, meaning that the resultant axial stiffness of the system should be very close to the stiffness of the universal joint.

The theoretical stiffness of the plate was calculated by assuming that the total displacement of the plate/fixture system was composed of a displacement (δ) from the universal joint and a displacement from the plate. Displacement from the shear pin was ignored since it was several orders of magnitude smaller than either the plate or the universal joint stiffnesses.

$$\delta_{TOTAL} = \delta_{PLATE} + \delta_{UNIV-JOINT} \quad (3.1)$$

Since the stiffness is defined as force divided by displacement and since the applied force can be assumed to be constant in this system, the following equation for the stiffness (K) of the plate/fixture system was derived:

$$\frac{1}{K_{TOTAL}} = \frac{1}{K_{PLATE}} + \frac{1}{K_{UNIV-JOINT}} \quad (3.2)$$

The overall stiffness of the plate/fixture system (K_{TOTAL}) was determined experimentally from the slope of the load versus crosshead displacement curve. The stiffness of the universal joint ($K_{UNIV-JOINT}$) was also determined experimentally. Thus the only unknown in equation 3.2 is the stiffness of the plate (K_{PLATE}). Solving for plate stiffness the following values were obtained:

Plate Type	Theo. Results (Backman) Plate Stiffness (N/mm)	Exp. Results (Yeadon) Plate Stiffness (N/mm)	Relative Error (%)
ROUND	13087.31	13782.17	5.04
PMMA030	7695.30	8312.89	7.43
PMMA059	5310.00	4220.85	25.8
LDPE030	1918.25	1624.93	18.05

Table 3.4: Theoretical Results of Plate Stiffness

3.5 Statistical Analysis

The data from the axial stiffness test was analysed statistically. In order to determine whether a statistically significant difference existed among the PMMA030, PMMA059 and the LDPE030 plate a one-way analysis of variance was conducted. The minimum sample size was five and a p value (the probability of incorrectly rejecting the null hypothesis) less than or equal to 0.05 was considered significant. One-way analysis of variance indicated that the data had two degrees of freedom and an F statistic of 57.11. The F statistic corresponded to a P value of 2.85×10^{-8} . Therefore it was possible to conclude that significant differences existed among the three ACP plates on the basis of axial stiffness. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that each ACP type was significantly different than the others on the basis of axial stiffness. Thus it can be concluded that the PMMA030 plate is stiffer than the PMMA059 plate and that the PMMA059 plate was stiffer than the LDPE030 plate. The post-hoc results are summarised in Table 3.5. The ROUND plate was not included in the test results since no polymer insert was used and thus only one ROUND plate was used in the testing.

(I) PLATE (J) PLATE		Mean Difference (I-J)	Sig.
PMMA030	PMMA059	46.0880	.005
	LDPE030	166.3780	.000
PMMA059	PMMA030	-46.0880	.005
	LDPE030	120.2900	.000
LDPE030	PMMA030	-166.3780	.000
	PMMA059	-120.2900	.000

Table 3.5: Post hoc test results for the polymer inserts

(Note: statistical significances (SIG) of .000 imply significance values less than 0.0001

3.6 Discussion

The results from the tensile tests clearly show that there is a significant variation in axial stiffness among the various pairs of polymer inserts of the same type. Post hoc testing also showed that there were significant differences in axial stiffness between each of the different types of polymer inserts. However, the testing data did show that there was a large variation in the stiffness results for the PMMA059 inserts. There was enough variation in the results to prohibit simple random selection of a PMMA059 insert for the main test series. It was entirely possible that an overly "stiff" pair of inserts could be chosen, thus confounding the experimental results. For testing purposes with the aluminium test fixture only one insert pair of each type would be needed. The results from the preliminary tests allowed for the selection of polymer inserts that most closely matched the group average. In this way it was possible to avoid using inserts with abnormal physical properties.

Since the test procedure followed was essentially the same as the one used by Yeadon et al ⁷⁸ it is useful to compare the results. Table 3.4 compares Yeadon's experimental results for plate stiffness with the theoretical results from the current

research. It is quite evident from the data that the theoretical results strongly support the experimental results obtained by Yeadon et al ⁷⁸. The stiffest plates (ROUND and PMMA030) show the best results with relative errors of 5.04 and 7.43 percent respectively. The larger differences seen with the least stiff plates (PMMA059 and LDPE030) may be attributable to unforeseen changes in material properties as a result of the in-vivo implantation process. The other more likely possibility is that in the experiment performed by Yeadon et al ⁷⁸ an insert was tested that had properties dissimilar from the majority of the other inserts. Working backwards, it is possible to show that for the PMMA059 inserts choosing the insert with a measured stiffness of 2146.04 N/mm would have reduced the relative error to approximately 10%. However, this insert had a stiffness value almost one standard deviation away from the group mean, making it less than the ideal choice. One additional factor that remained in common between the two data sets is the relative rank of the stiffness values. Both sets of data conclusively show that the ROUND plate is stiffest, followed by the PMMA030, the PMMA059 and then the LDPE030 plate.

4 EXPERIMENTAL DESIGN, BIOMECHANICAL TESTING AND RESULTS

4.1 Purpose

One purpose of the mechanical testing was to validate the results obtained with this test method against results from an independently validated source. The work done by Yeadon et al ⁷⁸ on the canine ACP plates was an easily accessible source. Due to differences in the experiment design and setup direct numerical comparisons could not be made between the two sets of data. Instead, the conclusions derived from Yeadon et al's ⁷⁸ study were used to form the basis of three testable hypotheses. These hypotheses are as follows:

- 1.) The axial stiffness of the canine plates was significantly different between the various plate types. The ROUND plate was stiffest, followed by the PMMA030 plate, the PMMA059 plate then the LDPE030 plate.
- 2.) There were no significant differences in bending stiffness between the various canine ACP plates
- 3.) Theoretical analysis of the torsional stiffness of the canine plates showed no significant differences between the various canine ACP plates.

In addition to testing the hypotheses generated by Yeadon et al ⁷⁸ the second purpose of using this new testing protocol was to generate additional important

information regarding plate performance. It will be possible to determine the amount of micromotion occurring for each of the canine ACP plates when subjected to a compressive load. Since the canine plates had already undergone in-vivo testing this information would allow for some conclusions to be drawn regarding the degree of micromotion and its effect on fracture healing. The current test methodology also stipulates the testing of the current standard in fracture plate fixation (DCP/LD-DCP) thus allowing for some analysis regarding performance differences between these plates and the canine ACP plates.

4.1.1 Experiment Design

The aluminium test fixtures were all designed and machined so that each test bone was identical to another within the prescribed manufacturing tolerance (± 0.127 mm). This allowed for a factorial experiment design set up in order to investigate differences in the mechanical response of the aluminium bones as a result of being plated with different fracture plate systems. The mechanical response of each plate/bone system was defined differently for both the GAP and the WHOLE bone models, where each model had different variables of interest.

4.1.2 Loading Modes

The aluminium bones were not designed to test a fracture plate to failure, instead the focus was on the performance of the plate/aluminium bone system under an applied load. A review of fracture mechanisms in long bones made it clear that axial

compression, offset bending, torsion and pure bending were all important failure modes for bone^{13: 18: 22: 25}. For the purpose of this test the three pure loading modes: axial compression, four-point bending and torsion were selected. Axial compression was one of the most critical loading modes since it allowed for the measurement of how much micromotion was occurring at the fracture gap. This type of movement is closely related to the ability of a fracture plate to enhance osteogenesis^{24: 27: 36: 37: 78}. An investigation of bending and torsional stiffness properties was also important since very low stiffness results in these loading modes have been linked to possible refracture in bone⁴⁰. Applying an axially compressive load to the WHOLE bone test fixtures allowed for the determination of the degree of stress shielding that occurred as a result of fracture plate application.

Testing the aluminium bones in four point bending allowed for the study of the behaviour of fracture plates under a loading mode that commonly results in bone fracture in-vivo. In order to accomplish this the bending stiffness of the GAP bone model was measured during four-point bending. Testing the WHOLE bone model in four-point bending enabled one to investigate the effect of stress shielding as a result of different plate designs.

Testing the aluminium bone models in torsion also allowed for the study of the behaviour of various fracture plates under a loading mode that routinely results in bone fracture. With the GAP bone model rotational stiffness was measured while with the WHOLE bone model the effect of stress shielding was determined.

4.1.3 Variables of Interest

4.1.3.1 Extensometer displacement

Extensometer displacement was measured only on the GAP bone model under axial compression. Extensometer displacement was measured with the use of an MTS 632.13N-20 extensometer (MTS Systems, Minneapolis, MN) having a gage length of 12.7-mm (0.5-in). The extensometer was placed in such a way that the gage length spanned the 3-mm fracture gap on one side of the aluminium bone fixture. The extensometer is in reality measuring the maximum amount of micromotion that occurs under an axial compressive load. It is this type of axial movement has been shown by many researchers to be able to stimulate fracture healing^{27: 62: 78: 80}.

4.1.3.2 Axial Stiffness

Axial stiffness was a derived value, defined as the slope of best fit, calculated using a least squares regression method, for the plot of load (N) versus extensometer displacement (mm). This yielded a local stiffness value in N/mm for the stiffness of the bone/plate in the area centred on the fracture gap. This variable is of interest since it serves as an indication of the performance of the plate as a load is applied.

4.1.3.3 Mid-Span Displacement

Mid-span displacement was measured only on the GAP bone model in four point bending. Mid-span displacement was measured with the use of a linear variable differential transformer (LVDT) mounted directly underneath the aluminium bone at its

geometrical midpoint. Given the symmetry of the bone model this allowed for an approximate measure of the maximum mid-span displacement for each of the GAP bone models. This gave an immediate indication as to the exact degree of deformation that occurred under a bending load.

