EVALUATION OF TRANSFIXATION CAST CONSTRUCTS IN HORSES

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By

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ABSTRACT

Transfixation pin casts have been used to manage a variety of different equine fracture configurations, but are particularly useful in comminuted fractures of the distal limb. The objectives of this study were to investigate strain at the bone-pin interface, the cast, and the fracture site, as well as load transfer between the bone and cast in different equine transfixation pin cast configurations. Three transfixation pin cast configurations (5 forelimbs per group) were evaluated: *Construct 1*: Two, 6.3-mm diameter pins spaced 4-cm apart in the cannon bone; *Construct 2:* Two, 6.3-mm diameter pins spaced 5-cm apart; *Construct 3:* Four, 4.8-mm diameter pins spaced 2-cm apart. Strain gauges were attached to the cast, cannon bone, and adjacent to a simulated fracture in the proximal phalanx. Limbs were subjected to single cycle compressive loading to failure as well as cyclic loading that simulated 6 weeks of wearing a cast. A simplified finite element (FE) model of Construct 1 and 3 was used to further evaluate strain and load transfer between the bone and cast during load to failure and cyclic loading. The results indicated that there was no difference in strain between the two 2-pin constructs in load to failure or cyclic loading. Relative to the 2-pin constructs, the 4-pin construct had less strain at the bone-pin interface and more strain in the cast, indicating that more load is transferred to the cast with the 4-pin construct. In-line with these findings, FE analyses indicated that the 4-pin system had less bone strain at the bone-pin interface, less strain adjacent to the fracture site, and less load transferred to the bone. These results suggest that the 4-pin cast is more effective at unloading the fractured bone.

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LIST OF ABBREVIATIONS

-cm	Centimeter
CT	Computed tomography
ESF	External skeletal fixator
E	Elastic modulus
E_S	Pin modulus
FE	Finite element
GPa	Gigapascal
Hz	Hertz
I	Pin area of moment of inertia
K_f	Axial stiffness
kN	Kilonewton
LVDT	Linear variable differential transformer
M	Number of pins used
-mm	Millimeter
MPa	Megapascal
NaCl	Sodium chloride
P1	Proximal phalanx
P2	Middle phalanx
P3	Distal phalanx
S	Distance from side bar to bone
SD	Standard deviation
TPC	Transfixation pin cast

CHAPTER 1

LITERATURE REVIEW

1.1 Introduction

When a fracture of the equine distal limb occurs, ideally it is repaired using internal fixation if surgical intervention is required. Internal fixation is achieved by applying implants, such as plates and screws, to the affected region to provide stability while the fractured bones heal. Certain fractures of the equine limb may not be amenable to internal fixation due to a lack of intact struts of bone, severe comminution, or a combination of both. Examples of these types of fractures include high impact fractures of the proximal or middle phalanx resulting in several small fracture fragments. Another example would be open long bone fractures of the 3rd metacarpal and metatarsal bones, where the use of implants to repair the fracture will inevitably result in osteomyelitis, secondary to contamination at the time of fracture. When internal fixation cannot be performed, external skeletal fixators may be appropriate.

External skeletal fixators consist of fixation pins, connecting bars (sidebars) or rods, and connecting clamps. They are divided into types based on the location of the sidebars and number of planes incorporated in the construct. Although there are many different ways to describe an external skeletal fixator, the commonly accepted nomenclature was described by Roe (1992). Within this nomenclature there are terms that are frequently referred to and should be understood. Unilateral and bilateral describe whether or not pins are inserted through one skin surface or two skin surfaces, respectively, while the term plane describes the direction the group of pins will adopt (Brinker 2016).

Roe (1992) describes Type I external skeletal fixators as unilateral fixators that can be further classified based on whether or not they are composed of one or two planes. If they are composed of a single plane, they are referred to as Type IA (Figure 1.1). Type IB external skeletal fixators are two; Type IA fixators placed 90 degrees from one another. The next configuration of external skeletal fixators are Type II fixators. These fixators are bilateral, but are within one plane (Figure 1.2). A third configuration is Type III and they are a combination of Type I and Type II configurations by being bilateral and biplanar.

Although these types of external skeletal fixator designs work well in smaller patients such as dogs and cats, their use in larger animals can prove problematic. In order to make the external skeletal fixator strong enough, the size of the materials utilized makes it difficult for the animal to ambulate and can cause significant trauma to the otherwise healthy limbs. An alternative design to external skeletal fixators is the transfixation pin cast. Transfixation pin casts consist of fiberglass casting tape and transcortical pins of various types placed above, and occasionally below, the fracture. The design and materials used in a transfixation pin cast make it a modification of the Type II external skeletal fixator. With a transfixation pin cast, the casting tape acts as the sidebars while the cast-pin interface replaces the connecting clamps. As with the Type II external skeletal fixator, the transfixation pin cast construct is bilateral and uniplanar.

Figure 1.1: Example of a Type 1A external skeletal fixator that is unilateral and uniplanar.

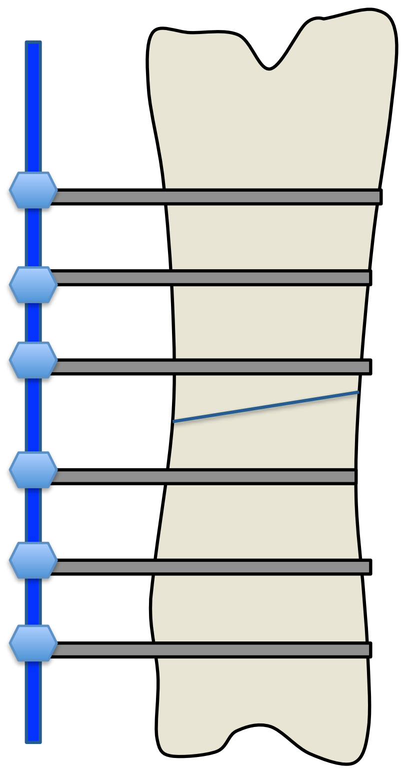
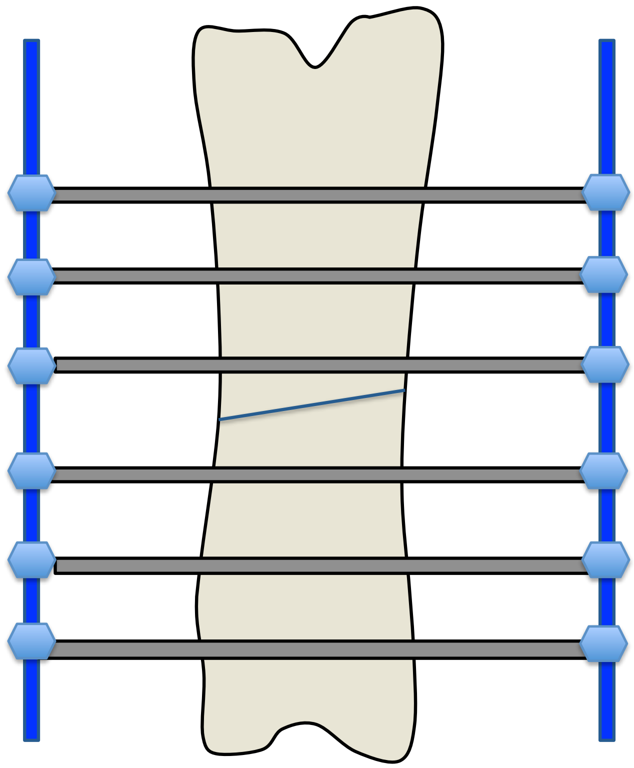


Figure 1.2: Example of Type II external skeletal fixator that is bilateral and uniplanar.



1.2 Purpose of the Transfixation Pin Cast

Transfixation pin casts have been shown to be a viable option, alone or in combination with internal fixation (Nunamaker et al. 1986; Németh et al. 1991). One of the most redeeming qualities of a transfixation pin cast is that it allows transfer of the axial weight bearing forces from the limb through the pins to the cast. This results in a significant decrease in the transfer of weight bearing forces to the bones distal to the pins (Schneider et al. 1998), allowing weight to be distributed away from the fracture site as it heals, and immediate weight bearing post-operatively. This is critical as horses have the best outcome when their weight is distributed over all four limbs. Horses that become non-weight bearing, or require a cast for whatever reason in one limb, have an increased risk in developing support limb laminitis. Between 17.5% and 27% of horses having a limb fracture develop supporting limb laminitis (Levine et al. 2007; Virgin et al. 2011). It is therefore important to get equine patients as comfortable as possible, as quickly as possible.

1.3 Complications associated with transfixation pin casts

As with many forms of external coaptation, there can be significant complications associated with the application and wearing of transfixation pin casts (Lescun et al. 2017). These include pin loosening, pin breakage, ring sequestra at the bone-pin interface, fracture through the pin tract, cast fracture, and simple cast sores related to an improperly fitting cast.

Pin loosening can occur secondary to pin tract infection, which results in severe discomfort and potential fracture at the bone-pin interface and or even collapse at the fracture site. Pin loosening can also be secondary to thermal necrosis at the time of pin placement where excessive drilling temperatures can be created (Matthews 1972; Matthews et al. 1984; Egger et al. 1986; Zaruby et al. 1995). Cyclic loading that occurs during the six to eight week time frame that a transfixation pin cast is worn results in pin loosening as a result of bone resorption in addition to local infection that may be present. This cyclic loading will ultimately impact the longevity of the bone-pin interface (actual site of contact between the bone and pin,) and subsequent stabilization offered by a transfixation pin cast (Clary et al. 1996). Thermal and microstructural damage are contributing factors that impact the length of time prior to pin loosening. Minimizing damage at the bone-pin interface at the time of drilling and pin insertion is particularly important when attempting to prevent premature pin loosening.

While similar research hasn't been performed in horses, Eriksson et al. (1983) determined the temperature threshold for bone tissue injury in the rabbit. They demonstrated that the critical threshold for bone necrosis is approximately 47° Celsius, while consistent osteocyte death occurs at 50° Celsius. These researchers also observed that the first signs of bone resorption were noted at least 20 days after the thermal injury occurred. This is consistent with clinical reports of transfixation pin casting in horses (Lescun et al. 2007). This retrospective study found that while on average there was pin loosening evident at 40 days post pin placement, a portion of horses treated for distal limb fractures with transfixation pin casts had evidence of pin loosening as early as 30 days. Additionally, two-thirds of those horses had clinical signs severe enough that pin removal was required.

Toews et al. (1999) suggested that premature pin loosening was likely associated with thermal necrosis secondary to heat created at the time of pin placement. They experimented with drill speed and feed rates (rate of drill tip advancement into bone) of 6.2-mm drill bits and found that a combination of slow drill speeds and high feed rates resulted in the lowest mean maximal temperature recorded 1-mm from the holes that were drilled. The conclusion was that low drill speeds should be used while applying sufficient axial force to advance the drill as rapidly as possible through the bone. In addition to low drill speed and increased axial force while drilling, Lescun et al. (2011) subsequently showed that sequential drilling with 3.2-mm, 4.5-mm, 5.5-mm, and 6.2-drill bits helped lower the amount of heat created during drilling. They concluded that the lower amount of heat within the cadaveric bones was related to the increased amount of time taken to perform the sequential drilling. Bubeck et al. (2009) tested the 4-step sequential drilling method against a single step drill bit method to see if there was a difference in heat generated. These researchers found that there was no significant difference in the amount of heat created and that the single step drill bit method was much faster. Additionally, they theorized that the single drill bit method would lower the opportunity for error that might occur during the 4-step sequential drilling method as they found a difference in the size of the holes created in the cis- and trans-sides of the third metacarpal bone during sequential drilling. This is thought to be due to the amount of instability of the drill bit on the cis-side of the bone as the pin-hole is initially being created. Care when creating the pre-hole is required to keep this difference in the cis- and trans-cortices to a minimum.

Cortical bone can only deform by 2% before breakage occurs, which is observed histologically as microfractures within the bone (Perrin et al. 1979). Zaruby et al. (1995) found that these microfractures occurred when there was as little as a 0.5-mm difference in pilot hole and pinhole size. When drilling pilot holes, the pre-hole diameter should be the same size as the inner diameter of the pin (Clary et al. 1996). This improves initial pin stability as well as reducing microstructural damage. The stability of the bone-pin interface is measured *in-vitro* as "holding power" and is defined as the peak axial tensile extraction force (McClure et al. 2000). Too small of a pin hole allows for damage to the bone at the time of pin insertion, which compromises the interlock between the bone and the implant (Clary et al. 1996). If the pre-hole is too large there is a reduction in the holding strength resulting in implant failure and premature pin loosening (Clary et al. 1996). The discrepancy in pre-hole and pinhole size can occur at the time of drilling dependent upon the type of drilling method used.

1.4 Evolution of the transfixation pin cast

Transfixation pin casts were initially described for use in large animals by Kirk (1952). He described a construct that consisted of placing two pins perpendicularly across the metacarpus that were connected to a horseshoe worn by the horse using two metal bars that were parallel to the limb. Reichel (1956) altered Kirks design to include a Ushaped metal bar with space between the hoof and metal bar. Németh and Numans (1972) later developed the modified walking cast which was used in 20 animals (horses and cattle) with a 75% success rate. In a retrospective study by Németh and Back (1991), in which the walking cast was used to treat 123 animals, including sheep and cattle. When only equine patients were taken into account, there was a 57% success rate. Fractures that were treated included long bone fractures of the radius, tibia, metacarpus, and metatarsus, as well as fractures of the first phalanx. One of a number of complications that occurred while treating these fractures was the inability to make the walking cast strong enough to allow some of the fractures to heal. The plaster of Paris walking cast was stressed under the weight of healthy animals in good condition even with the metal U-bar in place, resulting in bent or broken pins; fractures at the pin sites; and overloading of the contralateral limb. Although these complications were seen less frequently in the treatment of fractures associated with the phalanges, metacarpi and metatarsi, they were a significant problem in fractures of the radius and tibia. It is important to note that although 57% doesn't appear to be a very high success rate, it was considered to be acceptable by the authors given the severity of fractures treated at the time (Németh and Back 1991).

In an attempt to improve upon the external fixation methods available at the time, Nunamaker et al. (1986) introduced a skeletal fixation device that consisted of a foot support that attached to a shoe on the horses' foot, which was then incorporated into an external fixator. The transfixation pins placed across the metacarpus or metatarsus were 9.6-mm stainless steel pins that were partially threaded, and self-tapping, in an attempt to reduce thermal damage to the bone at the time of placement. These pins were incorporated into polyurethane steel-reinforced sidebars that were constructed at the time of surgery. The construct consisted of at least 3 pins placed 5-cm apart. Although patients seemed to exhibit a profound level of comfort following application of the device, there were a significant number of complications experienced. The most common of these was fracture of the third metacarpal bone through a pinhole. It was considered to be a function the size of the pinhole, which was necessarily large to accommodate the size of pins (Nunamaker et al. 1986).

The original walking bar cast and Nunmaker's external fixator device required that the transcortical pins be placed parallel to one another, which may have led to bone fracture. McClure et al. (1994) performed an *in-vitro* study to assess the breaking strength of equine third metacarpal bones when pins were placed parallel to one another vs. at thirty degrees divergence. This study revealed that metacarpal bones with parallel pin placement experienced an oblique fracture associated with the proximal pin tract, while the metacarpal bones that had pins placed thirty degrees from one another were stronger and subsequently resulted in comminuted fractures among multiple pin tracts at higher loads.

In an *in-vitro* experiment conducted by McClure et al. (1994b), comparing a standard short limb cast to three different transfixation cast constructs, there was no significant difference in performance among the 3 different transfixation casts. All of the constructs included two, 6.3-mm smooth Steinman pins placed through 4.8-mm holes drilled in the distal portion of the third metacarpal bone. All limbs were casted with fiberglass casting tape in which the pins were incorporated. The construct designs used were a transfixation cast with pins placed through the bone parallel to one another with a U-bar incorporated, a transfixation cast with pins placed through the bone parallel to one another without the use of an incorporated U-bar, and a transfixation cast with the pins placed through the bone with 30 degrees divergence from the frontal plane, without a U-bar. Given that transfixation pin casts without U-bars performed similarly to transfixation casts with the U-bar in place, the U-bar could subsequently be eliminated from the construct design. That allowed for divergent pin placement, but it also meant that the transfixation casts could be applied quicker with less difficulty, allowing for shorter anesthetic periods.

1.5 Pin placement

A number of factors must be considered in transfixation pin selection. Not only are specific characteristics of the pin important, but also size, and the location of where the pin is to be placed must be taken into account. Qualities that should be selected for provide strength and durability without causing harm to the patient.

Transfixation pins are available in a variety of configurations and come of the more commonly use are either smooth or threaded. Threaded pins are further described as positive-profile or negative-profile. The threads of positive-profile pins extend beyond the core diameter while negative-profile pins do not. Aron et al. (1986), proved that transfixation pins that are threaded have reduced osteolysis and pin-tract infection compared with smooth pins. They also determined that threaded transfixation pins have less medial to lateral migration as well as increased pull out strength. Unfortunately threaded pins are known to be weaker at the threaded-nonthreaded interface; however, this stress concentrator has been eliminated with the development of positive-profile pins (Morisset et al. 2000).

In previous studies and in clinical use, transfixation pins were placed in a variety of different locations throughout the length of bone. Németh and Back (1991) concluded that in order to decrease pin tract fracture, pins need to be placed as distally as possible and that the cast needed to go as far proximally as possible. When the pins were placed too proximal, there was a greater chance that a fracture was going to occur involving the pin tract at that location. They also noted that placing a pin near the proximal extent of the cast caused a greater amount of stress at that site, resulting in fracture through the pin tract. McClure et al. (1994) found that when placing the pins as distally as possible, there was increased risk of damage to the collateral ligament of the metacarpophalangeal joint. As a result they placed the distal pin within the metaphysis of the third metacarpal bone with the second pin 2-cm proximally. The second pin was placed at that location in an attempt to avoid diaphyseal bone where the diameter of bone lessens.

Although the amount of cortical bone is greater within the diaphysis, pin placement within the bone can subsequently weaken it, by creating a stress riser at the bone-pin interface. Edgerton et al. (1990) determined that defects within the bone greater than 10% of the bone's diameter act as a stress riser, and that defects greater than 20% of the bone's diameter decrease the structural stiffness linearly when the bone is placed in torsion. Although increased cortical bone thickness may provide more security when pins are placed within it, Egger et al. (1986) proved that with increased cortical thickness there is an increased likelihood of causing frictional heat to accumulate during pin placement since it takes longer to drill through thicker cortical bone. Whether or not the increased cortical bone thickness of the diaphysis of the equine third metacarpal bone is large enough compared to the cortical bone thickness of the metaphysis of the third metacarpal bone to create a significant change in temperature during drilling is unknown at this time.

