BIOMECHANICAL EVALUATION OF INCOMPLETE RINGS AND HYBRID FIXATOR CONSTRUCTS

By

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To my love, Maria

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Circular external skeletal fixation and linear-circular hybrid external skeletal fixation have become well established therapeutic modalities for correction of various orthopedic diseases in dogs and cats. Incomplete rings are often incorporated in fixator constructs when juxta-articular placement of a ring has the potential to interfere with joint range of motion. To date, no biomechanical studies have been performed which specifically evaluate the mechanical properties of incomplete rings or of hybrid fixator constructs. The purpose of this study was to quantify the mechanical properties of single ring constructs and linear-circular hybrid external skeletal fixator constructs, both utilizing incomplete rings.

Single ring constructs were constructed utilizing complete and incomplete rings. Construct variables included ring diameter and fixation wire crossing angle. Fixation wires were sequentially tensioned and the change in internal ring diameter due to wire tensioning was measured to quantify ring deformation. No complete rings were deformed by wire tensioning. Incomplete rings were elastically deformed by application of 30 kg or 60 kg of wire tension, while application of 90 kg wire tension resulted in permanent deformation or incomplete ring failure.

Single ring constructs were assembled using complete and incomplete rings. Construct variables included ring diameter and wire tension. Constructs were axially loaded to 375 N for 15 cycles. Load-displacement curves were obtained and used to calculate construct stiffness. Ring deformation as a result of wire tensioning and construct loading was sequentially quantified. Load-displacement curves from incomplete ring constructs had a more gradual increase in slope than comparable complete rings. Incomplete ring constructs were less stiff than comparable complete ring constructs. Fixation wire tensioning increased construct stiffness in incomplete ring constructs and decreased axial displacement in both complete and incomplete ring constructs. Complete rings experienced minimal deformation while incomplete rings experienced significant deformation as a result of both wire tensioning and construct loading.

Three designs of hybrid constructs were assembled utilizing a single incomplete ring with two tensioned olive wires, one or two hybrid rods, and three half pin fixation pins. Construct variables included number of hybrid rods and uniplanar vs biplanar fixation pin insertion. Constructs were tested in axial compression, craniocaudal and mediolateral four point bending, and torsion. Load-displacement curves were linear in axial compression and torsion while some curves obtained in bending were multiphasic linear. Addition of a secondary hybrid rod increased construct stiffness in axial compression. Incorporation of a secondary hybrid rod and biplanar fixation pin insertion increased construct stiffness in all modes of loading.

Our study demonstrates that single incomplete ring constructs deform as a result of wire tensioning and axial loading, and are less stiff in axial loading than comparable diameter complete ring constructs. The stiffness of hybrid fixator constructs can be increased by the addition of a secondary hybrid rod and biplanar fixation pin insertion. Future studies are warranted to quantify the effects of additional modifications to hybrid fixator design.

CHAPTER 1
AN INTRODUCTION TO LINEAR-CIRCULAR HYBRID FIXATION IN DOGS AND CATS

Background

External skeletal fixation is a method of bone stabilization which utilizes metallic fixation elements that are inserted percutaneously into bone segments.[1,2] The fixation elements are connected by an external framework which serves to stabilize the transfixed bone segments.[2] External skeletal fixation has been utilized as a method of fracture stabilization for over 100 years.[1] The first report of the application of these devices in animals was in 1934.[1] Since that time the use of external skeletal fixation has been steadily evolving with many advances in the area of implant design[3,4], biomechanical studies[5-18], and clinical experience in the utilization of these fixation systems.[19-32]

Today a number of fixator systems are available for use in dogs and cats including linear external skeletal fixation systems, circular external skeletal fixation systems and a combination of these two fixation systems which is known as linear-circular hybrid fixation.[4] Hybrid fixators were designed in an attempt to take advantage of the benefits of both linear fixation and circular fixation systems while avoiding some of the complications commonly encountered when using either system alone.[32] The purpose of Chapter 1 is to provide an overview of the development of hybrid fixator systems and a review of the biomechanical properties of the components of both linear and circular fixation which are utilized in hybrid fixation systems.

History

The first external skeletal fixator systems to become widely utilized in dogs and cats were linear external skeletal fixators.[1] Linear fixation typically utilizes pins as fixation elements.[33] The pins are smooth or threaded and can be inserted from one or more planes (uniplanar vs.

multiplanar).[34-37] The pins are connected using a system of linear rods and clamps or by means of an acrylic column, which can be custom molded to the desired shape or dimension.[33,35,37]

Circular external skeletal fixation was developed during the early 1950's in the Soviet Union by a surgeon named Gavriil Ilizarov.[38] Circular fixation systems utilize an external framework of metallic or carbon fiber rings which encircle a limb.[4,39] The fixation elements consist of small diameter wires which pass percutaneously through a limb, transfixing a bone segment perpendicular to the longitudinal axis and are bolted to the surface of the rings.[4,39,40] The fixation wires are generally tensioned to improve the stiffness characteristics of the construct. Various ring and fixation wire diameters are available to provide versatile clinical applicability.[39] Dr. Ilizarov initially utilized circular fixators to successfully correct angular limb deformities, lengthen bones, and stabilize fractures.[41] The principles and instrumentation developed by Dr. Ilizarov were introduced in Italy in 1981 and in the United States in 1987.[42] The system rapidly gained popularity as familiarity was acquired with application of the device and as clinical successes were achieved.[42]

In the early 1990's circular fixators began to be widely utilized in dogs and cats.[43] Several commercially available circular fixator systems have been developed for these species. Numerous biomechanical studies evaluating the mechanical parameters of various circular fixator systems designed for use in dogs and cats have been performed.[5-8,14,44] Circular fixators are currently used in dogs and cats to correct angular limb deformities,[28,31,45,46] lengthen appendicular limb segments,[45] perform bone transport,[24] arthrodese joints,[23,27,47] and to stabilize fractures.[19,30,47,48]

Bone Healing With External Skeletal Fixation

There are several important differences between linear external skeletal fixation and circular external skeletal fixation which may affect bone healing. Linear fixators are connected

to bone segments with relatively stiff fixation pins, often unilaterally.[1] This configuration leads to a rigid construct which experiences cantilever bending when loaded in axial compression.[49,50] Thus fracture or osteotomy gap axial displacement is highest at the trans cortex of the stabilized bone, opposite the fixator. Circular fixators utilize relatively small diameter tensioned wires rather than pins to stabilize bone segments.[4,51] The wires cross at divergent angles through bone segments but placement is generally confined to the medial to lateral plane in order to limit post-operative morbidity.[6,52] Divergent fixation wire insertion leads to uniform fracture gap displacement under axial loading.[52,53] Circular constructs have been shown to have a lower stiffness under axial loading than linear fixators.[49,50] The load-displacement curve slope of circular fixator constructs is non-linear and increases with increasing load application.[6,39,49,50,52-54] This exponentially increasing stiffness has been purported to allow controlled axial micromotion at the fracture site.[49,52] Axial micromotion has been shown to increase the rate of new bone formation and a circular fixator construct which allows axial micromotion may result in faster bone healing than would be expected with a stiffer linear fixator.[55-60]

Limitations of External Skeletal Fixation

There are several disadvantages associated with the use of any external fixator system. The framework and external portion of the fixation elements require daily cleaning and in some cases daily adjustments to the frame may need to be performed.[2] Tracts where fixation elements exit the skin commonly drain and may result in local inflammation or infection, leading to implant loosening and/or osteomyelitis.[61]

An additional problem associated more specifically with the use of circular fixator systems are the anatomic constraints associated with the application of rings in proximal limb segments or near joints. Full rings are difficult or impossible to use in the proximal humerus and femur

due to the resulting impingement of the frame on the axilla and groin, respectively. When placed near joints, rings may cause soft tissue impingement and inhibit range of motion.[8,20] To deal with the restrictions to range of motion imposed by full rings, incomplete rings have been developed. Incomplete rings typically have a "horseshoe" or "U" shape. When used near a joint, the open end is often oriented in the direction of joint flexion.[26]

Hybrid Fixator Development

As early as 1986, Italian surgeons started substituting threaded half-pins in place of some of the tensioned wires in circular fixator constructs.[42] The half-pins were inserted off of incomplete rings utilized as the proximal ring components of circular fixator constructs which utilized traditional fixation wires on the distal ring components. This technique allowed circular fixator systems to be used in proximal limb segments and avoided excessive soft tissue impingement and neuropraxia which had previously been noted when traditional circular fixator systems were applied in similar situations.[42] The successful substitution of half-pins for tensioned wires in circular fixator systems encouraged further experimentation in the integration of both linear and circular components. A new class of hybrid external skeletal fixation emerged which utilized components of both circular and linear fixator systems.[20] The early hybrid systems were necessarily somewhat make-shift because the available linear fixator components were not designed to be incorporated into the circular fixator systems. Once the use of hybrid fixators became more widespread, specialized hybrid fixation elements began to be produced which were compatible with components from both linear and circular fixator systems.[20,26]

Initially the term "hybrid fixator" had a relatively broad definition and included any fixator which used at least some circular fixator elements combined with elements of a linear fixator.[42] The term "hybrid fixator" in veterinary surgery typically refers to a linear-circular hybrid fixator which incorporates one or more rings articulated with a hybrid-linear connecting

rod or rods.[20,26] The ring(s) typically utilize small diameter wires as fixation elements, although partially threaded or smooth half-pins can also be inserted as fixation elements and bolted to the ring.[20,26] Hybrid-linear connecting rods have a smooth shaft that accepts fixation pin clamps and threads at one end to allow the rod to be bolted to a ring. Half- or full-pin fixation pins are utilized as fixation elements clamped to the hybrid-linear rod.[20,26]

A nomenclature system to group linear fixators by construct design has been previously described by Simon Roe.[62,63] The nomenclature system is based on the number of planes in which fixation pins are inserted, whether full- or half-pin splintage is used, and connecting rod position. The linear fixator nomenclature has been modified for application to linear-circular hybrid fixator constructs.[32] Hybrid construct grouping is based on the configuration of the linear portion of the fixator. A type Ia construct utilizes a single linear rod and unilateral insertion of half-pin fixation pins. Incorporation of two linear rods and biplanar insertion of half-pins results in classification as a type Ib hybrid construct. A type II hybrid construct utilizes two linear rods applied on opposite sides of the ring in the same plane and incorporation of at least one full-pin. Classification as a type III hybrid fixator is attained by the incorporation of at least three linear rods, each supporting insertion of at least one fixation pin.

Hybrid Fixator Biomechanics

The use of hybrid fixators is steadily increasing in orthopedic surgery performed on dogs and cats.[20,26] Despite the routine clinical use of hybrid fixators, little information is available regarding the biomechanical performance of hybrid fixator systems designed for use in dogs and cats. A biomechanical study comparing the performance of several early designs of hybrid fixator constructs to circular fixator constructs concluded that overall hybrid fixators were less stiff than circular fixators in axial compression, similar in stiffness in bending, and slightly stiffer in torsion.[18] The results of this study must be interpreted with caution because all hybrid

constructs tested utilized fixation pins inserted from unilaterally positioned 1/3 ring-arches as the linear components of the frames. Many of the individual components utilized in hybrid fixator constructs have been evaluated in previous biomechanical studies. Extrapolation of this data may prove useful in guiding hybrid fixator design to optimize mechanical performance and in offering guidelines on the utilization of hybrid fixators in clinical cases.

Linear Connecting Rods

Connecting rods are the supporting elements of the linear portion of hybrid fixator constructs.[33] The influence of connecting rods on linear fixator mechanical properties have been evaluated in several studies. Adding a second linear connecting rod to a type Ia fixator doubled the fixator strength when tested in axial compression; however, bending stiffness was only improved by 20% and torsional stiffness was not affected.[13] A 50% increase in connecting rod diameter has been shown to increase construct stiffness by up to 34% in axial compression, 25% in torsion, and 62% in bending.[64] Connecting rod number and positioning also affects construct stiffness. Incorporation of two connecting rods in a biplanar (type Ib) or uniplanar (type II) configuration can more than double construct stiffness in axial compression, torsion, and bending compared to the stiffness of a construct incorporating a single connecting rod (type Ia).[64] Construct bending stiffness is increased by incorporation of a linear rod in the plane in which bending testing is performed. Incorporation of a third linear rod in a type III configuration increases construct stiffness by approximately 50% in axial compression and 23% in torsion compared to constructs incorporating two linear rods.[64]

Fixation Pins

Fixation pins are the fixation elements utilized in the linear portion of hybrid fixator constructs.[20,26] Fixation pins are either smooth or partially threaded.[36] Partially threaded fixation pins can have a negative profile (core diameter is less in the threaded portion of the pin

compared to the diameter of the unthreaded portion of the pin) or positive profile (core diameter is constant throughout the length of the pin).[34,36,65] Partially threaded pins require greater force to be axially extracted from bone and are thus more resistant to loosening than are smooth pins.[13,34,65] Positive profile partially threaded fixation pins are stronger in bending than negative profile partially threaded fixation pins due to the positive profile pins' constant core diameter.[34,35,65,66] Biplanar insertion of half-pins can increase construct stiffness in axial compression, torsion, and bending by approximately 50% to 100% compared to constructs utilizing uniplanar insertion of half-pins.[9,11,16] Increasing fixation pin diameter can significantly improve construct performance because the bending stiffness of fixation pins is related to the radius of the pin raised to the 4th power.[35] Brinker et al. demonstrated that an increase in fixation pin diameter of 60% was sufficient to almost double overall construct stiffness.[12]

Rings

Circular rings are utilized as supporting elements in hybrid fixators.[20,26,29,32] The rings typically have a series of uniformly spaced holes around the ring circumference to facilitate articulation with other frame components and fixation elements. Several attributes which affect the mechanical properties of rings include whether the ring is complete or incomplete (has an open section), the diameter of the ring, and the material composition of the ring.[6-8,49,54,67] Ring diameter affects construct stiffness because the internal diameter of the ring determines fixation wire length.[49,54] To minimize fixation wire length, circular are usually constructed using the smallest ring diameter which allows two cm of space between the inside of the ring annulus (metal circumference of the ring) and the limb segment to which the ring is applied. Lewis et al. showed that increasing ring diameter from 50 mm to 118 mm resulted in a greater than 70% decrease in axial stiffness.[6] Bronson demonstrated a 30% reduction in axial stiffness and a 10% reduction in bending and torsional stiffness associated with a ring diameter increase of 40%.[5]

Circular rings are commonly made of aluminum, stainless steel, or carbon fiber. A comparison of the mechanical properties of stainless steel and carbon fiber rings found only small differences in construct stiffness based on ring material.[67] The incorporation of incomplete rings has been demonstrated to decrease construct stiffness by 25% to 30% in axial compression, bending, and torsion compared to similar constructs incorporating complete rings.[8]

Fixation Wires

Fixation wires are the fixation elements which bolt to the ring component of a hybrid fixator. Fixation wires are typically constructed of stainless steel and may be smooth or incorporate a stopper (olive wires).[51] Increasing the number of fixation wires has been demonstrated to increase construct stiffness. Cross demonstrated an increase in construct stiffness of 12% in axial compression, 12% in torsion, 35% in craniocaudal bending and 50% in mediolateral bending associated with increasing the number of fixation wires from two to three in a group of circular fixator constructs.[8] A 50% increase in fixation wire diameter has been shown to more than double the load to wire yield.[40] Fixation wire diameter also affects construct stiffness. An increase of 10% to 20% in overall construct stiffness was noted when fixation wire diameter was increased from 1.5 mm to 1.8 mm diameter.[40] Fixation wires are commonly tensioned, particularly when incorporated in constructs made with larger diameter rings. Fixation wire tensioning from 0 kg to 90 kg has been shown to increase circular fixator construct axial stiffness by 35% to 50% and decrease construct axial displacement by approximately 30%.[7] Increasing the divergency angle of fixation wire insertion may increase resistance to shear forces and promote uniform bending stiffness.[51,53,68] The incorporation of olive wires in place of smooth fixation wires has also been shown to increase construct bending stiffness and bone segment stability.[51,52,54]

Fixation Bolts

Fixation bolts are utilized to secure fixation wires to the surface of the ring.[40] Fixation bolts are typically slotted or cannulated or both.[40] Application of adequate torque to fixation bolts is important to prevent fixation wire slippage and the torque required is dependent on fixation wire diameter.[40] A 50% increase in fixation wire diameter requires that fixation bolt torque be doubled to prevent wire slippage.[40] A torque of approximately 8 Nm is required to prevent slippage of the commonly utilized 1.6 mm diameter fixation wires.[40] The ability of fixation bolts to prevent wire slippage declines with reuse.[40] Slotted fixation bolts should not be reused more than four times and cannulated fixation bolts should not be reused more than seven times.[40]

Assimilation of the information supplied by the studies evaluating the different components which are incorporated in hybrid fixators allows some general conclusions to be drawn regarding construct design. Incorporation of multiple linear connecting rods and multiplanar insertion of optimized diameter positive profile fixation pins is likely to increase the stiffness of the axial portion of a hybrid fixator. Stiffness in the circular portion of the hybrid fixator construct can likely be increased by selecting the smallest diameter complete ring appropriate to limb size and utilization of more than 2, large diameter, tensioned, olive fixation wires per ring inserted with a 90° wire crossing angle.

Clinical Reports of Hybrid Fixator Application

Despite the lack of biomechanical studies evaluating the performance of hybrid fixators, the body of evidence supporting the clinical use of hybrid fixators in dogs and cats is steadily increasing. Several case series have been published describing the application and efficacy of hybrid fixators for the correction of limb deformities[29,32] and stabilization of fractures.[20,22,26]

Fracture Fixation

Farese et al. reported the use of linear-circular hybrid fixators to stabilize long bone fractures in three dogs and three cats.[20] Patient age ranged from 10 to 108 months and weight ranged from 2 to 15 kg. All fractures were closed and in five out of six cases the fractures had a metaphyseal component. Fracture configuration was comminuted in five cases and simple in one case. All hybrid fixators utilized a single ring which was either incomplete (five cases) or complete (one case). Ring fixation elements included a single full-pin in two cases, two fixation wires in two cases and three fixation wires in the remaining two cases. Olive wires were used in three of the four cases which utilized fixation wires to increase bone segment stability. Two hybrid rods were used in all cases and three cases had a third linear rod attached as a diagonal strut between the two hybrid rods. Uniplanar fixation pin insertion was used in three cases and biplanar fixation pin insertion was used in the other three cases. The number of fixation pins used in the linear portion of the fixators ranged from three to five. Three cases experienced significant complications requiring a revision surgery. Two additional cases experienced minor complications including soft tissue impingement by the hybrid frame in one case and pin tract infection in the second case. All cases progressed to union with healing times ranging from 64 to 137 days and final outcome was assessed as excellent in all cases.