4.1.3.4 Bending Stiffness

Bending stiffness was a derived value defined as the slope of best fit, calculated using a least squares regression method, for the plot of load (N) versus LVDT displacement (mm). This yielded a bending stiffness value in N/mm for the entire bone/plate construct. This variable is of interest since it serves as another indicator of the stability of fixation of a particular fracture plate system.

4.1.3.5 Rotational Displacement

Rotational displacement was measured on the GAP bone model in torsional loading. Rotational displacement was measured with the aid of a rotational variable differential transformer (RVDT) integrated into the load actuator of the Instron 8510 bi-axial testing machine. Only the maximum rotational displacement was reported for each of the GAP bone models. This gave an immediate indication as to the exact degree of deformation incurred under a torsional load.

4.1.3.6 Rotational Stiffness

Rotational stiffness was a derived value defined as the slope of best fit. calculated using a least squares regression method. for the plot of torque (Nm) versus RVDT displacement (deg). This yields a torsional stiffness value in Nm/deg for the entire bone/plate construct. This variable is of interest since it serves as an indicator of the torsional stability of a particular fracture plate.

4.1.3.7 Strain Measurement

Surface strain measurements were only of interest in the WHOLE bone models. In compression and bending strain was measured in the anterior part of the bone. at the mid-point, directly below the fracture plate. In torsional loading strain was measured on the side of the aluminium bone, at 45 degrees to the long axis of the bone. The medial gage was placed at 45 degrees to the longitudinal axis since this is the principal direction for an element under pure torsion. The values for this gage were only recorded when a torsional load was applied to the bone. The strain values were of interest, since many researchers have hypothesised that the reduction in surface strain as a result of fracture plate placement can cause delayed fracture healing^{10-12; 14; 28}.

4.1.3.8 Percent Stress Shielding

The percent stress shielding is a derived value used to quantify the degree to which there is a reduction of surface strain in the bone as a result of fracture plate application^{10-12; 14; 28}. Percent stress shielding is defined mathematically as:

$$\%StressShield = 1 - \frac{MaxAnteriorStrain(plated)}{MaxAnteriorStrain(unplated)} \quad (4.1)$$

All of the tests on the WHOLE bone models included tests with an unplated aluminium bone. This allowed for the measurement of maximum anterior surface strain when no fracture plate was applied. Thus a plate that provides zero percent stress shielding would have a maximum anterior strain measurement equivalent to that of the unplated bone. Conversely a fracture plate that resulted in one hundred percent stress shielding would give an anterior strain value of zero.

4.1.4 Factorial Design

Two separate factorial type experiments were run. One set of experiments focussed on the GAP bone model and the other experiment focussed on the WHOLE bone model. The factor of interest in this experiment was plate type and the dependant variables changed according to the loading mode. Table 4.1 and 4.2 below detail the various combinations of plate systems, loading modes and dependant variables.

<i>GAP Bone</i>	<i>Dependant Variable</i>					
<i>Loading Mode</i>	<i>Ext. Disp.</i>	<i>Axial Stiffness</i>	<i>LVDT Disp.</i>	<i>Bending Stiffness</i>	<i>Rot. Disp.</i>	<i>Rot. Stiffness</i>
Comp	Yes	Yes	No	No	No	No
Bend	No	No	Yes	Yes	No	No
Torsion	No	No	No	No	Yes	Yes

Table 4.1: GAP bone dependant variables

<i>WHOLE Bone</i>	<i>Dependant Variable</i>	
<i>Loading Mode</i>	<i>Anterior Strain</i>	<i>Medial Strain</i>
Comp	Yes	No
Bend	Yes	No
Torsion	No	Yes

Table 4.2: WHOLE bone dependant variables

4.2 Experimental Methods

4.2.1 Torsion Testing - Procedure

The torsion experiment was conducted at the Stress Analysis Laboratory in the Department of Mechanical Engineering at Queen's University. Testing was performed on Instron 8521 biaxial testing machine and data analysis was conducted using a computerised data acquisition system running Labview 3.0 software. The 25-pin connector housing each of the strain gage leads was plugged directly into the Instron's control system that was in turn being driven by the Labview software.

The canine plates were affixed to the bone models using specially designed 3.5-mm cortical screws. The installation instructions with the canine plates specified that a torque screwdriver was to be used and that the maximum screw torque for the round plate was 0.170 Nm and the maximum screw torque for the other canine plates was to be 0.113 Nm. The DCP and LC-DCP plates were affixed to bone using standard 3.5-mm cortical screws. No maximum torque limit was specified in the operating instructions for the DCP plate and a standard surgical driver (Sulzer Medica – Part #109.01.020) was used. This installation protocol was unchanged for all loading modes.

Securing the WHOLE and GAP bones to the top and bottom crosshead of the Instron 8521 was accomplished using a 6.35-mm diameter shear pins. The Instron was run in displacement control, with a maximum applied torque of 15 Nm. and a rotational loading rate of 0.1 deg/sec. Data from the strain gages, rotational transducer and load cell were all acquired at 10 Hz and saved directly to a Pentium 166 workstation. Each plate type was mounted and then re-attached for a total of five times in order to ensure no undue variation existed as a result of the experimental procedure.

4.2.2 Compression Testing - Procedure

The compression experiment was conducted at the Orthopaedic Biomechanics Laboratory at the University of Ottawa. Testing was performed on a MTS Sintech 1G testing machine and data acquisition was performed using MTS Testworks 4.0 software.

The canine plates were affixed to bone using specially designed 3.5-mm cortical screws. The upper half of the aluminium bone models were rigidly secured to the moving crosshead using a specially designed adapter (see Chapter 2) that screwed into the upper portion of the aluminium bone. For the GAP bone model, the aluminium bone was rotated so that the fracture plate was facing to the front and an extensometer was then mounted straddling the fracture gap. Next, the female 25-pin connector housing each of the strain gage leads was plugged directly into the MTS machine.

The MTS machine was run under displacement control and a maximum load of 441.45 N was applied to each bone at a strain rate of 0.015-mm/sec. Two trials for each

compression test were run, while each test was repeated ten times per trial in order to ensure no undue variation existed in the results.

4.2.3 Four Point Bend Testing – Procedure

The four point bending experiment was also conducted at the Orthopaedic Biomechanics Laboratory at the University of Ottawa. Testing was performed on a MTS Sintech 1G testing machine and data acquisition was performed using MTS Testworks 4.0 software.

The four point bending jig was attached to the base of the MTS machine with a pair of ¼-20 socket head cap screws. Each aluminium bone model was then placed into the four point bending jig. Score marks on each end of the aluminium bone models helped to ensure that the proper alignment was maintained between trials. The aluminium bones were aligned such that the plated side of the bone was facing downwards. This ensured that the plate was on the tension side of the bone when it was put under load. The loading platen was attached to the load cell with a ½-20 socket head cap screw. An opening in the centre of the plate allowed for a LVDT to be securely fixed to the four-point bend fixture. This allowed for mid-span deflection to be measured with the GAP bone model. The LVDT was removed during testing of the WHOLE bone model and the female 25-pin connector housing each of the strain gage leads was plugged directly into the MTS machine.

The MTS machine was run under displacement control and a maximum load of 441.45 N was applied to each bone at a strain rate of 0.015 mm/sec. Two separate trials for each bend test were run, with each specimen tested ten times per trial in order to ensure no undue variation existed in the results. Appendix D includes some load versus deformation results for various tests.

4.2.4 Percent Stress Shielding - Procedure

Percent stress shielding was measured in compression, bending and torsion. The same procedure was used with the WHOLE bone model as was used with the GAP bone model in the various loading modes. The data acquisition system on the MTS machine was used to record strain signals received from the strain gage mounted directly underneath the fracture plate. In order to perform this test a special WHOLE bone model was manufactured. This bone model had the anterior flat but did not have any screw holes, effectively representing a fully healed bone. Each of the test plates was mounted on their respective WHOLE bone models and tested in either compression, torsion, or bending. The tests protocols were the same as with the GAP bone models except that only maximum strain values were recorded. The same procedure was followed for the special NO PLATE version of the WHOLE bone, which was also tested in compression, bending and torsion.

4.2.5 Statistical Analysis Methods

Each of the data sets was analysed using parametric statistical methods in order to determine whether differences existed among each of the plate types on the basis of the

various experimental variables. A p value (the probability of incorrectly rejecting the null hypothesis) less than or equal to 0.05 was considered significant. One-way analysis of variance was conducted in order to determine whether the mean value of a particular variable was statistically similar (null hypothesis) or dissimilar (alternate hypothesis) between the different plates. In cases where using one-way analysis of variance (ANOVA) rejected the null hypothesis, an additional post-hoc analysis was conducted using the Least Significant Difference (LSD) method. This type of statistical analysis involved performing a pair-wise comparison between each of the plates in order to determine whether a statistically significant difference existed between each possible plate pair combination.

4.3 Experimental Results

4.3.1 Compression Testing - GAP Aluminium Bone Model

4.3.1.1 Extensometer Displacement Results

Table 4.3 summarises the extensometer displacement results for the GAP bone model. These results showed excellent repeatability with the largest standard deviation of 0.014 mm. The results with the canine plates showed the ROUND plate to be the stiffest followed by the PMMA030, PMMA059, and the LDPE030 plate. Both the DCP and the LC-DCP show very little axial displacement when compared to the canine plates. Figure 4.1 shows a graphical summary of the results.