Larger diameter pins are stiffer given the fact that the pin's stiffness is proportional to the fourth power of its diameter (Chao and Pope 1982). However, the risk of creating a defect that is too large must be considered when using a larger diameter pin. Another characteristics of transfixation pins that has been taken into account is the fact that deflection plays a major factor in pin function. The distance from the bone-pin interface to the side bar impacts deflection to the third power (Chao and Pope 1982).

Decreasing deflection occurs when pin diameter is larger, or when the distance from the bone-pin interface to the side bar reduces. As the distance from the bone-pin interface and the side bar, or cast material in a transfixation pin cast, is minimal (Figure 1.3), the use of smaller diameter pin is possible.

1.6 Pin location

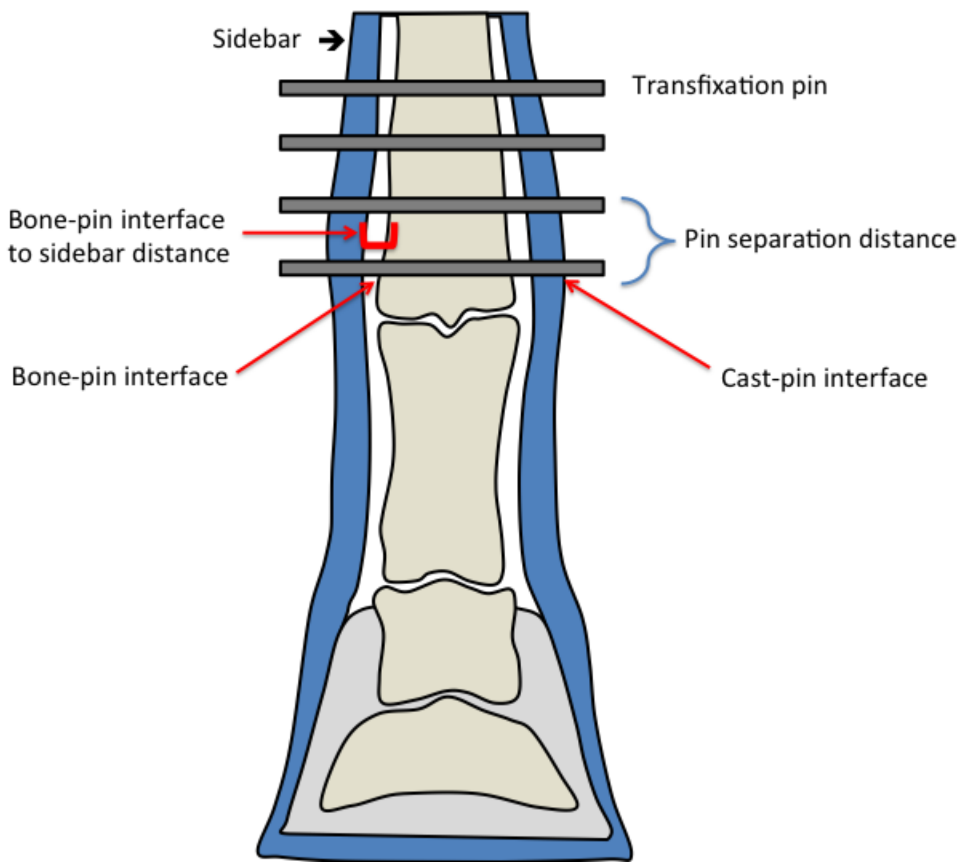
Placing pins further apart can potentially increase the stability of the external fixator (Chao and Kasman, 1982). Typically the principles of pin diameter (pins should not be placed closer than 6 times their diameter) are followed when determining the distance of separation (Vogel and Anderson 2014). However as mentioned above, when following those recommendations, it is easy to end up placing pins within the smaller diameter of the diaphyseal bone. Seemingly, the diaphysis with its small diameter is an unfavorable location for pin placement because of high risk of bone fracture through the pinholes. In order to determine how the diaphyseal bone would handle pin placement McClure et al. (2000) tested the behavior of 6.3-mm centrally threaded, positive profile transfixation pin within the mid-diaphysis and metaphysis of the equine third metacarpal bone. These pins are approximately 20% of the diameter of the mid-diaphysis of the equine third metacarpal bone, which is close to the limit determined by Edgerton et al. (1990).

McClure et al. (2000) found that the centrally threaded, positive profile pins placed within the diaphysis of the third metacarpal bone were more resistant to removal and subsequently had a higher axial extraction forces than those same pins placed within the metaphysis of the third metacarpal bone. These researchers concluded that the diaphysis provides significantly greater ultimate tensile strength than the metaphysis, and

Figure 1.3: Example of a transfixation pin cast using 4 smooth transcortical transfixation pins. Notice the minimal space between the bone-pin interface and the sidebar or casting material in the case of a transfixation pin cast. The stiffness of the external fixator can be described by:

$$K_f = 12ME_SI/S^3$$

K_f = axial stiffness; M = number of pins used; E_s = pin modulus; I = pin area of moment of inertia (proportional to the 4th power); and S = distance from the side bar (cast) to the bone.



placing transfixation pins within the diaphysis might strengthen the transfixation pin cast construct. Additional studies, such as torsional testing would need to be conducted in order to further assess that conclusion.

1.7 Size of pins

The pins used in the walking cast used by Németh and Back (1991) ranged in size from 3.96-mm to 8-mm depending on the size and weight of the horse and where the fracture they were treating was located. As pointed out previously, the bone to pin size ratio needs to be taken into account in order to decrease the risk of fracture through the pin tract secondary to a large cortical defect. Finding a balance between the size of pin, the cortical defect it creates, and the strength of the construct is challenging.

Seltzer et al. (1996) and McClure et al. (2000) pointed out that strength within an implant such as the transfixation pin increases with the size of the implant. This pertains to the fact that increasing the diameter of the transfixation pin increases the stiffness of the transfixation pin to the 4^{th} power (Chao et al. 1982). Unfortunately Seltzer et al. (1996) proved using larger transfixation pins result in larger cortical bone defects, subsequently reducing bone strength. As these researchers ultimately pointed out, the strength of a transfixation pin cast construct can be weakened by using small pins where the construct is weak overall, or by too large of a pin where the bone is also weakened.

When the pins within a transfixation pin cast are too small there is an increased bending moment occurring on the pins when the limb is fully loaded (Crippen et al. 1981; Huiskes et al. 1985; Aro et al. 1993). As the pin bending occurs, the bone at the bone-pin interface is taxed, especially during cyclic loading that occurs as the horse ambulates (Crippen et al. 1981; Huiskes et al. 1985; Nash et al. 2001). Over time, the bone surrounding the bone-pin interface can become resorbed leading to further instability, pain, and ultimately failure (Brianza et al. 2010). Nash et al. (2001) developed and tested a tapered-sleeve transcortical pin, in an attempt to reduce the stress at the bone-pin interface. They proposed that by decreasing the bending moment, they could eliminate much of the stress, and with it many of the problems, associated with the bone-pin interface. After application of the tapered-sleeve transcortical pin, the researchers concluded that the pins used in their study were able to withstand higher loads and had increased stiffness. Although the construct seemed promising, its use *in vivo* has not been evaluated and subsequent clinical use has not been adopted.

Similarly, Nutt et al. (2010) compared a solid sidebar external fixation device with sleeves covering the transcortical pins to a full limb transfixation pin cast utilizing two 7.94-mm transcortical pins. They found that under static and cyclic loading, the solid external fixation device was stronger and stiffer than the transfixation pin cast. Although this construct shows promising results it has not yet been tested *in vivo*. In a similar fashion Brianza et al. (2010), tested a novel pin-sleeve system against a transfixation pin cast. With their design, a pin-sleeve is inserted through the bone and subsequently houses a single 6.3-mm pin that crosses an external ring that would be incorporated into a cast.

The pin-sleeve is designed to act as the bone-pin interface within the sleeve. Since the "bone-pin" interface is maintained within the pin-sleeve that is securely placed within the bone, the pin is able to experience bending moment without the stress occurring at a bone-pin interface (Brianza et al. 2010). They found during in vitro testing, the novel pin-sleeve system reduced strain around the implant while having a similar axial displacement as the transfixation pin cast system it was compared to. Additional *in vivo* testing would need to be conducted in order to determine this constructs' usefulness in a clinical setting.

Williams et al. (2014) compared the performance of two transfixation pin cast constructs in the third metacarpus in which one construct contained two 6.3-mm positive profile pins and the other construct contained four, 4.8-mm smooth Steinmann pins. They found that both constructs provided a larger reduction in strain on the dorsal aspect of the distal limb compared to a non-casted control, but that neither transfixation pin cast construct was superior. Within that study they removed the most proximal pin of each construct and found that the strain measured on the dorsal aspect of the distal limb increased in both constructs, while the most significant decrease in strain occurred at higher loads.

Although fiberglass casting tape is the current casting material of choice, Rossignol at al. (2014) reported the use of a modified transfixation pin cast that utilized both plaster of Paris and fiberglass casting tape in adult horses. In addition to both plaster of Paris and fiberglass casting tape, two splints were incorporated within the layers of the cast. As both materials were being applied, they were placed in a figure of eight pattern around the transcortical pins. This was done to increase the contact of the cast material and pins. Plaster of Paris was chosen so that the cast could be molded around the distal limb, while still utilizing lighter fiberglass casting tape to incorporate fiberglass splints in an attempt to increase the overall strength and comfort of the cast (Rossignol et al. 2014). They found that the horses wore their casts well even though they ended up being heavier. In addition to changes in the application of the cast, the most distal transcortical pin was placed within the epiphysis of the third metacarpal bone while the proximal pin was placed in the distal metaphysis rather than the more traditional metaphyseal placement (Rossignol et al. 2014). The thought was that the epiphyseal and distal metaphyseal regions contain a higher content of cancellous bone that is known to be less brittle and fails at a higher strain, in addition to being tougher than cortical bone more proximally (Auer et al. 2012; Rossignol et al. 2014). They found that when transcortical pins were removed 6-8 weeks after placement there was no evidence of pin loosening. At the end of their study, Rossignol et al. (2014), determined that 82% of the horses that wore this modified transfixation pin cast survived. They concluded that although they could not comment on the superiority of their construct over others, the results of their study seemed promising and agreed additional cases would be needed.

1.8 General objectives and specific aims

The goal of this research was to determine the optimum placement of transcortical pins in an equine transfixation cast construct. This thesis constitutes a first approach to this goal. The specific aims of this thesis are:

1. To determine the load to failure of three different transfixation cast constructs tested in single cycle to failure. This would mimic recovering a horse from general anesthesia (Chapter 2).
2. To describe cyclic load parameters by loading the cast to mimic 6 weeks worth of wear. This is representative of the expected time period of transfixation cast use in fracture management (Chapter 3).

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CHAPTER 2

EVALUATION OF TRANSFIXATION CAST CONSTRUCTS IN HORSE FORELIMBS: A MECHANICAL STUDY

CHAPTER 2

Transition Page

Evaluation of transfixation cast constructs in horse forelimbs: A mechanical study

This chapter utilized strain measurements in addition to finite element analysis to determine the behavior of two different transfixation pin cast models after a single load to failure was applied. The load applied was similar to what a horse would experience while recovering from anesthesia.

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Author contribution: Thomas, Carmalt, and Burnett were responsible for the experimental design. Johnston was responsible for data analysis. Arjmand and Johnston were responsible for the finite element analysis. Thomas, Burnett, Arjmand, and Johnston wrote the manuscript with Carmalt providing editorial assistance.

2.1 Introduction

Transfixation pin casts are generally used for comminuted fractures where it is difficult to apply internal fixation for fracture repair (although it is not uncommon to use them in conjunction with internal fixation). Typically, the transfixation pin cast consists of pins that pass through the bone proximal to a fracture. The pins are then incorporated within a cast, taking the majority of the weight, thereby partially unloading the fracture. In fact, the main purpose of the transfixation pin cast is to allow an appropriate amount of load on bone distal to the pins, encouraging the fracture fragments to heal without an excessive amount of applied load and corresponding bone deformation (strain) (McClure et al. 1994).

Transfixation pin casts have been used to treat fractures of large animals for more than fifty years (Reichel 1956); though, there are associated complications, including pin loosening, secondary infection, or necrosis at the bone-pin interface (Mahan et al. 1991; McClure et al. 1995). The bone-pin interface is typically a site of excessive motion that ultimately results in loosening and pain (Nunamaker et al. 1986; Németh and Back 1991; Morisset et al. 2000). Consequently, it is often necessary to remove or replace pins in addition to replacing the cast. In an attempt to decrease complications at the bone-pin interface, many aspects have been investigated, such as pin size, smooth vs. positive-profile pins, self-tapping vs. non-tapping positive-profile pins, hydroxyapatite-coated positive-profile pins, and diaphyseal vs. metaphyseal pin placement (Aron et al. 1986; McClure et al. 2000; Zacharias et al. 2007; Bubeck et al. 2010). At present, the optimal fixation strategy for decreasing bone-pin interface complications, minimizing motion, and unloading the fracture site is unknown.

Currently, traditional transfixation pin casts incorporating the 3rd metacarpal or metatarsal bone consist of two 6.3-mm positive-profile pins placed proximal to the fracture location. Pins are placed with 30 degrees relative to one another in the transverse plane, offering improved resistance to torsional loading over parallel pin placement (i.e., metacarpals with parallel pins fractured at lower torques than metacarpals with diverging pins) (McClure et al. 1994). Recently, it was suggested that a transfixation pin cast construct consisting of four 4.8-mm smooth pins spaced 2-cm apart (rather than two 6.3-mm positive-profile pins spaced 4-cm to 5-cm apart), with some degree of divergence, would be a more appropriate construct design.[a] The idea for this alternate design arose from the supposition that 6.3-mm pin designs are too rigid, inhibiting normal biomechanical stimuli needed to achieve appropriate healing (Smith 1985). Alternately, smaller diameter pins will allow for some distribution of load across the fracture site in order to promote fracture healing (Perrin 1979), while still protecting it from excess strain. A study by Williams et al. (2014), indicated no difference in the amount of strain on the dorsal aspect of the proximal phalanx with two 6.3-mm centrally-threaded, positive-profile pins or with a transfixation pin cast with four 4.8-mm smooth Steinman pins. These results suggest that similar loading is occurring at the fracture site despite the use of different transfixation constructs; though, this information is currently unknown.

Using a combination of experimental testing and finite element (FE) modeling, the objective of this study was to investigate bone strain in the cannon bone (3^{rd} metacarpal) and proximal phalanx, as well strain in the surrounding cast and bone-pin interfaces with three different transfixation pin cast configurations: two 6.3-mm centrally-threaded positive-profile pins spaced 4-cm apart, two 6.3-mm centrally-threaded positive-profile pins spaced 5-cm apart, and four 4.8-mm centrally-threaded positive-profile pins spaced 2-cm apart. The null hypothesis of this study was that there are no differences in the amount of strain at the bone-pin interface or cast-pin interface between the three different transfixation cast constructs.

2.2 Materials and methods

2.2.1 Mechanical Testing

2.2.1.1 Specimens

Fifteen (n=15) forelimbs from 15 horses were collected from adult horses euthanized for reasons not related to the musculoskeletal system. All limbs were disarticulated at the carpometacarpal joint. The cannon bone was sectioned with a reciprocating saw at 25% of its length distal to the carpometacarpal joint, creating a flat surface for load application during mechanical testing. The distal limbs, with all soft tissue structures left intact, were wrapped in towels soaked in saline solution[b], sealed in plastic, and stored at -20° Celsius until tested. Limbs were thawed at room temperature (approximately 21° Celsius) for 24 hours prior to testing. After thawing, limbs were prepared for mechanical testing and randomly assigned to one of three different construct groups (with 5 limbs per group):

Construct 1: two 6.3-mm centrally-threaded positive-profile pins spaced 4-cm apart,

Construct 2: two 6.3-mm centrally-threaded positive-profile pins spaced 5-cm apart,

Construct 3: four 4.8-mm centrally-threaded positive-profile pins spaced 2-cm apart.

2.2.1.2 Constructs

Construct 1: a stab incision was placed on the lateral aspect of the metaphysis, just proximal to the lateral epicondyle of the 3rd metacarpal bone. A second stab incision was placed 4-cm proximal into the diaphysis with 30 degree divergence in the transverse plane (Figure 2.1A, Figure 2.2B). Although the conventional recommendation is to use 2-cm spacing, we elected to use 4-cm spacing as prior research indicates that the diaphysis provides greater pin stability than the metaphysis (measured by resistance to axial extraction) (McClure et al. 2000). Drill holes were performed in sequence starting with a 3.2-mm drill bit, followed by a 4.5-mm drill bit, then a 5.5-mm drill bit, and lastly a 6.2-mm drill bit. The drill holes were then tapped with a 6.3-mm tap. Two 6.3-mm centrally-threaded, positive-profile pins were placed under power. Figure 2.1A illustrates pin placement; Figure 2.2A illustrates pin divergence.

Figure 2.1: Testing configurations for transfixation pin casts, including: **A)** Construct 1, consisting of two 6.3-mm diameter pins spaced 4-cm apart; **B)** Construct 2, consisting of two 6.3-mm diameter pins spaced 5-cm apart; and **C)** Construct 3, consisting of four 4.8-mm diameter pins, each spaced 2-cm apart. Uniaxial strain gauges (orange) are placed medially and laterally proximal to the bone-pin and cast-pin interfaces, as well as proximal and distal to a manually created fracture in the proximal phalanx (P1).