Kirkby et al. reported a case series of 21 dogs and five cats with humeral or femoral fractures stabilized with linear-circular hybrid fixators.[26] Patient age ranged from 3 to 132 months and weight ranged from 3 to 54 kg. Five out of 26 fractures were open and 24 out of 26 fractures were comminuted. Fractures involved the metaphyseal or physeal region in 20 out of 26 cases. All hybrid fixators utilized a single incomplete ring except for one construct which utilized a single 1/3 arch in place of a ring. One construct also utilized an additional 1/3 arch in addition to a ring. Ring fixation elements included a single full-pin in nine cases, half-pins only

in two cases, fixation wires alone in 10 cases, and a combination of the aforementioned ring fixation elements in five cases. Olive wires were incorporated in eight cases which utilized fixation wires. The number of hybrid rods used in each construct ranged from one to four with a mode of two. In all constructs the primary hybrid rod was placed on the lateral aspect of the limb segment. In 14 cases at least one of the hybrid rods was incorporated as a diagonal strut between the ring and the proximal end of the primary hybrid rod. The number of fixation pins utilized in the linear portion of the hybrid fixators ranged from two to five with a mode of three pins. Fixation pin insertion was uniplanar in 13 cases and multiplanar in 13 cases. Postoperative complications occurred in 14 cases, with the most common complication being pin tract inflammation (13 cases). Fractures failed to heal in two cases, although in both the application of a hybrid fixator had been a revision to a previously failed surgical stabilization. Out of 26 fractures, 24 had complete radiographic follow up and 22 progressed to complete union with healing times ranging from 25 to 280 days. Limb function was assessed as excellent in 15 cases, good in six cases and fair in three cases.

The use of hybrid fixators to stabilize distal diaphyseal fractures in three dogs was reported by Clarke and Carmichael.[22] Patient age ranged from 20 to 48 months and weight ranged from 19 to 29 kg. Fractures were closed and comminuted in all cases. Two cases were revisions of a previous surgical stabilization which had failed to heal, one of which was complicated by osteomyelitis. Hybrid fixators utilized a single ring in two cases and two rings in the remaining case. Ring fixation elements consisted of two or three smooth fixation wires per ring. All hybrid constructs utilized two hybrid rods and biplanar fixation pin insertion with five fixation pins. Pin tract inflammation and fixation element loosening occurred in two out of the three cases. All three fractures went to union with healing times ranging from 49 to 70 days.

Angular Deformity Correction

Sereda et al. described the correction of 18 developmental antebrachial deformities in 17 dogs using linear-circular hybrid fixators.[32] Patient age ranged from 7 to 69 months and weight ranged from 6 to 39 kg. The primary component of the antebrachial deformity was valgus deviation in 13 cases, varus deviation in two cases and rotational in two cases. Hybrid fixators incorporated a single ring in 14 cases, two rings in one case, and three rings in two cases which required a proximal ulnar osteotomy. Two or three fixation wires were attached to each ring as fixation elements and a combination of smooth wires and olive wires were used in most cases. The majority of cases utilized two hybrid rods (16 fixators) and the remainder utilized a single hybrid rod (two fixators). The number of fixation pins incorporated in the linear portion of the fixator ranged from three to six with a either three or five most commonly used. Biplanar fixation pin insertion was used in 16 fixators and uniplanar fixation pin insertion in two fixators. Complications occurred in 13 cases with pin tract inflammation being the most common complication and occurring in nine limbs. Radial fracture requiring revision surgery occurred in two cases. Successful correction of antebrachial deformity and bone union was achieved in all cases. Time to union ranged from 50 to 123 days. Final functional outcome was assessed as excellent in 13 cases, and good in four cases.

Radasch et al. utilized hybrid fixators to correct 14 pes varus deformities in 13 dachshunds.[29] Patient age ranged from 3 to 15 months and weight ranged from 4 to 7 kg. All deformities were corrected with a single, opening wedge osteotomy. All hybrid fixators utilized a single complete or incomplete ring to stabilize the metaphyseal bone segment. Ring fixation elements consisted of two fixation wires in 13 cases and three fixation wires in one case. A combination of both smooth Kirschner wires and olive wires were incorporated in all hybrid fixator constructs. A single, medially placed hybrid rod and two half-pin fixation pins were used

as the linear component of all hybrid fixator constructs. Complications were noted in five cases including pin tract drainage in two cases, fibular fracture and concurrent translational deformity in two cases and tibial re-fracture post fixator removal in one case. All osteotomies achieved successful bone union with healing times ranging from 7 to 12 weeks. Outcome was assessed as good to excellent in 13 out of 14 deformity corrections.

The published case series describing the use of hybrid fixators in dogs and cats provide a summary of the range of clinical applications and the typical designs utilized in the assembly of hybrid constructs. Based on a compilation of the information contained in the reports; the most commonly used hybrid fixator construct incorporates a single ring and three fixation wires as the circular component while two hybrid rods with three positive profile fixation pins inserted in multiple planes make up the linear component. This "typical" hybrid construct would be termed a type Ib hybrid construct. The majority of hybrid fixator constructs incorporate olive fixation wires and utilize incomplete rings. The overall incidence of major complications (requiring revision surgery) reported was 12% and overall incidence of minor complications was 52%. The most common minor complication was pin tract inflammation and drainage. In most cases the final functional outcome was rated as excellent.

Summary

Hybrid fixators are made using both circular and linear fixator components. Hybrid fixation has been purported to combine the benefit of rapid stimulation of bone healing traditionally associated with circular fixation with the reduction in soft tissue impingement afforded by linear fixation. While no biomechanical studies have evaluated the performance of hybrid fixator construct designs similar to those commonly utilized in dogs and cats today, many of the individual components incorporated in hybrid fixators have been evaluated in previous biomechanical studies. Several clinical case series have described the application of hybrid

fixators for fracture stabilization and correction of limb deformities. Despite the successful clinical use of hybrid fixators, biomechanical testing of hybrid fixator constructs is needed to validate the purported benefits of hybrid fixation. Additional biomechanical studies may also determine guidelines for frame or fixation element modification to allow optimization of construct performance and facilitate rapid bone union.

CHAPTER 2
EFFECT OF WIRE TENSION ON INCOMPLETE AND COMPLETE RING COMPONENT DEFORMATION

Background

The use of circular external skeletal fixation and hybrid external skeletal fixation has become a well established treatment modality in dogs and cats over the past decade.[19,20,23-32,45-47] Circular fixator constructs are composed of a series of complete or incomplete extracorporeal rings that are interconnected by multiple rods.[41] Hybrid fixator constructs typically use ring components to stabilize one of the major bone segments while an articulated linear rod or rods utilizing half- or full-pins are used as fixation elements to stabilize the other major bone segment.[20,26,29,32] In previous clinical reports, small diameter wires are used as fixation elements securing the stabilized bone segments within the ring components of both circular and hybrid fixator constructs.[20,26,29,32,37,41,52,69,70] In human patients, fixation wires are always tensioned to improve construct mechanics and mitigate excessive displacement of the secured bone segments during weight-bearing. Studies performed to determine the appropriate tension for wires utilized in complete ring components specifically designed for use in dogs have suggested that wires on 50 mm rings do not require tensioning, 66 mm rings require a wire tension between 0 and 90 kg depending on patient body weight, and for 84 mm and 118 mm rings the maximum wire tension of 90 kg should be utilized.[7]

While the use of complete rings confers biomechanical advantages to a circular or hybrid fixator construct, anatomic constraints often prohibit the use of complete rings in a number of locations.[8,47] Complete rings are not typically used proximal to the elbow or stifle in dogs and cats and the use of complete rings adjacent to joints can interfere with range of motion in the distal extremities.[26] Incomplete rings which have an open section in the ring circumference are often utilized in circular and hybrid constructs to avoid impingement or interference with the

regional anatomy.[8,20,30,47] It has been suggested that less tension be applied to wires on incomplete rings than to wires on comparable diameter complete rings to prevent excessive incomplete ring deformation.[8,39] Small wire crossing angles have also been associated with ring deformation in a study utilizing complete rings.[53] To date, no studies have been performed which evaluate the effect of wire tensioning or wire crossing angle on ring deformation using ring components designed for use in dogs and cats. The purpose of this study was to characterize the deformation in complete and incomplete (5/8 circumference) rings as a result of fixation wire tensioning in 50 mm, 66 mm, 84 mm and 118 mm diameter rings utilizing three different wire crossing angles. Our hypotheses were that wire tensioning would result in measurable deformation of incomplete but not complete rings and that ring deformation would increase with decreasing wire crossing angle.

Materials and Methods

Construct Preparation

Single ring constructs (IMEX Veterinary Inc., Longview, Tx) were made using 50 mm, 66 mm, 84 mm and 118 mm (inner diameter) complete and incomplete (5/8 circumference) rings. A 50 mm long segment of 16 mm diameter Delrin rod (Acetal polymer, MSC Industrial Supply, Melville, New York) was used as a bone model. Two 1.6 mm diameter Kirschner wires were used to stabilize the Delrin rod vertically within each ring, each wire secured to an opposing surface of the ring. A 1.58 mm twist drill bit mounted in a drill press was used to create pilot holes in the Delrin rod to facilitate accurate insertion of the Kirschner wires. The Kirschner wires were inserted using a variable speed cordless drill (Bosch 14.4 Volt drill, Robert Bosch LLC, Farmington Hills, MI) and crossed through the Delrin rod at a divergence angle of either 45° or 90° for all constructs. A third wire crossing angle of 67.5° for 50 mm and 118 mm rings, 72° for 66 mm rings, and 60° for 84 mm rings was also utilized. The Kirschner wires were

secured to the rings using cannulated/slotted wire fixation bolts (IMEX Veterinary Inc., Longview, Tx).

All constructs were assembled and mounted on a solid base which consisted of a 200 mm x 200 mm x 76 mm aluminum block with a single 6 mm diameter threaded rod protruding vertically from the block. The block was clamped firmly to the work surface. Constructs made with incomplete rings were secured to the block by inserting the threaded rod through the ring fixation hole located in the center of the closed end of the ring. Constructs made with complete rings were secured through an analogous hole based on wire position. Two nuts were tightened against opposing surfaces of the ring to secure the construct's position on the threaded rod. This system was designed to minimize inhibition of ring deformation in all planes during testing.

Ring Deformation Measurement

To quantify ring deformation during testing, two measurement points were scribed on opposite sides of the inner surface of each ring. The location of the scribed points was defined as the point on the inner surface of each ring which was bisected by the mediolateral axis crossing through the center of the ring perpendicular to the craniocaudal axis which bisected both the center of the ring and the hole in the center of the closed end of the ring (Figure 2-1). Ring measurements were collected from the two scribed points during the testing cycle using the articulating arm of a three dimensional digitizing system (Microscribe, CNC Services Inc., Amherst, VA). Initial ring measurements were collected before applying tension to the Kirschner wires. The two wire fixation bolts farthest from the open end of the ring were tightened to 10.5 Nm using a factory calibrated torque wrench (Craftsman, KCD IP, LLC, Hoffman Estates, IL) prior to tensioning the Kirschner wires.[40] The Kirschner wires were tensioned simultaneously throughout the study using two calibrated dynamometric wire tensioners (Smith and Nephew Inc., Memphis, TN). The Kirschner wires were sequentially

tensioned from 0 kg to 30 kg, to 60 kg, and to 90 kg. Once the first pre-determined wire tension was reached, the two remaining wire fixation bolts were tightened to 10.5 Nm, the tensioners were removed, and both ring measurement points were sequentially recorded using the Microscribe. After ring measurements were obtained, the fixation bolts farthest from the ring fixation point on the threaded rod, were loosened, releasing wire tension. Ring measurements were again obtained with wire tension released. Wires were then tensioned to the next higher tension level, fixation bolts were re-tightened to 10.5 Nm, tensioners were removed, and ring measurements were collected with the Microscribe. This process was repeated until the final wire tension of 90 kg was achieved. Visual observations regarding ring deformation were also recorded. Three replications of each construct were tested for a total of 72 constructs, utilizing new rings and Kirschner wires for each construct. The outcome measure for our study was ring deformation, defined as a statistically significant decreased in measured internal ring diameter as a result of wire tensioning. Ring deformation was further classified as elastic deformation if ring diameter returned to pre-wire tensioning diameter once wire tension was released or as plastic deformation if measured ring diameter remained residually decreased from initially measured ring diameter once wire tension was released. Ring failure was defined as catastrophic plastic deformation resulting in failure to achieve the predetermined wire tension levels.

Data Analysis

The Microscribe exports the location in space of a measurement point as X, Y, and Z coordinates based on a coordinate system located in the base of the Microscribe unit. Data from the Microscribe was exported to a spreadsheet program (Microsoft Office Excel 2003, Microsoft Corporation, Redmond, WA) and analyzed using the Euclidean distance formula

$$\sqrt{(X_1-X_2)^2 + (Y_1-Y_2)^2 + (Z_1-Z_2)^2} \qquad (2\text{-}1)$$

to determine the inner ring diameter between the two scribed ring points at each time ring measurements were collected.

Statistical analysis was performed utilizing commercially available statistical software (SPSS 18, IBM Corp. Somers, NY). The effect of ring type, wire crossing angle, and fixation wire tension on measured ring diameter within each ring size was evaluated using a repeated measures analysis of variance (ANOVA). A separate ANOVA was performed for each of the four tested ring sizes (50 mm, 66 mm, 84 mm, and 118 mm diameters). Each group of ring constructs within a single ring size was further subdivided into two groups based on ring type. Separate repeated measures ANOVAs were performed to identify the effect of wire tension on measured ring diameter by ring type. A post hoc Bonferoni correction was utilized to account for multiple comparisons. Percentage changes in ring diameter as a result of fixation wire tensioning were calculated and descriptive statistics (mean and standard deviation) were performed on these percentages. A P value of ≤ 0.05 was set as significant.

Results

Observations

Visible deformation was not apparent in any of the complete rings at any of the applied wire tensions. Visible deformation occurred in all incomplete rings when the wires were tensioned to 60 kg or greater, regardless of wire crossing angle. Incomplete ring failure was noted in one, 50 mm diameter ring with a 67.5° wire crossing angle; all three, 66 mm diameter rings with 45° wire crossing angles; one, 66 mm diameter ring with a 72° wire crossing angle; one, 66 mm diameter ring with a 90° wire crossing angle and one, 84 mm ring with a 60° wire crossing angle when we attempted to tension fixation wires to 90 kg.

Ring Deformation

Fixation wire tension had a significant effect on measured ring diameter for all four ring diameters tested. In addition, all interactions including 2-way interactions between wire tension and wire crossing angle and between wire tension and ring type, and 3-way interactions between wire tension, wire crossing angle, and ring type were significant for all four ring diameters. With all wire tensions and wire crossing angles pooled, measured ring diameter was significantly different between constructs based on ring type in all four ring diameters tested. Individual analysis of the effect of wire tensioning on measured ring diameter by ring type revealed that wire tensioning did not significantly decrease measured ring diameter in complete ring constructs. Wire tensioning to 30 kg, 60 kg, or 90 kg significantly decreased measured ring diameter in all sizes of incomplete ring constructs. Wire tensioning to 90 kg resulted in a permanent decrease in measured ring diameter in 50 mm, 66 mm (most failed), and 118 mm diameter incomplete ring constructs (Table 2.1 and 2.2). Wire crossing angle had a significant but inconsistent effect on the measured decrease in ring diameter at sequential wire tensions between the four ring diameters.

Discussion

Fixation wires utilized in circular fixator constructs in dogs are commonly tensioned to increase construct stiffness and improve stability of transfixed bone segments.[6,7,37,39] Additionally, incorporation of fixation wires inserted at the maximum crossing angle of 90° has been shown to improve circular fixator construct performance by increasing bending stiffness and decreasing bone segment slippage when smooth fixation wires are utilized.[51-53,68] In previous studies it was speculated that fixation wire tensioning, particularly when combined with a small wire crossing angle, could result in deformation of incomplete rings.[39,71] For this reason, applying less tension to fixation wires attached to incomplete rings than to fixation wires

attached to complete rings has been recommended.[8,39] The results of our study support the previous theories and our hypothesis by demonstrating that wire tensioning results in measurable deformation of incomplete rings. Different than our hypothesis, the amount of ring deformation which was induced by wire tensioning in this study was not consistently influenced by wire crossing angle.

Effect of Wire Tensioning

Wire tensioning resulted in statistically significant deformation in incomplete ring constructs even at the lowest applied tension of 30 kg. Continued wire tensioning resulted in permanent ring deformation or ring failure in some rings of all four ring diameters tested. The clinical significance of ring deformation is currently unknown. Podolsky et al. attributed a decrease in circular fixator construct axial stiffness to ring deformation, but the magnitude of the ring deformation was not measured.[53] Incorporation of incomplete rings in circular fixator constructs has been shown to decrease construct stiffness in axial compression, bending, and torsion compared to the incorporation of similar diameter complete rings by Cross et al.[8] The decrease in stiffness associated with incorporation of incomplete rings in the study by Cross et al. was attributed to deformation of the incomplete rings, although ring deformation was not objectively measured. Nele et al. demonstrated that permanent deformation of complete rings can result in decreased construct stiffness.[67] The residual decrease in ring diameter noted in our study when incomplete rings were tensioned to 90 kg suggests that the rings underwent a small amount of plastic deformation at the maximum tension we applied. Based on the study by Nele et al. we suspect that tensioning wires attached to incomplete rings to 90 kg may induce enough deformation to decrease ring stiffness.

Description of Ring Failure

The incomplete ring failure noted when we attempted to tension the fixation wires to 90 kg occurred via a similar pattern each time. As the Kirschner wires were tensioned above 60 kg force, one arm of the ring would begin to bend out of the plane of the ring. Out of plane bending of the ring initiated adjacent to the central hole in the closed end of the ring through which the threaded rod had been secured to mount the ring. The initial bending of one ring arm was rapidly followed by a similar bending in the opposing ring arm and continued application of tension to the Kirschner wires resulted in rapid collapse of the ring and loss of wire tension (Figure 2-2). Ring failure occurred consistently in 66 mm diameter rings (five out of nine rings failed), but a single 50 mm diameter ring and a single 84 mm diameter ring also failed. The metal annulus, the circumferential body of the ring, in the 66 mm diameter rings has the same width and thickness as the 50 mm diameter rings. The thickness of the ring annulus increases in the 84 mm diameter rings and increases again in the 118 mm diameter rings. Thus the annulus of the 66 mm rings has the same cross sectional area as the smaller, 50 mm rings while the larger 84 mm and 118 mm rings have an increased annulus cross sectional area. We hypothesize that the 66 mm rings failed a higher percentage of the time than the other diameter rings due to a smaller cross sectional area relative to the ring diameter, resulting in decreased ability to sustain tensile load across the ring span. The point of ring failure was noted to be near the mounting hole on the threaded rod. We are unsure what influence the location of the threaded rod may have played in the point of ring failure. When incomplete rings are utilized in circular fixator constructs, several connecting rods are commonly used to articulate rings in the frame.[8,30,47,48] Placement of multiple connecting rods would likely mitigate the deformation of incomplete rings we observed in this study. The deformation observed when incomplete rings are incorporated in hybrid fixator constructs may be closer to that noted in our study as it is common practice to

articulate the ring component of a hybrid fixator to a single hybrid rod, which serves as the linear connecting element of the hybrid fixator construct.[26,29,32]

Effect of Wire Crossing Angle

We chose to evaluate three wire crossing angles in our study. The smallest crossing angle evaluated was 45° as this is the minimum crossing angle which is recommended for clinical use to avoid excessive bone segment instability.[5,39,51,54,68,70] The largest angle tested was 90° as this is the maximum wire crossing angle achievable and the most optimal from a biomechanical standpoint.[5,39,51,52,68,70] The minimum and maximum wire crossing angles tested did not allow fixation wires to intersect in the center of the incomplete ring constructs. A third wire crossing angle was also tested which allowed fixation wires to intersect in the center of the incomplete rings; however, this angle varied between rings of different diameters. In the clinical setting wire crossing angle is typically maximized to the extent possible without compromising important neurovascular structures.[39] The maximum fixation wire crossing angle achievable clinically is often much less than the biomechanically optimal angle of 90°.[30,41,54]

No consistent trend between wire crossing angle and ring deformation was observed in our study. The wire crossing angle which resulted in the greatest decrease in ring diameter after wire tensioning varied between ring sizes (Table 2.3) and subjectively varied between different wire tensions within the same ring size (Table 2.4). Optimally the effect of wire crossing angle on ring deformation would have been analyzed by evaluating the effect of wire tensioning on measured ring diameter within each ring size by ring type at each of the three wire crossing angles; unfortunately, this analysis was not performed due to small sample size. Our findings are different than the results of a previous study which suggested that utilization of small wire crossing angles and the resultant asymmetric ring loading could result in complete ring deformation which would be mitigated by the incorporation of wire crossing angles approaching

90°.[53] The inconsistencies between our findings and the conclusions of previous studies[53,71] may be related to inherent differences between complete and incomplete rings and the method of ring failure we observed in our study. Our ring constructs failed by catastrophic plastic deformation which occurred near the closed end of the incomplete rings. As wire crossing angle increases, wires must be attached closer to the gap in the incomplete ring circumference which will increase the moment arm through which the tension in the fixation wires acts on the incomplete ring and may result in increased ring deformation. In some ring diameters or at certain wire tensions the larger moment arm associated with large wire crossing angles may have been able to cause greater ring deformation than could the asymmetrical force exerted by a small wire crossing angle in our incomplete ring constructs.