Plate Type	Extensometer Displacement (mm)	Std. Dev	N
Round	0.16	0.005	20
PMMA059	0.28	0.014	20
LDPE030	0.53	0.05	20
PMMA030	0.25	0.011	20
DCP	0.085	0.005	20
LC-DCP	0.093	0.005	20

Table 4.3: Extensometer displacement results

4.3.1.2 Extensometer - ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had five degrees of freedom and an F statistic of 1171.25. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of extensometer displacement. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that each of the canine ACP's were significantly different from the others on the basis of extensometer displacement. Thus it can be concluded that the LDPE030 allows for the most micromotion, followed by the PMMA059 plate, then the PMMA030 and then the ROUND plate. The post-hoc results are summarised in appendix C. Only the LC-DCP and DCP plates, showed no significant differences on the basis of extensometer displacement.

Figure 4.1: Extensometer displacement versus plate type

4.3.1.3 Axial Stiffness - Results

The axial stiffness results showed good repeatability with the largest standard deviation of 467.0 N/mm occurring with the DCP plate. The results showed that of the canine plates, the ROUND plate is stiffest followed by the PMM030, PMMA059 and then the LDPE030 plate. Both the DCP and the LC-DCP plate are significantly stiffer than the canine plates in axial compression. Table 4.4 shows a graphical summary of the axial stiffness results.

Plate Type	Stiffness (N/mm)	Std. Dev	N
Round	2428.0	157.9	20
PMMA059	1320.4	65.7	20
LDPE030	765.2	62.8	20
PMMA030	1549.0	55.9	20
DCP	5215.0	467.0	20
LC-DCP	4671.5	161.9	20

Table 4.4: Axial stiffness results

4.3.1.4 Axial Stiffness – Post Hoc Results

One-way analysis of variance indicated that the data had five degrees of freedom and an F statistic of 1469.731. The F statistic corresponded to a P value of less than 0.001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of axial stiffness. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that each of the canine ACP's was significantly different than the others on the basis of stiffness. Thus it can be concluded that the PMMA030 plate is stiffer than the PMMA059 plate and that the PMMA059 plate is stiffer than the LDPE030 plate. The post-hoc results are summarised in Appendix C.

Figure 4.2: Axial stiffness versus plate type

4.3.2 Compression Testing - WHOLE Aluminium Bone Model

4.3.2.1 Anterior Stress Shielding Results

The results from the anterior strain gages also show excellent repeatability with the largest standard deviation of only 3.5 microstrain. According to the stress shielding results the stiffest plate shows the most stress shielding while the least stiff plate shows the least amount of stress shielding. Figure 4.2 shows a graphical summary of the axial stiffness results.

Plate Type	Mean Strain ($\mu\text{mm}/\mu\text{mm}$)	Percent Stress Shielding	Std. Dev	N
No plate	-64	N/A	3	20
Round	-24	63	0.5	20
PMMA059	-33	48	1.5	20
LDPE030	-37	42	0.5	20
PMMA030	-30	53	1	20
DCP	-36	44	3	20
LC-DCP	-31	52	3.5	20

Table 4.5: Percent stress shielding result summary

4.3.2.2 Anterior Stress Shielding– ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had six degrees of freedom and an F statistic of 662.74. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of anterior microstrain. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that significant differences existed between each of the canine ACP's on the basis of anterior microstrain. However the pair-wise comparison showed that no significant difference existed between the LDPE030 and the DCP plate or between the PMMA030 and the LC-DCP plate. The post-hoc results are summarised in Appendix C.

Figure 4.3: Anterior microstrain versus plate type

4.3.3 Four Point Bending Test - GAP Aluminium Bone Model

4.3.3.1 LVDT Displacement Results

The results for mid-span LVDT displacement showed good repeatability with the largest standard deviation of 0.06 mm occurring with the ROUND plate. The results showed that of the canine plates, the ROUND plate had the least amount of mid-span displacement followed by the PMM030, PMMA059 and then the LDPE030 plate. Both the DCP and the LC-DCP plate had significantly more mid-span displacement than the canine plates. Figure 4.4 shows a graphical summary of the LVDT displacement results.

Plate Type	Max. LVDT Displacement (mm)	Std. Dev	N
Round	0.99	0.06	20
PMMA059	1.13	0.03	20
LDPE030	1.26	0.03	20
PMMA030	1.07	0.03	20
DCP	1.50	0.05	20
LC-DCP	1.50	0.0009	20

Table 4.6: Maximum LVDT Displacement results

4.3.3.2 LVDT Displacement - ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had five degrees of freedom and an F statistic of 846.65. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of mid-span deflection. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that each of the canine ACP's were significantly different from the others on the basis of stiffness. Thus it can be concluded that the PMMA030 plate is stiffer than the PMMA059 plate and that the PMMA059 plate was stiffer than the LDPE030 plate. Pair-wise comparison showed that there was no significant difference between the DCP and the LC-DCP plate on the basis of mid-span deflection. The post-hoc results are summarised in Appendix C.

Figure 4.4: Maximum LVDT displacement versus plate type

4.3.3.3 Bending Stiffness Results

The bending stiffness results show excellent repeatability with the largest standard deviation of 12.8 N/mm occurring with the PMMA059 plate. The results show that the ROUND plate had the highest bending stiffness followed by the PMM030, PMMA059 and then the LDPE030 plate. Both the DCP and the LC-DCP plate have significantly less bending stiffness than the canine plates. Figure 4.5 shows a graphical summary of the bending stiffness results.

Plate Type	Bending Stiffness (N/mm)	Std. Dev	N
Round	475.8	8.8	20
PMMA059	423.8	12.8	20
LDPE030	384.6	10.8	20
PMMA030	445.4	4.9	20
DCP	224.6	3.9	20
LC-DCP	196.2	5.9	20

Table 4.7: Bending stiffness result summary

4.3.3.4 Bending Stiffness - ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had five degrees of freedom and an F statistic of 3686.61. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of bending stiffness. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that each of the canine ACP's were significantly different from the others on the basis of bending stiffness. The results confirm that the ROUND plate is stiffest followed by the PMMA030, the PMMA059 plate and then the LDPE030 plate. Pair-wise comparison also showed that there was a significant difference between the DCP and the LC-DCP plate with the DCP plate being stiffer in bending than the LC-DCP plate. The post-hoc results are summarised in Appendix C.

Figure 4.5: Bending stiffness versus plate type

4.3.4 Four Point Bending - WHOLE Aluminium Bone Model

4.3.4.1 Anterior Stress Shielding Results

The results from the anterior strain gages also show good repeatability with the ROUND plate results having the largest standard deviation of 10 microstrain. The stress shielding results show that the stiffest plates have the most stress shielding while the less stiff plates have the least amount of stress shielding. The DCP and the LC-DCP plates show significantly more stress shielding than the canine plates. Figure 4.6 shows a graphical summary of the average anterior microstrain results and table 4.8 shows the results in tabular format.

Plate Type	Mean Strain ($\mu\text{mm}/\mu\text{mm}$)	Std. Dev	Percent Stress Shielding	N
No plate	270	1.5	N/A	20
Round	180	10	33	20
PMMA059	198	2	27	20
LDPE030	213	4	21	20
PMMA030	190	3	30	20
DCP	140	3	48	20
LC-DCP	120	2	56	20

Table 4.8: Anterior stress shielding result summary

4.3.4.2 Anterior Strain - Post Hoc Results

One-way analysis of variance indicated that the data had six degrees of freedom and an F statistic of 2351.83. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of anterior microstrain. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that each of the canine ACP's were significantly different than the others on the basis of anterior microstrain. The pair-wise comparison showed statistically significant differences among all the plates on the basis of maximum anterior microstrain. Pair-wise comparison confirmed that the LDPE030 plate had the most anterior microstrain followed by the PMMA059 plate, the PMMA030 and then the ROUND plate. The DCP and the LC-DCP plate had maximum anterior strain values well below the canine plates with the DCP plate having a slightly higher maximum strain value than the LC-DCP. The post-hoc results are summarised in Appendix C.

Figure 4.6: Maximum anterior microstrain versus plate type

4.3.5 Torsional Testing- GAP Aluminium Bone Model

4.3.5.1 Torsional Displacement Results

The torsional displacement results also show very good repeatability with the largest standard deviation of only 1.63 degrees occurring with the DCP plate. The rotational displacement results seem to suggest that there is some variation between the plates on the basis of torsional displacement. Figure 4.7 shows a graphical summary of the maximum rotation results and table 4.9 shows the results in tabular format.

Plate Type	Torsional Displacement (deg)	Std. Dev	N
Round	-5.47	0.49	5
PMMA059	-8.45	0.24	5
LDPE030	-11.70	0.34	5
PMMA030	-8.64	0.52	5
DCP	-10.61	1.63	5
LC-DCP	-8.81	0.47	5

Table 4.9: Torsional displacement result summary

4.3.5.2 Torsional Displacement - ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had five degrees of freedom and an F statistic of 3.84. The F statistic corresponded to a P value of 0.034. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of anterior microstrain. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that significant differences existed between the majority of the plate types except between the PMMA059 and the PMMA030 plate. The full post-hoc results are summarised in Appendix C.

Figure 4.7: Torsional displacement versus plate type

4.3.5.3 Torsional Stiffness Results

The torsional stiffness results also show very good repeatability with the largest standard deviation of only 0.38 Nm/deg occurring with the ROUND plate. Figure 4.8 shows a graphical summary of the torsional stiffness results.