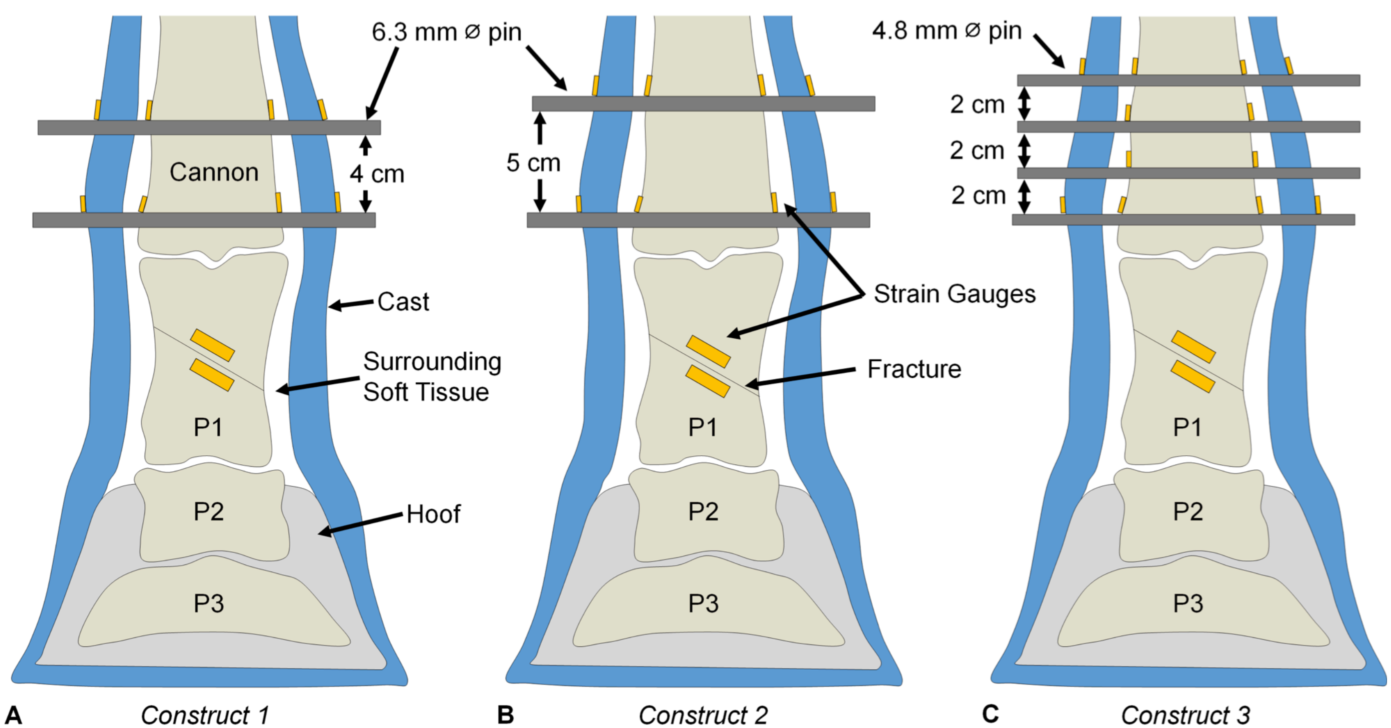
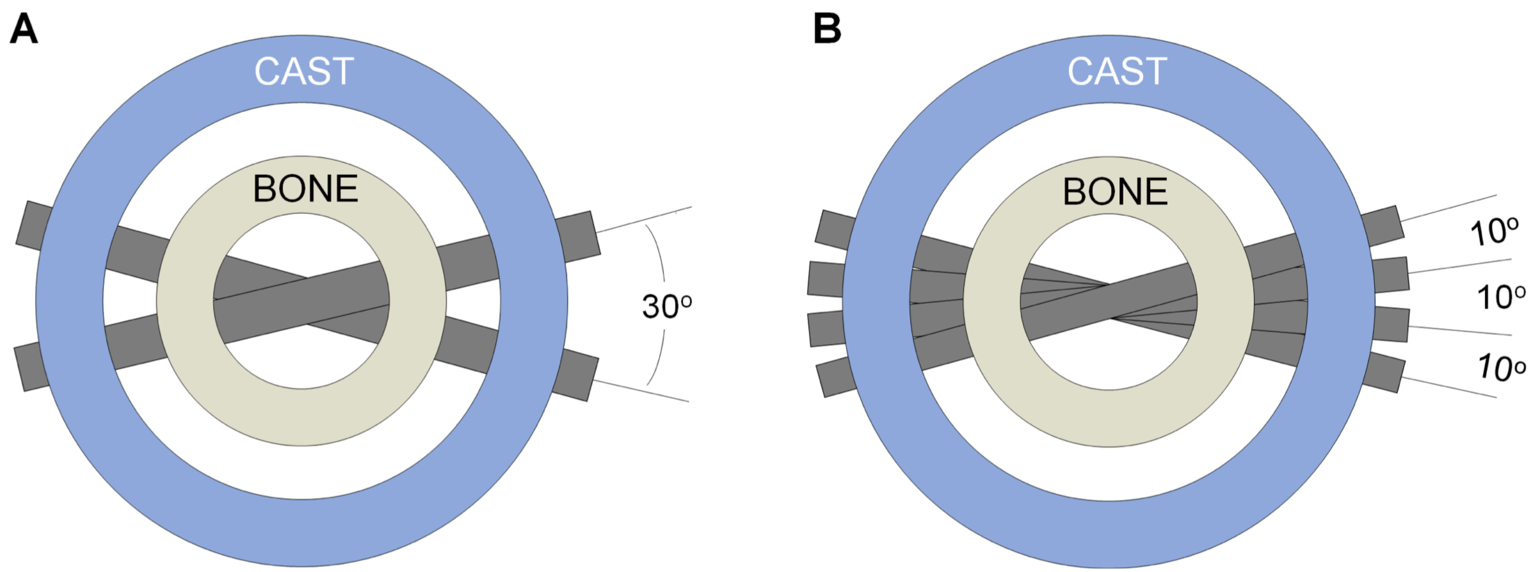


Figure 2.2: A) 30^{o} pin divergence arrangement for Construct 1 and 2 in the transverse plane; **B)** 10^{o} pin divergence arrangement for Construct 3 in the transverse plane.



Construct 2: a stab incision was placed on the lateral aspect of the distal metaphysis, just proximal to the lateral epicondyle of the 3rd metacarpal bone. A second stab incision was placed 5-cm proximal with 30 degree divergence (Figure 2.1B, Figure 2.2A). Drill holes and pins were placed as previously described.

Construct 3: a stab incision was placed on the lateral aspect of the distal metaphysis, just proximal to the lateral epicondyle of the 3rd metacarpal bone. A second stab incision was placed 2-cm proximal with 10 degree divergence (Figure 2.1C, Figure 2.2B). Third and fourth stab incisions were placed proximally 2-cm between one another in a 10 degree divergence from the previous pin orientation, so that adjacent pins were not in the same plane as one another. With this approach, the first and fourth pins were oriented at 30 degrees to one another, mimicking divergence with Constructs 1 and 2. Drill holes were performed in sequence starting with a 3.5-mm drill bit, followed by a 4.5-mm drill bit. The drill holes were tapped with a 4.8-mm tap. Four 4.8-mm centrally-threaded, positive-profile pins were placed under power.

For all specimens, an osteotomy was created in the proximal phalanx to simulate a fracture where a transfixation pin cast would commonly be used as a treatment option. The osteotomy was oriented at 30 degrees in the frontal plane of the phalanx, from proximolateral to distomedial, and was created with a reciprocating saw (Figure 2.1). Lastly, all pins were cut with the use of a bolt cutter to leave 4-cm of length protruding from the specimens.

2.2.1.3 Strain Gauge Instrumentation

Each specimen was dissected to expose the bone proximal to each pin medially and laterally as well as the dorsal surface of the osteotomy. The subcutaneous tissues, common digital extensor tendon, and periosteum were dissected and freed from the bone with the use of a #3 scalpel handle and #10 scalpel blade. The exposed bone was debrided lightly with sand paper, cleaned with 70% ethyl alcohol and allowed to dry in room air. Uniaxial strain gauges[c] were secured with cyanoacrylate following manufacturer's recommendations and best practices for application to bone *in vitro* (Cordey and Gautier 1999). A gauge was applied to the bone with the distal edge of the gauge located 1- to 2-mm proximal to each pin (Brianza et al. 2010), both medially and laterally, as well as proximal and distal to the osteotomy (Figure 2.1). Gauges were applied in a proximal-to-distal direction except adjacent to the osteotomy, where the gauges were applied parallel to the oblique osteotomy (Figure 2.1). Strain gauges and surrounding area (~1-cm radius) were sealed with polyurethane coating[d] and leads were further secured to the limb with electrical tape up to the carpometacarpal joint. The skin over the gauges and exposed bone was sutured in a simple continuous pattern with nylon suture material in order to protect the strain gauges during testing.

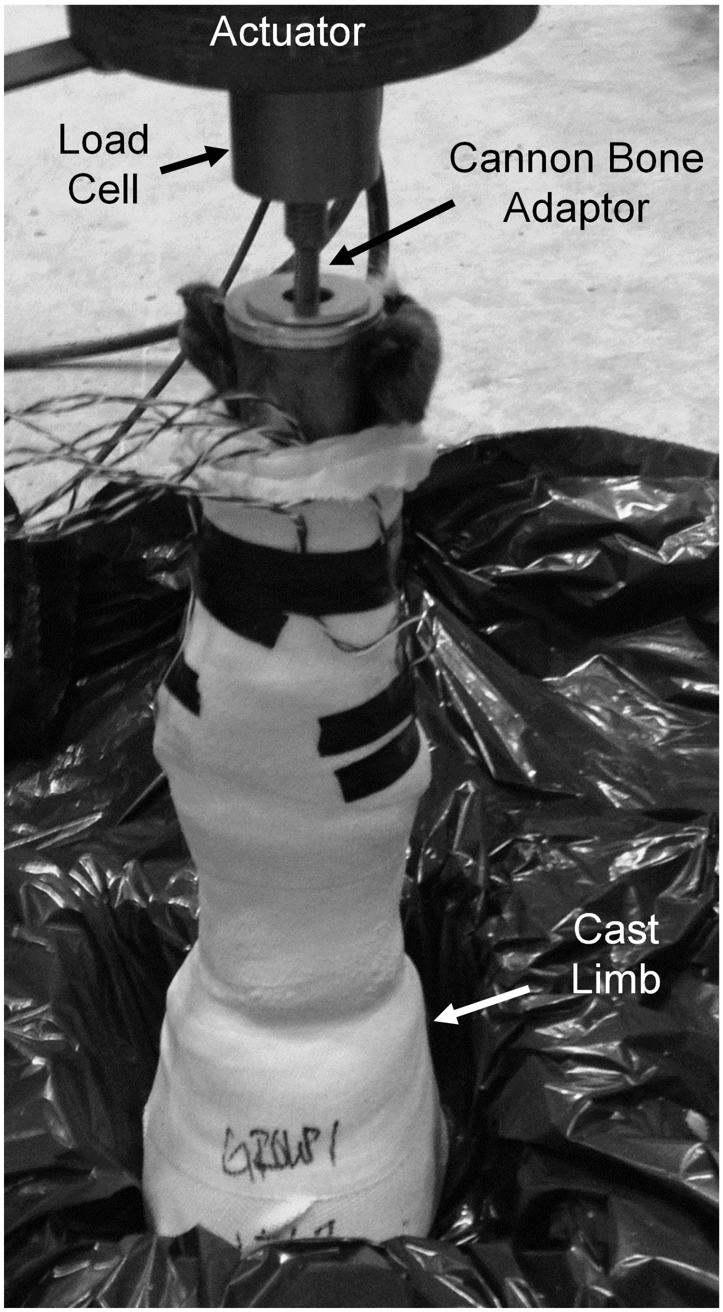
2.2.1.4 Cast Application

Casts were constructed in a standard fashion, incorporating the foot and extending proximal to 2-cm below the carpometacarpal joint. Two layers of stockinette were applied followed by 5 rolls of 4-inch fiberglass cast material[e]. Rationale for this choice was based upon a previous study by McClure et al. (1994), which used 2 rolls of 4-inch fiberglass casting tape and 2 rolls of 5-inch fiberglass casting tape. The pins were incorporated into the cast by making incision through the fiberglass cast material. Uniaxial strain gauges were attached to the cast with the distal edge of the gauge located 1- to 2-mm proximal to the pin in a proximal-to-distal direction, following manufacturer's recommendations for porous materials. Leads were secured to the cast with electrical tape to the proximal extent of the cast. All casts were allowed to cure and dry for approximately 1.5 hours prior to testing.

2.2.1.5 Biomechanical Testing

Limbs were axially loaded to failure using a hydraulic actuator[f] instrumented with a linear variable differential transformer (LVDT)[g] and load cell.[h] Each specimen was placed under the actuator and secured in place with the use of a custom frame (Figure 2.3). The toe of each cast made contact with the frame, which prevented forward sliding once axial compressive loading was applied. The load cell was fitted with an adaptor and plate which applied load to the sectioned surface of the cannon bone. The adaptor sat in

Figure 2.3: Photograph of compression testing setup consisting of a hydraulic actuator, 250 kN load cell, and adaptor rigidly connected to the cannon bone of the cast limb.



the marrow cavity to a depth of 1-cm, and was used to ensure that the loading plate was centered across the cannon bone.

Limbs were loaded in a single cycle to failure with axial compression at a constant rate of 2-mm/s (Nash et al. 2001; Elce et al. 2006; Nutt et al. 2010). Load, displacement, and strain data were collected at 250 Hz. Axial load was applied until catastrophic failure was observed, defined as fracture of the proximal 3rd metacarpal bone, cast breakage, or bowing of the load cylinder. The limb was returned to a freezer to be dissected at a later date.

2.2.1.6 Data Analysis

Strain at 2.5 kN (corresponding with standing (Turner et al. 1975)) and 7.5 kN (corresponding with walking (Turner et al. 1975)) was used to evaluate load transfer between the different constructs.

2.2.1.7 Statistical Analysis

Descriptive data from each test was reported as the raw, mean, and standard deviation (SD) for strain measures. The maximum compressive strain between medial and lateral strain gauges at each pin was selected for analysis. To permit comparison between the different constructs, only strain proximal to the first and last pins were used in the statistical analysis. All variables were tested for normality using Shapiro-Wilk tests. The null hypothesis was that there was no significant difference in the amount of strain at the bone-pin interface or cast-pin interface between the different constructs. For normally distributed variables, the null hypothesis was tested using an analysis of variance. If an overall significant difference was found, pairwise comparisons were conducted using Student's t-tests. For variables that were not normally distributed, non-parametric tests were used: the overall effect was tested using Kruskal-Wallis and pairwise comparisons were conducted using Mann-Whitney U tests. For all tests, a p-value of less than 5% ($p<0.05$) was considered statistically significant. Statistical analyses were performed using commercial software[i].

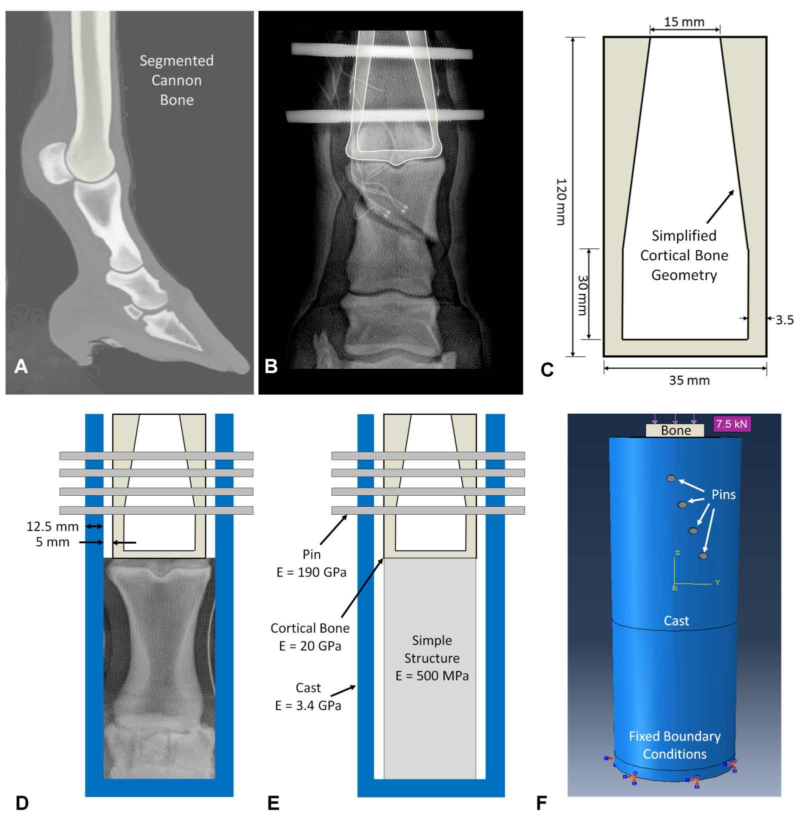
2.2.2 Finite Element (FE) Modeling

Finite element (FE) modeling[j] was used to evaluate mechanical behavior (e.g., load transfer, bone strain) in Construct 1 and Construct 3. Construct 2 was not evaluated as preliminary analyses indicated nearly identical results as Construct 1.

2.2.2.1 Geometry

A computed tomography (CT) image of a horse forelimb was used (Figure 2.4A) to derive a simplified circular geometry representing the cannon bone (Figure 2.4B, Figure 2.4C). Since trabecular bone has a low elastic modulus (~1/40 that of cortical bone), only cortical bone was modeled, transitioning from thick cortical bone at the midshaft (10-mm thick) to thin (3.5-mm thick) subchondral cortical bone at the distal site

Figure 2.4: A) Computed tomography image used to approximate geometry of the cannon bone; **B)** Outlined cannon bone from x-ray, further used to approximate geometry of cannon bone; **C)** Simplified cortical bone geometry used for finite element (FE) modeling; **D)** Illustration of the cast, pins and cannon cortical bone geometry, including physical dimensions; **E)** Defined elastic moduli for the cast, pins, and cortical bone. As all structures distal to the fetlock joint were similar between the 2-pin and 4-pin configurations, these structures (which included P1, P2 and P3 as well as articulating cartilage and hoof) were modeled with a single elastic modulus equal to 500 MPa; **F)** Simple FE model of the cast configuration. The distal portion of the cast was rigidly fixed and a 7.5 kN equivalent load was applied to the proximal portion of the cannon bone.



(fetlock joint) (Figure 2.4C). The cast was modeled with 12.5-mm thickness with a 5-mm gap between the bone and the cast (Figure 2.4D). Pins were transversely placed 30 degrees to one another for Construct 1 and 10 degrees to one another for Construct 3 (Figure 2.4D, Figure 2.2). The distance between the bottom pin and fetlock joint was 3.5-cm for both models. Since the tissues below the fetlock joint were the same between the two cast constructs, and thus would have the same effect on load transfer, they were modeled as a simple cylindrical structure (Figure 2.4E). The simple cylindrical structure was assumed to be completely bonded to the bone and cast. The fracture site was assumed to be located 3-cm distal to the fetlock joint, sufficiently far from the joint to achieve a uniform strain distribution. We did not model the 30-degree osteotomy; instead, we monitored strain and load at the fracture site for a relative comparison of the different transfixation casts in unloading bone.

2.2.2.2 FE Material Properties

Quadratic, 10-node tetrahedral elements were used to mesh the model. Isotropic, homogeneous and linearly elastic materials were used for modeling cortical bone (E = 20 GPa, Poisson's ratio = 0.3) (Rauber 1876), pins (E = 190 GPa, Poisson's ratio = 0.305) (Budynas and Nisbett 2011) and cast (E = 3.4 GPa, Poisson's ratio = 0.3) (Wytch et al. 1987) (Figure 2.4E). Cast material properties were taken to be the mean of longitudinal and flexural modulus properties reported by Wytch et al (1987), with the assumption that flexural modulus is equal to elastic modulus. A single 'effective' elastic modulus (E = 500 MPa, Poisson's ratio = 0.3) was used to model all bony and soft tissue structures distal to the fetlock joint (including the fracture site). This modulus was derived iteratively until FE-derived compressive strain results at three sites (proximal and distal cast sites, proximal bone site) mimicked experimental strain measures found with the 2-pin and 4-pin configurations. FE strain predictions were within ±13% of average experimental values reported in this study. We deemed this level of accuracy sufficient for application in this study.