Summary

In conclusion, fixation wire tensioning results in minimal deformation of complete rings but significant deformation of incomplete rings. The amount of ring deformation observed in incomplete ring constructs increases with increasing wire tension, regardless of ring size. Application of 90 kg wire tension may result in residual ring deformation or ring failure. Varying the wire crossing angle is not likely to have a predictable effect on the amount of ring deformation which occurs. While our study demonstrates that ring deformation does occur during wire tensioning, it does not address the effect of this deformation on ring performance. Additional mechanical testing studies are needed to determine if ring deformation occurs during load application to circular or hybrid fixator constructs which incorporate incomplete rings and to quantify the effect this deformation on construct performance.

Table 2-1. Ring deformation as a result of wire tensioning

Construct	0 kg tension	30 kg tension	30 kg released	60 kg tension	60 kg released	90 kg tension	90 kg released
50 mm incomplete	49.5 ± 0.7[a]	49.2 ± 0.6[b]	49.5 ± 0.7[a]	48.7 ± 0.5[c]	49.3 ± 0.6[ab]	47.3 ± 0.9[d]	48.1 ± 0.7[c]
50 mm complete	49.9 ± 1.1[a]	49.9 ± 1.1[a]	49.9 ± 1.1[a]	49.8 ± 1.2[a]	49.9 ± 1.1[a]	49.8 ± 1.2[a]	49.8 ± 1.2[a]
66 mm incomplete	66.5 ± 0.8[a]	65.7 ± 0.7[b]	66.4 ± 0.7[a]	64.6 ± 0.6[c]	66.0 ± 0.8[ab]	N/A	N/A
66 mm complete	66.6 ± 0.8[a]	66.5 ± 0.7[a]	66.5 ± 0.8[a]	66.5 ± 0.8[a]	66.5 ± 0.8[a]	66.5 ± 0.8[a]	66.6 ± 0.9[a]
84 mm incomplete	84.6 ± 0.9[a]	83.6 ± 0.7[b]	84.5 ± 0.9[a]	82.2 ± 0.7[cd]	84.4 ± 0.8[a]	77.7 ± 4.1[d]	80.3 ± 3.6[abc]
84 mm complete	84.8 ± 0.7[ab]	84.7 ± 0.8[ab]	84.7 ± 0.7[ab]	84.7 ± 0.8[ab]	84.7 ± 0.8[ab]	84.6 ± 0.9[a]	84.8 ± 0.8[b]
118 mm incomplete	118.6 ± 0.7[a]	116.7 ± 0.5[b]	118.6 ± 0.8[ac]	114.7 ± 0.7[d]	118.5 ± 0.7[ac]	109.0 ± 5.5[e]	114.1 ± 4.5[bcd]
118 mm complete	118.9 ± 1.2[a]	118.9 ± 1.2[a]	119.0 ± 1.2[a]	118.8 ± 1.2[a]	118.9 ± 1.2[a]	118.8 ± 1.2[a]	119.0 ± 1.2[a]

Values are mean ± standard deviation (mm) for internal ring diameter measurements during the wire tensioning sequence. Significant differences are designated by different letter superscripts. Statistical comparisons are between wire tensions (along rows) within a single construct type. Sixty-six mm diameter incomplete ring constructs tensioned to 90 kg were not included in the statistical analysis (N/A) because of small sample size due to ring failure.

Table 2-2. Percent decrease in measured inner ring diameter arranged by ring size and type

Construct size and ring type	30 kg tension	30 kg released	60 kg tension	60 kg released	90 kg tension	90 kg released
50 mm incomplete	-0.8	-0.1	-1.6	-0.4	-3.9	-2.5
50 mm complete	0.0	0.0	-0.1	0.0	-0.1	-0.1
66 mm incomplete	-1.3	-0.1	-2.9	-0.9	-4.9	-1.7
66 mm complete	-0.1	-0.1	-0.1	0.0	-0.2	0.0
84 mm incomplete	-1.2	-0.1	-2.8	-0.2	-7.9	-4.9
84 mm complete	-0.1	-0.1	-0.1	0.0	-0.2	0.0
118 mm incomplete	-1.6	0.0	-3.3	-0.1	-8.1	-3.7
118 mm complete	0.0	0.1	-0.1	0.0	-0.1	0.0

Percentage decreases in measured ring diameter are based on initial measured inner ring diameter at 0 kg wire tension. Measurements obtained during sequential wire tensioning.

Table 2-3. Mean decrease in measured ring size due to wire tensioning, arranged by wire crossing angle.

Ring size	45.0°	60°, 67.5°, or 72.0°	90.0°
50 mm	-0.43 ± 0.50[a]	-0.74 ± 1.37[b]	-0.14 ± 0.27[c]
66 mm	N/A	-0.61 ± 0.94[a]	-0.69 ± 1.15[a]
84 mm	-1.81 ± 3.17[a]	-1.56 ± 2.90[a]	-0.45 ± 0.73[b]
118 mm	-1.42 ± 2.43[a]	-2.91 ± 5.39[b]	-0.72 ± 1.39[a]

Values are mean ± standard deviation (mm) of the decrease in measured ring diameter due to sequential wire tensioning. Measurements from both complete and incomplete rings at all wire tensions are combined within each ring size. The middle wire crossing angle utilized was 67.5° in 50 mm and 118 mm rings, 72° in 66 mm rings, and 60° in 84 mm rings. Significant differences are designated by different letter superscripts. Statistical comparisons are between wire crossing angles (along rows) within a single ring size. Sixty-six mm diameter incomplete ring constructs with a wire crossing angle of 45° were not included in the statistical analysis (N/A) due to ring failure.

Table 2-4. Percent decrease in measured inner ring diameter arranged by ring size and wire crossing angle.

Ring size	Wire crossing angle	30 kg tension	30 kg released	60 kg tension	60 kg released	90 kg tension	90 kg released
50 mm	45.0°	-0.6	-0.2	-1.1	-0.3	-1.9	-1.1
	67.5°	-0.5	-0.1	-1.0	-0.4	-3.2	-2.4
	90.0°	-0.1	0.1	-0.5	-0.1	-0.8	-0.4
66 mm	45.0°	-0.6	0.0	-1.6	-0.5	-0.3	-0.1
	72.0°	-1.0	-0.1	-1.8	-0.7	-1.7	-0.1
	90.0°	-0.6	-0.1	-1.1	-0.2	-2.3	-1.2
84 mm	45.0°	-0.7	-0.2	-1.7	-0.2	-6.1	-4.1
	60.0°	-0.7	-0.1	-1.6	-0.2	-4.2	-2.8
	90.0°	-0.5	-0.1	-1.1	-0.1	-1.4	-0.2
118 mm	45.0°	-0.8	0.1	-1.9	-0.1	-3.4	-1.1
	67.5°	-1.0	0.0	-2.0	-0.1	-7.1	-4.4
	90.0°	-0.6	0.0	-1.2	0.0	-1.8	-0.1

Percentage decreases in measured ring diameter are based on initial measured inner ring diameter at 0 kg wire tension. Measurements obtained during sequential wire tensioning.

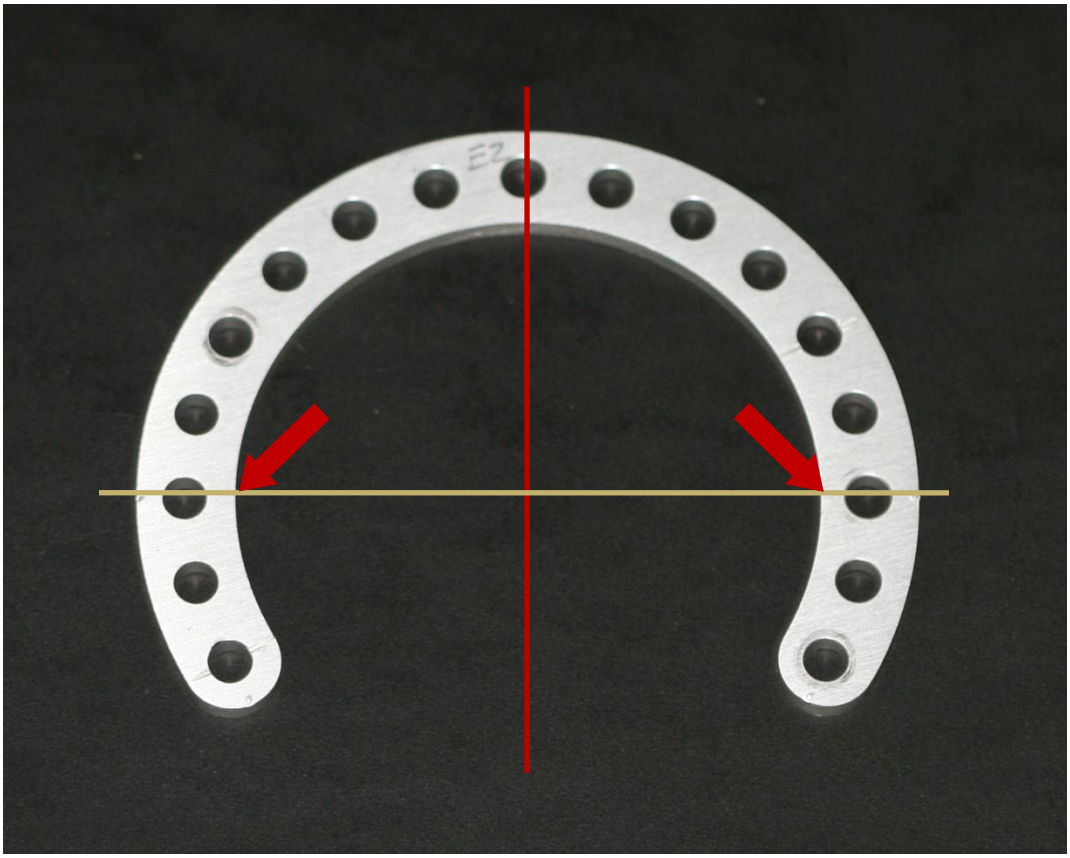


Figure 2-1. Incomplete ring 84 mm diameter. The red line indicates the craniocaudal axis (bisecting the center of the ring and the hole at the closed end of the ring) and the gold line indicates the mediolateral axis (perpendicular to the craniocaudal axis and passing through the ring center). Red arrows indicate the intersection of the mediolateral axis with the inside of the ring annulus which is the location of the scribed points used to collect ring measurements during the testing sequence and subsequently calculate the measured internal ring diameter.

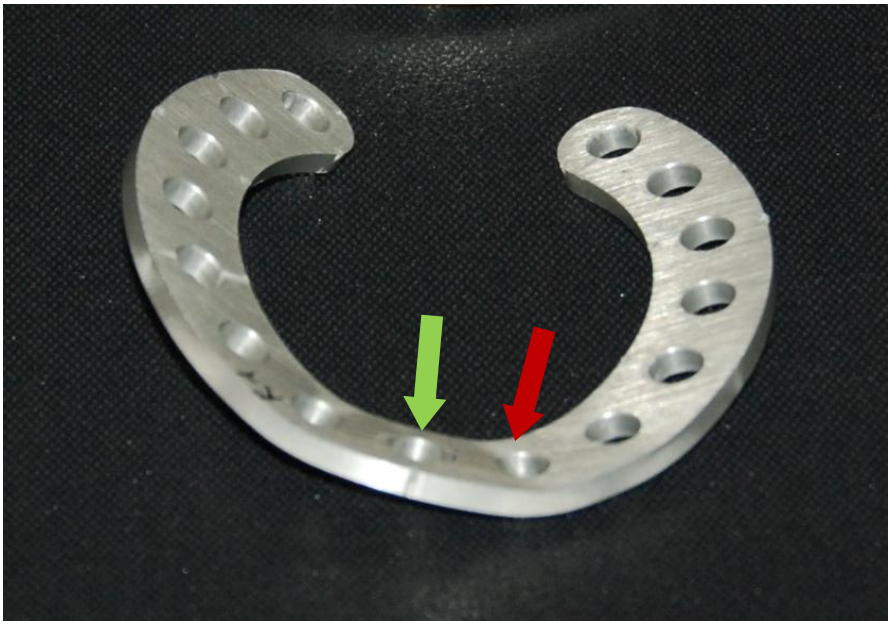


Figure 2-2. Failed 66 mm diameter incomplete ring. Red arrow indicates the location of initial ring deformation near the hole in the center of the closed end of the ring (green arrow) through which the ring was secured to the threaded rod during the testing sequence

CHAPTER 3
AXIAL STIFFNESS AND RING DEFORMATION OF COMPLETE AND INCOMPLETE SINGLE RING CIRCULAR EXTERNAL SKELETAL FIXATOR CONSTRUCTS.

Background

Circular external skeletal fixation and hybrid linear-circular external skeletal fixation has become a well-established therapeutic modality for addressing a variety of orthopedic problems affecting dogs and cats. [19,20,24-28,30,32,45,47,48] Circular fixators and the circular components of hybrid fixator constructs typically utilize small diameter, transosseous wires as fixation elements.[41] The fixation wires are secured to ring components which are the supporting element unique to circular and hybrid fixators.[4,20,26,37,39] The ring components can be complete or incomplete, containing an open section in the circumference of the ring.[8,20,30,39] The fixation wires are generally tensioned and impart unique biomechanical properties to circular fixators which differ considerably from linear external skeletal fixators.[6,18,39]

Several studies have evaluated the biomechanical properties of circular fixator systems designed specifically for use in dogs and cats.[6-8,15,17] These studies have primarily evaluated the mechanical characteristics of constructs utilizing complete rings. While complete rings confer biomechanical advantages,[8] incomplete rings are often incorporated in fixator constructs utilized in clinical situations due to anatomic constraints.[8,47] Incomplete rings are often used when a ring is positioned adjacent to a joint with the open section of the ring positioned to allow a greater range of motion in a stabilized limb segment.[20,26,48] Despite the routine use of incomplete rings in circular and hybrid fixator constructs in dogs and cats,[30] little is known regarding the biomechanical properties of incomplete rings. Cross et al. evaluated six configurations of distal ring blocks, three of which included incomplete rings.[8] The study by Cross et al. demonstrated that constructs incorporating incomplete rings were less stiff than comparable constructs utilizing

complete rings, but this study did not evaluate the effect of varying fixation wire tension and only evaluated 84 mm diameter rings.

The purpose of our study was to compare the biomechanical properties of complete and incomplete (5/8 circumference) single ring constructs subjected to axial loading. We specifically evaluated axial stiffness, axial displacement, and ring deformation resulting from wire tensioning and application of an axial force to single ring constructs. Construct variables included ring diameter (50 mm, 66 mm, 84 mm, and 118 mm) and fixation wire tension (0 kg, 30 kg, 60 kg, and 90 kg). We hypothesized that complete ring constructs would be significantly stiffer than incomplete ring constructs and that axial stiffness would increase with increasing wire tension and decrease with increasing ring diameter. We also hypothesized that significant ring deformation would occur in incomplete but not in complete rings with wire tensioning and axial loading.

Materials and Methods

Construct Preparation

Single ring constructs were designed incorporating components manufactured by IMEX Veterinary Inc. (IMEX Veterinary Inc., Longview, Tx). Constructs were made using 50 mm, 66 mm, 84 mm, and 118 mm (inner diameter) complete and incomplete (5/8 circumerference) rings. A 60 mm length of 16 mm diameter Delrin rod (Acetal polymer, MSC Industrial Supply, Melville, New York), was utilized as a bone model.[6,8] Two pilot holes were made in the Delrin rod using a 1.58 mm twist drill bit mounted in a drill press. Pilot holes were oriented transversally to the longitudinal axis of the Delrin rod and had crossing angles analogous to the wire divergence angles utilized for each specific ring diameter. Wire divergence angles utilized were 67.5° (50 mm and 118 mm rings), 72° (66 mm rings), and 60° (84 mm rings). Wire crossing angles varied between ring diameters to allow the Kirschner wires to cross at the center

of each ring. Two, 1.6 mm diameter Kirschner wires were inserted through the pilot holes using a variable speed cordless drill (Bosch 14.4 Volt drill, Robert Bosch LLC, Farmington Hills, MI). Kirschner wires were secured to opposing surfaces of each ring using cannulated fixation bolts. The fixation bolts positioned farthest from the open section of the ring for incomplete rings and the analogous bolts on complete ring constructs were tightened to 10.5 Nm torque[40] using a factory calibrated torque wrench (Craftsman, KCD IP, LLC, Hoffman Estates, IL). Kirschner wires were tensioned simultaneously using two calibrated dynamometric wire tensioners (Smith and Nephew Inc., Memphis, TN). Wire tensions of 0 kg, 30 kg, 60 kg, and 90 kg were utilized. After wire tensioning, the nuts on the fixation bolts adjacent to the tensioners were tightened to 10.5 Nm torque.

Constructs, once assembled, were mounted on a custom fixture for testing. The fixture consisted of a 200 mm x 200 mm x 76 mm aluminum block with a complete base ring, the same diameter as the ring to be tested, firmly bolted one cm off the surface of the aluminum block by means of five, 6 mm diameter threaded rods which screwed into the aluminum block. Two blocks of milled Delrin were secured to the upper surface of the base ring using bolts which were inserted through lower surface of the base ring and screwed into the Delrin blocks. One of the 6 mm threaded rods which attached the complete base ring to the aluminum block was longer than the other rods and protruded three cm above the top of the Delrin blocks (Figure 3-1). Test constructs were mounted to the fixture by passing the protruding section of threaded rod through one of the holes on the side arm of an incomplete ring or through an analogous hole in a complete ring. The construct to be tested was aligned directly over the base ring so that the construct ring rested on the Delrin blocks attached to the base ring. Each construct was fixed in place by tightening nuts on the protruding threaded rod against opposing surfaces of the

construct ring (Figure 3-2). Five replicates of each construct were evaluated, for a total of 160 constructs. New Kirschner wires were utilized for each construct and new rings were used for each construct utilizing incomplete rings. Complete rings without visible evidence of deformation were reused to make subsequent constructs.