Plate Type	Torsional Stiffness (Nm/deg)	Std. Dev	N
Round	3.28	0.38	5
PMMA059	2.26	0.12	5
LDPE030	1.50	0.11	5
PMMA030	2.23	0.16	5
DCP	1.00	0.15	5
LC-DCP	1.48	0.12	5

Table 4.10: Torsional stiffness results

4.3.5.3.1 Torsional Stiffness - ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had six degrees of freedom and an F statistic of 83.25. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of anterior microstrain. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that the majority of the fracture plates were different from one another except for the PMMA030 and the PMMA059 as well as the LDPE030 and the LC-DCP. The full post-hoc results are summarised in Appendix C.

Figure 4.8: Torsional stiffness versus plate type

4.3.6 Torsional Testing - WHOLE Aluminium Bone Model

4.3.6.1 Medial Stress Shielding Results

The results from the medial strain gages also show excellent repeatability with the largest standard deviation of only 3.5 microstrain. The stress shielding results show that all the canine plates have almost exactly the same amount of stress shielding. It is interesting to note that both the DCP and the LC-DCP seem to provide less stress shielding in this area than the canine plates. Figure 4.9 shows a graphical summary of the medial microstrain results with table 4.11 detailing the results in detail.

Plate Type	Mean Strain ($\mu\text{mm}/\mu\text{mm}$)	Percent Stress Shielding	Std. Dev	N
No plate	316	N/A	3.0	5
Round	281	11	4.5	5
PMMA059	280	11	3.5	5
LDPE030	285	10	4.0	5
PMMA030	280	11	2.0	5
DCP	299	5	5	5
LC-DCP	311	2	4.5	5

Table 4.11: Maximum medial strain results

4.3.6.2 Medial Stress shielding - ANOVA and Post Hoc Results

One-way analysis of variance indicated that the data had six degrees of freedom and an F statistic of 84.29. The F statistic corresponded to a P value of less than 0.0001. Therefore it was possible to conclude that significant differences existed among all the fracture plates on the basis of medial microstrain. Further analysis using the Least Significant Difference (LSD) for pair-wise post-hoc comparison indicated that no significant difference existed between the ROUND plate and any other canine ACP. In

addition there was no significant difference between the PMMA059 - PMMA030 plate or between the LDPE030 - PMMA030 plates. The full post-hoc results are summarised in Appendix C.

Figure 4.9: Maximum medial microstrain results

5 DISCUSSION AND ANALYSIS

5.1 Verification of the new Mechanical Testing System

One the goals of this research project was to verify whether the results obtained using a new test methodology for fracture plates would correspond to those obtained using a previously developed and verified test method ⁷⁸. Although it would have been useful to make a direct comparison between the results obtained by Yeadon et al ⁷⁸ and the results obtained with the current test methodology, this was impossible. The procedure used in chapter three to generate theoretical stiffness results would not be effective here. Unlike with Yeadon et al ⁷⁸ where an attempt was made to test the properties of each fracture plate in isolation, what was of most interest here was the performance of the plate/bone system. The fact that the plates were secured to the aluminium bone using actual cortical screws implies that the substrate plays a role in the load sharing of the system. Inherent in the stiffness equation for the system would be a term representing the frictional forces between the plate and bone. Although comparing numerical results would not be possible, Yeadon et al ⁷⁸ put forward three general conclusions from their research that formed the basis of the testable hypotheses used to validate our results. The hypothesises that were tested are as follows:

- 1.) The axial stiffness of the canine plates was significantly different between the various plate types. The ROUND plate was stiffest, followed by the PMMA030 plate, the PMMA059 plate then the LDPE030 plate.
- 2.) There were no significant differences in bending stiffness between the various canine ACP plates.
- 3.) Theoretical analysis of the torsional stiffness of the canine plates showed no significant differences between the various canine ACP plates.

5.1.1 Hypothesis 1: Axial Stiffness

The results from the axial stiffness test conclusively showed that there were significant differences in axial stiffness among the various canine ACP plates. The ROUND plate was determined to be the stiffest followed by the PMMA030, the PMMA059 and the LDPE030 plate. These results fully support the conclusion reached by Yeadon et al ⁷⁸.

5.1.2 Hypothesis 2: Bending Stiffness

The analysis of the data from the four point bending tests contradicted Yeadon et al's ⁷⁸ finding in that a statistically significant difference was detected between the plates. Further post hoc testing showed that there was a pair-wise difference among all the plates on the basis of bending stiffness. However, it is important to note that results that are statistically significant are not always significant in real world terms. The differences in bending stiffness between the various plates ranged from approximately 29 - 98 N/mm while the differences in axial stiffness ranged from approximately 883 - 1668 N/mm. In

addition several researchers have made it clear that axial stiffness, not bending stiffness, is the key variable for improving fracture healing^{7: 64: 76}. Thus it is reasonable to conclude that differences in bending stiffness between the canine plates are negligible compared to the differences in axial stiffness. These results may be a consequence of the increased sensitivity of this new test method.

5.1.3 Hypothesis 3: Torsional Stiffness

The theoretical analysis performed by Yeadon et al⁷⁸ on the geometry of the plates led to the conclusion that there was no significant difference between the canine ACP plates on the basis of torsional stiffness. However, the one way analysis of variance on the data from the torsion tests indicates that there is a statistically significant difference. The pair-wise testing showed a difference between all the canine plates except between PMMA030 and PMMA059 plates. The non-trivial differences in torsional stiffness ranged from 0.72 - 1.77 Nm/deg. These results are not unexpected since the theoretical analysis took into account the geometric configuration of the plate and made the assumption that due to the design of the canine ACP plates the different stiffnesses of the inserts would not play a significant role in torsional stiffness. Thus, the question remains, how significant are the differences in torsional stiffness? Although torsional stiffness is not recognized as a significant contributor to fracture healing, it is clear that some degree of torsional stiffness is required in order to prevent loss of reduction and possible refracture. All the canine plates show a torsional stiffness higher than either the DCP or the LC-DCP so in this context the canine ACP plates provide an adequate amount of torsional stiffness and pose no safety concerns. Considering that

researchers have consistently shown axial and not torsional stiffness to be the key variable to improving fracture healing^{7: 64: 76}, it is reasonable to conclude that differences in torsional stiffness between the canine plates are negligible compared to the differences in axial stiffness.

5.1.4 Axial micromotion

The second reason for testing the canine axially compressible plates was to gain a better understanding regarding exactly how much micromotion at the fracture site creates the optimal environment for fracture healing. One of the great benefits of studying the Canine ACP system is that all four Canine ACP plates have been subjected to both in-vitro and in-vivo testing. The stiffness index results obtained by Yeadon⁷⁸ and Foux^{23: 24} are reprinted in Table 5.1 along with results of the extensometer displacement. The higher the value of the stiffness index (SI), the closer the mechanical properties of the operated bone approached those of the control bone. Analysis of the stiffness index results by Yeadon⁷⁸ showed that the optimal properties for a fracture plate would be in between the PMMA030 and the PMMA059 plate, which corresponds to 0.25-0.28 mm of micromotion and a corresponding axial stiffness of between 1320 - 1549 N/mm. It is interesting to note that neither the most stiff fracture plate (ROUND) or the least stiff fracture plate (LDPE030) provided optimal healing.

Plate Type	Stiffness Index	Axial Stiffness (N/mm)	Extensometer Displacement (mm)
ROUND	0.70	2428.0	0.16
PMMA030	0.80	1549.0	0.25
PMMA059	0.85	1320.4	0.28
LDPE030	0.72	765.2	0.53

Table 5.1: Comparison of stiffness index and compression results

5.1.5 Bending and Torsional Stability

With respect to the GAP bone model, the issue of greatest concern was whether the canine plates would provide enough stability in bending such that the healing process would not be interrupted. Terjesen⁶⁴, using a rabbit model, showed that fracture plates with extremely low bending stiffnesses resulted in increased instances of loss of reduction, and refracture in the rabbit tibias. In order to put the results into the proper context, bending and torsional stability were compared to that achieved by the DCP and the LC-DCP plate, both of which are considered to be the current standard for fracture fixation. From the results it was quite clear that the canine ACP plates were significantly stiffer than the DCP/LC-DCP plates. Terjesen's study concluded that the use of fracture plates with extremely high bending stiffnesses did not enhance the environment for fracture healing. Unfortunately, more quantitative conclusions could not be drawn from Terjesen's study⁶⁴ as a result of their decision to use plates that varied by orders of magnitude in both their axial and bending stiffness. This is in direct contrast to studies by Yeadon⁷⁸ and Foux^{23:24} where the fracture plates used only varied essentially in the degree of axial stiffness. As a result it is not possible to conclude from Terjesen's⁶⁴ study whether it is the reduction in bending stiffness or the reduction in axial stiffness or

the reduction in both that influence fracture healing. Overall, Terjesen's study⁶⁴ certainly supports the concept that plates with both intermediate bending and intermediate axial stiffness can improve fracture healing. The fact that the canine ACP plates are approximately twice as stiff as the DCP/LC-DCP plates, would seem to suggest that a reduction in their bending stiffness may prove to be beneficial.

The discussion of loading modes on human long bones, made it clear that torsional loading is very common in long bones. In addition, bone failure as a result of torsional loading is also common. This was the impetus for testing the fracture plates in this loading mode using both the GAP and WHOLE aluminium bone models. The main concern in examining the results from the GAP testing was to determine whether the canine plates exhibited an adequate level of torsional stability. For the purposes of this research, the DCP/LC-DCP were used for comparison purposes. A plate that did not provide a minimum level of torsional stability could result in malunion at the fracture site and delay fracture healing. Post hoc testing showed that there were no statistically significant differences between the PMMA059 and the PMMA030 plate or between the LC-DCP and the LDPE030 plate. Overall the canine ACP plates were at least as torsionally stiff as the DCP/LC-DCP plates.