2.2.2.3 Loads and Boundary Conditions

The most distal section of the cast was fully constrained in all directions. Additionally, the pin-bone and pin-cast contact surfaces were completely bonded to avoid any relative movement between them. We deemed this reasonable as sliding should not have occurred between the threaded holes and threaded pins during experimental testing. A uniform pressure equivalent to the force during walking (7.5 kN) was applied to a rigid plate which was placed on top of the most proximal section of the cannon bone to simulate experimental test loading (Figure 2.4F).

2.2.2.4 FE Outcomes

FE-derived mechanical outcomes included average bone compressive strain at the distal bone-pin interface and fracture site for both the 2-pin and 4-pin transfixation pin cast constructs, as well as relative amount of load transferred to the cast and fracture site. For the bone-pin interface strain measure, we report the average of the strain values of the proximal elements around the pin. For load measures, load was calculated by summating the product of stress and area in the compressive direction for each element at the fracture site. Stress values were assumed to be constant over each element volume.

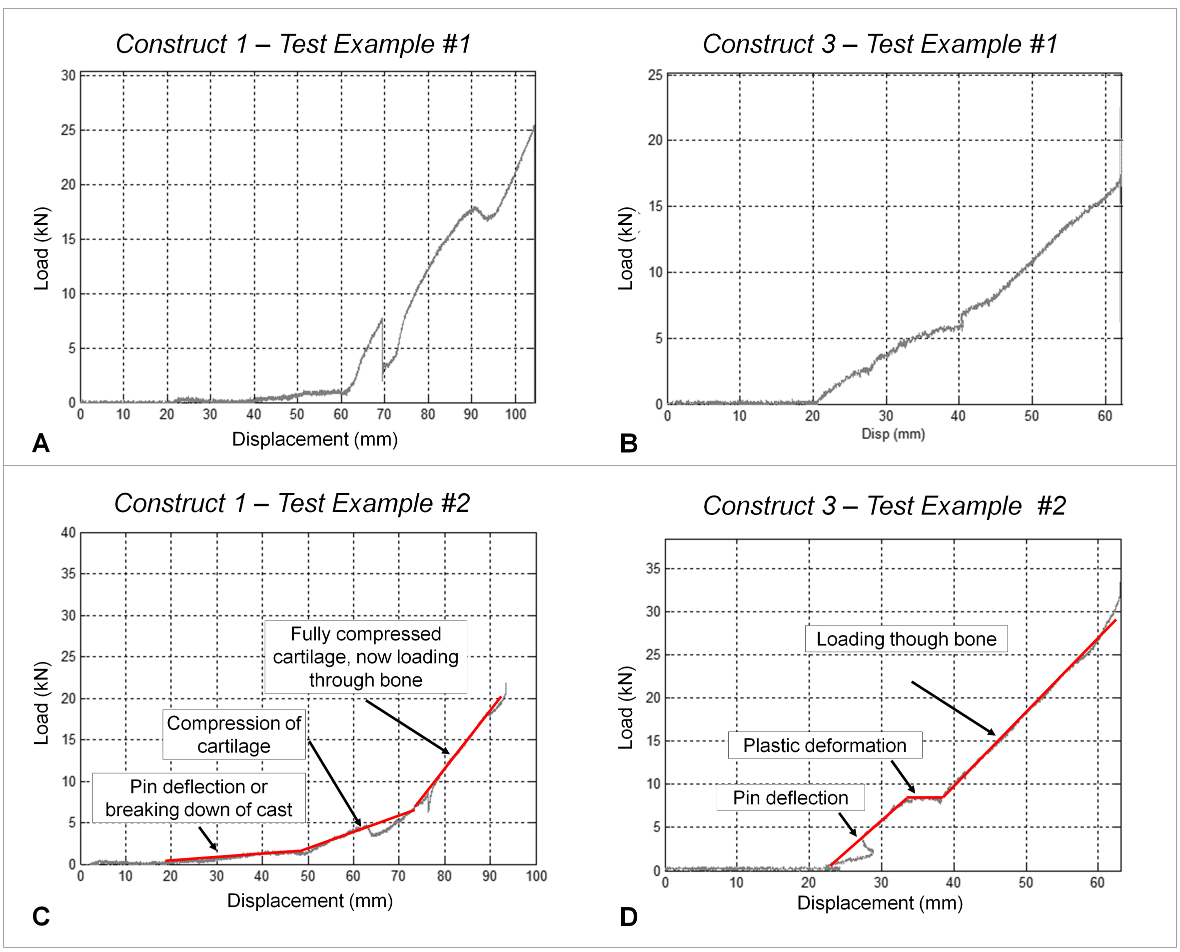
2.3 Results

Qualitative analyses of the load-displacement data indicated that the 2-pin constructs (Constructs 1 & 2) exhibited erratic behavior (Figure 2.5A). The 4-pin constructs, however, exhibited fairly stable, linear mechanical behavior (Figure 2.5B).

Strain gauge data across the different constructs was somewhat variable, more so for the 2-pin constructs. In particular, strain data adjacent to the osteotomy was highly erratic and somewhat unusable (e.g., in some cases load was transferred on the dorsal side in-line with the strain gauges, in other cases load was transferred on the palmar side). Accordingly, we did not include osteotomy strain data in our statistical analysis.

All included variables, except for strain at the proximal cast pin, were normally distributed. Although strain data was variable, comparison of the main effect of treatment was statistically significant ($p<0.05$); therefore, the null hypothesis was rejected. Pairwise comparisons indicated no difference in strain between the 2-pin construct with 5-cm spacing and the 2-pin construct with 4-cm spacing ($p>0.05$) (Figure 2.6); however, differences in strain were noted between the 2-pin and 4-pin constructs ($p<0.05$) (Figure 2.6). When compared alongside the 2-pin constructs, the 4-pin construct had less strain at

Figure 2.5: Representative load-displacement results for **(A)** Constructs 1 & 2, and **(B)** Construct 3**. (C)** With Constructs 1 & 2, results sometimes indicated that the cast and/or pins broke down, leading to compression of the cartilage and load transfer through the bone after 7.5 kN. **(D)** With Construct 3, after 7.5 kN there were two specimens which endured plastic deformation of the pins, leading to load being transferred through the bone (as opposed to the pins and cast).



the bone-pin interface at the most proximal site (-69% and -75%) and higher strain in the cast at the most distal site (+269% and +430%), (Figure 2.6). Similar findings were noted at 7.5 kN (Figure 2.6). In particular, there was higher strain in the cast at the most distal site (+218% and +386%).

An analysis of the relative strain between the different pins indicated that, with the 2-pin constructs, the most proximal pin carried the most load (Figure 2.6); whereas, with the 4-pin construct, the two most proximal pins carried the most load (Figure 2.6).

FE results indicated that, when compared alongside the 2-pin construct, the 4-pin construct had less load transferred to the fracture site (-6.4%), lower bone strain at the distal bone-pin interface (-18%), and lower strain adjacent to the fracture site (-22%) (Table 2.1). Local bone strain at the bone-pin interface was quite high, reaching approximately -15000 microstrain for both the 2-pin and 4-pin constructs (Figure 2.7). The region experiencing high local strain was smaller with the 4-pin construct than the 2-pin construct (Figure 2.7).

One particularly interesting finding regarding to the FE analysis and mechanical testing pertained to stress in the pins. With the 4-pin construct, pin stress (specifically von Mises stress) at 7.5 kN was 280 MPa, which is quite close to the yield strength of stainless steel (260 MPa (Budynas and Nisbett 2011)). Analysis of the mechanical testing

Figure 2.6: Maximum compressive strain (μstrain) in equine limbs with different transfixation pin cast strategies (2-pin 4-cm spacing, 2-pin 5-cm spacing, and 4-pin 2-cm spacing) under 2.5 kN and 7.5 kN compressive loading. For the pin and cast measures, results pertain to the maximum (negative) value between medial and lateral strain gauges. For the osteotomy, results pertain to the maximum (negative) value between proximal and distal strain gauges. Statistically significant between-group differences ($p<0.05$) are marked with brackets. Error bars represent 95% confidence intervals. Analysis of variance and Student's t-tests were employed for all variables apart from the Proximal Cast at 2.5-kN, which was assessed using Kruskal-Wallis and Mann-Whitney U tests.

Figure 2.6:

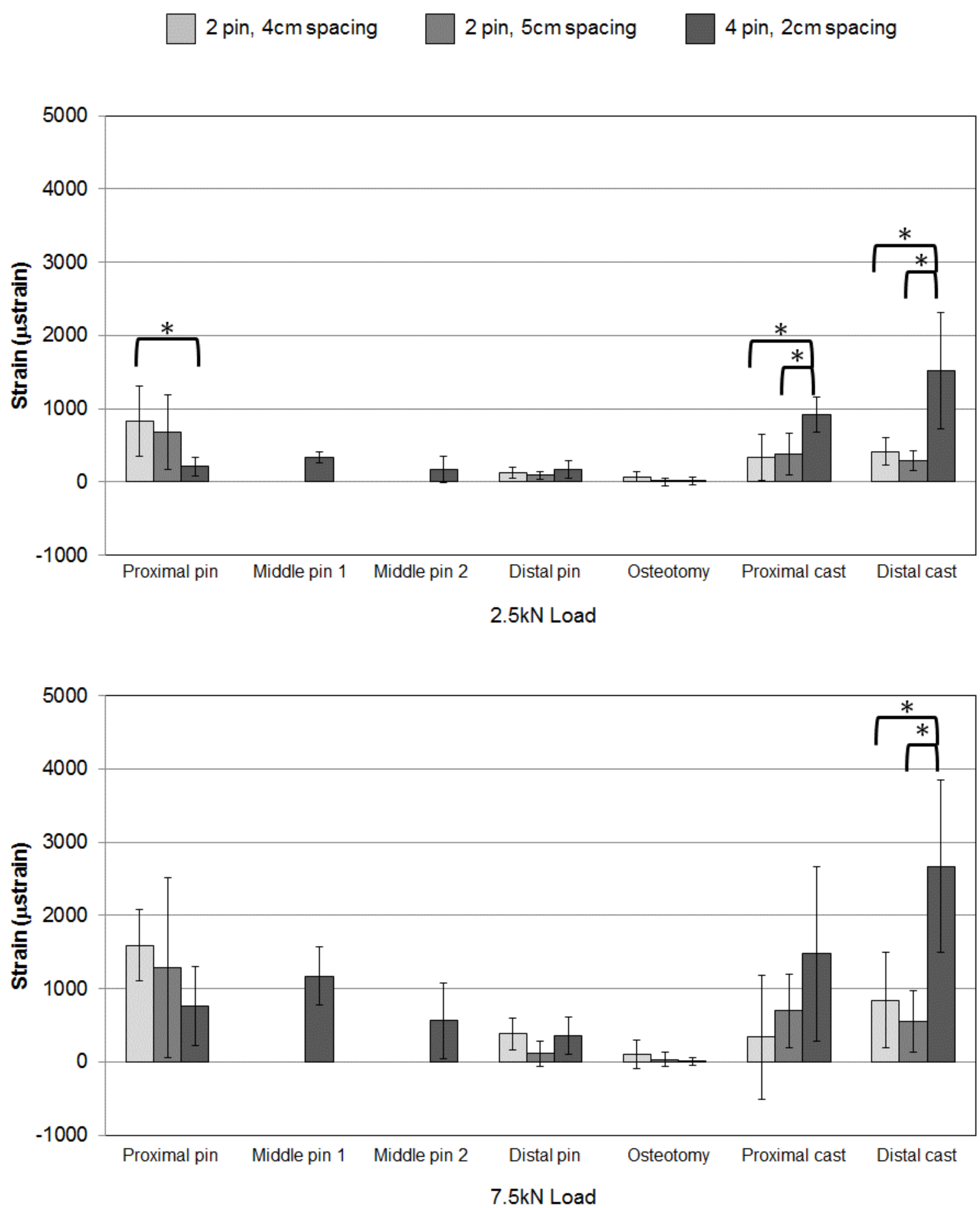
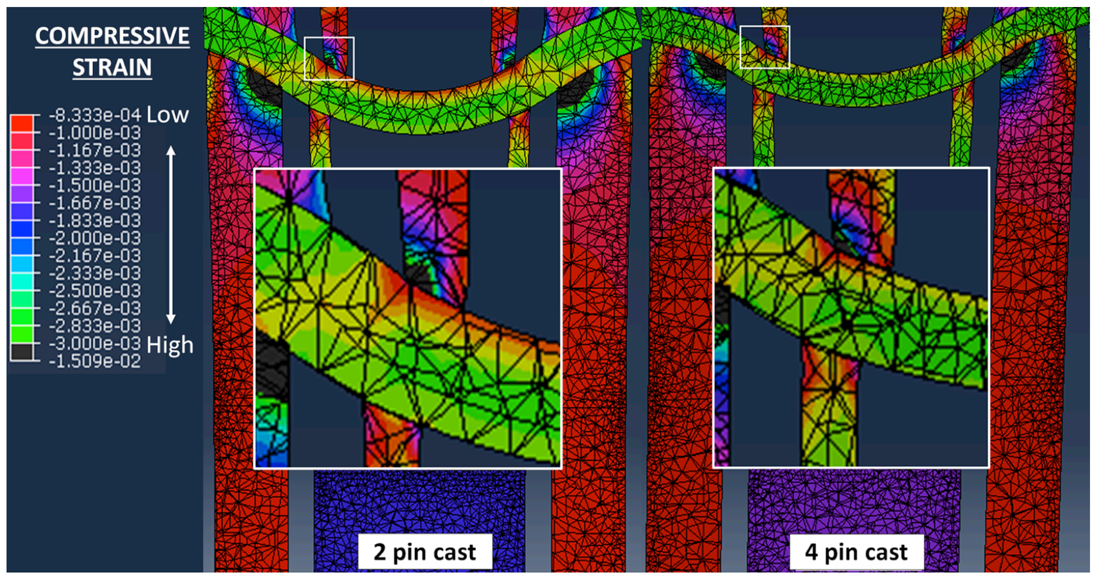

2 pin, 4cm spacing
2 pin, 5cm spacing
4 pin, 2cm spacing
5000
4000
3000
2000
1000
0
-1000
Strain (µstrain)
Proximal pin
Middle pin 1
Middle pin 2
Distal pin
Osteotomy
Proximal cast
Distal cast
2.5kN Load
Strain (µstrain)
7.5kN Load
*

Table 2.1: Summary of finite element (FE) results for the 2-pin construct (with 4 cm spacing) and the 4-pin construct under 7.5-kN axial loading.

FE Outcome	2-pin construct	4-pin construct
Compressive strain at bone-pin interface (μstrain)	-2667	-2222
Compressive strain at fracture site (μstrain)	-1875	-1460
Load going into the fracture site (N)	788	737
Load going into the cast (N)	6712	6763

Figure 2.7: Finite element (FE) results illustrating compressive strain (specifically minimum principal strain) at the bone-pin interface for the 2-pin, 4-cm spacing construct and the 4-pin, 2-cm spacing construct. Zoomed-in results indicate less compressive strain on the 4-pin construct relative to the 2-pin construct at the bone-pin interface. Please note that pin and construct deformation is exaggerated for visualization purposes.



data confirmed this finding, with plastic deformation occurring around 7.5 kN in two of the five 4-pin constructs (Figure 5D). Visual examination of the legs following testing confirmed plastic deformation.

2.4 Discussion

Using experimental testing and FE modeling, this study compared the mechanical performance of different transfixation pin cast constructs at unloading a fracture site. Our results indicate that the two 2-pin constructs transfer similar load to the cast whereas the 4-pin construct transfers more load to the cast, suggesting that the 4-pin construct may provide more fracture site protection than 2-pin constructs. Also, the 2-pin construct had higher strain at the bone-pin interface, possibly explaining pin loosening complications associated with this treatment strategy.

Our study findings conflict with those of Williams et al. (2014), who indicated no difference in the amount of strain with two 6.3-mm centrally-threaded positive-profile pins or with four 4.8-mm smooth Steinman pins. In their study, strain was measured on the dorsal aspect of P1, sans osteotomy, under a 5 kN compressive load. The differences in results could be related to our use of centrally-threaded positive-profile pins, with previous studies showing that centrally-threaded positive-profile pins are stiffer than smooth pins. It is important to note though that our observed differences are small (e.g., the FE analysis indicated ~50N difference in load transfer between the 2-pin and 4-pin constructs), which is in-line with the findings of Williams et al (2014). It is also important to note that although the distal cast sites exhibited large differences in strain between the 2-pin and 4-pin constructs, the 4-pin strain measures reflect strain due to loading as well as strain imposed by proximal pins under bending. As such, they somewhat overestimate the degree of loading being transferred to the cast. Nevertheless, our experimental and FE results suggest that the 4-pin construct may be more protective of the fracture site (albeit modestly). This study found that the 2-pin constructs had higher strain at the bone-pin interface relative to the 4-pin construct. This is important because we believe high strain, in combination with factors associated with the 2-pin surgical technique, may possibly explain complications associated with the 2-pin treatment strategy, specifically necrosis at bone-pin interfaces and pin loosening. To clarify, Chamay and Tschantz (1972) found that excessive strain could lead to local bone failure and subsequent avascular necrosis (due to limited blood supply). They also concluded that local bone failure would also lead to pin loosening, which will further damage bone due to repetitive (fatigue) loading. Tissue irritation at the site of the loosened pin would lead to swelling and draining at the pin track, thereby further encouraging loosening. Although the external ends of the pins are typically covered, they are exposed to dirt, debris, and fecal material. If coming in contact with compromised tissue (say through a larger pin track due to excess motion associated with a loosened pin), this could result in infection and osteitis. Irrespective of loading, the 2-pin surgical technique involves multiple sequential size increases for hole creation. As such, the bone-pin interface may experience thermal necrosis (Matthews and Hirsch 1972) due to inappropriate lubrication and cooling while drilling. Also, larger drill sizes may cause additional damage to bone's blood supply (Perren 2002), further encouraging necrosis. Taken together, these factors, could presumably lead to necrotic ring sequestra formation, pin loosening and subsequent pain experienced by the horse.