Axial Stiffness and Displacement Measurement

The fixture with an attached construct was positioned in a servohydraulic testing system (858 Mini Bionix II, MTS Systems Corp., Eden Prairie, MN). Load was applied axially to the proximal end of the Delrin rod at a rate of 200 N/s to a maximum load of 400 N and utilizing a 5 N preload. Each construct was axially loaded for 15 cycles at a rate of 0.25 Hz. Servohydraulic testing system software recorded axial force (N) and axial actuator displacement (mm) for all testing cycles. Data was collected at a rate of 100 points per second. Load-displacement curves were plotted from the axial force and displacement data for each construct. Data from the fifteenth cycle was utilized for analysis. The secant stiffness was calculated for each load/displacement curve by calculating the slope of a line connecting our preload value of 5 N and our chosen endpoint of 375 N.[72] Maximum displacement was defined as the total axial displacement occurring from 0 N load (prior to preload application) to the displacement at 375 N axial load.

Ring Deformation Measurement

During construct preparation, two measurement points were scribed on each ring using a steel punch. Ring measurement points were scribed at the outer edge of the upper surface of each ring at the location bisected by a plane which passed through the center of the ring, perpendicular to a plane passing through both the center of the ring and the hole through which the threaded rod was inserted to stabilize the ring during testing. A three dimensional digitizing system with an articulating arm (Microscribe, CNC Services Inc., Amherst, VA) was utilized to

record the position of the ring measurement points at specific times during the testing sequence. Measurements were obtained with the digitizing system for each construct before application of tension to fixation wires, after tensioning the fixation wires but prior to construct loading, during maximal axial load application for the fifteenth cycle, after unloading for the fifteenth cycle, and after releasing the tension on fixation wires. The ring measurements were digitally exported to a spreadsheet program (Microsoft Office Excel 2003, Microsoft Corporation, Redmond, WA) as X, Y, and Z coordinates which described the location of the scribed points in space in relation to the digitizing system. Ring measurement data was analyzed using the Euclidean distance formula (Eq. 2-1) to calculate the distance between the two scribed points (outer ring diameter) obtained for each ring during testing.

Data Analysis

Statistical analysis was performed using commercially available statistical software (SPSS 18, IBM Corp. Somers, NY). The effect of ring type and wire tension on axial stiffness and on axial displacement was analyzed by ring size using a multivariate analysis of variance (ANOVA). When significant interactions existed, a univariate ANOVA was utilized to establish significant differences between constructs based on ring type and wire tension. A Sidak's correction was performed to adjust our P value to account for multiple comparisons. This correction decreased the P value for determining significance from our initially selected value of $P \leq 0.05$ to $P \leq 0.013$ for constructs utilizing 50 mm, 66 mm, and 84 mm diameter rings and to $P \leq 0.01$ for constructs utilizing 118 mm diameter rings.

The effect of fixation wire tensioning and construct loading on measured outer ring diameter was analyzed within each construct type using a repeated measures ANOVA. A post hoc Bonferroni correction was utilized when significant differences in ring diameter were identified during a testing cycle. A P value of ≤ 0.05 was considered significant.

Results

Ring Deformation

The 50 mm, 66 mm, and 84 mm diameter incomplete ring constructs failed by catastrophic plastic deformation when we attempted to tension the fixation wires to 90 kg. The ring deformation started as in-plane deformation, resulting in a decrease in the width of the open section of the ring and progressed to catastrophic out-of-plane plastic deformation resulting in folding of the ring and loss of wire tension (Figure 3-3). We did not test any of the 50 mm, 66 mm, or 84 mm diameter ring constructs with the fixation wires tensioned to 90 kg because we could only effectively tension the fixation wires to 90 kg in 118 mm diameter incomplete ring constructs.

Ring diameter of complete ring constructs was nominally and not consistently affected by wire tensioning or construct loading, nor was a permanent change in ring diameter induced in any complete ring construct as a result of wire tensioning or construct loading. A consistent decrease was noted in the diameter of the incomplete ring constructs associated with wire tensioning and axial loading. Deformation was greater at higher wire tensions and in larger diameter constructs. Application of 60 kg of wire tension to 66 mm diameter incomplete ring constructs and 90 kg of wire tension to 118 mm incomplete ring constructs resulted in a significant residual decrease in ring diameter of 1% and 5%, respectively, after fixation wire tension was released, at the end of testing (Table 3-1, Figure 3-4).

Axial Stiffness

All load-displacement curves displayed a characteristic initial exponential increase in stiffness with the slope of the curves becoming more linear as the applied load increased. This initial phase of exponential increase in stiffness was more prolonged in the load-displacement curves of constructs utilizing incomplete rings than in comparable diameter and similarly

tensioned complete ring constructs. Decreasing ring diameter as well as increasing wire tension resulted in load-displacement curves with a more linear appearance (Figure 3-4). Significant differences in construct stiffness were noted between similar constructs utilizing incomplete versus complete rings. Axial stiffness increased with sequential tensioning of the fixation wires, with the increase being significant in most incomplete ring constructs and some complete ring constructs (Table 3-2).

Maximum Displacement

All constructs exhibited an inverse relationship between displacement at 375 N and fixation wire tension. This relationship was more obvious in larger diameter rings. Displacement was significantly less in complete ring constructs than comparable diameter incomplete ring constructs. Maximum displacement decreased as a result of sequential wire tensioning in both complete and incomplete ring constructs. The decreases in displacement were significant in most incomplete ring constructs and some complete ring constructs (Table 3-3).

Discussion

The nonlinear load-displacement behavior of both the complete and incomplete ring constructs observed in this study is consistent with the results of previous studies evaluating the axial stiffness characteristics of circular fixators.[5-8,53] Axial elasticity is a property inherent to circular fixators utilizing small diameter wires as fixation elements.[5-7] This axial elasticity allows for controlled axial micromotion of the secured bone segments which is purported to provide an environment conducive to bone healing.[6,49,53,54,68] The precise amount of axial micromotion which optimizes bone healing is unknown, but likely varies from case to case and is dependent on a number of variables, including species, the configuration and location of the fracture or osteotomy, as well as the patient's body weight, age and activity level.[57]

Effect of Wire Tensioning on Construct Performance

Tensioning of the fixation wires is performed to increase construct stiffness and prevent excessive motion of the secured bone segments during weight-bearing.[6,7,39] As expected, we found that axial stiffness increased and axial displacement decreased as wire tension was sequentially increased in both the complete and incomplete ring constructs. While the differences were not always significant, incomplete ring constructs consistently had a lower stiffness and allowed greater displacement than comparable diameter and similarly tensioned complete ring constructs. The lower stiffness and increased displacement of the incomplete ring constructs when compared to analogous complete ring constructs is the result of deformation of the incomplete rings which occurred during wire tensioning and subsequent loading.[53,71] The decrease in axial displacement associated with increasing wire tension highlights the benefit of fixation wire tensioning, even in constructs in which fixation wire tensioning does not increase construct stiffness.

A preload of 5 N was utilized for all constructs in this study. Axial displacement of the Delrin rod occurred when applying this preload and was greater in constructs utilizing larger diameter rings and lower fixation wire tensions. The preload resulted in a "self tensioning effect" on the fixation wires and negated differences in the initial toe region of the load-displacement curve between similar diameter complete ring constructs with different fixation wire pretension values. Similar findings have been reported previously in studies evaluating the effects of fixation wire tension on construct stiffness.[5,49,53,73]

Ring Deformation

Our objective measurements of ring diameter showed that all incomplete rings and 118 mm diameter complete rings temporarily deformed when the fixation wires were tensioned and/or the construct was loaded. Ring diameter returned to a value statistically similar to initial

ring diameter once fixation wire tension was released in all constructs, with the exception of the 66 mm diameter incomplete ring constructs which had been tensioned to 60 kg and 118 mm diameter incomplete ring constructs which had been tensioned to 90 kg (Table 3-1, Figure 3-3). The clinical relevance of the elastic ring deformation noted in our constructs associated with application of tension to fixation wires and construct loading is unknown.[71] We would suggest that permanent ring deformation may decrease construct performance[53] and should probably be avoided if possible. Deformation leading to catastrophic ring failure as observed in our 50 mm, 66 mm, and 84 mm incomplete rings when we attempted to tension fixation wires to 90 kg is of academic interest but is not clinically relevant as wires are not typically tensioned to 90 kg when using these diameter rings.[7]

Wire Tensioning Guidelines

Based on the data obtained in this study some basic guidelines can be developed regarding appropriate tensioning of wires on incomplete rings. Tensioning of fixation wires in 50 mm diameter incomplete ring constructs didn't increase construct stiffness. 50 mm diameter incomplete ring constructs, had greater stiffness and less axial displacement than any of the constructs utilizing larger diameter rings, irrespective of what tension was applied to the fixation wires. Thus tensioning wires attached to 50 mm diameter incomplete rings is unnecessary, which is consistent with the recommendation based on previous biomechanical studies not to tension wires attached to 50 mm diameter complete rings.[6,7] Tensioning wires to 60 kg in constructs utilizing 66 mm diameter incomplete rings increased stiffness but caused permanent deformation of the rings. Based on biomechanical studies and clinical experience, wires on 66 mm diameter complete rings are usually not tensioned or are tensioned to 30 kg.[7,31,74] We would advocate that based on the results of this study, wires attached to 66 mm diameter incomplete rings should not be tensioned more than 30 kg. Stiffness of 84 mm diameter

incomplete ring constructs was significantly increased by increasing wire tension to 60 kg and this wire tension did not result in significant residual ring deformation. It has previously been recommended that wires attached to 84 mm diameter complete rings should be tensioned to 60 kg or 90 kg.[7,8,17,74] Catastrophic ring failure occurred when we attempted to tension the fixation wires to 90 kg in the 84 mm diameter incomplete ring constructs and thus we would recommend tensioning wires on 84 mm diameter incomplete rings to 60 kg. Constructs utilizing 118 mm diameter incomplete rings could be made significantly stiffer by tensioning wires to 60 kg. Tensioning of wires to 90 kg resulted in permanent deformation of the 118 mm diameter incomplete rings and there was no significant increase in axial stiffness when wire tension was increased from 60 kg to 90 kg. The current recommendation for 118 mm diameter complete rings is to tension fixation wires to 90 kg,[7,74] but we would recommend that wires on 118 mm diameter incomplete rings should not be tensioned beyond 60 kg.

Limitations

This study has several limitations. We designed our testing jig so that constructs were only constrained at a single fixation hole. The remainder of the construct was supported by two Delrin blocks. Delrin has a low coefficient of friction and allowed the construct to deform during axial loading. By constraining the ring at a single hole we replicated connecting elements typical of a type IA hybrid construct. Ring deformation would be more restricted if an incomplete ring was used in a traditional circular fixator construct which would typically have multiple connecting rods positioned around the circumference of the ring.

The digitizing system used to measure ring diameter in this study has a manufacturer reported accuracy of 0.2 mm. The accuracy of the digitizing system limited our ability to detect very small changes in ring diameter which might have been apparent had a measuring system with a greater accuracy been utilized.

Traditional circular fixator constructs typically consist of three or four rings, preferably with two or more rings stabilizing each major bone segment. We tested only single ring constructs, but did so to isolate differences between complete and incomplete rings.[5,6,51] We only tested our constructs in axial compression. Further studies evaluating the biomechanical differences between constructs utilizing complete and incomplete rings subjected to bending, shear, and torsional loading should be performed to more fully elucidate the forces affecting fractures and osteotomies stabilized with circular and hybrid fixator constructs. Only five replicates of each construct were tested. This minimal number of constructs per group may have hidden significant differences between constructs which would have been evident had more constructs been tested.

We utilized a single diameter of Delrin rod as our bone model, irrespective of ring diameter, to allow us to make discreet comparisons between constructs of different diameters, a precedent established in previous studies.[5-7] Altering the diameter of Delrin rod for each ring diameter to approximate the diameter of the bone segment that each ring diameter would be applied to in a clinical setting would likely yield different results.[7]

Summary

The results of this study document that incomplete ring constructs subjected to axial loading exhibit non-linear load-displacement behavior, similar to that of complete ring constructs.[6,7] In support of our initial hypotheses, this study demonstrated that complete ring constructs are stiffer than analogous incomplete ring constructs and the differences become more noticeable with increasing ring diameter. For both complete and incomplete ring constructs, axial stiffness increases with increasing fixation wire tension and decreases as ring diameter is increased. Incomplete rings deform to a much greater extent than complete rings as a result of fixation wire tensioning and axial loading, although the clinical significance of this ring

deformation is unknown. Based on our findings we recommend that wire tensioning in incomplete ring constructs should not exceed 30 kg in 66 mm diameter rings and 60 kg in 84 mm and 118 mm diameter rings. Future studies designed to evaluate the biomechanical properties of incomplete rings when loaded in bending and torsion are warranted.

Table 3-1. Ring deformation of each construct type in all four modes of loading

Ring diameter	Wire tension	Ring type	Pre-wire tensioning	Pre-loading	Under load cycle 15	Post-loading	Wire tension released
50 mm	0 kg	Complete	76.1 ± 0.3[a]	76.3 ± 0.5[a]	76.0 ± 0.5[a]	76.1 ± 0.6[a]	76.2 ± 0.5[a]
		Incomplete	76.4 ± 0.4[a]	76.3 ± 0.4[a]	76.2 ± 0.3[a]	76.4 ± 0.3[a]	76.4 ± 0.3[a]
	30 kg	Complete	76.0 ± 0.5[a]	76.1 ± 0.5[a]	75.9 ± 0.5[a]	76.2 ± 0.4[a]	76.2 ± 0.5[a]
		Incomplete	76.3 ± 0.4[a]	75.8 ± 0.5[b]	75.7 ± 0.3[b]	75.9 ± 0.4[b]	76.4 ± 0.3[a]
	60 kg	Complete	76.1 ± 0.2[a]	76.0 ± 0.4[a]	76.0 ± 0.2[a]	76.0 ± 0.3[a]	75.9 ± 0.4[a]
		Incomplete	76.3 ± 0.5[a]	75.1 ± 0.5[b]	75.0 ± 0.5[b]	75.1 ± 0.5[b]	76.1 ± 0.5[a]
66 mm	0 kg	Complete	91.9 ± 0.3[a]	91.9 ± 0.4[a]	91.8 ± 0.4[a]	91.9 ± 0.3[a]	91.8 ± 0.3[a]
		Incomplete	92.5 ± 0.4[a]	92.4 ± 0.3[a]	91.7 ± 0.3[b]	92.4 ± 0.3[a]	92.2 ± 0.3[a]
	30 kg	Complete	92.2 ± 0.4[a]	92.2 ± 0.4[a]	92.3 ± 0.5[a]	92.2 ± 0.4[a]	92.0 ± 0.6[a]
		Incomplete	92.3 ± 0.5[a]	91.3 ± 0.5[b]	91.1 ± 0.2[b]	91.5 ± 0.3[b]	92.2 ± 0.4[a]
	60 kg	Complete	91.9 ± 0.2[a]	91.9 ± 0.4[a]	91.9 ± 0.3[a]	91.9 ± 0.2[a]	91.8 ± 0.3[a]
		Incomplete	92.4 ± 0.6[a]	90.0 ± 0.4[b]	89.6 ± 0.6[b]	90.0 ± 0.6[b]	91.5 ± 0.6[c]
84 mm	0 kg	Complete	110.4 ± 0.4[a]	110.5 ± 0.4[a]	110.2 ± 0.4[a]	110.4 ± 0.3[a]	110.3 ± 0.5[a]
		Incomplete	110.8 ± 0.4[a]	110.8 ± 0.4[a]	109.6 ± 0.5[b]	110.7 ± 0.4[a]	110.8 ± 0.4[a]
	30 kg	Complete	110.3 ± 0.3[a]	110.0 ± 0.2[a]	109.8 ± 0.3[a]	110.1 ± 0.3[a]	110.1 ± 0.3[a]
		Incomplete	110.8 ± 0.3[a]	109.5 ± 0.3[b]	108.8 ± 0.2[c]	109.5 ± 0.2[b]	110.8 ± 0.2[a]
	60 kg	Complete	109.9 ± 0.4[a]	109.8 ± 0.4[a]	109.7 ± 0.3[a]	109.8 ± 0.4[a]	109.7 ± 0.4[a]
		Incomplete	111.0 ± 0.2[a]	108.3 ± 0.3[b]	107.9 ± 0.4[c]	108.4 ± 0.4[b]	110.5 ± 0.5[a]
118 mm	0 kg	Complete	144.3 ± 0.3[a]	144.3 ± 0.3[a]	144.0 ± 0.2[a]	144.2 ± 0.2[a]	144.1 ± 0.4[a]
		Incomplete	144.6 ± 0.5[a]	144.4 ± 0.4[a]	142.0 ± 0.4[b]	144.3 ± 0.4[a]	144.5 ± 0.4[a]
	30 kg	Complete	144.3 ± 0.4[a]	144.0 ± 0.5[b]	144.0 ± 0.4[b]	144.2 ± 0.4[ab]	144.2 ± 0.4[ab]
		Incomplete	144.3 ± 0.7[a]	141.8 ± 0.5[b]	140.6 ± 0.6[c]	141.8 ± 0.6[b]	144.4 ± 0.5[a]
	60 kg	Complete	144.4 ± 0.5[a]	144.2 ± 0.5[a]	144.2 ± 0.4[a]	144.3 ± 0.4[a]	144.2 ± 0.6[a]
		Incomplete	144.8 ± 0.5[a]	140.2 ± 0.4[b]	139.3 ± 0.5[c]	140.2 ± 0.4[b]	144.2 ± 0.6[a]
	90 kg	Complete	144.3 ± 0.4[a]	144.1 ± 0.2[b]	143.9 ± 0.4[b]	144.2 ± 0.2[ab]	144.1 ± 0.3[ab]
		Incomplete	147.4 ± 2.4[a]	135.6 ± 0.6[b]	134.8 ± 0.8[c]	135.4 ± 0.6[bc]	140.9 ± 0.6[d]

Values are mean ± standard deviation (mm) for outer ring diameter measured during the testing sequence. Statistical comparisons are between cycle phases (along rows) within a single construct type. Significant differences are designated by different letter superscripts, $P \leq 0.05$.

Table 3-2. Construct axial stiffness

Ring diameter	Ring type	0 kg wire tension	30 kg wire tension	60 kg wire tension	90 kg wire tension
50 mm	Incomplete	190.5 ± 11.7[a]	218.8 ± 16.4[a]	223.7 ± 32.8[a]	N/A
	Complete	205.9 ± 21.2[b]	230.5 ± 58.3[b]	232.1 ± 26.4[b]	N/A
66 mm	Incomplete	108.7 ± 2.2[c]	118.8 ± 15.4[c]	142.4 ± 13.9[d]	N/A
	Complete	153.7 ± 9.5[e*]	153.7 ± 11.8[e*]	176.1 ± 15.9[f*]	N/A
84 mm	Incomplete	78.8 ± 2.5[g]	89.1 ± 1.6[h]	101.0 ± 6.8[i]	N/A
	Complete	123.9 ± 5.1[j*]	124.0 ± 1.1[j*]	135.0 ± 11.9[j*]	N/A
118 mm	Incomplete	45.0 ± 2.1[k]	51.2 ± 1.3[k]	61.3 ± 3.1[l]	68.4 ± 7.7[l]
	Complete	88.2 ± 3.0[m*]	85.3 ± 2.0[m*]	87.8 ± 3.9[m*]	89.4 ± 9.0[m*]

Axial stiffness at 0 kg, 30 kg, 60 kg, and 90 kg wire tension arranged by construct ring size and ring type. Values are mean ± standard deviation axial secant stiffness (N/mm) for all construct groups tested. Significant differences between wire tensions within a single construct type (along a row) are indicated by different superscript letters. Significant differences between complete and incomplete ring constructs of the same diameter are indicated by the presence of an asterisk in the complete ring column ($P \leq 0.013$ for 50 mm, 66 mm, and 84 mm diameter constructs, $P \leq 0.01$ for 118 mm diameter constructs). N/A indicates that values were not assessed.