5.1.6 Stress Shielding

The stress shielding results in compression are quite interesting in that they do not seem to have a direct correlation to the axial stiffness results. When the results from the canine plates are examined in isolation the results confirm our initial hypothesis. The

stiffest canine plates results' in the highest degree of stress shielding. while the least stiff plate results in the least amount of stress shielding. Post hoc comparisons show that all the canine plates have statistically significant differences on the basis of axial strain. What is unexpected is that the DCP and the LC-DCP plates, which were much stiffer in compression than the canine plates, do not necessarily provide for more stress shielding. In fact, the plates that provide the least amount of stress shielding in compression are the LDPE030 and the DCP plates. According to the post hoc testing no statistically significant difference can be seen between these two plates. In addition, the post hoc testing also shows that no statistically significant difference exists between the PMMA030 and the LC-DCP plates, even though the axial stiffness results show that the LC-DCP plate is more than twice as stiff as the PMMA030 plate in compression. It is in this regard that axial stiffness does not appear to be a good predictor of stress shielding. Since load transfer between the underlying bone and the plate is a primary contributor to stress shielding, it would appear that there is another mechanism at work, which effectively is transferring more load to the bone. The answer to this question may lie in the method used to secure the DCP/LC-DCP plates to bone. The correct surgical technique involved in implanting DCP/LC-DCP plates' places no limit on the maximum screw torque that should be used in securing plate to bone. The emphasis here is on enhancing the bone/plate contact. The same method was used in the experimental methodology for both the DCP and the LC-DCP plates. However, due to the design of the canine ACP plates, a special torque screwdriver was used and the maximum torque was limited to 0.170 Nm (ROUND) and 0.113 Nm (PMMA030, PMMA059 and LDPE030). This significantly reduced the normal force at the bone/plate interface thus

reducing the maximum frictional force between the plate and bone and impacting on the load transfer mechanism. Extrapolating the stress shielding results from the DCP/LC-DCP bones to the physiological situation is also difficult due to the actual mechanism of action of the dynamic compression plate. Physiologically, DCP/LC-DCP plates are often implanted in such a way that they can provide compression at the fracture site. However, due to the stiff nature of the aluminium bone models used, the DCP and the LC-DCP plate were installed in neutral mode for the WHOLE bone models. Neutral mode for these plates means that they do not provide for any axial compression to the bone. This additional compression provides a huge increase in bone/fracture plate rigidity and could have significantly affected the stress shielding results in-vivo.

In terms of stress shielding during the four point bending tests, the canine plates seem to exhibit significant reductions in stress shielding, especially when compared to the DCP/LC-DCP plates. This result is also unexpected given that the canine plates are all significantly stiffer in bending than either the DCP or the LC-DCP plates. One way analysis of variance and additional post hoc testing of the canine plates shows that there is a statistically significant difference between all of the plates on the basis of anterior strain. The results of the pair-wise comparison are not unexpected, with the stiffest canine plate, the ROUND plate, showing the most stress shielding followed by the PMMA030, PMMA059 and the LDPE030 plate. It is somewhat surprising that the canine plates which proved to have higher bending stiffness than the DCP/LC-DCP plates actually have less stress shielding than these plates. Once again the reason for this may be the difference in screw torque between the canine and the DCP/LC-DCP plates. The

low screw torque used with the canine ACP plates may have prevented the canine plates from adequately reinforcing the WHOLE bone when a bending load was applied. The opposite would have been true with the DCP/LC-DCP plates, which were secured rigidly against the bone models with the cortical screws tightened to maximum torque.

Additional work using finite element analysis might be able to address this issue. This information would certainly be important to anyone looking to improve current fracture plate designs.

In terms of torsional stress shielding, there was very little statistically significant differences among the various canine plates. In general both the canine and the DCP/LC-DCP plates showed very little stress shielding in the medial region of the aluminium bone. From an in-vivo perspective it is really the stress shielding directly underneath the fracture plate that has the greatest implications for fracture healing^{16: 65: 68}. The argument can be made that from a design perspective medial stress shielding has relatively little importance.

5.1.7 Performance of DCP/LC-DCP fracture plates

The tertiary goal of this study was to be able to comment on the operation of other fracture plates once the new test method was verified. The DCP and the LC-DCP plates are currently the standard for fracture fixation and as such they were suitable candidates for testing. Although Yeadon⁷⁸ did not test DCP or LC-DCP plates in-vivo, it is well known from the literature^{3: 4: 6: 14: 16} that these types of plates, which provide for extremely rigid fixation, do not necessarily provide for optimal fracture healing. As can

be seen from Table 5.2, the DCP and LC-DCP plates allow for very little axial movement, implying that their ability to improve fracture healing would be decreased when compared to plates with decreased axial stiffness, such as the canine ACP plates.

Plate Type	Axial Stiffness (N/mm)	Extensometer Displacement (mm)
DCP	5215.0	0.085
LC-DCP	4671.5	0.093

Table 5.2: Compression results for DCP/LC-DCP plates

These results would seem to be supported by Foux²³ who performed biomechanical testing on canine femurs plated with DCP plates. Although the results showed that the DCP/LC-DCP plates did not cause as much stress shielding as was initially hypothesised, it must be noted that the plates in this research were installed in neutral mode and did provide any compression to the bone fragments as in the in-vivo situation.

6 CONCLUSIONS AND RECOMMENDATIONS

The new testing methodology/test fixtures were designed to increase the amount of information that could be obtained from the in-vitro testing of different types of fracture plates. In order to verify that the new testing system worked, four plates that had previously undergone both in-vivo and in-vitro testing were compared to the current standard in fracture fixation, the DCP and the LC-DCP plate.

The previous work done on the canine ACP drew several conclusions regarding the mechanical performance of these plates that could be verified using the new testing methodology. Although the values of axial stiffness that were obtained for the canine ACP plates did not correspond to those obtained previously, this was not surprising given the difference in testing methods and test fixtures. However, post hoc statistical testing of the results did confirm significant differences between the canine ACP plates on the basis of axial stiffness. As well the testing showed that the ROUND plate was the stiffest followed by the PMMA030 the PMMA059 and the LDPE030 plate, confirming the findings of Yeadon et al ⁷⁸. As a result of significant differences in testing methodology for the four point bending tests, the values that were obtained did not correspond to those from the previous experiment. One way analysis of variance was used in order to try and confirm the hypothesis that no significant difference existed among the canine plates on

the basis of bending stiffness. Due to the increased sensitivity of our test method, post hoc testing determined that statistically significant differences existed among the canine ACP plates on the basis of bending stiffness. However, the results of the testing show that the differences in axial stiffness among the canine ACP plates are significantly greater than the differences in bending stiffness among the canine ACP tests. In this regard the results from this study are similar to those obtained by Yeadon et al ⁷⁸. The mechanical testing of the canine plates in torsion, did not correspond with the hypothesis drawn by Yeadon et al ⁷⁸. The analysis performed by Yeadon did not include any mechanical testing in torsion and relied instead on theoretical results that could not take into account the different insert materials employed with the canine ACP plates. Although statistically it is not possible to validate the theoretical findings of Yeadon et al ⁷⁸, it should be noted that all the canine plates have a greater torsional stiffness than the DCP/LC-DCP plates. Since torsional stiffness does not play a significant role in influencing fracture healing, this would imply that the small differences in torsional stiffness would not influence fracture healing. Thus, it can be concluded that overall the new testing methodology not only confirms the results of Yeadon et al ⁷⁸, but also provides for increased testing sensitivity and accuracy.

The fact that the new testing methodology has been validated allowed for some conclusions to be derived from the rest of the mechanical results, including those from the WHOLE bone models. By comparing the test results to the in-vivo results obtained by Yeadon et al ⁷⁸, it was possible to obtain a profile of the mechanical characteristics that provide an optimum environment for fracture healing. The results from the

compression testing show that all the canine ACP plates allowed for significantly more axial movement than either the DCP or the LC-DCP plates. In addition all the canine ACP plates show increased amounts of micromotion. Yeadon et al.⁷⁸ were able to conclude that a plate with properties in between those of the PMMA030 and the PMMA059 plate would have optimal mechanical properties. From the test results it was possible to determine the corresponding mechanical properties of these plates. According to the test results, in order to optimise fracture healing, a fracture plate should have an axial stiffness of between 1324 - 1530 N/mm, and it should allow for approximately 0.25-0.28 mm of micromotion. In addition the plate should also be able to provide a degree of torsional and bending stability at least on par with the DCP and the LC-DCP plates.

In terms of stress shielding, both the canine and the DCP/LC-DCP plates result in some degree of stress shielding. The canine plates reduced the effect of stress shielding quite significantly under a bending load. This effect is less pronounced when the aluminium bones are subjected to purely compressive loads.

6.1 Recommendations

This investigation presented a new methodology for testing fracture fixation plates. The aluminium bone models used in this study attempted to simulate bone during both the union and the remodelling phase of fracture healing. The ultimate goal of this research was to develop a test procedure that would provide more information about how a fracture plate behaves in-vivo without the expense of using an in-vivo model or the uncertainty involved in using computer simulations. Although the current investigation

with the aluminium bones provided valuable information, several improvements could be made in future investigations.

The aluminium bone models used in this investigation simulated both the early union phase and the late remodelling phase of fracture healing. However, the aluminium models did not provide any information regarding the effect of fracture plates' during the intermediate stage of fracture healing. This type of model could be similar to the aluminium models except that instead of a fracture gap there would be a material with an intermediate stiffness filling this void. This type of model could theoretically provide information regarding the strain field beneath the plate as well as information regarding fixation stiffness and micromotion at the fracture site. The difficulty with producing this type of model is in quantifying what mechanical properties constitute "intermediate" fracture healing and in actually integrating this material into the bone model in such a way as to simulate the physiologic condition.