This study found that bone strain with the 2-pin construct with 4-cm separation was similar to the 2-pin construct with 5-cm separation. Although these results may not seem necessarily surprising, they are relevant. To explain, bone located at the 5-cm site has thicker cortical bone than bone located at the 4-cm site (as verified from radiographs). Presumably, pins at the 5-cm site would be more rigidly constrained and thus more effective at unloading bone. As such, we investigated whether a 2-pin construct containing a pin placed more proximal to the metaphysis would perform similarly as a 2-pin construct with only pins located in the metaphysis of the 3rd metacarpal bone. Our results suggest that both constructs perform similarly with regards to bone strain and load transfer. It is important to note though that prior research has indicated a higher risk of fracture when placing pins in the diaphysis (Joyce et al. 2006; Lescun et al. 2007). This is thought to be because the overall bone diameter is smaller (even though cortical bone is thicker) and placement of larger diameter pins ($\geq$ 10% of the diameter of bone) (Edgerton et al. 1990) result in stress concentrations. To minimize stress concentrations, prior research has recommended 2-cm or 3-cm spacing, with the distal pin placed near the metaphysis (Joyce et al. 2006). However, the higher stability associated with thick cortical bone may offset negatives associated with high stress concentrations and higher risk of fracture. Conversely, thin cortical bone found at the metaphysis may not offer sufficient fixation, leading to pin loosening and failure. Further testing between the different 2-pin constructs (2-cm and 3-cm vs 4-cm and 5-cm spacing) is needed with fatigue and torsional testing to identify the more effective construct.

In the past, two centrally-threaded positive-profile pins have been recommended for transfixation pin casts incorporating the third metacarpal/metatarsal bones in an attempt to provide a pin that is adequately strong without creating what could be a catastrophic defect in the cortical bone (Lescun et al. 2007). Interestingly, the mechanical testing and FE results of this study indicated high strain at the bone-pin interface with the 2-pin construct (when compared to the 4-pin construct). Importantly, this high strain may contribute to pin loosening complications associated with this treatment strategy. Accordingly, perhaps an alternative design is needed to avoid high strain at the bone-pin interface, such as with 3 positive-profile pins with a diameter between 4.8-mm and 6.3-mm. In line with this point, because the two most proximal pins carried the majority of load with the 4-pin construct, a 3-pin configuration should exhibit similar effectiveness in unloading the fracture site.

This study evaluated bone and cast strain at 2.5 kN and 7.5 kN load limits, which correspond with standing and walking, respectively (Turner et al. 1975). We had hoped to evaluate bone strain at higher loads, but strain readings at loads of 10 kN and above were lost either due to gauges loosening from the constructs or from reaching the maximum allowable readings. Interestingly however, both 2-pin and 4-pin constructs were able to support loads greater than 25 kN before catastrophic failure. This is reassuring given that the 3rd metacarpal bone endures a load of almost 21 kN at the time of recovery from anesthesia (Turner et al. 1975). This finding, as well as clinical experience, suggests all three constructs are capable of withstanding the load applied to the transfixation pin cast construct at the time of recovery as long as the attempts to rise are not numerous; however, a method of assisted recovery should be considered post-operatively.

While the different constructs were able to support high loads before failure, they were not necessarily effective at transferring load away from the fracture site. As indicated by the FE analysis and experimental testing, the pins of the 4-pin construct suffered permanent, plastic deformation around 7.5 kN. This meant that more load was transferred through the fracture site instead of through the pins and cast. In fact, just beyond 7.5 kN, both the 2-pin and 4-pin constructs endured excess pin deflection, pin deformation, or breakage of the cast, with more load being transferred through the cartilage, bone and the fracture site (Figure 2.5C, Figure 2.5D). These results indicate that additional research is needed, perhaps with 3 or 4 pins with diameters between 4.8-mm and 6.3-mm. Such research is important as appropriate transfixation pin cast constructs will endure multiple weeks of use, with fewer complications resulting in optimal comfort, functionality, and ultimately, fracture healing.

Limitations of this study relate to the number of specimens, strain gauge placement, strain gauge selection, mechanical testing procedures, and FE modeling assumptions. First, this study was limited to 5 samples per group. As such, statistically significant outcomes need to be evaluated with caution and future research is needed to confirm study findings. Second, in this study strain gauges were placed parallel and close to the osteotomy. This approach proved problematic due to how load was transferred across the fracture gap, which led to high variability in strain measures. In hindsight, a more suitable approach would have been to apply a strain gauge closer to the hoof, or even a load cell under the hoof to estimate load transfer with the different constructs. Similar to this point, it would have also been prudent to put gauges on the pins themselves to estimate load transfer as opposed to numerous bone and cast sites in-between pins. Continuing along these lines, we chose strain gauge placement above the pins for ease of comparison to similar existing studies available in the literature (Brianza et al. 2010). However, in hindsight, it would have been more appropriate to place gauges distal to the pins, as this would have provided a more accurate assessment of the compressive strain experienced in the cast due to load transfer from the bone and pins. Third, the strain gauges used in this study were limited to a maximum compressive strain of ± -5% (-50,000 microstrain), which limited our analysis to loads under 10 kN. Also, the gauges were unidirectional and only permitted measurements in a single direction, in this case along the compressive axis. Although we attempted to align the gauges with the long-axis of the bone, there was some degree of malalignment; thus, strain measures may not fully reflect maximum compressive strain. Future research in this area should employ multi-directional strain gauge rosettes to derive maximum/minimum principal strain measures and thereby account for strain gauge placement error. Fourth, we did not precondition the specimens prior to failure testing. We did this as we wished to assess the response of each cast to immediate loading. This omission likely explains some of our erratic strain data, as inclusion would have removed some of the highly compliant parts of the system prior to testing. Fifth, the FE model used in this study applied simplified geometry and material properties (e.g., ignored trabecular bone and modeled bone as an isotropic material, whereas it has been shown to be at least orthotropic in anisotropy (Ashman et al. 1989)). Although our model could be further advanced to account for such factors, it would be more advantageous to develop a subsequent equine-specific FE model, integrating equine geometry; orientations and contact locations; heterogeneous material properties as well as trabecular orientation and anisotropy. Of note, the fracture site was not modeled separately to avoid complications associated with the fracture mechanics (e.g., lateral movement of fractured bone, crack propagation, cohesive elements). Instead, a single material was used to model structures distal to the fetlock joint. Since this structure was the same for all FE models, and the FE-based fracture site strain was only used to compare different configurations in regards to unloading, we believe this modeling simplification is justified.

In conclusion, this study indicated that, in comparison with 2-pin transfixation pin cast constructs, the 4-pin construct transferred more load to the cast thereby protecting the fracture site. Despite this, at loads above 7.5 kN, the 4-pin construct may behave similar to 2-pin constructs due to plastic deformation of the pins. Results indicated that the 2-pin construct has high strain at both the bone-pin interface and the cast-pin interface, possibly explaining (at least to some degree) observed cases of pin loosening. To address limitations of both constructs, further research evaluating the use of 3 positive-profile pins that range in size between 4.8-mm and 6.3-mm should be evaluated as an alternative transfixation pin cast construct.

2.5 Footnotes

[1] Personal Communication with Dr. Larry Bramlage

[b] 0.9% NaCl

[c] CEA-06-125UW-120, Vishay Micro-Measurements, Raleigh, NC

[d] M-Coat A, Vishay Micro-Measurements, Raleigh, NC

[e] Delta-Lite Plus, BSN medical, Luxembourg

[f] Model RRH-10010, Enerpac, Milwaukee, WI

[g] LDI-119-200-A020A, Omega, Norwalk, CT

[h] Model 1220-AF, 250 kN capacity, Interface, Scottsdale, AZ

[i] SPSS 18.0; SPSS Inc, Chicago, IL

[j] ABAQUS, 3DS, Waltham, MA

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CHAPTER 3

EVALUATION OF TRANSFIXATION CAST CONSTRUCTS IN HORSE FORELIMBS: CYCLIC LOADING

CHAPTER 3

TRANSITION PAGE

Evaluation of transfixation cast constructs in horse forelimbs: cyclic loading

The previous study indicated that, in comparison with 2-pin transfixation pin cast constructs, the 4-pin construct transferred more load to the cast thereby protecting the fracture site. Despite this, at loads above 7.5 kN, the 4-pin construct may behave similar to 2-pin constructs due to plastic deformation of the pins.

This chapter utilized strain measurements to investigate the behavior of the previously described transfixation pin cast models after a cyclic load was applied. The load applied was similar to that which a horse would experience while standing in a stall for approximately six weeks, as apposed to a single load similar to that occurring during recovery from anesthesia.

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Author contribution: Thomas, Carmalt, and Burnett were responsible for the experimental design. Johnston was responsible for data analysis. Thomas and Burnett wrote the manuscript with Carmalt providing editorial assistance.

3.1 Introduction

Transfixation pin casts have been used for comminuted fractures of the distal limb for a number of years. They were first described as early as 1956 (Reichel) and have since evolved. Typically the transfixation pin cast consists of pins that pass through the bone proximal to a fracture. The pins are then incorporated within a cast that takes the majority of the weight, offloading the fracture. Although it is not uncommon to use them in conjunction with internal fixation, transfixation pin casts are generally used for comminuted fractures where it is difficult to utilize internal fixation for fracture repair. The main purpose of the transfixation pin cast is to minimize the load on bone distal to the pins, allowing the fracture pieces to heal without an excessive amount of strain. Traditional casts provide stability and decrease motion, but do not prevent excessive compression across a fracture gap. When comparing a traditional cast to three different transfixation pin casts, it was found that there was a significant difference in the ability of transfixation pin casts to decrease the amount of strain endured by the underlying bone when compared to the traditional cast (McClure et al. 1994b).

One complication associated with transfixation pin casting is pin loosening, secondary to infection or necrosis at the bone-pin interface (Mahan et al. 1991; McClure et al. 1995). There is generally a significant amount of pain associated with pin loosening and it is often necessary to replace the transfixation pin cast which may be difficult due to lack of space for additional pins to be placed. The weakest point of the transfixation pin cast is the bone-pin interface and deflection of the transfixation pins at this site is proportional to the cube of the distance between the bone and sidebars or cast (Nunamaker et al. 1986; Németh and Back 1991; McClure et al. 1994; Morisset et al. 2000; Oberg et al. 2012; Figure 1.3). It is thought that a greater distance between the bone and side bar, can result in excessive motion at the bone-pin interface creating pin loosening (Nunamaker et al. 1986). In an attempt to decrease the strain at the bone-pin interface a variety of factors have been investigated, such as the size of the pins, smooth vs. positive profile pins, self-tapping vs. non-tapping positive profile pins, hydroxyapatite coated positive profile pins, and diaphyseal vs. metaphyseal pin placement (Aron et al. 1985; McClure et al. 2000; Zacharias et al. 2007; Bubeck et al. 2010). The stiffness of the pin is proportional to the fourth power of its diameter and the larger the pin diameter, the lower the stresses that are generated in the pin as well as the bone-pin interface (Nunamaker et al. 1986; McClure et al. 1994b).

Currently, traditional transfixation pin casts consist of 2, 6.3-mm positive profile pins placed proximal to the location of the fracture. Initially pins were placed parallel to one another in the same plane. However pins placed in a 30-degree divergence to the frontal plane of the bone are reported to be a stronger construct, while preventing damage to the bone (McClure et al. 1994). There is concern that two larger diameter pins (6.3-mm) prevent healing of the fracture site and so it has been suggested that a transfixation pin cast construct that consisted of 4, 4.8-mm smooth pins placed 2-cm apart be utilized.[1]

A recent study determined that there was no significant difference in the amount of strain on the dorsal aspect of the proximal phalanx after load as applied to a transfixation pin cast with two 6.3-mm centrally-threaded, positive-profile pins or in a transfixation pin cast with four 4.8-mm smooth Steinman pins (Williams et al. 2014).

Thomas et al. (2018; Chapter 2) found that in comparison to transfixation pin cast constructs using two 6.3-mm centrally-threaded, positive-profile pins, a construct using four, 4.8-mm centrally-threaded, positive-profile pins transferred more load to the cast thereby protecting a distal fracture site. Despite this, at loads above 7.5 kN, the 4-pin construct may behave similar to 2-pin constructs due to plastic deformation of the pins. Additionally, the 2-pin construct has high strain at both the bone-pin interface and the cast-pin interface, possibly explaining (at least to some degree) observed cases of pin loosening.

The objective of this study was to describe the behavior of three different transfixation pin cast constructs in cyclic axial loading simulating a 6-week duration, similar to that which would be expected *in vivo*.

3.2 Materials and Methods

3.2.1 Mechanical Testing

3.2.1.1 Specimens

Fifteen (n=15) forelimbs from 15 horses were collected from adult horses euthanized for reasons unrelated to the musculoskeletal system. All limbs were disarticulated at the carpometacarpal joint and frozen at -20 $^{\circ}$C wrapped in saline soaked towels. Limbs were thawed at room temperature prior to testing; after which they were prepared for mechanical testing and randomly assigned to one of three different construct groups (with 5 limbs per group), as per Chapter 2.

3.2.1.2 Constructs

Constructs were prepared as per Thomas et al. (2018, Chapter 2):

Construct 1: two 6.3-mm centrally threaded positive-profile pins spaced 4-cm apart, 30 degree divergence.

Construct 2: two 6.3-mm centrally threaded positive-profile pins spaced 5-cm apart, 30 degree divergence.

Construct 3: four 4.8-mm centrally threaded positive-profile pins spaced 2-cm apart, 10 degree divergence.

An oblique osteotomy was created in the proximal phalanx in all specimens to simulate a fracture (Figure 2.1).

3.2.1.3 Strain Gauge Instrumentation

A uniaxial strain gauge[b] was applied to the bone proximal to each pin (Chapter 2) in a proximal-to-distal direction, except adjacent to the osteotomy where the gauges were applied parallel to the oblique osteotomy (Figure 2.1).

3.2.1.4 Cast Application

Casts[c] were constructed in a standard fashion, incorporating the foot and extending proximal to 2-cm below the carpometacarpal joint. The pins were incorporated into the cast by making incision through the fiberglass cast material. Uniaxial strain gauges[b] were attached to the cast in a proximal-to-distal direction, following manufacturer's recommendations for porous materials. Leads were secured to the cast with electrical tape and gathered at the proximal extent of the cast.

3.2.1.5 Biomechanical Testing

Limbs were axially loaded using a hydraulic actuator[d] instrumented with a linear variable differential transformer (LVDT)[e] and load cell.[f] Each specimen was placed under the actuator and secured in place with the use of a custom frame (Figure 2.3), to prevent forward sliding under axial compressive loading. The load cell was fitted with custom adaptor that connected with the inner cortex (i.e, marrow cavity) of the cannon bone (proximal 3rd metacarpal).

Limbs underwent an axial load that ranged between 5.5kN and 7.5kN at a cyclic rate of 2 Hz. Axial load was applied until 192,000 cycles were performed which took approximately 23 hours, representing 6 weeks of cast wear. Cyclic axial loading was then discontinued, the limb removed from the load cylinder, and returned to a freezer to be dissected at a later date.

3.3 Results

Investigation of data comparing the 4-pin construct to the 2-pin constructs reveal similar results between the two construct types in that the proximal pin(s) takes the majority of the load in each construct (Figure 3.1). When evaluating the load distribution amongst the pins in the 4-pin construct, there is dramatic increase in strain associated with the proximal pin around 90,000 cycles. These anomalies were observed solely in specimen number four, and when this is excluded, a clearer picture of strain over time is seen (Figure 3.2).

Figure 3.1: The mean microstrain associated with each pin demonstrating that the proximal pin carries the majority of the load in each construct. Note the abrupt increase in microstrain at approximately 90,000 cycles in the first pin of the 4-Pin construct (4-Pin 1st Pin).

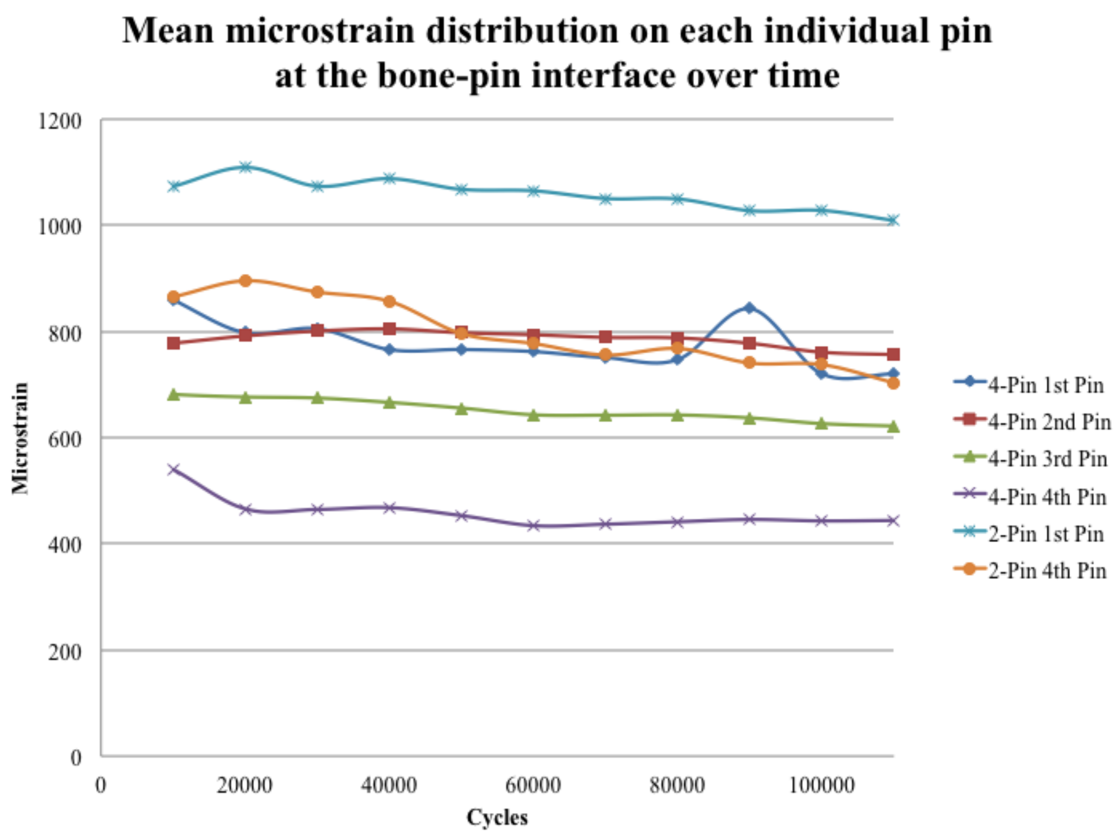
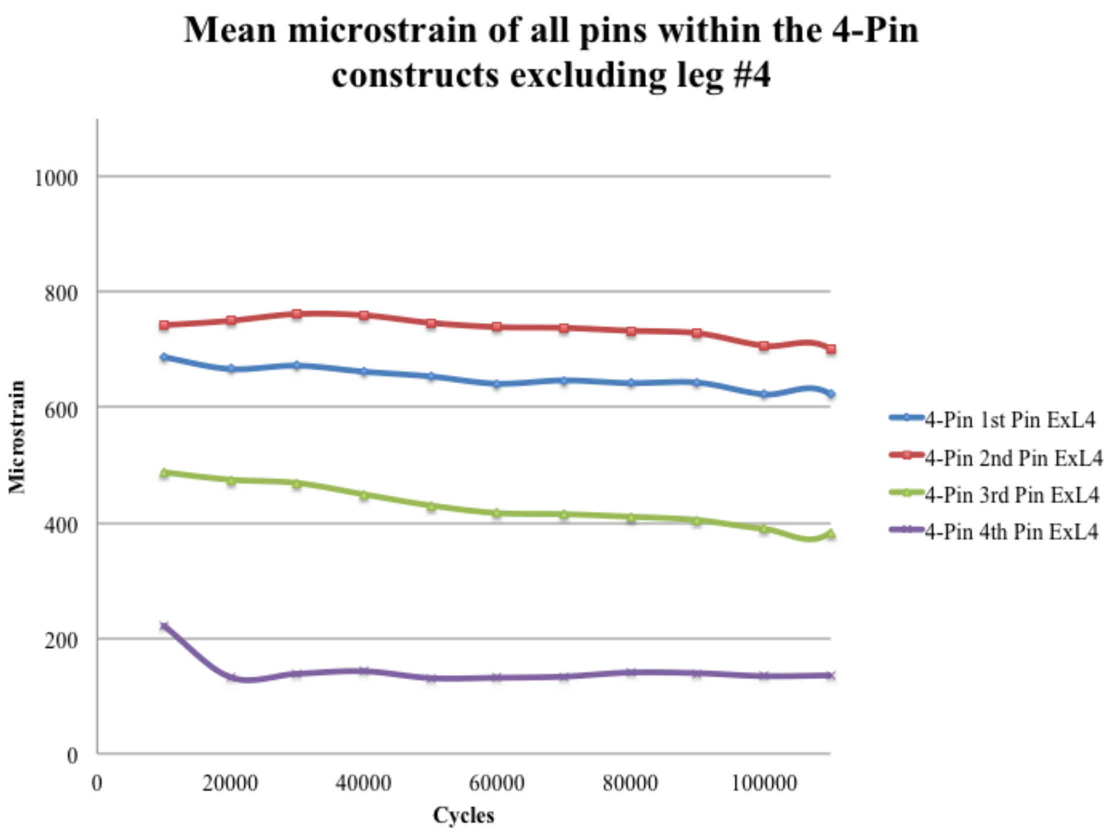


Figure 3.2: The mean microstrain associated with each pin within the 4-Pin construct after eliminating the fourth specimen.