Table 3-3. Construct axial displacement

Ring diameter	Ring type	0 kg wire tension	30 kg wire tension	60 kg wire tension	90 kg wire tension
50 mm	Incomplete	2.9 ± 0.3^{a}	2.2 ± 0.2^{ab}	2.1 ± 0.4^{b}	N/A
	Complete	$2.1 \pm 0.3^{c*}$	2.0 ± 0.6^{c}	2.0 ± 0.3^{c}	N/A
66 mm	Incomplete	4.3 ± 1.0^{d}	3.9 ± 0.7^{d}	3.0 ± 0.2^{d}	N/A
	Complete	3.1 ± 0.2^{e}	2.9 ± 0.2^{ef}	2.6 ± 0.2^{f}	N/A
84 mm	Incomplete	6.1 ± 1.1^{g}	4.8 ± 0.7^{gh}	4.2 ± 0.5^{h}	N/A
	Complete	$3.9 \pm 0.6^{i*}$	$3.4 \pm 0.4^{i*}$	$3.1 \pm 0.5^{i*}$	N/A
118 mm	Incomplete	9.9 ± 1.3^{j}	7.8 ± 0.6^{k}	6.7 ± 0.8^{kl}	5.7 ± 0.6^{l}
	Complete	$5.8 \pm 1.3^{m*}$	$5.2 \pm 0.1^{m*}$	$5.0 \pm 0.6^{m*}$	4.7 ± 0.6^{m}

Axial displacement at 0 kg, 30 kg, 60 kg, and 90 kg wire tension arranged by construct ring size and ring type. Values are mean ± standard deviation for axial displacement (mm) at 375 N axial load. Significant differences between wire tensions within a single construct type (along a row) are indicated by different superscript letters. Significant differences between complete and incomplete ring constructs of the same diameter are indicated by the presence of an asterisk in the complete ring column ($P \leq 0.013$ for 50 mm, 66 mm, and 84 mm diameter constructs, $P \leq 0.01$ for 118 mm diameter constructs). N/A indicates that values were not assessed.

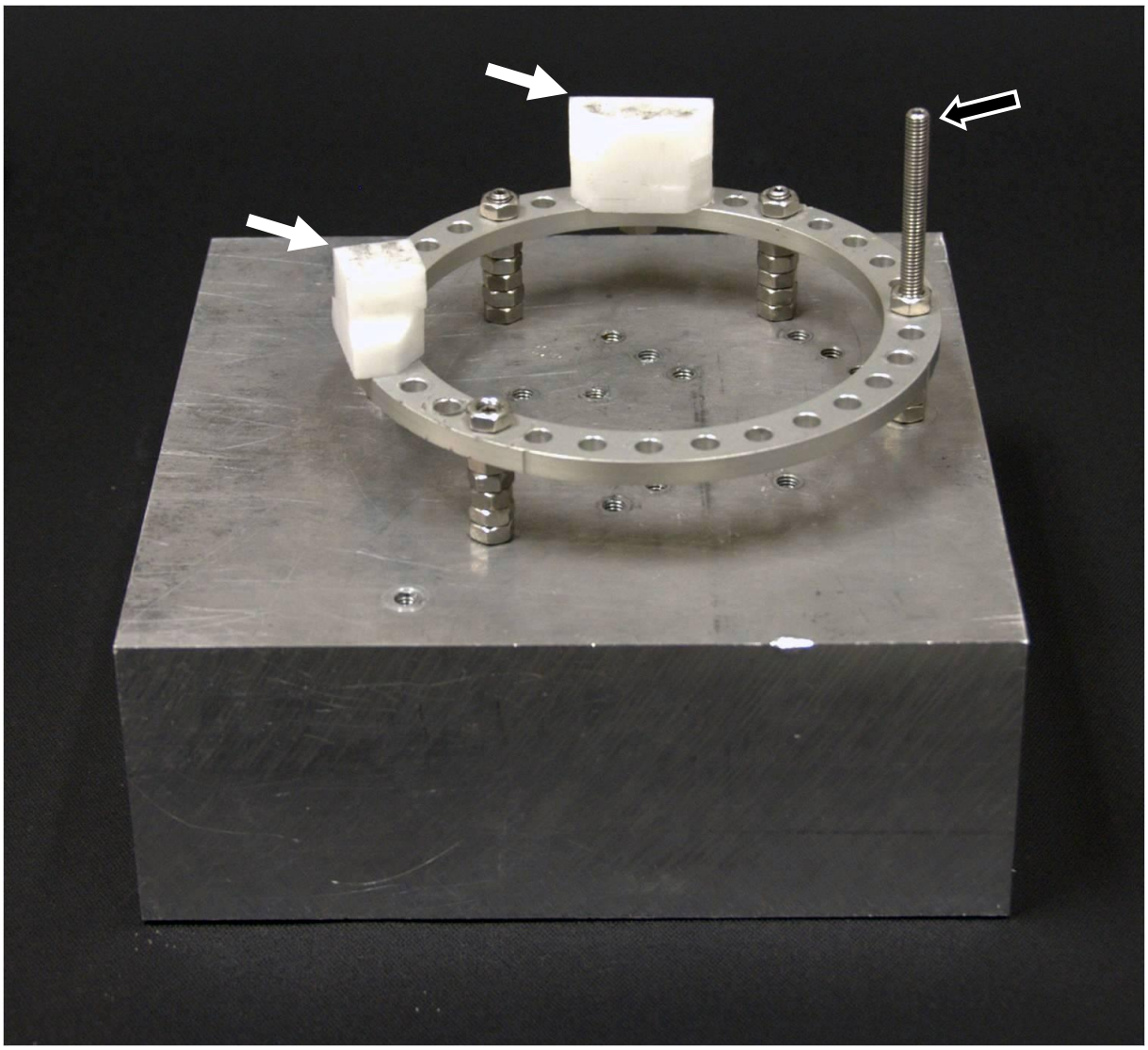


Figure 3-1. Testing jig consisting of an aluminum block with a complete base ring bolted to the surface of the block. Two delrin spacers (solid white arrows) and a segment of threaded rod (white outline arrow) support the test construct.

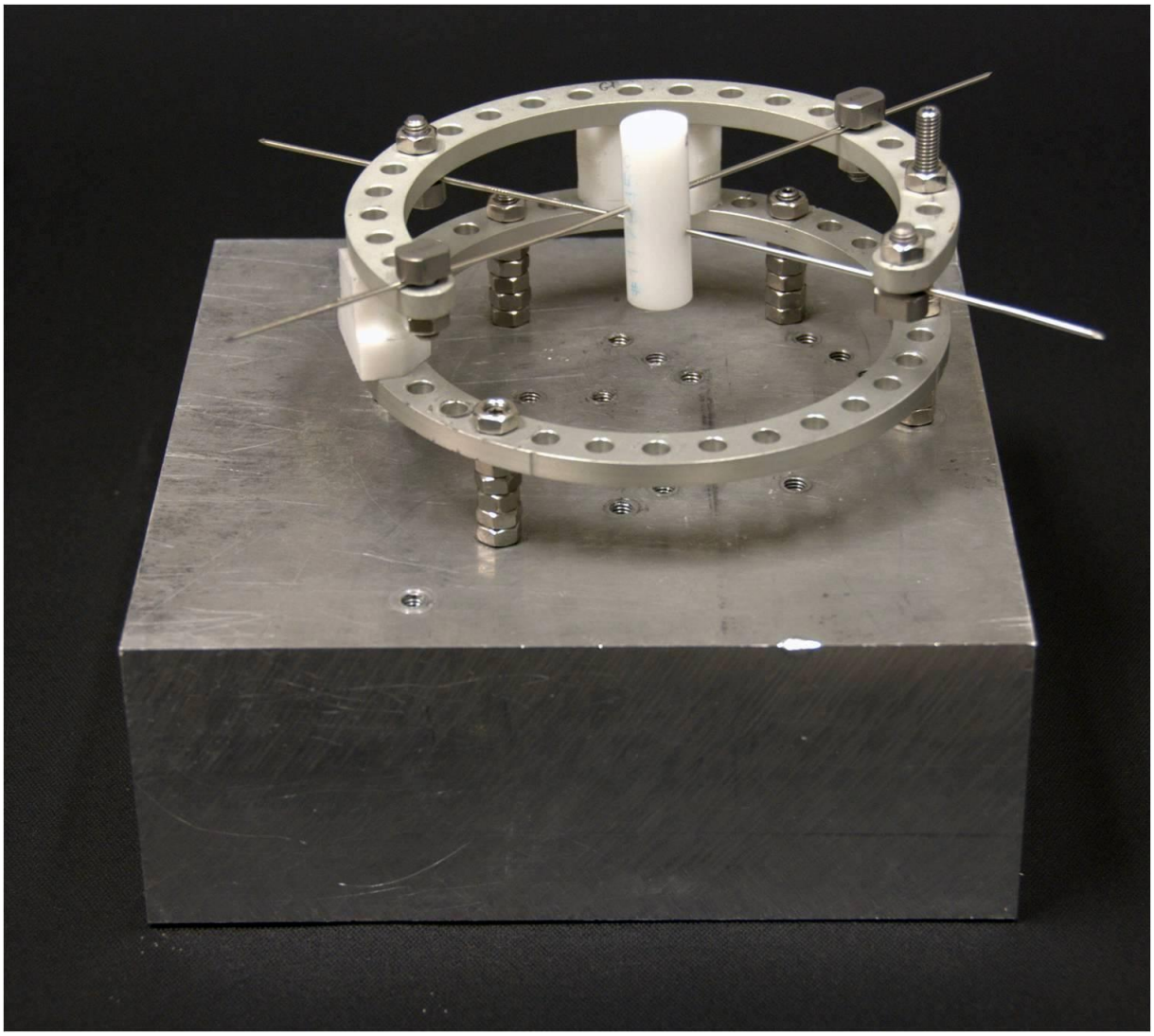


Figure 3-2. Testing jig with a 118 mm diameter incomplete ring construct secured in place prior to axial loading.

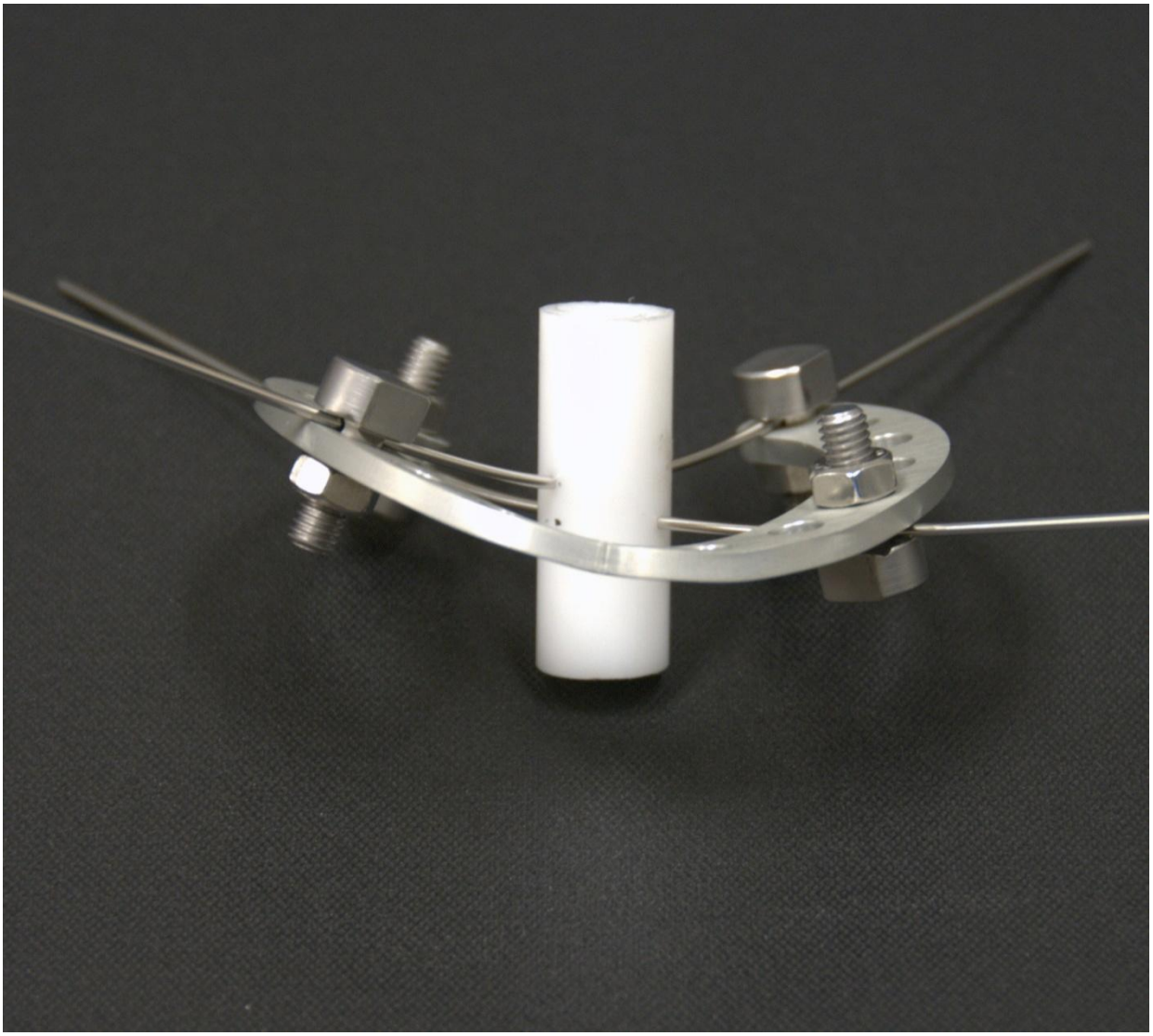


Figure 3-3. Ring failure in a 66 mm diameter incomplete ring construct which occurred during an attempt to tension fixation wires to 90 kg. Note the out of plane deformation (folding) which was characteristic of the failure pattern noted in all failed 50 mm, 66 mm and 84 mm diameter incomplete ring constructs.

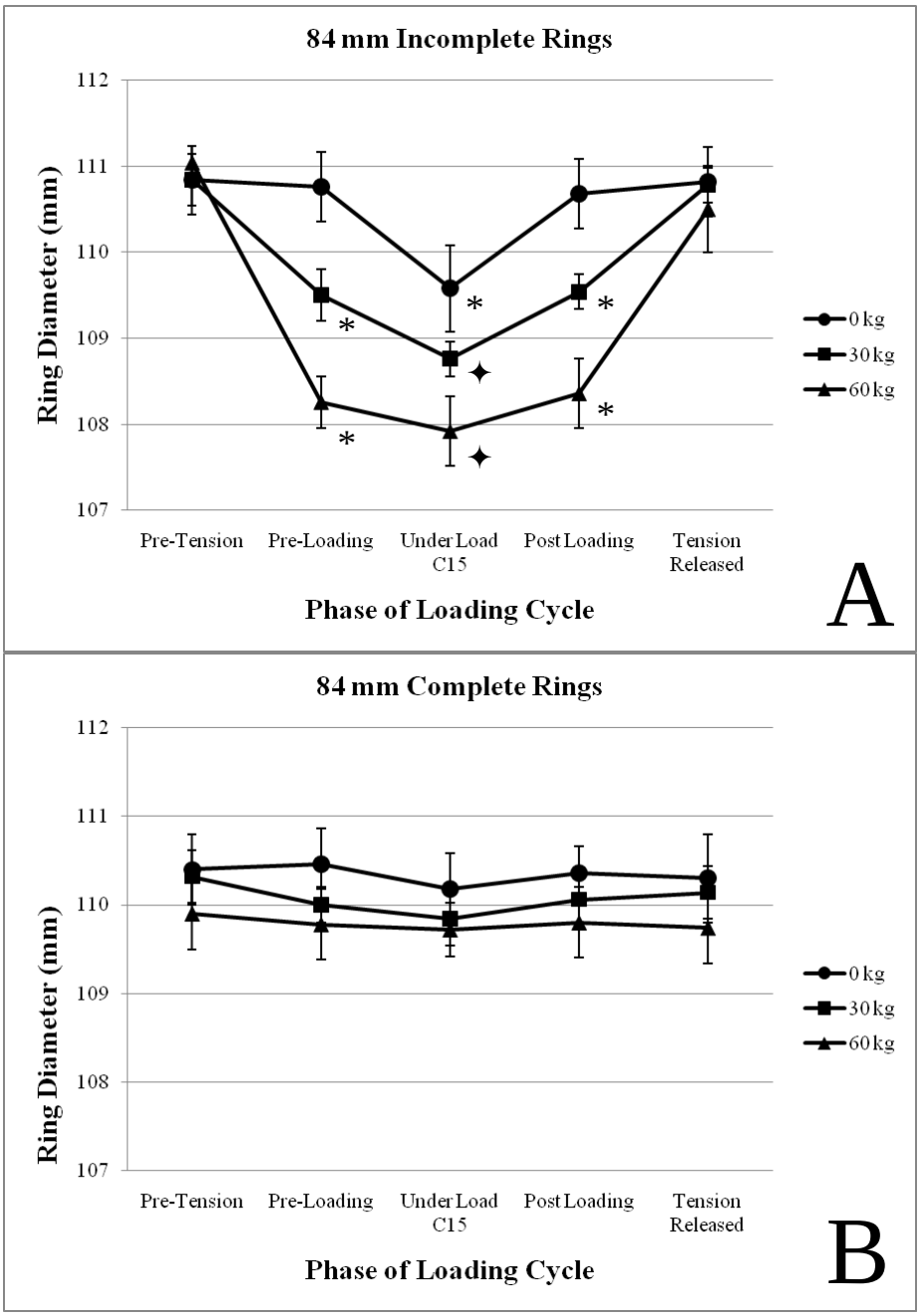
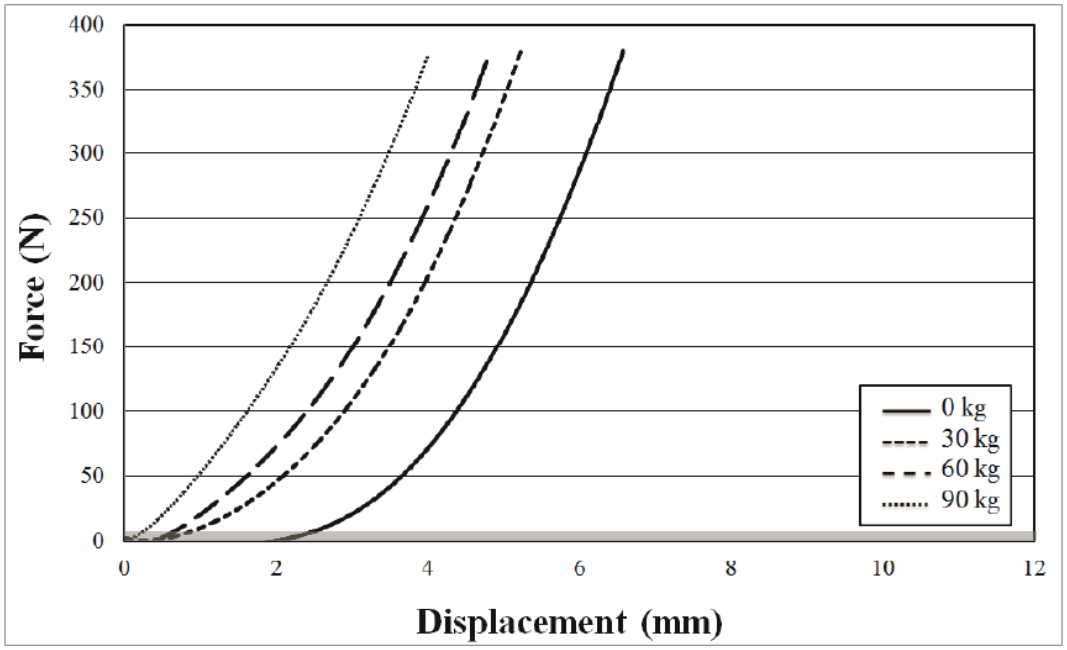


Figure 3-4. Ring deformation during the testing sequence. Mean with standard deviation bars of measured outer ring diameter (mm) in A) 84 mm incomplete and B) 84 mm complete ring constructs. Pre-tension is before fixation wire tensioning. Pre-loading is after fixation wire tensioning but before construct loading. Under Load C15 is while the test construct is loaded on cycle 15. Post Loading is once axial load is released after cycle 15. Tension Released is after fixation wire tension is released, post loading. Statistical comparisons are between measurements within a single wire tension. Presence of an * or ✦ indicates significant differences ($P \leq 0.05$) between measured diameters.