Overall the experimental procedure was robust and the data that was obtained had good repeatability. However, the addition of extra strain gages around the circumference of the aluminium bone, aligned with the longitudinal axis would be very useful in allowing one to confirm the alignment of the bone during compression testing.

Subjecting both the WHOLE and GAP bone models to fatigue loads, would provide important information regarding the performance of the plates under conditions that more closely simulate in-vivo use. This type of testing would be especially

important for testing the next generation of fracture plates that are designed with bioresorbable components that reduce undergo changes in elastic modulus over time. Their would be no difficulty in adapting the current models for use in fatigue testing. except for the logistics involved in obtaining machine time from a lab that is so equipped.

Although an in-depth review of available materials was conducted before production of the prototypes began, a new material has recently become available. The new material is a third generation glass epoxy material produced by Sawbones USA. The glass epoxy material has a tensile strength of 132 MPa and a tensile modulus of 19.0 GPa and it can be machined and instrumented in a similar manner to the aluminium bones. Although this new glass epoxy bone substitute is not as economical or as machinable as aluminium, the fact that its material properties are closer to cortical bone would prove very useful. Using this material it would be possible to normalize the axial/bending/torsional stiffness results to the axial/bending/torsional stiffness of the glass epoxy bone model. Also, the normalized results would be more directly applicable to the in-vivo situation. As an example, in the current study it was not possible to comment on the bending and torsional stiffnesses of the plates, except to say how they compared to the current surgical standard - the DCP/LC-DCP plates. Having normalized results would have allowed one to comment on how the bending, torsional and axial stiffnesses of the plates compared to that of the underlying bone. That said, this in no way negates the value of the aluminium bone models, which are useful as a platform for comparing the axial/bending/torsion properties of various fracture plates.

Given the robust nature of the test protocol it would be worthwhile to test additional types of fracture plates. This type of test platform would also prove to be excellent for testing new concepts and new designs of fracture plates. With the information amassed from this research it would be possible to make some general predictions regarding the ability of specific plate design to improve fracture healing. The mechanical test results that have been obtained to date would also be valuable to anyone working to create a finite element model of a bone/fracture plate system. One of the problems with computer analysis is verifying that the results of the simulation replicate what occurs in real life. The test procedure could easily be recreated for computer simulation purposes. If the computer results were verified, this would open other interesting avenues of research. One investigation would be of particular interest - the differences in load transfer characteristics between the GAP and the WHOLE bone model. As well the issue of load sharing between the screws could be addressed using finite element analysis. This type of computer simulation might also be an appropriate approach to ascertain what factors have the greatest influence on stress shielding.

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Appendix A: Sample Calculation

Given Variables:

Applied load (P) = 441.45N

Aluminium Modulus (E) = 68.9 Gpa

Gage length of strain gage (L) = 0.00157 m

Outer diameter (OD) = 0.02m

Inner diameter (ID) = 0.014m

Cross-sectional area

$$A = \pi \cdot (OD^2 - ID^2)$$

$$A = 1.6 \times 10^{-4} \text{ m}^2$$

Axial Deformation and Strain formula

$$1) \delta = \frac{P \cdot L}{A \cdot E}$$

$$2) \varepsilon = \frac{\Delta L}{L}$$

Results

$$\delta = 6.29 \times 10^{-8} \text{ m}$$

$$\varepsilon = 4.0 \times 10^{-5} \text{ mm/mm (40 microstrain)}$$

Appendix C: Post-Hoc Testing Results

Post Hoc Results for Extensometer Displacement

Dependent Variable: Extensometer

LSD

(I) Plate Type	(J) Plate Type	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
Round	PMMA059	-.1200*	.007	.000	-.1337	-.1063
	LDPE030	-.3680*	.007	.000	-.3817	-.3543
	PMMA030	-8.6500E-02*	.007	.000	-.1002	-7.2830E-02
	DCP	8.050E-02*	.007	.000	6.683E-02	9.417E-02
	LCDCP	7.250E-02*	.007	.000	5.883E-02	8.617E-02
PMMA059	Round	.1200*	.007	.000	.1063	.1337
	LDPE030	-.2480*	.007	.000	-.2617	-.2343
	PMMA030	3.350E-02*	.007	.000	1.983E-02	4.717E-02
	DCP	.2005*	.007	.000	.1868	.2142
	LCDCP	.1925*	.007	.000	.1788	.2062
LDPE030	Round	.3680*	.007	.000	.3543	.3817
	PMMA059	.2480*	.007	.000	.2343	.2617
	PMMA030	.2815*	.007	.000	.2678	.2952
	DCP	.4485*	.007	.000	.4348	.4622
	LCDCP	.4405*	.007	.000	.4268	.4542
PMMA030	Round	8.650E-02*	.007	.000	7.283E-02	.1002
	PMMA059	-3.3500E-02*	.007	.000	-4.7170E-02	-1.9830E-02
	LDPE030	-.2815*	.007	.000	-.2952	-.2678
	DCP	.1670*	.007	.000	.1533	.1807
	LCDCP	.1590*	.007	.000	.1453	.1727
DCP	Round	-8.0500E-02*	.007	.000	-9.4170E-02	-6.6830E-02
	PMMA059	-.2005*	.007	.000	-.2142	-.1868
	LDPE030	-.4485*	.007	.000	-.4622	-.4348
	PMMA030	-.1670*	.007	.000	-.1807	-.1533
	LCDCP	-8.0000E-03	.007	.249	-2.1670E-02	5.670E-03
LCDCP	Round	-7.2500E-02*	.007	.000	-8.6170E-02	-5.8830E-02
	PMMA059	-.1925*	.007	.000	-.2062	-.1788
	LDPE030	-.4405*	.007	.000	-.4542	-.4268
	PMMA030	-.1590*	.007	.000	-.1727	-.1453
	DCP	8.000E-03	.007	.249	-5.6701E-03	2.167E-02

Based on observed means.

*. The mean difference is significant at the .05 level.

Post Hoc Results for Axial Stiffness

Dependent Variable: Stiffness (Ext)

LSD

(I) Plate Type	(J) Plate Type	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
Round	PMMA059	112.910*	6.984	.000	99.075	126.745
	LDPE030	169.430*	6.984	.000	155.595	183.265
	PMMA030	89.535*	6.984	.000	75.700	103.370
	DCP	-284.140*	6.984	.000	-297.975	-270.305
	LCDCP	-228.705*	6.984	.000	-242.540	-214.870
PMMA059	Round	-112.910*	6.984	.000	-126.745	-99.075
	LDPE030	56.520*	6.984	.000	42.685	70.355
	PMMA030	-23.375*	6.984	.001	-37.210	-9.540
	DCP	-397.050*	6.984	.000	-410.885	-383.215
	LCDCP	-341.615*	6.984	.000	-355.450	-327.780
LDPE030	Round	-169.430*	6.984	.000	-183.265	-155.595
	PMMA059	-56.520*	6.984	.000	-70.355	-42.685
	PMMA030	-79.895*	6.984	.000	-93.730	-66.060
	DCP	-453.570*	6.984	.000	-467.405	-439.735
	LCDCP	-398.135*	6.984	.000	-411.970	-384.300
PMMA030	Round	-89.535*	6.984	.000	-103.370	-75.700
	PMMA059	23.375*	6.984	.001	9.540	37.210
	LDPE030	79.895*	6.984	.000	66.060	93.730
	DCP	-373.675*	6.984	.000	-387.510	-359.840
	LCDCP	-318.240*	6.984	.000	-332.075	-304.405
DCP	Round	284.140*	6.984	.000	270.305	297.975
	PMMA059	397.050*	6.984	.000	383.215	410.885
	LDPE030	453.570*	6.984	.000	439.735	467.405
	PMMA030	373.675*	6.984	.000	359.840	387.510
	LCDCP	55.435*	6.984	.000	41.600	69.270
LCDCP	Round	228.705*	6.984	.000	214.870	242.540
	PMMA059	341.615*	6.984	.000	327.780	355.450
	LDPE030	398.135*	6.984	.000	384.300	411.970
	PMMA030	318.240*	6.984	.000	304.405	332.075
	DCP	-55.435*	6.984	.000	-69.270	-41.600

Based on observed means.

*. The mean difference is significant at the .05 level.