There was only data recorded for 120,000 cycles from the fifth specimen as apposed to 190,000 with all of the rest of the specimens. As such Figures 3.1 and 3.2 reflect data to 120,000 not 190,000 cycles. Load distribution curves (Appendix A) reveal that there was significant individual variation between specimens.

During cyclic loading, there is more load being distributed through the pins to the cast in the 4-pin construct; a finding observed in the previous, static, experiment (Chapter 2). At 100,000 cycles, the compressive strain with the 4-pin construct (at the distal cast site) is around 850 microstrain, while in the 2-pin construct it is approximately 200 microstrain (Figure 3.3). This indicates that the cast of the 4-pin construct is carrying more load than the cast of the 2-pin construct.

As load was applied to the TPC constructs, displacement of the transcortical pins occurred. In the 2-pin construct there was an initial spike in displacement, followed by a steady decrease. In contrast, the 4-pin construct had a gradual rise in displacement that continued to occur over time under the same load conditions. Evaluation of the specimens post-testing revealed that there was evidence of cast damage associated with the cast-pin interface in the 2-pin constructs, while the transcortical pins of the 4-pin construct were bent (Figure 3.4).

Figure 3.3: Microstrain distribution at the cast-pin interface, showing that there is more microstrain, and subsequent load, being distributed to the cast of the 4-pin construct.

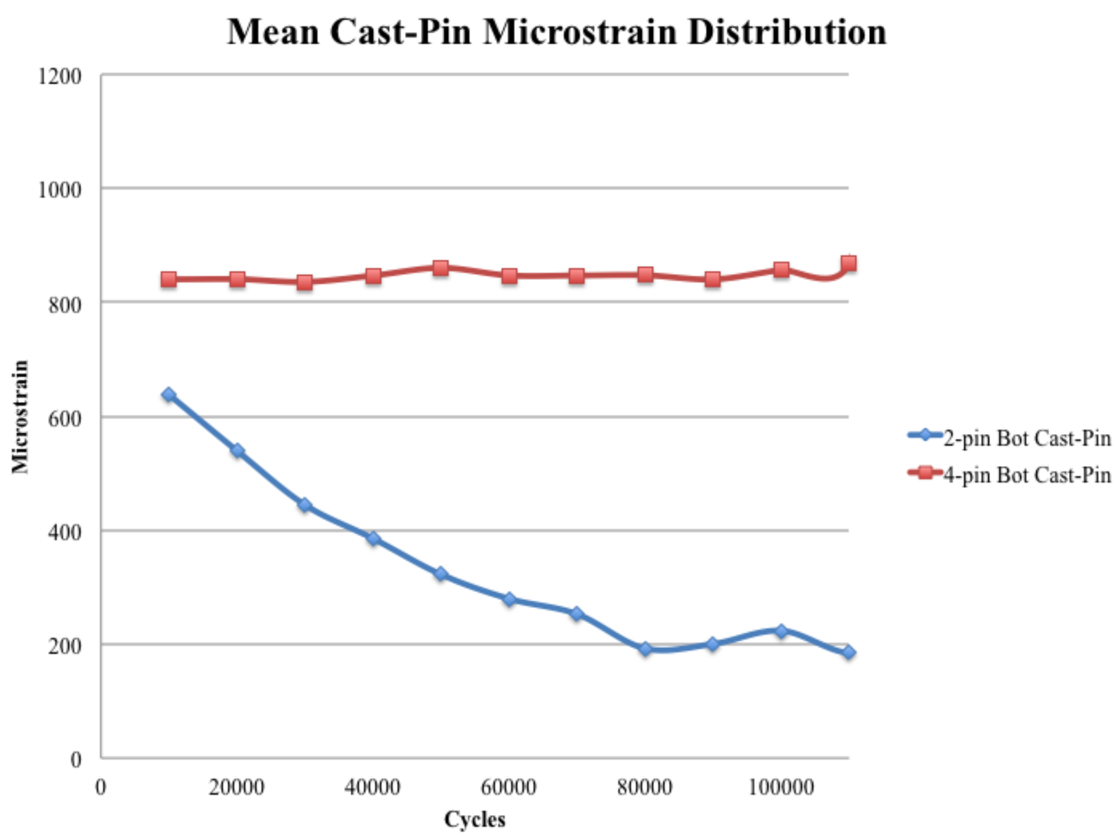
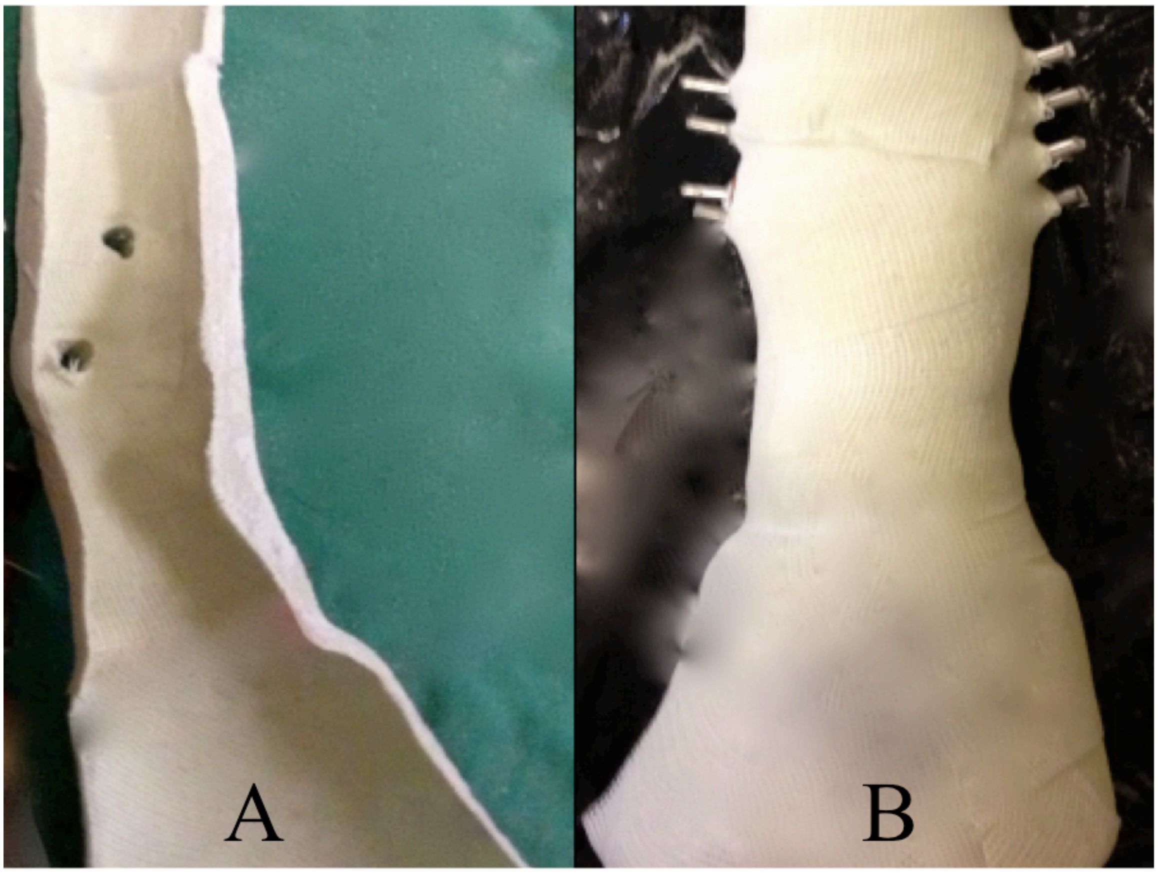


Figure 3.4: Example images of the defects that occurred secondary to a cyclic load of 5.5 kN – 7.5kN. Damage at the cast-pin interface associated with the 2-pin construct (A) and bent pins of the 4-pin construct (B).



3.4 Discussion

Initially it had been our intention to determine how a 2-pin construct with a greater distance between the pins would compare to a 2-pin construct with a lesser distance between the two, based on results of earlier work looking at pin location in regards to complications and longevity of transfixation pin constructs. Pin placement in the mid-diaphysis was thought to result in decreased breaking strength and catastrophic failure subsequent to increased stress-concentration of the bone-pin interface due to inappropriate pin diameter in comparison to bone circumference. However McClure et al. (2000) found that 6.3-mm positive profile pins placed at this location had greater resistance to axial extraction than similar pins placed in the metaphysis. Given this information, our intention was to determine if a 2-pin construct containing a pin placed more proximal to the metaphysis, located in the distal diaphysis, would perform similarly to a 2-pin construct with pins located in the metaphysis of the 3rd metacarpal bone. The aim had been to demonstrate that a pin could be placed in a more proximal location where it may hold up to continuous load application for a longer period of time without ending in catastrophic failure. Our results indicate that it may be appropriate, although not necessarily superior, to place pins slightly more proximal within the diaphysis of the 3rd metacarpal bone. In addition, we chose to report information as it relates to a 2-pin construct vs. a 4-pin construct. Further testing between the different 2-pin constructs, such as torsional testing, would need to be conducted to determine which was superior.

As found in the previous experiment (Chapter 2), the most proximal pin in each construct carries the majority of the load. However, the 4-pin construct transfers more of the load to the cast than the 2-pin construct does. This may indicate that the 2-pin construct is too stiff and doesn't transfer as much load to the cast and subsequently has a higher amount of strain at the bone-pin interface.

The amount of displacement that occurred secondary to load application differed between constructs. In the 2-pin construct there was a spike in the amount of displacement that occurred initially and based on assessment of the constructs after testing, we believe that this is a representation of the 6.3-mm pins cutting through the cast at the cast-pin interface. Once this occurred, the amount of displacement was negligible. In comparison, the 4-pin construct had continued displacement over time. This was probably related to the 4.8-mm pins bending in response to the cyclic load. Because the pins bend secondary to the load applied over time, this may result in an increased amount of strain at the fracture site. Unfortunately, due to the way that the strain gauges were applied at the fracture site, there was no way to measure this.

While the 6.3-mm pins did not deform, the 6 week cyclic load resulted in permanent deformation of the 4.8-mm pins in the 4-pin construct. This was most likely due to the yield stress of the 4-pin construct being above the yield strength of the 4.8-mm pins. The smaller pins had been used to increase the total stiffness of the construct, but decrease the risk of fracture through a single pin tract as a result of load sharing. However, our results show the 4.8-mm pins are not strong enough to withstand the cyclic loading. It is possible that three positive profile pins with a diameter between 4.8-mm and 6.3-mm may be preferable for 3^{rd} metacarpal bone transfixation pin casting. This may allow for increased stiffness without increasing the cortical bone defect to a level that would result in an increased stress concentrator at the bone-pin interface.

A limitation of this study was the number of specimens that were used to determine the results of the study, strain gauge placement, and strain gauge selection. As previously stated, individual load distribution curves revealed large differences between specimens over time. More specimens may have improved our findings. Additionally, given the large number of strain gauges, it wasn't uncommon for strain gauge readings to become unreliable, or for them to not produce any useful data. The strain gauges were placed parallel, and close, to the osteotomy. This proved problematic due to how load was transferred across the fracture gap, leading to variability in strain measurements, and making conclusions regarding the strain across the fracture site difficult to determine.

In conclusion, we found that the 4-pin construct transfers more load to the cast, distributing the strain through the bone more evenly, but permanently deforms when cyclically loaded to mimic a horse walking for a 6 week period. The 2-pin construct has high stress at both the bone-pin interface and the cast-pin interface, with both the 4-pin and 2-pin constructs carrying the majority of the load in the most proximal pin. To address the faults of both constructs, further research evaluating the use of positive profile pins that ranging in size between 4.8-mm and 6.3-mm, the use of 3 positive profile pins, or additional layers of cast material should be evaluated to develop a superior transfixation pin cast construct. Additional studies may help determine the most appropriate transfixation pin cast construct; one that is capable of enduring up to 6 weeks of use with fewer complications allowing for optimal comfort, functionality, and fracture healing.

3.5 Footnotes

[1] Personal Communication with Dr. Larry Bramlage

[b] CEA-06-125UW-120, Vishay Micro-Measurements, Raleigh, NC

[c] Delta-Lite Plus, BSN medical, Luxembourg

[d] Model RRH-10010, Enerpac, Milwaukee, WI

[e] LDI-119-200-A020A, Omega, Norwalk, CT

[f] Model 1220-AF, 250 kN capacity, Interface, Scottsdale, AZ

[g] SPSS 18.0; SPSS Inc, Chicago, IL

3.6 References

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CHAPTER 4

GENERAL DISCUSSION

4.1 Introduction

Transfixation pin casts have been shown to be a viable option, alone or in combination, with internal fixation in the treatment of equine fractures (Nunamaker et al. 1986; Németh et al. 1991). As with many forms of external coaptation, there can be significant complications associated with the application and wearing of these types of casts (Lescun et al. 2007). Examples include pin loosening, pin breakage, ring sequestra at the bone-pin interface, fracture through the pin tract, cast fracture, and simple cast sores related to an improperly fitting cast. A number of factors must be considered in transfixation pin selection. Not only are specific characteristics of the pin important, but also size, and the location of where the pin is to be placed must be taken into account. Qualities that should be selected for provide strength and durability without causing harm to the patient.

4.2 General results and future studies

The goal of this research was to determine the optimum number and placement of transcortical pins in an equine transfixation cast construct.

The first specific aim, to determine the load to failure of three different transfixation cast constructs tested in single cycle to failure was determined by recording strain at the bone-pin interface, the cast-pin interface, and at a distal fracture site while an axial load of up to 25kN was applied (Chapter 2). Data showed that there was no significant difference between the two 2-pin constructs and that at loads greater than 7.5kN all three constructs behaved similarly due to plastic deformation of the pins. At loads under 7.5kN, the 4-pin construct was found to transfer more strain to the cast, subsequently protecting the fracture, while the 2-pin constructs had high strain at both the bone-pin interface and the cast-pin interface, possibly explaining (at least to some degree) observed cases of pin loosening.

The second specific aim of this thesis was to describe cyclic load parameters by loading the transfixation pin constructs with an axial load that ranged between 5.5kN and 7.5kN at a cyclic rate of 2 Hz. Axial load was applied until 192,000 cycles were performed. This represented 6 weeks of cast wear which is an expected period of transfixation cast use in equine fracture management (Chapter 3). Although the amount of data that was useful was limited, we believe that the 4-pin constructs transfer more load to the cast protecting the fracture site, but permanently deform at a load similar to that of walking over this period. The 2-pin construct has high stress at both the bone-pin interface, and the cast-pin interface, with the proximal pin carrying the majority of the load in both the 4-pin and 2-pin constructs. Additional testing with an increased number of specimens would need to be performed in order to prove if our assumptions are correct.

Strain gauge placement and strain gauge selection were factors in this limitation. Given the large number of strain gauges, it wasn't uncommon for strain gauge readings to become unreliable or for them to not produce any useful data. As the strain gauges were placed parallel and close to the osteotomy, it was difficult to determine how strain was transferred across the fracture gap. These discrepancies lead to variability in strain measurements and data interpretation.

Future studies should be conducted using a suitable strain gauge placed in a more appropriate location for accurate data collection. Additionally, it would be beneficial to test a construct that includes positive profile pins that are sized between 4.8-mm and 6.3-mm. Unfortunately, pins in this size range are not commercially available for use at the current time. An alternative method of testing would be finite element analysis. Utilizing finite element analysis, additional construct designs incorporating pins within the suggested size range, as well as a variable number of pins, could be tested. If the results were promising, perhaps an alternative pin size could be manufactured and used for additional *ex vivo* testing with the intention of testing the construct *in vivo*.

4.3 Conclusion

In conclusion, we have demonstrated that improvements can be made to the commonly used transfixation pin cast constructs in order to improve their use in fracture management, as well as to limit complications associated with their use. Further research is required to evaluate the use of positive profile pins which range in size between 4.8-mm and 6.3-mm, the use of 3 positive profile pins, or additional layers of cast material to develop a superior transfixation pin cast construct.