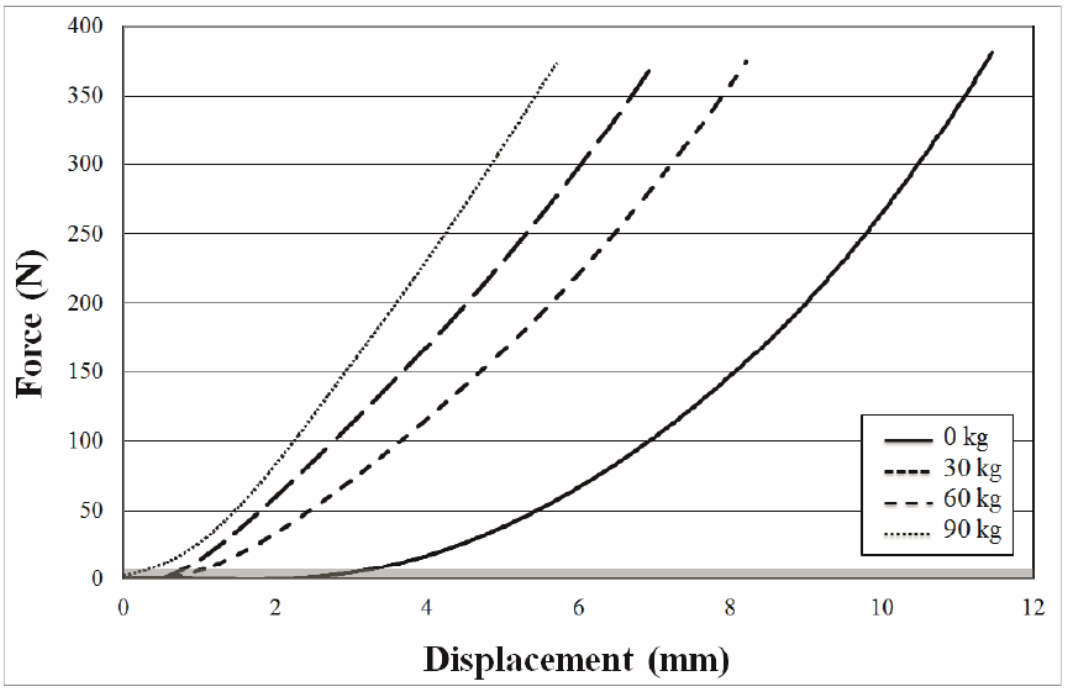


Figure 3-5. Construct load-displacement curves. A) complete and B) incomplete 118 mm diameter ring constructs at all wire tensions. The gray bar at the bottom of each graph represents the 5 N preload applied to all constructs. Overall the slope of the load-displacement curves increases more rapidly in the complete ring constructs compared to the incomplete ring constructs. In both complete and incomplete ring constructs, increasing fixation wire tension shifts the load-displacement curve to the left, resulting in less displacement for any given axial force value.

CHAPTER 4
A BIOMECHANICAL COMPARISON OF THREE HYBRID LINEAR-CIRCULAR EXTERNAL FIXATOR CONSTRUCTS

Background

The Russian physician Gavril Ilizarov pioneered the development of circular external skeletal fixation.[38,41,70] Circular fixators consist of a series of extra-corporeal rings, which serve as the supporting elements of the frame.[39,41,52] The rings are articulated by a series of threaded rods that serve as the frame connecting elements.[42,52] Small diameter wires are utilized as the fixation elements to stabilize the secured bone segments.[4,6,39,43] The fixation wires, which are generally tensioned to improve construct stiffness, allow axial micro-motion of the secured bone segments, which enhances bone healing.[39,53,54,68] The application of circular external skeletal fixation and the methods of Ilizarov in veterinary medicine was first reported by the Italian veterinary surgeon Antonio Ferreti over 20 years ago[75] and the clinical use of circular fixators in dogs and cats is becoming more common.[19,28,30,39,45,47,48]

Hybrid external skeletal fixation was developed to simplify the application and address some of the inherent limitations associated with the use of traditional circular fixators.[42] In human orthopedics, the term “hybrid fixator” is used to describe any circular fixator construct which also incorporates fixation pins as fixation elements.[42] In veterinary surgery, the term “hybrid fixator” typically refers to a linear-circular hybrid fixator construct which utilizes one or more rings as supporting elements, but the ring or rings are articulated with one or more hybrid (linear) connecting rods.[4,20,37] The rods are secured into the holes in the ring.[20,26,37,74] Specifically manufactured hybrid connecting rods have a smooth shaft that accommodates pin fixation clamps. Threads are milled into one end of the hybrid rod which allows the rod to be secured in one of the holes of a ring component.[20,26,74] Half- and full-pin fixation pins are used as the fixation elements secured to the hybrid rods.[32,74] Small diameter wires are traditionally

used as fixation elements attached to the rings; however, half- and full-pin fixation pins can also be used as fixation elements on the ring components.[20,26,74]

Hybrid fixators are being used with increased frequency in small animal practice for the management of fractures[20,22,26] and the correction of limb deformities;[29,32] however, there is surprisingly little information available regarding the biomechanics of these systems.[18]

The purpose of this study was to perform a biomechanical analysis of three linear-circular hybrid construct configurations which are used commonly in dogs and cats.[20,22,26,29,32] Specifically, the study evaluated the effect of adding a diagonal strut and varying fixation pin position on the stiffness of hybrid constructs loaded in axial compression, four point bending, and axial torsion. Motion of bone model segments at the simulated fracture gap and ring deformation which occurred as a result of loading were also evaluated. We hypothesized that the addition of a diagonal strut and orthogonal placement of the fixation pins would increase construct stiffness in all modes of loading evaluated.

Materials and Methods

Construct Assembly

Three configurations of linear-circular hybrid fixator constructs were constructed using components manufactured by IMEX Veterinary Inc. (IMEX Veterinary Inc., Longview, Tx). Constructs incorporated 84 mm internal diameter incomplete (5/8 circumference) aluminum rings, 6.3 mm diameter 150 mm long titanium hybrid rods, 1.6 mm diameter 200 mm long stainless steel fixation wires with stoppers (olive wires), stainless steel cannulated/slotted wire fixation bolts, 10 mm stainless steel nuts, stainless steel single hole posts, 3.2 mm diameter stainless steel, positive profile, partially (end) threaded fixation pins, and small (SK™) external fixator pin clamps. Delrin rod (Acetal polymer, MSC Industrial Supply, Melville, New York) 22 mm in diameter and 250 mm in length was utilized as a bone model.

Anatomic axes were defined relative to application of the hybrid constructs on the Delrin rods. The craniocaudal axis was defined as the axis which bisected the hole in the center of the closed end of the ring and passed through the center of the incomplete section of the ring. The mediolateral axis was defined as the axis which bisected the ring at 90° to the craniocaudal axis. The proximodistal axis was defined as the axis which bisected the previous two axes and parallels the shaft of the bone model (Figure 4-1).

A milling machine (XLT Ramill, Ramco, Taiwan Machine Tool Import Division) was used to prepare the Delrin rod bone models. Pilot holes were pre-drilled in the Delrin rod to facilitate accurate placement of all fixation wires (pilot holes 1.6 mm) and pins (pilot holes 2.8 mm). Four additional 7.9 mm diameter holes centered 15 mm from each end of each Delrin rod (220 mm apart) were drilled in the craniocaudal and mediolateral planes for suspending the construct during application of bending forces. A 10 mm segment of the Delrin rod was removed to simulate a fracture gap. This resulted in two asymmetric segments: An 82 mm long segment of Delrin rod which was designated the “distal” segment and a 158 mm long segment of Delrin rod which was designated the “proximal” segment.

All fixation wires and pins were inserted in the pre-drilled pilot holes using a commercially available cordless drill (Bosch 14.4 Volt drill, Robert Bosch LLC, Farmington Hills, MI). Stabilization of the distal segment was consistent for all construct configurations: The distal Delrin rod segment was stabilized with a ring and two fixation wires. Fixation wires had a crossing angle of 60° bisecting the mediolateral plane and were attached directly to opposing surfaces of the ring using the cannulated hole in the fixation bolts. All three construct configurations also had a single, primary hybrid rod, bolted to the ring through the hole bisected by the mediolateral axis and designated to be the medial side of the ring. Constructs were

designated Ia, Ia_d, and Ib based on the configuration of the proximal, linear portion of the construct. Construct Ia had three fixation pins inserted from the medial side of the proximal segment of the Delrin rod and then secured to the primary hybrid rod with three external fixator pin clamps. The fixation pins were spaced 3 cm apart. The distal pin was placed one cm from the simulated fracture gap (Figure 4-2). Construct Ia_d was similar to construct Ia except that an additional secondary hybrid rod traversed diagonally from the proximal end of the primary hybrid rod craniolaterally to the closed end of the ring. The secondary rod was secured to the primary hybrid rod with a double external fixator clamp and to the ring using two, articulated one-hole posts and three nuts (Figure 4-3). Construct Ib had two fixation pins separated by 6 cm inserted from the medial side into the proximal segment of Delrin rod which were secured to the primary hybrid rod. A secondary hybrid rod was attached in a craniolateral position on the closed end of the ring using two, articulated one-hole posts and three nuts and was directed diagonally towards the proximal end of the primary hybrid rod. A single fixation pin was inserted in the cranial surface of the proximal Delrin rod segment, midway between the two medially placed fixation pins and secured to the diagonal secondary hybrid rod (Figure 4-4).

Fixation wires on each construct were simultaneously tensioned to 60 kg using two calibrated dynamometric wire tensioners (Smith and Nephew Inc., Memphis, TN). The tensioners were calibrated using the load cell of a servohydraulic testing system (858 Mini Bionix II, MTS Systems Corp., Eden Prairie, MN). A factory calibrated torque wrench (Craftsman, KCD IP, LLC, Hoffman Estates, IL) was used to tighten all of the nuts and bolts to 10.5 Nm with the exception of the external fixator clamps, to which a torque of 8 Nm was applied.[3,40] All constructs were assembled prior to initiation of mechanical testing.

Construct Mechanical Testing

Mechanical testing was performed utilizing the servohydraulic testing system operated under displacement control. Eight replicates of each of the three construct configurations were tested for a total of 24 constructs. Each construct replicate was tested in four modes of loading: axial compression, craniocaudal bending, mediolateral bending, and axial torsion. Each construct was loaded through 10 cycles performed at 0.25 Hz for each mode of loading.

Axial compression was performed by placing the constructs in a custom jig (Figure 4-5A) and loading each construct at a rate of 2.5 mm/second to a maximal displacement of 5 mm. Load-displacement curves were generated by plotting the axial load applied (N) against the cross head displacement (mm).

Mediolateral and craniocaudal bending testing was performed by suspending each construct horizontally in a custom four point bending jig using pins placed through holes located 15 mm from each end of the Delrin rod. Force was applied symmetrically at points located 25 mm central to each of the construct suspension pins (Figure 4-5B). Bending force was applied at a rate of 1 mm/second to a maximal cross head displacement of 2 mm. Bending moment was calculated as the applied load (N) multiplied by the moment arm length (m). Angular displacement on each side of the fracture gap was calculated using trigonometry. A load deformation curve was generated by plotting the bending moment (Nm) against the angular displacement (degrees).

Torsional testing was performed by securing the constructs in a custom jig and applying torsional force to the proximal segment while restraining the distal segment of the Delrin rod (Figure 4-5A). Torsional force was applied at a rate of 5°/second to a maximum angular displacement of 10°. A load displacement curve was generated by plotting the applied torque (Nm) against the displacement (degrees).

For all modes of loading, stiffness was defined as the slope of the linear portion of the load/displacement curve. The linear portion of the curve was designated as the segment of the curve which could be described using a linear equation with a coefficient of determination (r^2) value greater than or equal to 0.9985. For load-displacement curves with a triphasic slope, the middle, more linear phase of the curve, was utilized for the stiffness calculation.

Ring Deformation and Bone Model Motion

Prior to initiation of testing, two points were scribed on each ring, using a steel punch, to serve as measurement points. The two points were located symmetrically at the outer edge of the proximal surface of each ring at the point bisected by the mediolateral axis. Eight measurement points were scribed on the Delrin rod, four on the proximal segment and four on the distal segment. Points on the Delrin rod were located segmentally along an arc which encircled the circumference of the Delrin rod, 4.5 mm from the end of each rod adjacent to the fracture gap. Measurements were collected from these scribed points at specified times in the testing sequence, using the articulating arm of a three dimensional digitizing unit (Microscribe, CNC Services Inc., Amherst, VA). The three dimensional digitizing unit records points in space and exports the data as X, Y, and Z coordinates, based on a reference coordinate system located at the base of the unit. Measurements were collected prior to the application of tension to the olive wires, after tensioning the olive wires but prior to load application, under maximum load during cycle 10; after the load was released following cycle 10; and after releasing the tension on the olive wires. Coordinates from the two ring points bisected by the mediolateral axis were analyzed using the Euclidean distance formula (Eq. 2-1) which allowed calculation of the outer ring diameter at each sequential point in the testing cycle for which measurements were collected. Coordinates from the four arc points located on each of the bone model segments were used to define the anatomic coordinate system. The origin positions were determined as the

centers of two best-fit circles which were fit to the arc points and then translated to the ends of each bone model at the simulated fracture gap. Craniocaudal, mediolateral and proximodistal directions were established as previously described. The position and orientation of the two bone model ends were calculated at construct preloading and then again with the construct under load during the tenth cycle for each construct in each mode of loading. The relative change in the bone models' position, from preloading to under maximum load during the tenth cycle, for each construct was expressed as a rotation in degrees around and a translation in mm along each of the three anatomic axes.

Data Analysis

Statistical analysis was performed using commercially available statistical software (SPSS 18, IBM Corp. Somers, NY). Stiffness from the tenth loading cycle for each mode of loading was compared between construct configurations utilizing a multivariate ANOVA with a post hoc bonferroni correction. A repeated measures ANOVA with a post hoc Bonferroni correction was used to evaluate the effect of fixation wire tensioning and construct loading on ring diameter within each construct type for each mode of loading. Rotations and translations occurring between bone model segments were individually evaluated between construct configurations for each mode of loading using a multivariate ANOVA with a post hoc Bonferroni correction. A P value ≤ 0.05 was considered significant for all analyses performed.

Results

None of the constructs or individual components failed during testing. Deformation of the olive wires was obvious during loading of the constructs. Angular deformation at the connection of the primary hybrid rod to the ring was observed when all three constructs configurations were loaded in axial compression. Load-displacement curves had a relatively linear slope for all three construct configurations tested (Figure 4-6). Some load-displacement

curves obtained in craniocaudal and mediolateral bending had a triphasic appearance (Figure 4-7). After the first loading cycle in each mode of loading, the load displacement curve shifted slightly to the right along the X axis (displacement) in all constructs. Significant differences in stiffness between construct configurations were noted in all modes of loading (Table 4-1).

Ring diameter decreased significantly in all constructs after application of tension to fixation wires. Construct loading did not result in any additional decrease in ring diameter. All rings returned to pre-wire tensioning diameter once fixation wire tension was released post-testing (Table 4-2).

Overall motion occurring between bone model segments, including rotations and translations, were fairly similar between construct types. During application of mediolateral bending loads, constructs Ia_d and Ib significantly mitigated translation along the proximodistal axis over that allowed by construct Ia. During application of axial torsion loads, construct Ib significantly mitigated translation along the craniocaudal axis over that allowed by constructs Ia and Ia_d (Table 4-3, 4-4).

Discussion

Load-Displacement Curve Shapes

Traditional circular fixator constructs display a characteristic initial non-linear increase in stiffness when loaded in axial compression.[6,7,49,53] The fixation wires secured to the ring components initially allow displacement of the secured bone segments when axial loads are applied, but this displacement increases the tension in the fixation wires which mitigates further displacement with increasing axial loads.[6,39,49,53,54] This self-tensioning effect purportedly allows axial micro-motion at the fracture or osteotomy.[6,39,53] Displacement is mitigated as the applied axial load increases because the increased tension in fixation wires increases the stiffness of the

construct.[49,54,76] Axial micromotion is presumed to be beneficial because it has been shown to increase callus formation and fracture healing.[55-58,60,77] The load-displacement curves we obtained during axial loading of each of the hybrid fixator construct configurations tested in our study were linear or multiphasic linear (Figures 4-6, 4-7). The linear load-deformation curves suggest that gap motion of bone segments stabilized with a hybrid fixator construct may be different than gap motion of bone segments stabilized with a traditional circular fixator construct. A linear load-displacement curve implies that construct stiffness does not change with increasing load. In this scenario, displacement of the stabilized bone segments would not be mitigated by increasing construct stiffness with an increased loading force as would occur with a circular fixator. Our results are similar to the findings of a study by Khalily et al., which compared an Ilizarov fixator to two configurations of linear-circular hybrid fixator constructs designed for use in human patients.[78] Khalily et al. found that the linear-circular hybrid fixators did not exhibit the non-linear increase in stiffness which the traditional Ilizarov construct displayed when loaded in axial compression. Khalily et al. likened the axial load-displacement behavior of linear-circular hybrid fixators to that of unilateral linear fixators.[78] When a linear fixator is loaded in axial compression the load-displacement curve is typically linear until the construct begins to undergo plastic deformation.[12,35]

The use of incomplete, rather than complete rings, may have contributed to the linear slope of the load-deformation curves noted in this study. Data from Chapter 3 demonstrated that constructs made with incomplete rings have a less pronounced increase in stiffness with increasing load application than comparable diameter constructs made with complete rings.[79] The load-displacement curves of our hybrid fixator constructs did not reach a yield point which

suggests that the maximal loads we applied to the constructs were not sufficient to cause significant plastic deformation in the fixator components.[35]

The conformation of load-displacement curves we obtained when all three constructs were loaded in torsion were linear, similar to the curves obtained in axial compression. Most of the load-displacement curves obtained when the constructs were loaded in craniocaudal and mediolateral bending were linear, but a few displayed a biphasic or triphasic slope. We attribute the multiphasic load-displacement curves obtained in bending to slippage of the distal bone model segment on the fixation wires. Several previous circular fixator biomechanical studies have reported biphasic or triphasic load-displacement curves in bending, with the multiphasic nature of the curves attributed to bone model translation on the fixation wires.[8,80] We chose to use olive wires in all of our constructs, which were positioned so that at least one of the olives opposed translation of the Delrin rod along the wire in each mode of loading in an attempt to minimize bone model slippage along fixation wires. Based on the multiphasic nature of several of the load-displacement curves we obtained, the olive wires were partially but not completely successful in mitigating bone model slippage.