Post Hoc Results Compression Anterior Strain

Dependent Variable: ANTERIOR

LSD

(I) Plate Type	(J) Plate Type	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
No plate	Round	-39.2000*	.697	.000	-40.5790	-37.8210
	PMMA059	-30.6500*	.697	.000	-32.0290	-29.2710
	LDPE030	-26.9500*	.697	.000	-28.3290	-25.5710
	PMMA030	-33.5000*	.697	.000	-34.8790	-32.1210
	DCP	-27.5000*	.697	.000	-28.8790	-26.1210
	LC-DCP	-32.7500*	.697	.000	-34.1290	-31.3710
Round	No plate	39.2000*	.697	.000	37.8210	40.5790
	PMMA059	8.5500*	.697	.000	7.1710	9.9290
	LDPE030	12.2500*	.697	.000	10.8710	13.6290
	PMMA030	5.7000*	.697	.000	4.3210	7.0790
	DCP	11.7000*	.697	.000	10.3210	13.0790
	LC-DCP	6.4500*	.697	.000	5.0710	7.8290
PMMA059	No plate	30.6500*	.697	.000	29.2710	32.0290
	Round	-8.5500*	.697	.000	-9.9290	-7.1710
	LDPE030	3.7000*	.697	.000	2.3210	5.0790
	PMMA030	-2.8500*	.697	.000	-4.2290	-1.4710
	DCP	3.1500*	.697	.000	1.7710	4.5290
	LC-DCP	-2.1000*	.697	.003	-3.4790	-.7210
LDPE030	No plate	26.9500*	.697	.000	25.5710	28.3290
	Round	-12.2500*	.697	.000	-13.6290	-10.8710
	PMMA059	-3.7000*	.697	.000	-5.0790	-2.3210
	PMMA030	-6.5500*	.697	.000	-7.9290	-5.1710
	DCP	-.5500	.697	.432	-1.9290	.8290
	LC-DCP	-5.8000*	.697	.000	-7.1790	-4.4210
PMMA030	No plate	33.5000*	.697	.000	32.1210	34.8790
	Round	-5.7000*	.697	.000	-7.0790	-4.3210
	PMMA059	2.8500*	.697	.000	1.4710	4.2290
	LDPE030	6.5500*	.697	.000	5.1710	7.9290
	DCP	6.0000*	.697	.000	4.6210	7.3790
	LC-DCP	.7500	.697	.284	-.6290	2.1290
DCP	No plate	27.5000*	.697	.000	26.1210	28.8790
	Round	-11.7000*	.697	.000	-13.0790	-10.3210
	PMMA059	-3.1500*	.697	.000	-4.5290	-1.7710
	LDPE030	.5500	.697	.432	-.8290	1.9290
	PMMA030	-6.0000*	.697	.000	-7.3790	-4.6210
	LC-DCP	-5.2500*	.697	.000	-6.6290	-3.8710
LC-DCP	No plate	32.7500*	.697	.000	31.3710	34.1290
	Round	-6.4500*	.697	.000	-7.8290	-5.0710
	PMMA059	2.1000*	.697	.003	.7210	3.4790
	LDPE030	5.8000*	.697	.000	4.4210	7.1790
	PMMA030	-.7500	.697	.284	-2.1290	.6290
	DCP	5.2500*	.697	.000	3.8710	6.6290

Based on observed means.

*. The mean difference is significant at the .05 level.

Post-Hoc Results: Maximum LVDT displacement

Dependent Variable: MAX_LVDT

LSD

(I) PLATE	(J) PLATE	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
ROUND	PMMA059	-.13280*	.011	.000	-.15367	-.11193
	LDPE030	-.26100*	.011	.000	-.28187	-.24013
	PMMA030	-7.90000E-02*	.011	.000	-9.98726E-02	-5.81274E-02
	DCP	-.50450*	.011	.000	-.52537	-.48363
	LC-DCP	-.50430*	.011	.000	-.52517	-.48343
PMMA059	ROUND	.13280*	.011	.000	.11193	.15367
	LDPE030	-.12820*	.011	.000	-.14907	-.10733
	PMMA030	5.3800E-02*	.011	.000	3.2927E-02	7.4673E-02
	DCP	-.37170*	.011	.000	-.39257	-.35083
	LC-DCP	-.37150*	.011	.000	-.39237	-.35063
LDPE030	ROUND	.26100*	.011	.000	.24013	.28187
	PMMA059	.12820*	.011	.000	.10733	.14907
	PMMA030	.18200*	.011	.000	.16113	.20287
	DCP	-.24350*	.011	.000	-.26437	-.22263
	LC-DCP	-.24330*	.011	.000	-.26417	-.22243
PMMA030	ROUND	7.9000E-02*	.011	.000	5.8127E-02	9.9873E-02
	PMMA059	-5.38000E-02*	.011	.000	-7.46726E-02	-3.29274E-02
	LDPE030	-.18200*	.011	.000	-.20287	-.16113
	DCP	-.42550*	.011	.000	-.44637	-.40463
	LC-DCP	-.42530*	.011	.000	-.44617	-.40443
DCP	ROUND	.50450*	.011	.000	.48363	.52537
	PMMA059	.37170*	.011	.000	.35083	.39257
	LDPE030	.24350*	.011	.000	.22263	.26437
	PMMA030	.42550*	.011	.000	.40463	.44637
	LC-DCP	2.0000E-04	.011	.985	-2.06726E-02	2.1073E-02
LC-DCP	ROUND	.50430*	.011	.000	.48343	.52517
	PMMA059	.37150*	.011	.000	.35063	.39237
	LDPE030	.24330*	.011	.000	.22243	.26417
	PMMA030	.42530*	.011	.000	.40443	.44617
	DCP	-2.00000E-04	.011	.985	-2.10726E-02	2.0673E-02

Based on observed means.

*. The mean difference is significant at the .05 level.

Post-Hoc Results: Bending Stiffness

Dependent Variable: Stiffness

LSD

(I) PLATE	(J) PLATE	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
ROUND	PMMA059	5.3035*	.281	.000	4.7463	5.8607
	LDPE030	9.2205*	.281	.000	8.6633	9.7777
	PMMA030	3.0780*	.281	.000	2.5208	3.6352
	DCP	25.5445*	.281	.000	24.9873	26.1017
	LC-DCP	28.4205*	.281	.000	27.8633	28.9777
PMMA059	ROUND	-5.3035*	.281	.000	-5.8607	-4.7463
	LDPE030	3.9170*	.281	.000	3.3598	4.4742
	PMMA030	-2.2255*	.281	.000	-2.7827	-1.6683
	DCP	20.2410*	.281	.000	19.6838	20.7982
	LC-DCP	23.1170*	.281	.000	22.5598	23.6742
LDPE030	ROUND	-9.2205*	.281	.000	-9.7777	-8.6633
	PMMA059	-3.9170*	.281	.000	-4.4742	-3.3598
	PMMA030	-6.1425*	.281	.000	-6.6997	-5.5853
	DCP	16.3240*	.281	.000	15.7668	16.8812
	LC-DCP	19.2000*	.281	.000	18.6428	19.7572
PMMA030	ROUND	-3.0780*	.281	.000	-3.6352	-2.5208
	PMMA059	2.2255*	.281	.000	1.6683	2.7827
	LDPE030	6.1425*	.281	.000	5.5853	6.6997
	DCP	22.4665*	.281	.000	21.9093	23.0237
	LC-DCP	25.3425*	.281	.000	24.7853	25.8997
DCP	ROUND	-25.5445*	.281	.000	-26.1017	-24.9873
	PMMA059	-20.2410*	.281	.000	-20.7982	-19.6838
	LDPE030	-16.3240*	.281	.000	-16.8812	-15.7668
	PMMA030	-22.4665*	.281	.000	-23.0237	-21.9093
	LC-DCP	2.8760*	.281	.000	2.3188	3.4332
LC-DCP	ROUND	-28.4205*	.281	.000	-28.9777	-27.8633
	PMMA059	-23.1170*	.281	.000	-23.6742	-22.5598
	LDPE030	-19.2000*	.281	.000	-19.7572	-18.6428
	PMMA030	-25.3425*	.281	.000	-25.8997	-24.7853
	DCP	-2.8760*	.281	.000	-3.4332	-2.3188

Based on observed means.

*. The mean difference is significant at the .05 level.

Post-Hoc Results: Bending Anterior Strain

Dependent Variable: Max Strain

LSD

(I) PLATE	(J) PLATE	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
NO_PLATE	ROUND	89.3000*	1.425	.000	86.4822	92.1178
	PMMA059	71.7000*	1.425	.000	68.8822	74.5178
	LDPE030	57.0000*	1.425	.000	54.1822	59.8178
	PMMA030	80.0500*	1.425	.000	77.2322	82.8678
	DCP	129.4500*	1.425	.000	126.6322	132.2678
	LC-DCP	149.5000*	1.425	.000	146.6822	152.3178
ROUND	NO_PLATE	-89.3000*	1.425	.000	-92.1178	-86.4822
	PMMA059	-17.6000*	1.425	.000	-20.4178	-14.7822
	LDPE030	-32.3000*	1.425	.000	-35.1178	-29.4822
	PMMA030	-9.2500*	1.425	.000	-12.0678	-6.4322
	DCP	40.1500*	1.425	.000	37.3322	42.9678
	LC-DCP	60.2000*	1.425	.000	57.3822	63.0178
PMMA059	NO_PLATE	-71.7000*	1.425	.000	-74.5178	-68.8822
	ROUND	17.6000*	1.425	.000	14.7822	20.4178
	LDPE030	-14.7000*	1.425	.000	-17.5178	-11.8822
	PMMA030	8.3500*	1.425	.000	5.5322	11.1678
	DCP	57.7500*	1.425	.000	54.9322	60.5678
	LC-DCP	77.8000*	1.425	.000	74.9822	80.6178
LDPE030	NO_PLATE	-57.0000*	1.425	.000	-59.8178	-54.1822
	ROUND	32.3000*	1.425	.000	29.4822	35.1178
	PMMA059	14.7000*	1.425	.000	11.8822	17.5178
	PMMA030	23.0500*	1.425	.000	20.2322	25.8678
	DCP	72.4500*	1.425	.000	69.6322	75.2678
	LC-DCP	92.5000*	1.425	.000	89.6822	95.3178
PMMA030	NO_PLATE	-80.0500*	1.425	.000	-82.8678	-77.2322
	ROUND	9.2500*	1.425	.000	6.4322	12.0678
	PMMA059	-8.3500*	1.425	.000	-11.1678	-5.5322
	LDPE030	-23.0500*	1.425	.000	-25.8678	-20.2322
	DCP	49.4000*	1.425	.000	46.5822	52.2178
	LC-DCP	69.4500*	1.425	.000	66.6322	72.2678
DCP	NO_PLATE	-129.4500*	1.425	.000	-132.2678	-126.6322
	ROUND	-40.1500*	1.425	.000	-42.9678	-37.3322
	PMMA059	-57.7500*	1.425	.000	-60.5678	-54.9322
	LDPE030	-72.4500*	1.425	.000	-75.2678	-69.6322
	PMMA030	-49.4000*	1.425	.000	-52.2178	-46.5822
	LC-DCP	20.0500*	1.425	.000	17.2322	22.8678
LC-DCP	NO_PLATE	-149.5000*	1.425	.000	-152.3178	-146.6822
	ROUND	-60.2000*	1.425	.000	-63.0178	-57.3822
	PMMA059	-77.8000*	1.425	.000	-80.6178	-74.9822
	LDPE030	-92.5000*	1.425	.000	-95.3178	-89.6822
	PMMA030	-69.4500*	1.425	.000	-72.2678	-66.6322
	DCP	-20.0500*	1.425	.000	-22.8678	-17.2322

Based on observed means.