4.4 References

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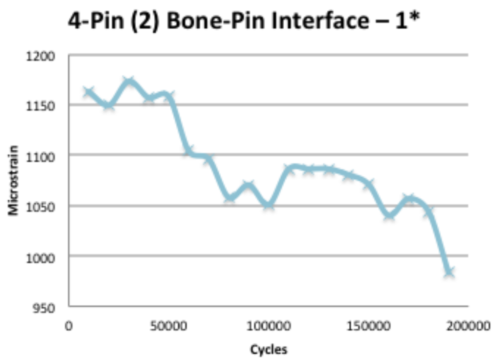
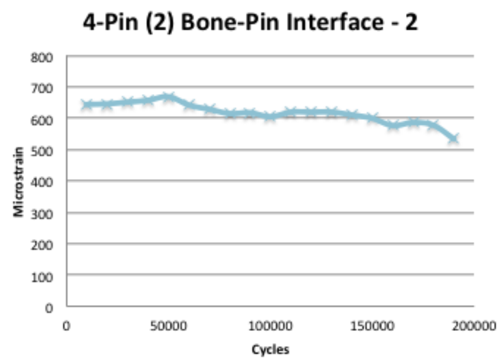
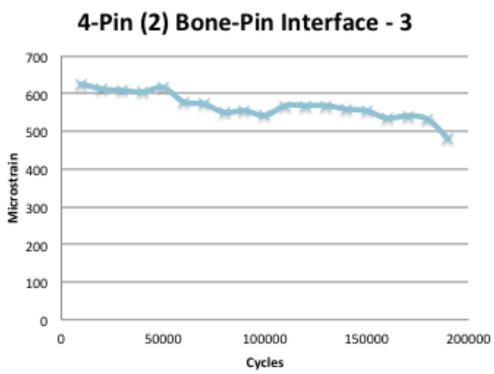
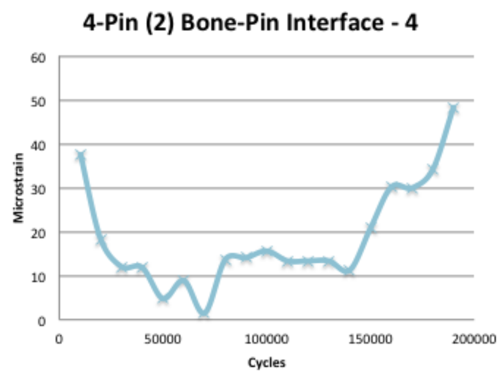
Nunamaker, DM; Richardson, DW; Butterweck, DM; et al (1986). A new external skeletal fixation device* that allows immediate full weightbearing application in the horse. Vet Surg.15:345-355

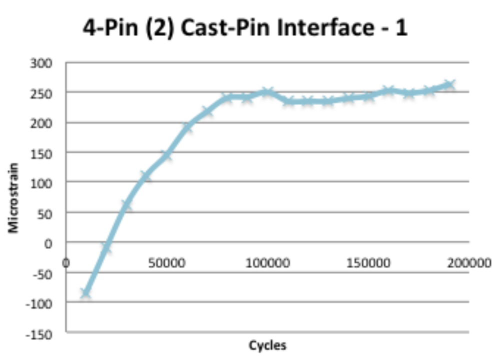
APPENDIX A

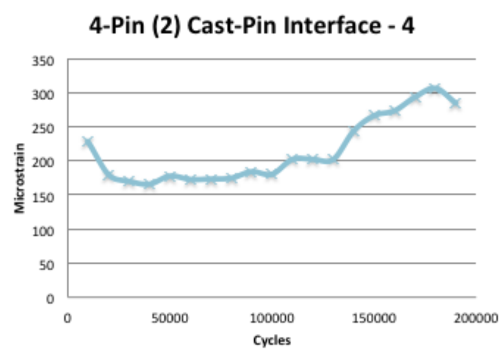


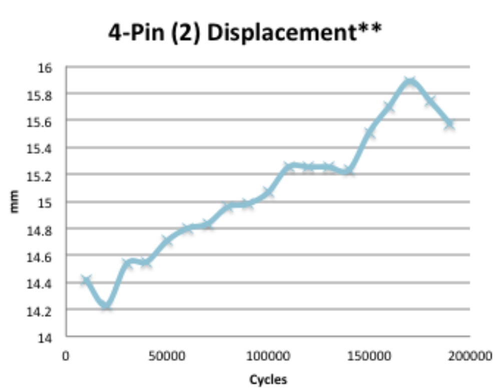




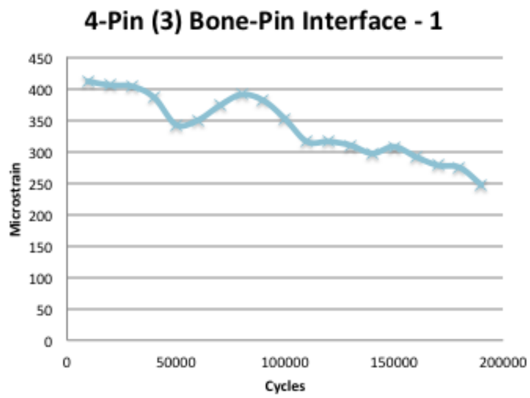








*Legend = 4-Pin (2) Bone-Pin Interface–1 = 4-Pin construct (specimen 2) Bone-Pin Interface One
**Legend = Displacement of entire construct.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Microstrain
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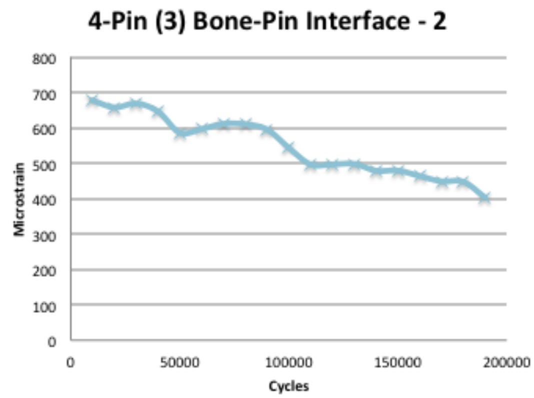
400
350
300
250
200
150
100
50
0
0
50000
100000
150000
200000
Cycles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Microstrain
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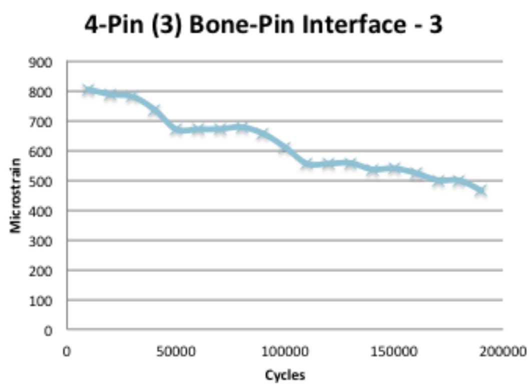
700
600
500
400
300
200
100
0
0
50000
100000
150000
200000
Cycles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Microstrain
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800
700
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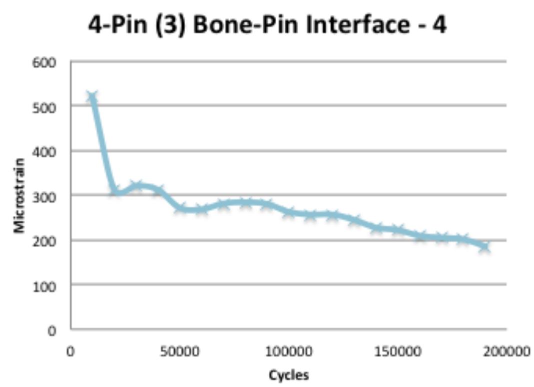
400
300
200
100
0
0
50000
100000
150000
200000
Cycles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Microstrain
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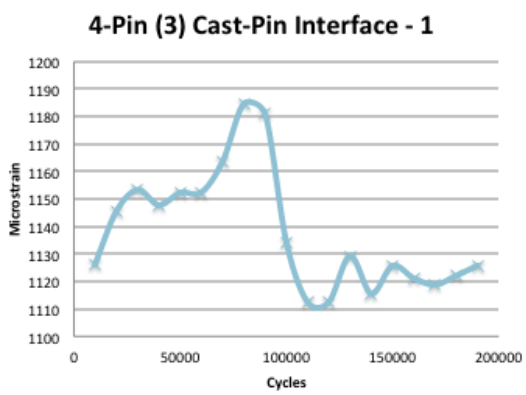
400
300
200
100
0
0
50000
100000
150000
200000
Cycles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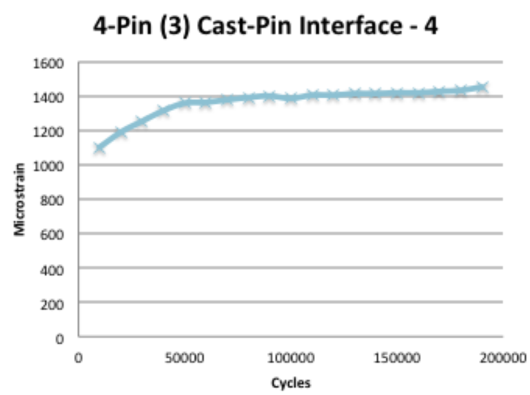
Microstrain
1200
1190
1180
1170
1160
1150
1140
1130
1120
1110
1100
0
50000
100000
150000
200000
Cycles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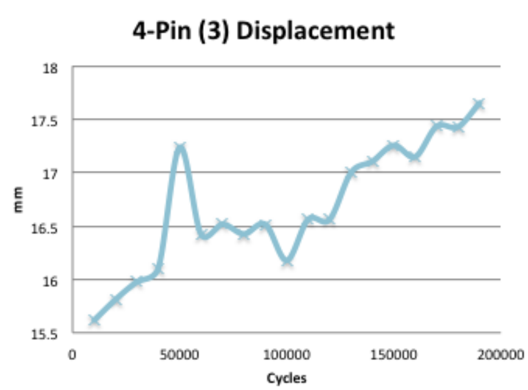
Microstrain
1600
1400
1200
1000
800
600
400
200
0
0
50000
100000
150000
200000
Cycles


4-Pin (3) Displacement
mm
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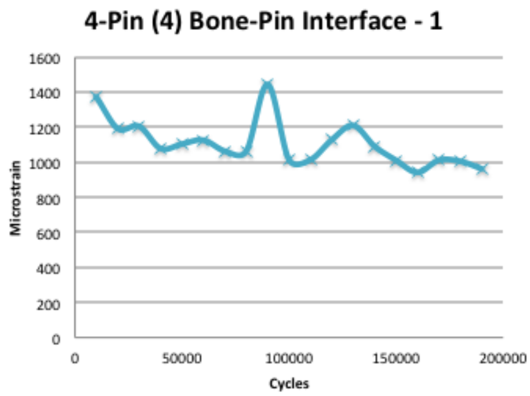
17.5
17
16.5
16
15.5
0
50000
100000
150000
200000
Cycles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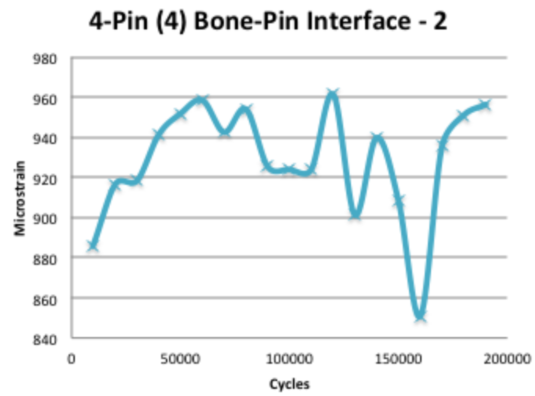
Microstrain
Cycles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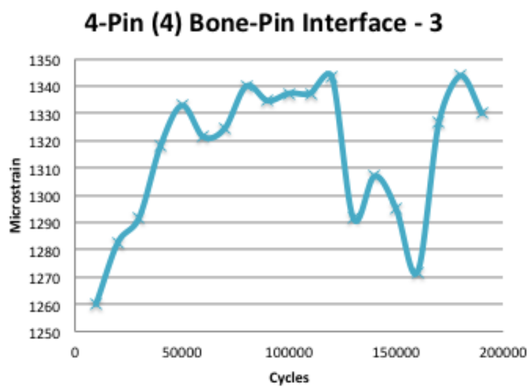
Microstrain
Cycles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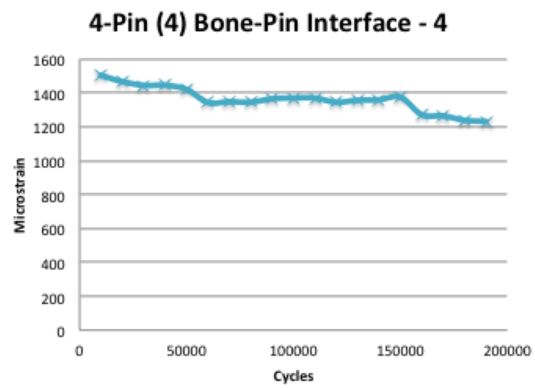
Microstrain
Cycles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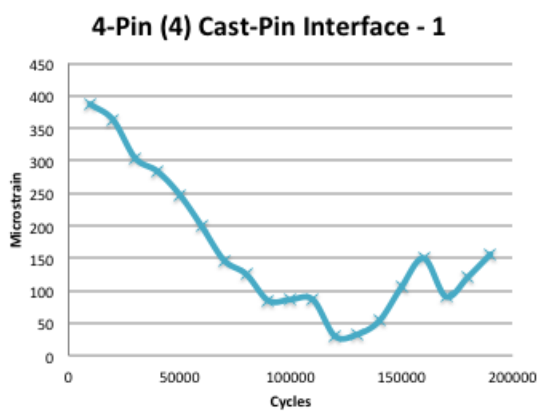
Microstrain
Cycles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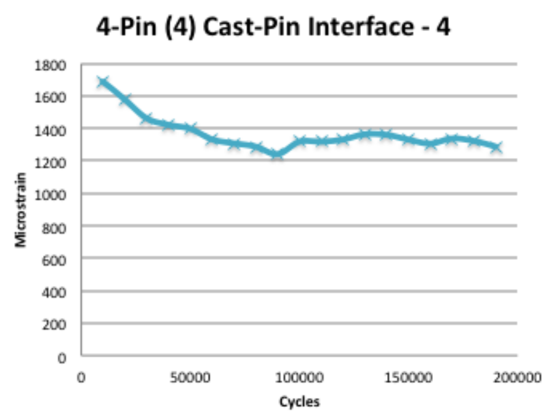
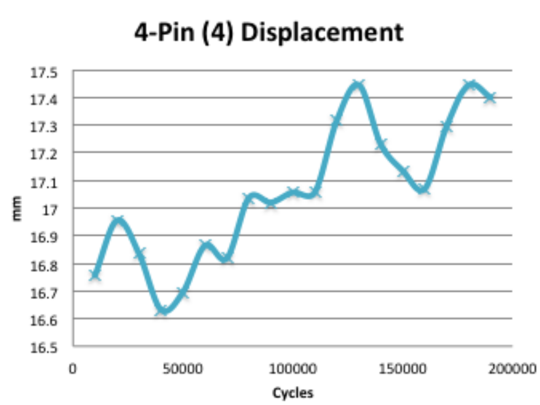
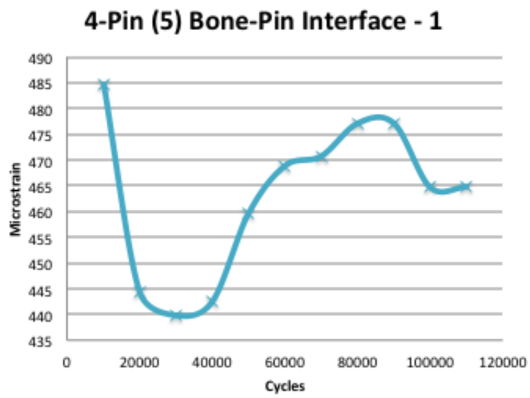
Microstrain
Cycles