Stiffness Results

The circular portion of all three constructs evaluated in this study was the same. This uniformity implies that differences in construct performance are due to the design of the proximal, linear portion of the constructs. If the proximal linear portion of the hybrid fixator constructs tested in our study were classified according to standard linear fixator nomenclature[62,63] then construct Ia would be most like a single bar type Ia fixator, construct Ia_d like a double bar type Ia fixator, and construct Ib like a type Ib fixator. Comparison of stiffness trends in each of the modes of loading utilized in our study to previously published data on the mechanical performance of various designs of linear fixators shows some similarities in

performance between our hybrid fixator constructs and linear fixator constructs based on construct design.

Our study found that in axial compression, the addition of a diagonal secondary hybrid rod and the incorporation of a cranially inserted fixation pin secured to a secondary hybrid rod both increased construct axial stiffness. Stiffness of constructs Ia_d and Ib was increased by 65% and 69% respectively over that of construct Ia. This 4% stiffness difference between constructs Ia_d and Ib was not significant, demonstrating that the addition of a diagonal strut to support a primary hybrid rod is just as beneficial as biplanar fixation pin insertion in axial compression. A similar trend of increasing stiffness from single bar (type Ia) to double bar (type Ia_d) to quadrilateral (type Ib) was reported by Egger et al. when loading linear fixator constructs in axial compression.[13] Brinker et al. also noted an overall trend of increasing stiffness from a unilateral, single connecting bar (type Ia) to a unilateral, single bar with an accessory bar (type Ia_d) to a biplanar (type Ib) with several exceptions, in a study evaluating linear fixators in axial compression.[12]

In torsion, our Ia_d construct was not significantly (4%) stiffer than our Ia construct, while our Ib construct was 25% more stiff than our Ia and 20% more stiff than our Ia_d constructs. Our findings suggest that biplanar fixation pin insertion is a valid strategy to increase stiffness in torsion, while the addition of a diagonal supporting hybrid rod is not. Egger et al. found similar increasing stiffness trends based on frame modification in a group of linear fixator constructs when loaded in torsion.[13]

In our study, craniocaudal and mediolateral bending stiffness was increased by 11% and 9% respectively in construct Ia_d by the addition of a diagonal support over the basic hybrid design (Ia); however, this stiffness increase was not statistically significant. A statistically

significant stiffness increase of 19% and 27% was noted in craniocaudal and mediolateral bending respectively with biplanar pin placement in construct Ib over the stiffness of the basic hybrid design (Ia). Previous studies evaluating linear fixators have demonstrated that bending stiffness is increased when force is applied in the plane of fixation pin insertion.[11,16,76] Similar to the linear fixator constructs in previous studies, our hybrid fixator constructs were stiffer when bending occurred in the plane of fixation pin insertion. Based on the results we obtained, biplanar fixation pin insertion is a valid strategy to increase hybrid fixator construct stiffness in both craniocaudal and mediolateral bending.

Numerical stiffness values obtained in our study may be useful to compare our results to results of a previously published mechanical study. We modeled our testing protocol after methods used in a study by Cross et al. which evaluated the effect of different distal ring block configurations on circular fixator construct stiffness.[8] We used similar loading modes, rates, and cycles as well as testing jigs. The study by Cross et al. utilized six construct designs all made with 84 mm rings and 1.6 mm Kirschner wires. All constructs had an identical proximal ring block construct consisting of two complete rings each with two associated Kirschner wires. One of the circular constructs used a distal ring segment consisting of a single incomplete ring and two Kirschner wires, similar to our hybrid constructs. Cross et al. reported stiffness results for this circular construct of 90 N/mm in axial compression, 0.35 Nm/degree in craniocaudal bending, 0.8 Nm/degree in mediolateral bending, and 0.85 Nm/degree in torsion. Our hybrid fixator constructs were uniformly less stiff than the circular fixator construct in the study by Cross et al. in axial compression and torsion, but similar in stiffness when loaded in craniocaudal and mediolateral bending. These findings are similar to the findings of a study performed by Pugh et al., which reported that Ilizarov constructs were stiffer than single ring

linear-circular hybrid fixator constructs loaded in axial compression and torsion, but that bending stiffness was more comparable between groups.[81] Cantilever bending, presumably occurring at the interface between the ring and linear rod in linear-circular hybrid fixator constructs during axial compression has been reported to result in increased fracture gap motion and decreased stiffness over that of comparable circular fixator constructs. This effect appears to be less pronounced in bending.[76] While hybrid fixator construct design can vary greatly, the relatively basic designs of hybrid fixator constructs we tested likely have lower axial and torsional stiffness than comparable circular fixator constructs. This generalization may be useful in guiding construct decision making in the clinical setting.

Ring Deformation

Elastic ring deformation occurred consistently in our three hybrid fixator construct designs, as demonstrated by the decrease in outside ring diameter associated with application of tension to fixation wires. The significance of incomplete ring deformation in the clinical setting is currently unknown. Circular fixator constructs utilizing incomplete rings have been demonstrated to be less stiff than comparable constructs utilizing complete rings.[8,79] Ring deformation is most likely the reason for the decreased stiffness noted when incomplete rings are utilized in fixator constructs and this decrease in stiffness is one of the disadvantages of using incomplete rings.[8,53,79] We chose to incorporate incomplete rings in our hybrid constructs because incomplete rings are commonly used when fixators are positioned near joints or used to secure proximal limb segments due to anatomic constraints.[8,20,26,47,48]

Bone Model Kinematics

Hybrid construct design is driven by an attempt to neutralize forces acting on a fracture or osteotomy site and prevent excessive motion which could delay or compromise bone healing. While axial micromotion has been shown to enhance fracture healing, the effects of shear and

bending on bone healing have not been as clearly demonstrated.[56-59,82] Our purpose in quantifying bone model movements during loading was two-fold: First, to objectively report the motions occurring at the simulated osteotomy site and second, to determine the extent to which variations in construct design would affect osteotomy site motion. A small but significant decrease in translation along a single axis was noted in mediolateral bending with the addition of a secondary hybrid rod and in mediolateral bending and torsion with orthogonal fixation pin insertion. The remainder of the translations and rotations experienced by the bone models as a result of construct loading were not significantly different between hybrid construct configurations. The similarity between measured bone model motions among our hybrid construct configurations is probably due to the distal ring segment design which was conserved between constructs. Strategies designed to increase distal ring segment stability such as incorporation of drop wires or utilization of multiple rings would be more likely to decrease motion between bone model segments than would the incorporation of additional linear fixator components in the proximal segment.

The majority of the biomechanical differences observed between linear and circular fixation have been attributed to the use of small diameter tensioned wires as fixation elements in circular fixators. The small diameter wires, which are inherently less stiff than larger diameter fixation pins, allow some dynamic motion at the fracture site when load is applied.[49,53] Not surprisingly, in our study, the majority of visible motion during loading appeared to result from deflection of the fixation wires stabilizing the distal bone model segment in the circular ring component of the hybrid fixators. A lesser amount of movement also appeared to occur at the ring/primary hybrid rod interface. This movement is most likely due to the cantilever bending effect typically associated with loading unilateral external fixators and which has been reported

as well for linear-circular hybrid fixator constructs.[39,52,76] The rotations measured between bone model segments during axial loading do not demonstrate any appreciable amount of cantilever bending (rotation about the craniocaudal axis) in any of our hybrid construct designs. The lack of cantilever bending noted between bone model segments was probably due to the design of the jig utilized for axial compression testing. The proximal and distal bone model segments were both constrained to prevent angulation out of the axis of force application (proximodistal axis). A jig design utilizing a universal joint to allow bone model angulation during axial loading might have produced different results.

Limitations

All displacement values utilized in this study to calculate load-displacement curves were obtained from the recorded displacement of the actuator of the servohydraulic testing system. This means that the stiffness values reported for constructs in this study are not fracture gap stiffness values, but rather construct stiffness values. Assuming that the Delrin rod utilized as a bone model does not deform a significant amount during loading, then construct stiffness should closely approximate fracture gap stiffness. A more accurate calculation of gap stiffness could have been performed if extensometers had been utilized to measure changes in fracture gap displacement during loading. Both methods have been reported in previous studies.[8,76,83,84] We chose to utilize displacement data from the servohydraulic testing system based on our assumption that the bone model deformation during loading would be relatively unappreciable.

The three hybrid fixator constructs tested in this study are relatively similar in design. The similarity in design was intentional, with the objective of being able to differentiate the contributions of individual components to frame performance. More elaborate fixators are often used in dogs and cats in the clinical settings.[20,26,32] The results obtained here may not directly extrapolate to cover the wide variety of hybrid fixator designs that are possible. In future studies,

development and validation of finite element models which could be used to accurately predict hybrid fixator performance might be the next logical step in towards a more complete understanding of hybrid fixator biomechanical properties as it would allow for rapid and economical assessment of variations in hybrid fixator construct design.

Summary

The results of this study supported our initial hypotheses. The addition of a secondary hybrid rod as a diagonal strut increased the axial stiffness of the type Ia_d hybrid fixator, and orthogonal fixation pin insertion increased hybrid fixator stiffness in all modes of loading evaluated. Incomplete rings incorporated in hybrid fixator constructs underwent elastic deformation associated with fixation wire tensioning; however, permanent ring deformation was not observed. Motion between the two bone model segments was minimally affected by changes to the linear portion of our hybrid fixator constructs and thus strategies designed to increase stability of the distal ring component may be more effective in reducing osteotomy site motion. Future studies are indicated to provide additional biomechanical information about a wider variety of hybrid fixator configurations.

Table 4-1. Construct stiffness

Construct	Axial compression	Craniocaudal bending	Mediolateral bending	Axial torsion
Ia	46.00 ± 1.68^{a}	0.54 ± 0.05^{a}	0.98 ± 0.22^{a}	0.48 ± 0.01^{a}
Ia_d	75.72 ± 3.85^{b}	0.60 ± 0.06^{ab}	1.07 ± 0.17^{ab}	0.50 ± 0.02^{a}
Ib	77.74 ± 2.66^{b}	0.64 ± 0.08^{b}	1.24 ± 0.17^{b}	0.60 ± 0.06^{b}

Values are mean ± standard deviation of construct stiffness in axial compression (N/mm), craniocaudal bending (Nm/degree), mediolateral bending (Nm/degree), and axial torsion (Nm/degree) for the three configurations of linear-circular hybrid constructs tested. Significant differences between constructs for each mode of loading (within a column) are indicated by different superscript letters ($P \leq 0.05$)

Table 4-2. Ring deformation for each construct type in all four modes of loading

Loading Mode	Construct	Pre-wire tensioning	Pre-loading	Under load cycle 15	Post-Loading	Wire tension released
Axial compression	Ia	110.0 ± 0.6^{a}	108.4 ± 0.5^{b}	108.3 ± 0.3^{b}	108.5 ± 0.5^{b}	110.0 ± 0.4^{a}
	Ia_d	109.6 ± 0.3^{a}	107.9 ± 0.4^{b}	107.9 ± 0.4^{b}	108.2 ± 0.3^{b}	109.6 ± 0.4^{a}
	Ib	110.1 ± 0.4^{a}	108.5 ± 0.3^{b}	108.4 ± 0.2^{b}	108.7 ± 0.2^{b}	110.1 ± 0.3^{a}
Craniocaudal bending	Ia	109.9 ± 0.6^{a}	108.1 ± 0.5^{b}	108.2 ± 0.6^{b}	108.1 ± 0.6^{b}	110.0 ± 0.5^{a}
	Ia_d	109.6 ± 0.3^{a}	107.9 ± 0.4^{b}	107.8 ± 0.3^{b}	107.7 ± 0.4^{b}	109.6 ± 0.4^{a}
	Ib	110.1 ± 0.4^{a}	108.2 ± 0.3^{b}	108.1 ± 0.3^{b}	108.3 ± 0.3^{b}	110.1 ± 0.3^{a}
Mediolateral bending	Ia	110.0 ± 0.6^{a}	108.6 ± 0.5^{b}	108.6 ± 0.6^{b}	108.7 ± 0.5^{b}	110.0 ± 0.4^{a}
	Ia_d	109.6 ± 0.3^{a}	107.9 ± 0.4^{b}	107.8 ± 0.3^{b}	107.7 ± 0.4^{b}	109.6 ± 0.4^{a}
	Ib	110.1 ± 0.4^{a}	108.2 ± 0.4^{b}	108.1 ± 0.3^{b}	108.3 ± 0.4^{b}	110.1 ± 0.3^{a}
Axial torsion	Ia	110.0 ± 0.6^{a}	108.5 ± 0.5^{b}	108.5 ± 0.5^{b}	108.5 ± 0.7^{b}	110.0 ± 0.4^{a}
	Ia_d	109.6 ± 0.3^{a}	108.2 ± 0.3^{b}	108.0 ± 0.4^{b}	108.0 ± 0.3^{b}	109.6 ± 0.4^{a}
	Ib	110.1 ± 0.4^{a}	108.7 ± 0.2^{bc}	108.5 ± 0.3^{b}	108.9 ± 0.2^{c}	110.1 ± 0.3^{a}

Values are mean ± standard deviation (mm) for outer ring diameter measured during the testing sequence. Significant differences are designated by different letter superscripts. Statistical comparisons are between cycle phases (along rows) within a single construct type ($P \leq 0.05$)

Table 4-3. Rotational motion between the ends of the bone model segments occurring at the simulated fracture gap as a result of construct loading

Loading mode	Rotational axis	Ia	Ia_d	Ib
Axial compression	Craniocaudal	1.8 ± 1.7^a	1.1 ± 1.5^a	1.3 ± 0.4^a
	Mediolateral	2.5 ± 1.4^a	3.5 ± 3.2^a	2.1 ± 1.0^a
	Proximodistal	1.4 ± 1.5^a	0.9 ± 0.5^a	1.0 ± 0.5^a
Craniocaudal bending	Craniocaudal	0.7 ± 0.3^a	1.1 ± 0.6^a	2.0 ± 1.5^a
	Mediolateral	6.2 ± 1.6^a	6.9 ± 2.1^a	6.7 ± 1.3^a
	Proximodistal	0.7 ± 0.5^a	1.2 ± 0.8^a	0.8 ± 0.6^a
Mediolateral bending	Craniocaudal	5.3 ± 1.3^a	4.9 ± 0.9^a	5.1 ± 0.7^a
	Mediolateral	3.5 ± 3.0^a	1.6 ± 0.9^a	2.4 ± 2.2^a
	Proximodistal	1.6 ± 1.7^a	1.9 ± 1.7^a	1.0 ± 0.9^a
Axial torsion	Craniocaudal	1.4 ± 1.4^a	0.7 ± 0.5^a	0.6 ± 0.4^a
	Mediolateral	3.5 ± 3.0^a	1.7 ± 1.7^a	1.2 ± 0.8^a
	Proximodistal	7.5 ± 1.1^a	7.0 ± 1.3^a	7.2 ± 0.9^a

Values are mean ± standard deviation for rotations in degrees occurring around the craniocaudal, mediolateral, and proximodistal axes. Statistical comparisons are between construct types within each mode of loading (along rows). Significant differences are designated by different letter superscripts ($P \leq 0.05$).

Table 4-4. Translation between the ends of the bone model segments occurring at the simulated fracture gap as a result of construct loading

Loading Mode	Translation axis	Ia	Ia_d	Ib
Axial compression	Craniocaudal	0.7 ± 0.9^a	0.5 ± 0.3^a	0.6 ± 0.5^a
	Mediolateral	0.5 ± 0.5^a	0.2 ± 0.1^a	0.2 ± 0.2^a
	Proximodistal	5.0 ± 0.6^a	4.8 ± 0.8^a	4.6 ± 0.3^a
Craniocaudal bending	Craniocaudal	0.7 ± 0.3^a	1.1 ± 0.6^a	2.0 ± 1.5^a
	Mediolateral	6.2 ± 1.6^a	6.9 ± 2.1^a	6.7 ± 1.3^a
	Proximodistal	0.7 ± 0.5^a	1.2 ± 0.8^a	0.8 ± 0.6^a
Mediolateral bending	Craniocaudal	0.9 ± 0.6^a	0.4 ± 0.3^a	0.6 ± 0.6^a
	Mediolateral	2.0 ± 0.4^a	1.6 ± 0.5^a	1.9 ± 0.3^a
	Proximodistal	1.7 ± 0.6^a	0.8 ± 0.3^b	0.8 ± 0.3^b
Axial torsion	Craniocaudal	1.8 ± 1.0^a	1.7 ± 0.6^a	0.5 ± 0.3^b
	Mediolateral	0.3 ± 0.3^a	0.2 ± 0.1^a	0.2 ± 0.1^a
	Proximodistal	0.8 ± 0.9^a	0.2 ± 0.3^a	0.2 ± 0.1^a

Values are mean ± standard deviation for translations in mm occurring along the craniocaudal, mediolateral, and proximodistal axes. Statistical comparisons are between construct types within each mode of loading (along rows). Significant differences are designated by different letter superscripts ($P \leq 0.05$).

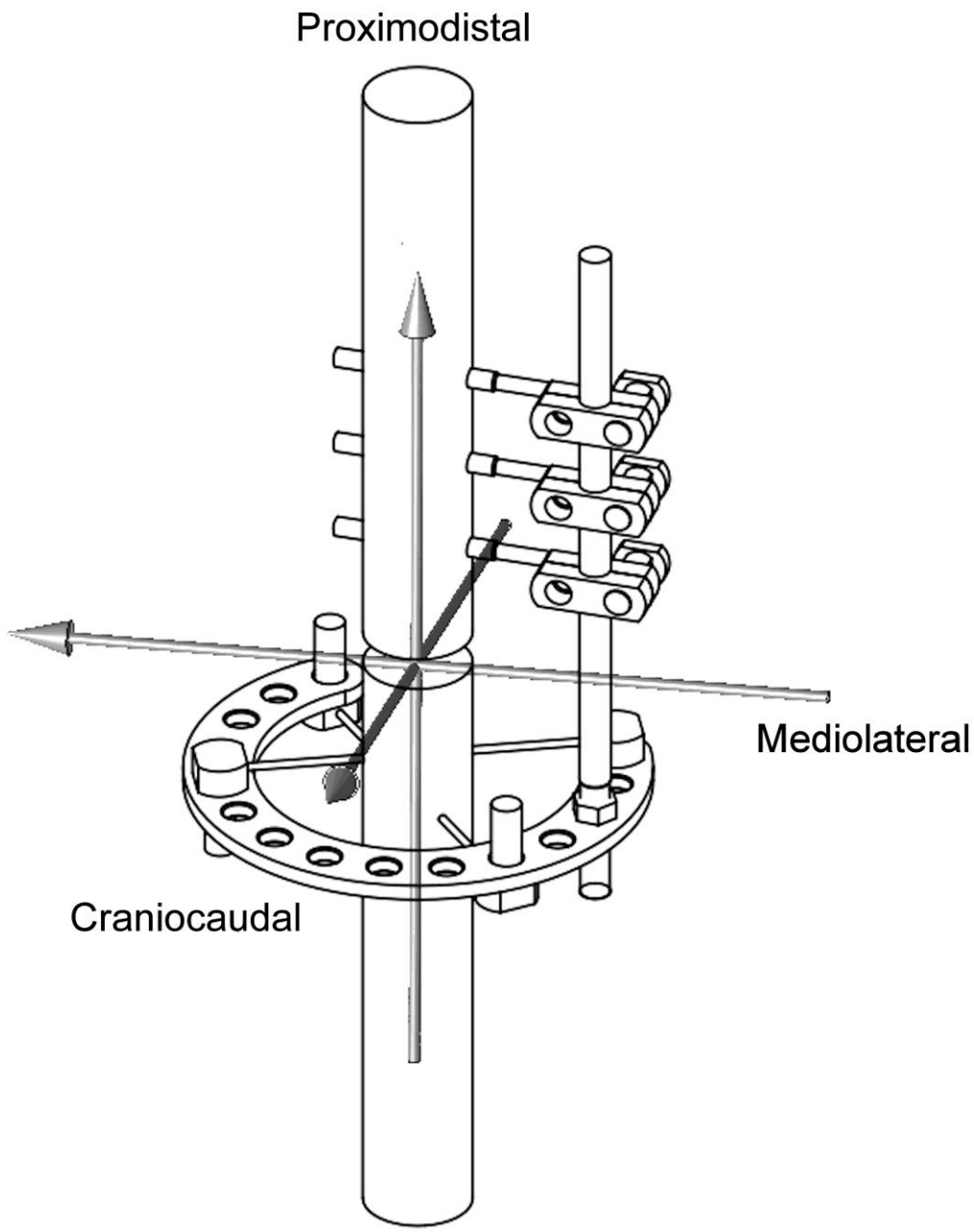


Figure 4-1. Anatomic axes defined for hybrid fixator constructs. The craniocaudal axis bisects the center of the ring and the hole at the midpoint of the closed end of the ring. The mediolateral axis bisects the center of the ring perpendicular to the craniocaudal axis. The proximodistal axis is oriented perpendicular to the 2 previous axes and parallels the shaft of the bone model longitudinally.