*. The mean difference is significant at the .05 level.

Post-Hoc Results: Maximum Rotation (Torsion)

Multiple Comparisons

Dependent Variable: Max_Rotate

LSD

(I) Plate	(J) Plate	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
ROUND	PMMA059	2.9780*	.4868	.000	1.9733	3.9827
	LDPE030	6.2326*	.4868	.000	5.2279	7.2373
	PMMA030	3.1700*	.4868	.000	2.1653	4.1747
	DCP	5.1420*	.4868	.000	4.1373	6.1467
	LC-DCP	1.8100*	.4868	.001	.8053	2.8147
PMMA059	ROUND	-2.9780*	.4868	.000	-3.9827	-1.9733
	LDPE030	3.2546*	.4868	.000	2.2499	4.2593
	PMMA030	.1920	.4868	.697	-.8127	1.1967
	DCP	2.1640*	.4868	.000	1.1593	3.1687
	LC-DCP	-1.1680*	.4868	.025	-2.1727	-.1633
LDPE030	ROUND	-6.2326*	.4868	.000	-7.2373	-5.2279
	PMMA059	-3.2546*	.4868	.000	-4.2593	-2.2499
	PMMA030	-3.0626*	.4868	.000	-4.0673	-2.0579
	DCP	-1.0906*	.4868	.035	-2.0953	-8.5861E-02
	LC-DCP	-4.4226*	.4868	.000	-5.4273	-3.4179
PMMA030	ROUND	-3.1700*	.4868	.000	-4.1747	-2.1653
	PMMA059	-.1920	.4868	.697	-1.1967	.8127
	LDPE030	3.0626*	.4868	.000	2.0579	4.0673
	DCP	1.9720*	.4868	.000	.9673	2.9767
	LC-DCP	-1.3600*	.4868	.010	-2.3647	-.3553
DCP	ROUND	-5.1420*	.4868	.000	-6.1467	-4.1373
	PMMA059	-2.1640*	.4868	.000	-3.1687	-1.1593
	LDPE030	1.0906*	.4868	.035	8.586E-02	2.0953
	PMMA030	-1.9720*	.4868	.000	-2.9767	-.9673
	LC-DCP	-3.3320*	.4868	.000	-4.3367	-2.3273
LC-DCP	ROUND	-1.8100*	.4868	.001	-2.8147	-.8053
	PMMA059	1.1680*	.4868	.025	.1633	2.1727
	LDPE030	4.4226*	.4868	.000	3.4179	5.4273
	PMMA030	1.3600*	.4868	.010	.3553	2.3647
	DCP	3.3320*	.4868	.000	2.3273	4.3367

Based on observed means.

*. The mean difference is significant at the .050 level.

Post-Hoc Results: Torsional Stiffness

Dependent Variable: Compliance

LSD

(I) Plate	(J) Plate	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
ROUND	PMMA059	1.0148*	.125	.000	.7571	1.2725
	LDPE030	1.7712*	.125	.000	1.5135	2.0289
	PMMA030	1.0480*	.125	.000	.7903	1.3057
	DCP	2.2736*	.125	.000	2.0159	2.5313
	LC-DCP	1.7920*	.125	.000	1.5343	2.0497
PMMA059	ROUND	-1.0148*	.125	.000	-1.2725	-.7571
	LDPE030	.7564*	.125	.000	.4987	1.0141
	PMMA030	3.320E-02	.125	.793	-.2245	.2909
	DCP	1.2588*	.125	.000	1.0011	1.5165
	LC-DCP	.7772*	.125	.000	.5195	1.0349
LDPE030	ROUND	-1.7712*	.125	.000	-2.0289	-1.5135
	PMMA059	-.7564*	.125	.000	-1.0141	-.4987
	PMMA030	-.7232*	.125	.000	-.9809	-.4655
	DCP	.5024*	.125	.000	.2447	.7601
	LC-DCP	2.080E-02	.125	.869	-.2369	.2785
PMMA030	ROUND	-1.0480*	.125	.000	-1.3057	-.7903
	PMMA059	-3.3200E-02	.125	.793	-.2909	.2245
	LDPE030	.7232*	.125	.000	.4655	.9809
	DCP	1.2256*	.125	.000	.9679	1.4833
	LC-DCP	.7440*	.125	.000	.4863	1.0017
DCP	ROUND	-2.2736*	.125	.000	-2.5313	-2.0159
	PMMA059	-1.2588*	.125	.000	-1.5165	-1.0011
	LDPE030	-.5024*	.125	.000	-.7601	-.2447
	PMMA030	-1.2256*	.125	.000	-1.4833	-.9679
	LC-DCP	-.4816*	.125	.001	-.7393	-.2239
LC-DCP	ROUND	-1.7920*	.125	.000	-2.0497	-1.5343
	PMMA059	-.7772*	.125	.000	-1.0349	-.5195
	LDPE030	-2.0800E-02	.125	.869	-.2785	.2369
	PMMA030	-.7440*	.125	.000	-1.0017	-.4863
	DCP	.4816*	.125	.001	.2239	.7393

Based on observed means.

*. The mean difference is significant at the .05 level.

Post-Hoc Results: Medial Strain

Dependent Variable: MED_STRN

LSD

(I) Plate	(J) Plate	Mean Difference (I-J)	Std. Error	Sig.	95% Confidence Interval	
					Lower Bound	Upper Bound
No Plate	Round	35.20*	2.598	.000	29.94	40.46
	PMMA059	36.20*	2.598	.000	30.94	41.46
	LDPE030	31.00*	2.598	.000	25.74	36.26
	PMMA030	36.40*	2.598	.000	31.14	41.66
	DCP	17.60*	2.250	.000	13.05	22.15
	LC-DCP	5.60*	2.250	.017	1.05	10.15
Round	No Plate	-35.20*	2.598	.000	-40.46	-29.94
	PMMA059	1.00	2.598	.702	-4.26	6.26
	LDPE030	-4.20	2.598	.114	-9.46	1.06
	PMMA030	1.20	2.598	.647	-4.06	6.46
	DCP	-17.60*	2.250	.000	-22.15	-13.05
	LC-DCP	-29.60*	2.250	.000	-34.15	-25.05
PMMA059	No Plate	-36.20*	2.598	.000	-41.46	-30.94
	Round	-1.00	2.598	.702	-6.26	4.26
	LDPE030	-5.20	2.598	.053	-10.46	5.93E-02
	PMMA030	.20	2.598	.939	-5.06	5.46
	DCP	-18.60*	2.250	.000	-23.15	-14.05
	LC-DCP	-30.60*	2.250	.000	-35.15	-26.05
LDPE030	No Plate	-31.00*	2.598	.000	-36.26	-25.74
	Round	4.20	2.598	.114	-1.06	9.46
	PMMA059	5.20	2.598	.053	-5.93E-02	10.46
	PMMA030	5.40*	2.598	.044	.14	10.66
	DCP	-13.40*	2.250	.000	-17.95	-8.85
	LC-DCP	-25.40*	2.250	.000	-29.95	-20.85
PMMA030	No Plate	-36.40*	2.598	.000	-41.66	-31.14
	Round	-1.20	2.598	.647	-6.46	4.06
	PMMA059	-.20	2.598	.939	-5.46	5.06
	LDPE030	-5.40*	2.598	.044	-10.66	-1.14
	DCP	-18.80*	2.250	.000	-23.35	-14.25
	LC-DCP	-30.80*	2.250	.000	-35.35	-26.25
DCP	No Plate	-17.60*	2.250	.000	-22.15	-13.05
	Round	17.60*	2.250	.000	13.05	22.15
	PMMA059	18.60*	2.250	.000	14.05	23.15
	LDPE030	13.40*	2.250	.000	8.85	17.95
	PMMA030	18.80*	2.250	.000	14.25	23.35
	LC-DCP	-12.00*	1.837	.000	-15.72	-8.28
LC-DCP	No Plate	-5.60*	2.250	.017	-10.15	-1.05
	Round	29.60*	2.250	.000	25.05	34.15
	PMMA059	30.60*	2.250	.000	26.05	35.15
	LDPE030	25.40*	2.250	.000	20.85	29.95
	PMMA030	30.80*	2.250	.000	26.25	35.35
	DCP	12.00*	1.837	.000	8.28	15.72

Based on observed means.

*. The mean difference is significant at the .05 level.

Appendix D: Sample Load-Deformation Curves

1) Results of DCP GAP test - Four Point Bending

2) Results of DCP GAP test - Compression

3) Results of DCP GAP test - Torsion

4) Results of DCP WHOLE test - Four Point Bending

5) Results of DCP WHOLE test - Compression

6) Results of DCP WHOLE test - Torsion