4-Pin (4) Cast-Pin Interface - 4
Microstrain
Cycles


4-Pin (4) Displacement
mm
Cycles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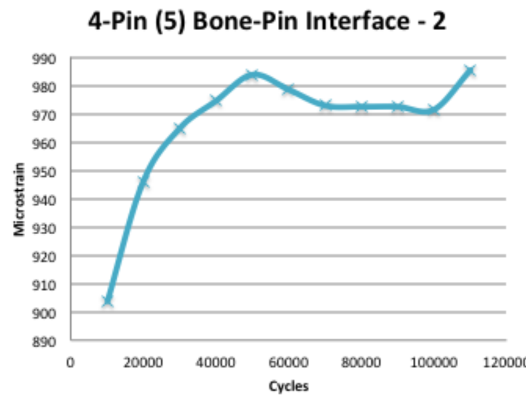
Microstrain
Cycles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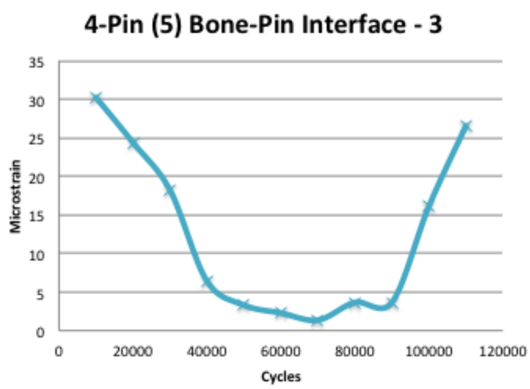
Microstrain
Cycles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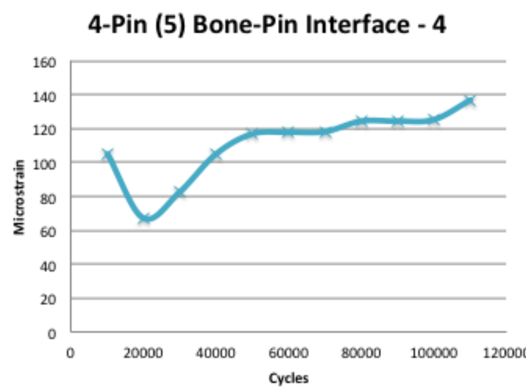
Microstrain
Cycles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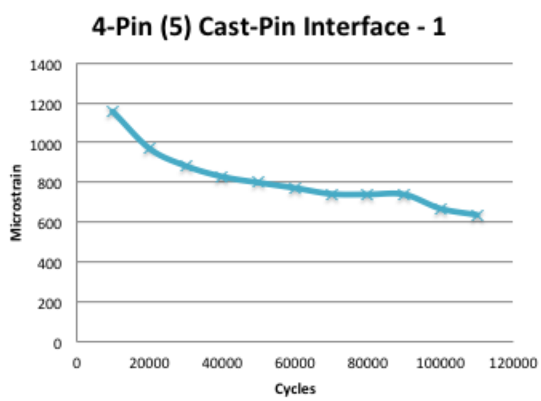
Microstrain
Cycles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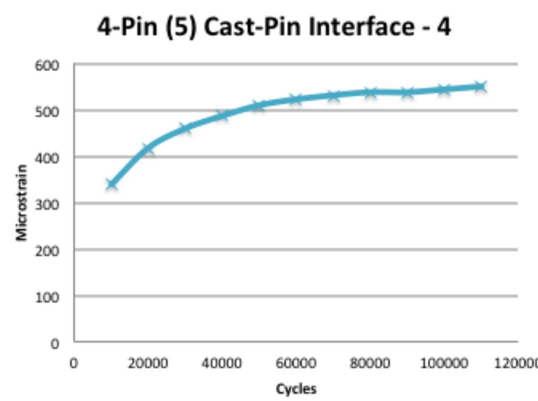
Microstrain
Cycles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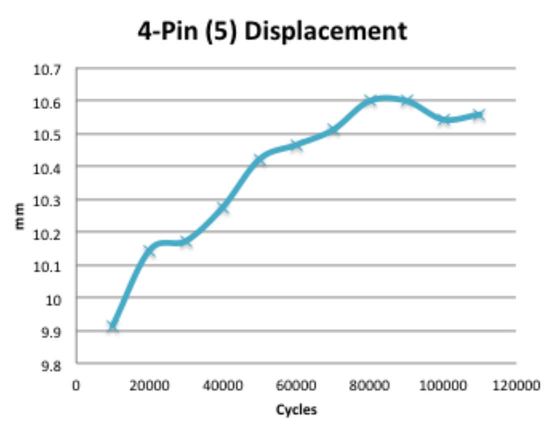
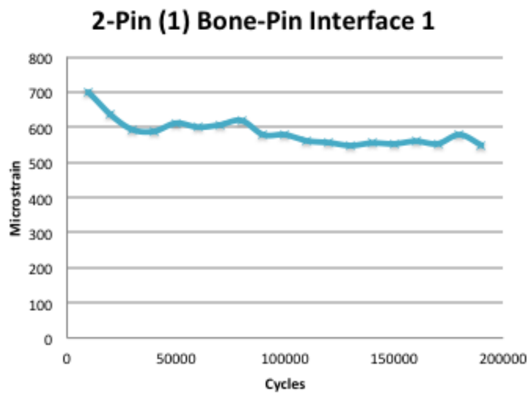
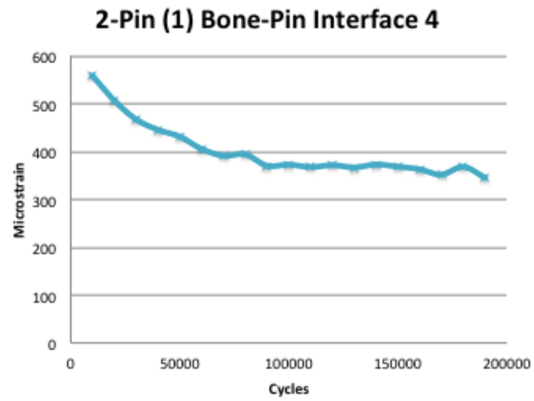
Microstrain
Cycles


4-Pin (5) Displacement
mm
Cycles


2-Pin (1) Bone-Pin Interface 1
Microstrain
Cycles
0
100
200
300
400
500
600
700
800
50000
100000
150000
200000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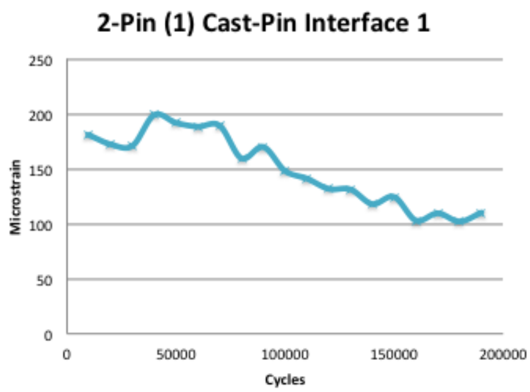
Microstrain
Cycles
0
100
200
300
400
500
600
50000
100000
150000
200000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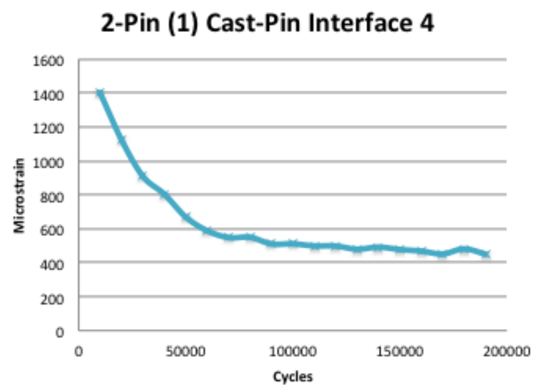
Microstrain
Cycles
0
50
100
150
200
250
50000
100000
150000
200000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Microstrain
Cycles
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200
400
600
800
1000
1200
1400
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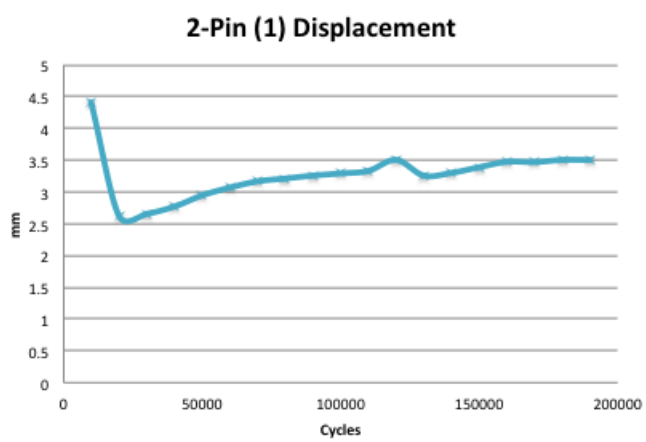
100000
150000
200000


2-Pin (1) Displacement
mm
Cycles
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1.5
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50000
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150000
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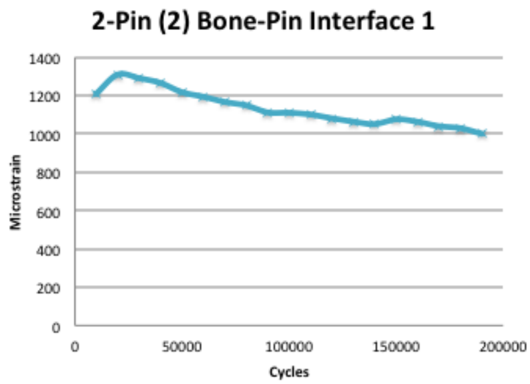
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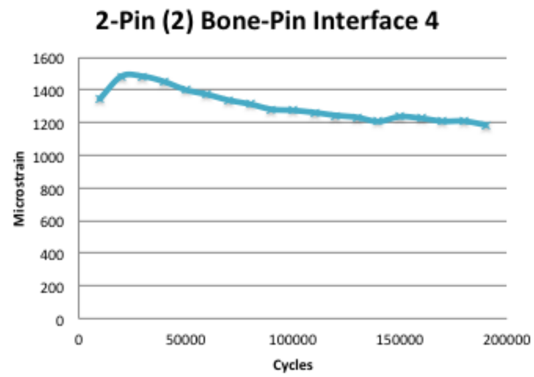
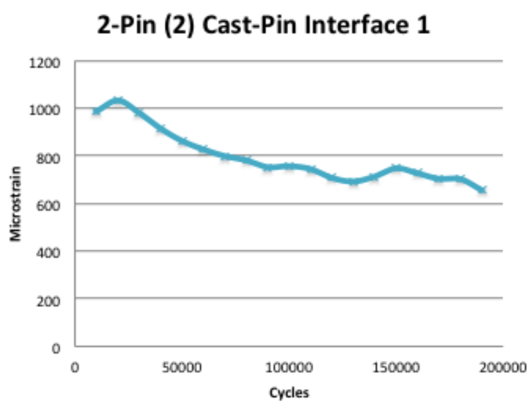
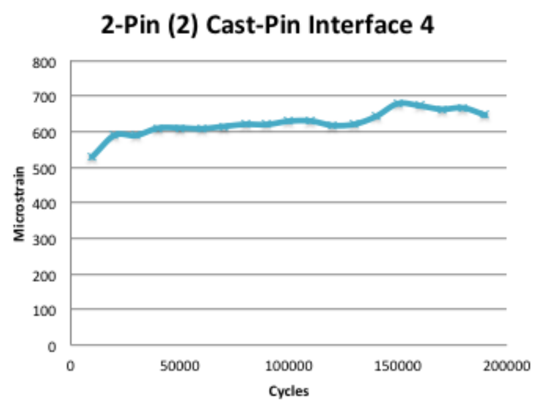
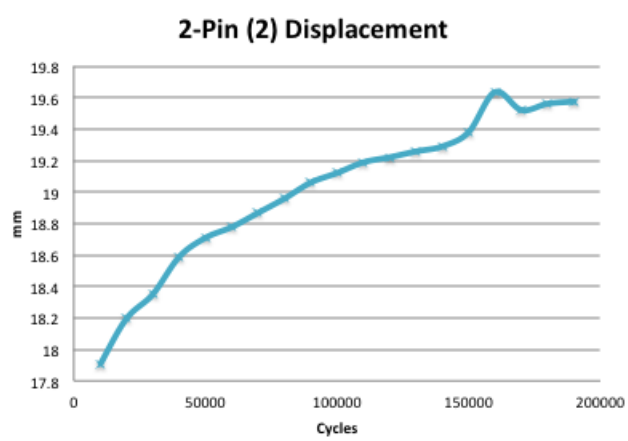
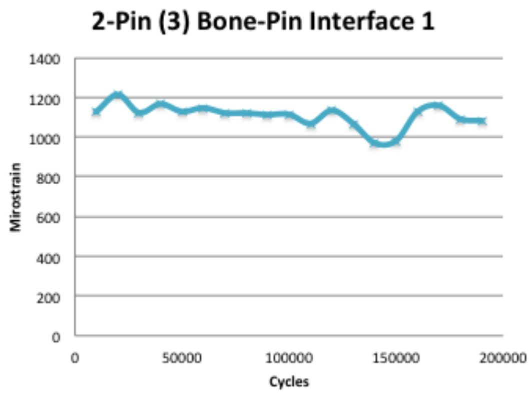
Microstrain
Cycles

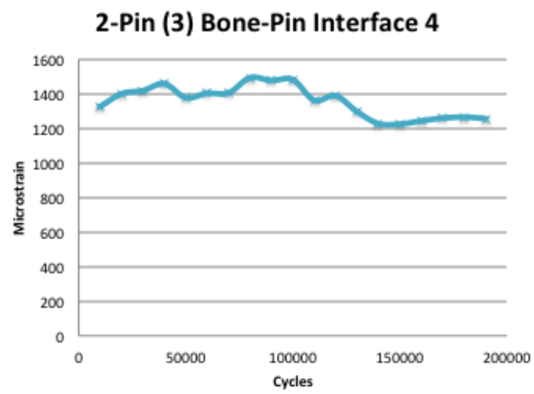

2-Pin (2) Bone-Pin Interface 4
Microstrain
Cycles

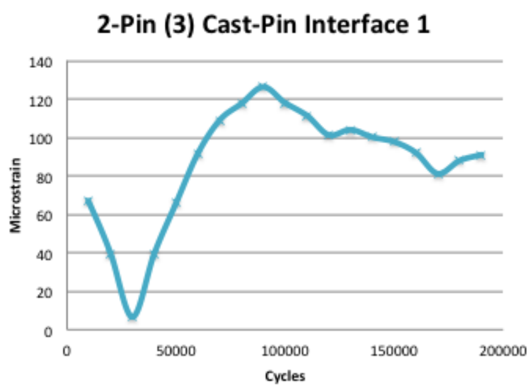

2-Pin (2) Cast-Pin Interface 1
Microstrain
Cycles

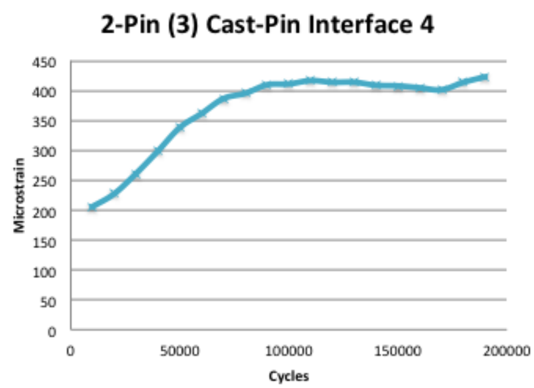

2-Pin (2) Cast-Pin Interface 4
Microstrain
Cycles

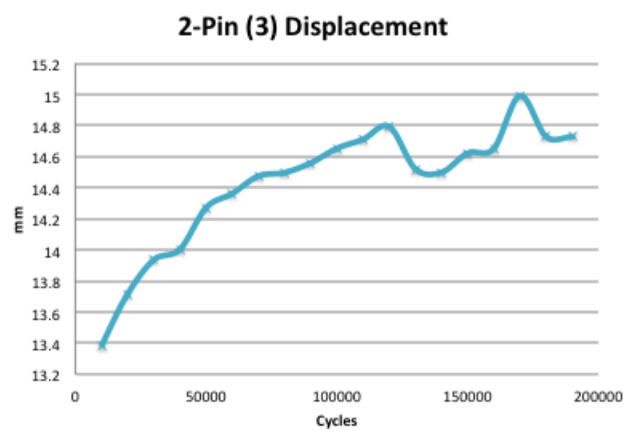

2-Pin (2) Displacement
mm
Cycles

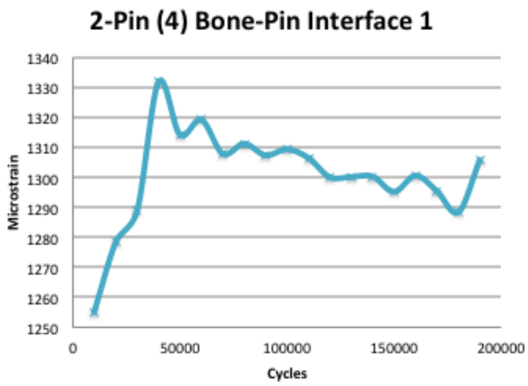

2-Pin (3) Bone-Pin Interface 1
Microstrain
Cycles
0
200
400
600
800
1000
1200
1400
0
50000
100000
150000
200000

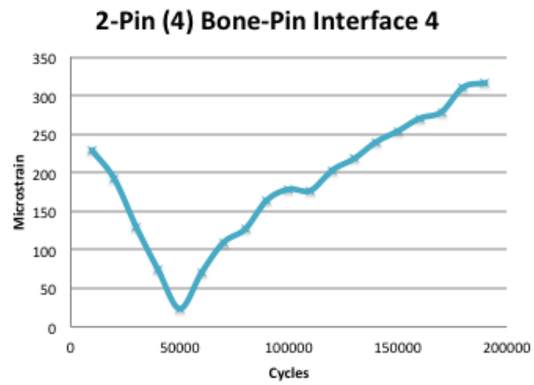

2-Pin (3) Bone-Pin Interface 4
Microstrain
Cycles
0
200
400
600
800
1000
1200
1400
1600
0
50000
100000
150000
200000

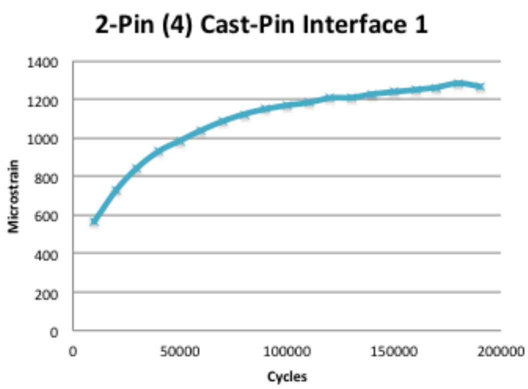

2-Pin (3) Cast-Pin Interface 1
Microstrain
Cycles
0
20
40
60
80
100
120
140
0
50000
100000
150000
200000

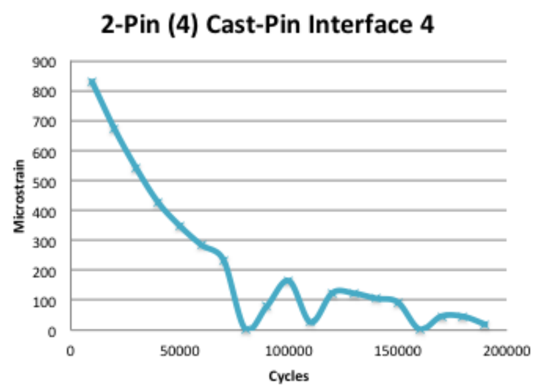

2-Pin (3) Cast-Pin Interface 4
Microstrain
Cycles
0
50
100
150
200
250
300
350
400
450
0
50000
100000
150000
200000

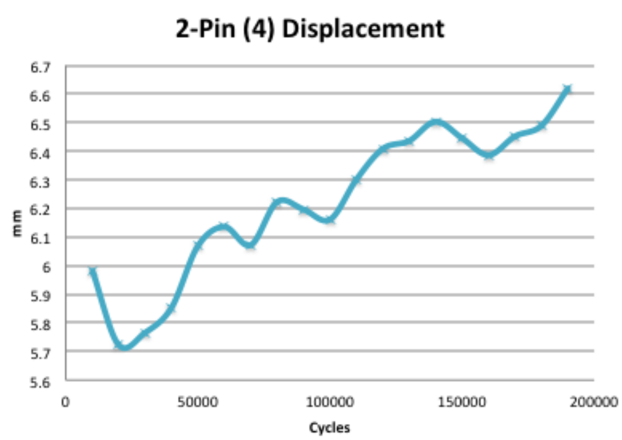

2-Pin (3) Displacement
mm
Cycles
13.2
13.4
13.6
13.8
14
14.2
14.4
14.6
14.8
15
15.2
0
50000
100000
150000
200000


2-Pin (4) Bone-Pin Interface 1
Microstrain
1340
1330
1320
1310
1300
1290
1280
1270
1260
1250
0
50000
100000
150000
200000
Cycles


2-Pin (4) Bone-Pin Interface 4
Microstrain
350
300
250
200
150
100
50
0
0
50000
100000
150000
200000
Cycles


2-Pin (4) Cast-Pin Interface 1
Microstrain
1400
1200
1000
800
600
400
200
0
0
50000
100000
150000
200000
Cycles


2-Pin (4) Cast-Pin Interface 4
Microstrain
900
800
700
600
500
400
300
200
100
0
0
50000
100000
150000
200000
Cycles


2-Pin (4) Displacement
mm
6.7
6.6
6.5
6.4
6.3
6.2
6.1
6
5.9
5.8
5.7
5.6
0
50000
100000
150000
200000
Cycles