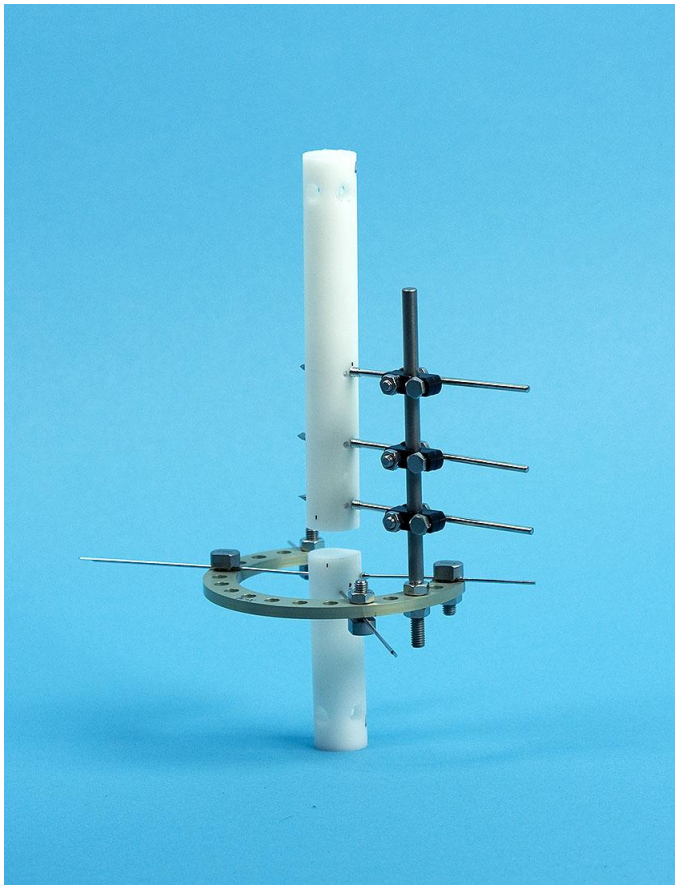


Figure 4-2. Construct Ia incorporates unilateral insertion of three half-pin fixation pins attached to a single hybrid rod.

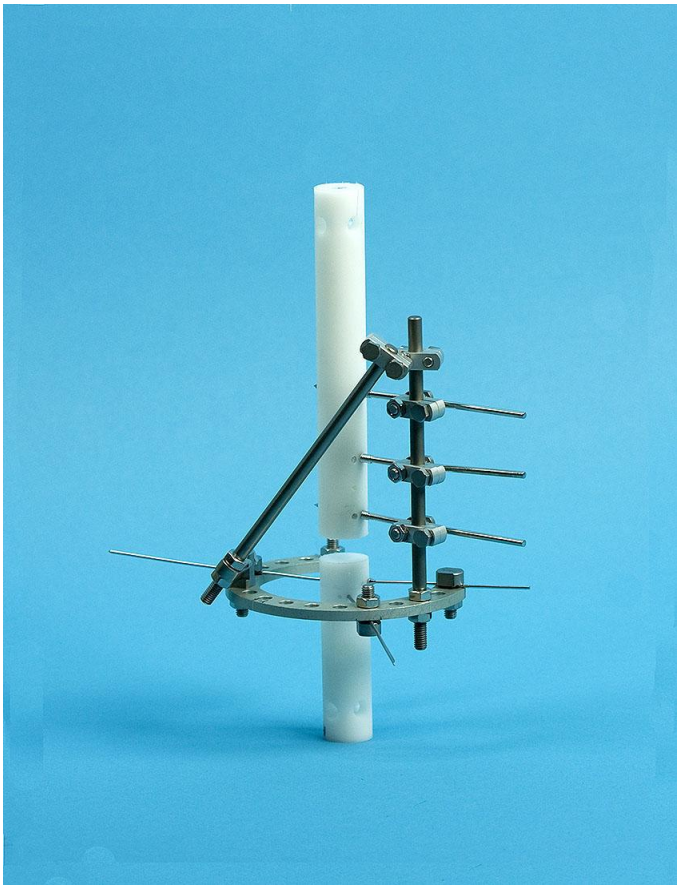


Figure 4-3. Construct Ia_d incorporates a primary, medial hybrid rod with three half-pins inserted in uniplanar fashion. A secondary hybrid rod runs articulates with both the ring and the proximal end of the primary hybrid rod as a diagonal strut.

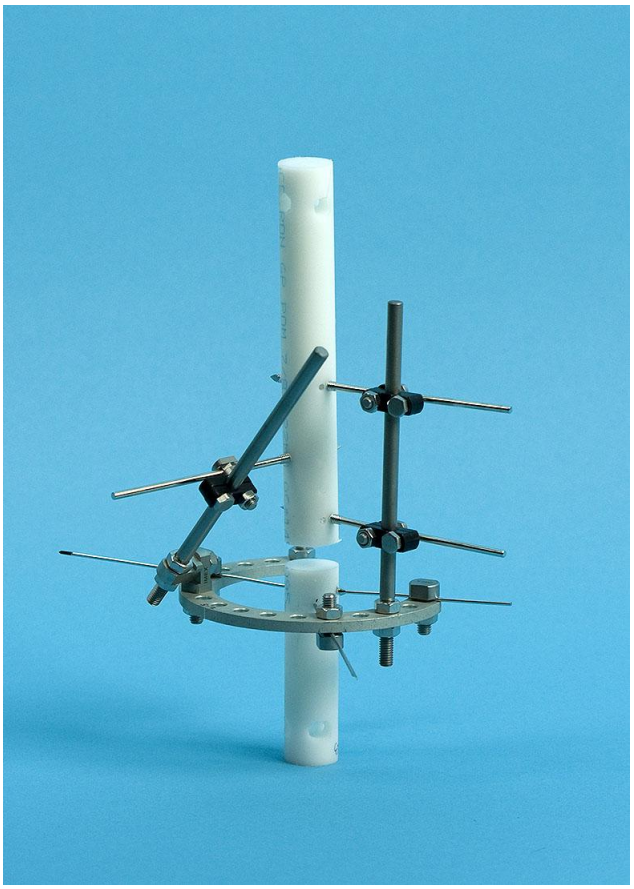


Figure 4-4. Construct Ib incorporates 3 half-pins inserted in biplanar fashion. Two fixation pins attach to the primary, medially placed hybrid rod and one attaches to the secondary cranial hybrid rod.

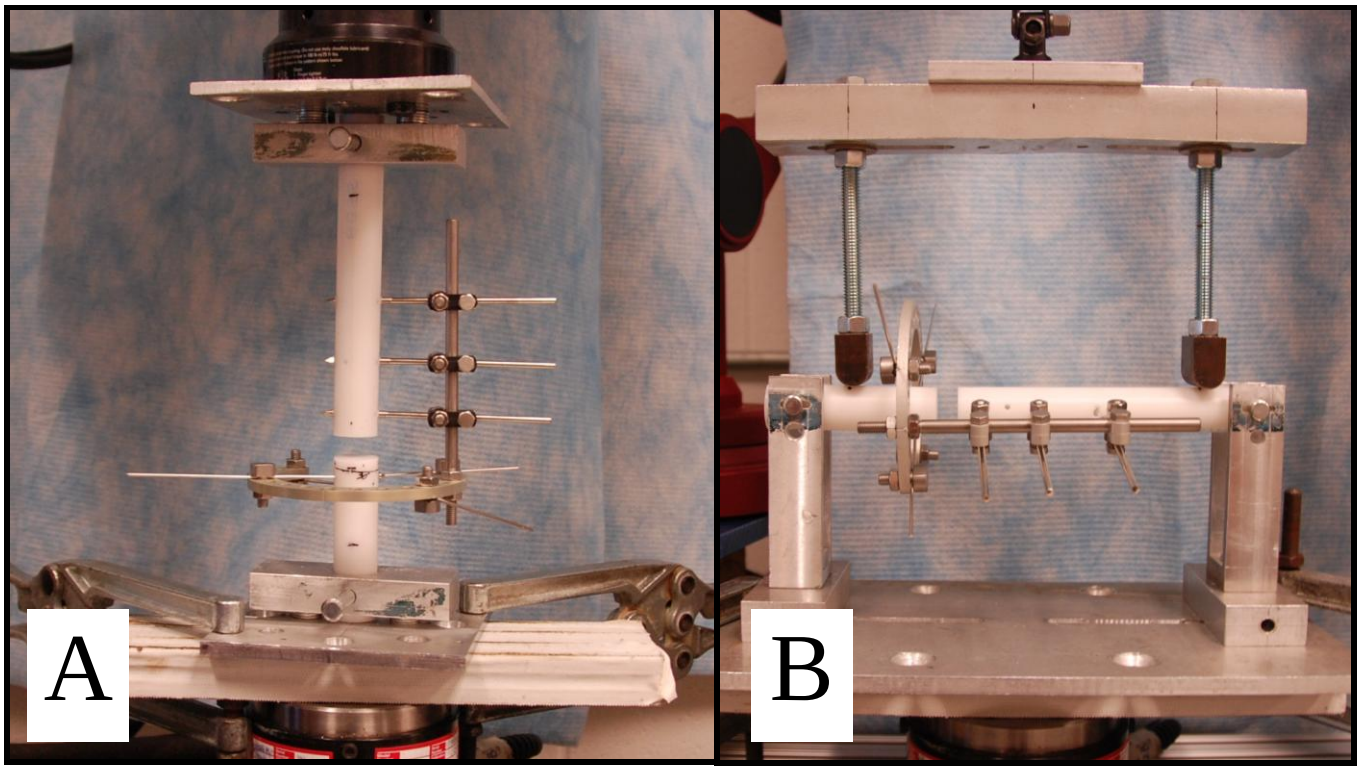


Figure 4-5. Custom jigs used for hybrid fixator construct testing. A) Construct Ia positioned in custom jig for axial compression and axial torsion testing. B) Construct Ia positioned in custom jig for craniocaudal bending. Mediolateral bending was performed in the same jig used for craniocaudal bending with the construct rotated 90° around the proximodistal axis.

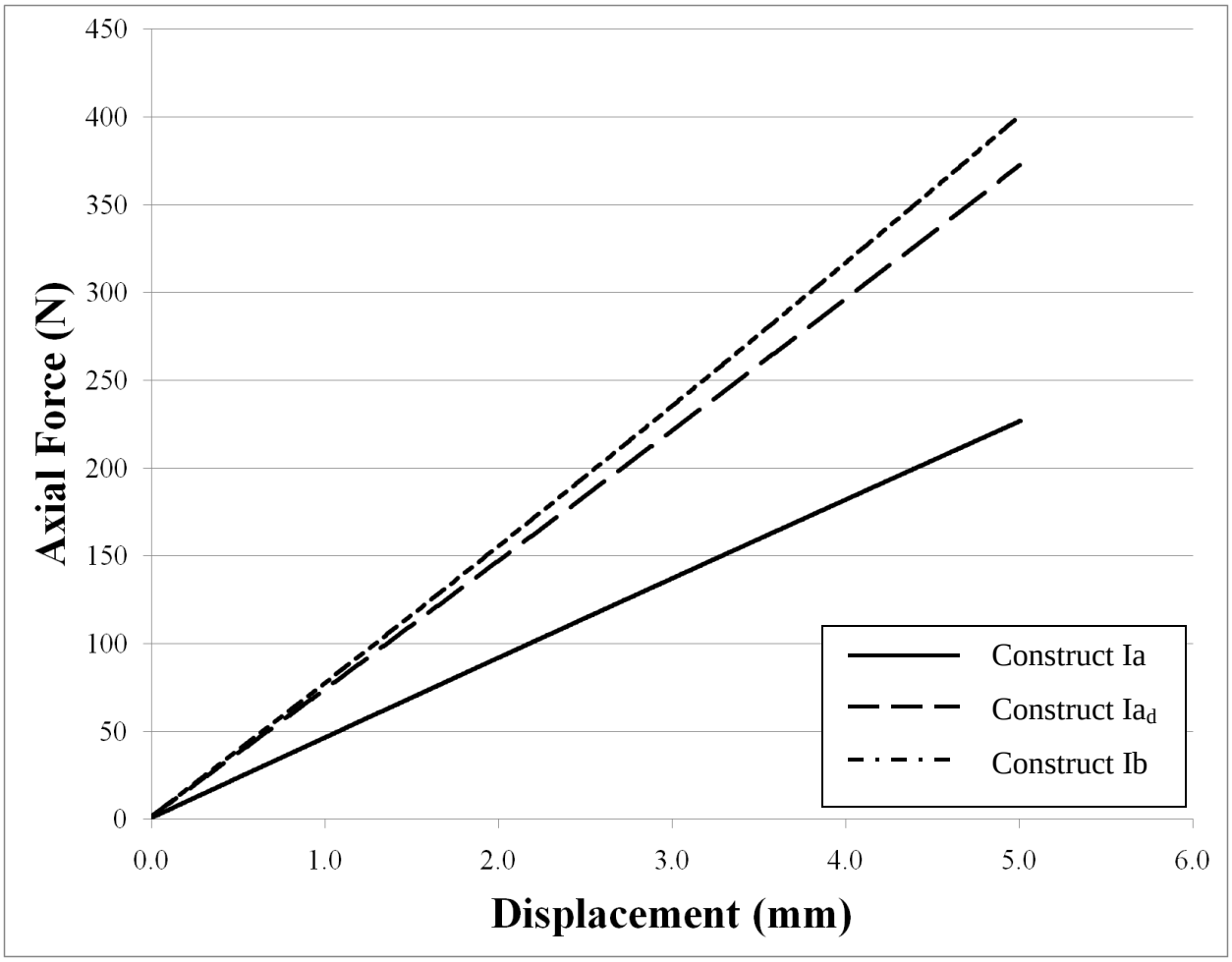


Figure 4-6. Axial compression load-displacement curves. All three constructs a relatively linear slope.

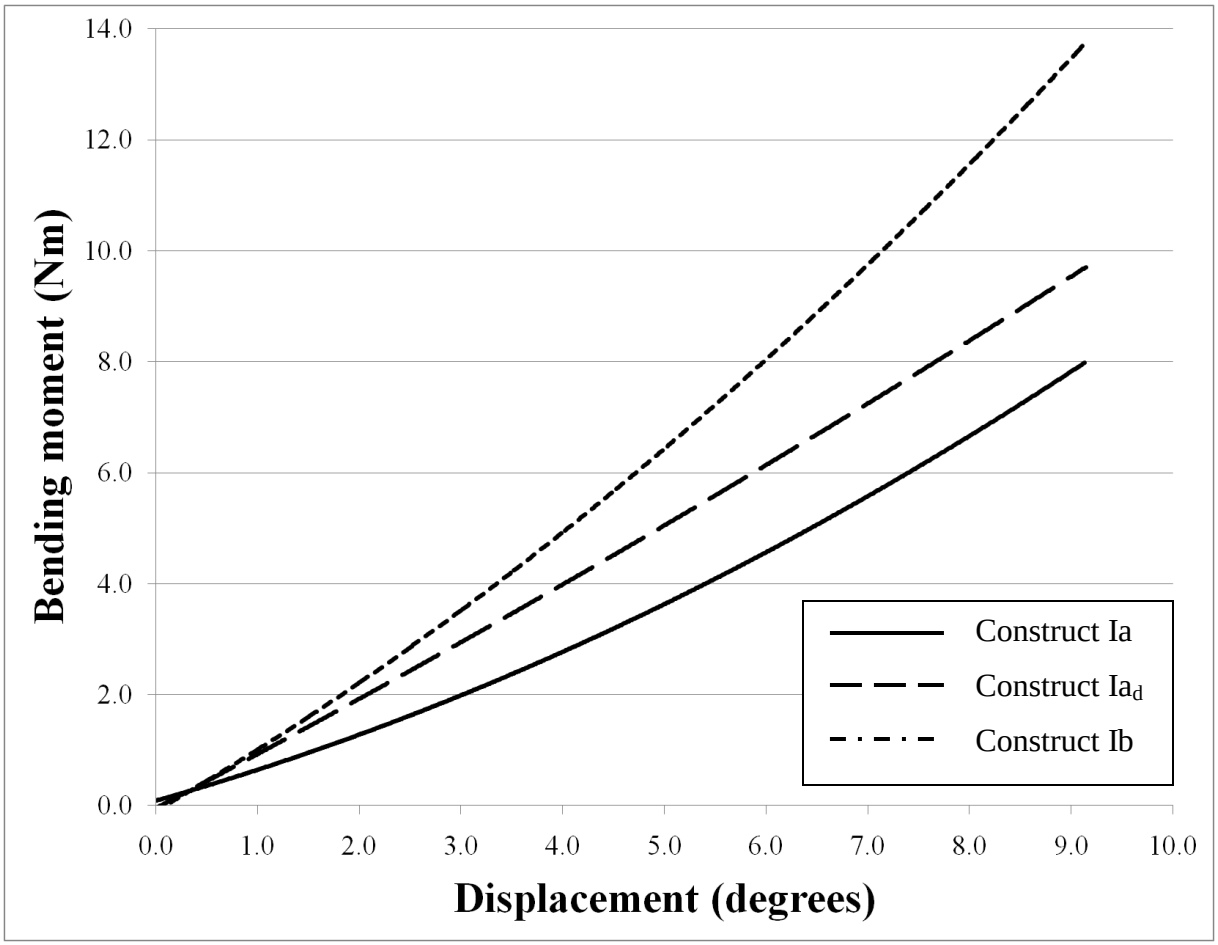


Figure 4-7. Mediolateral bending load-displacement curves. These curves have a multiphasic slope.

CHAPTER 5
CONCLUSION

External skeletal fixation is a system of bone stabilization which is commonly utilized in dogs and cats to stabilize fractures, correct angular limb deformities and perform bone transport.[4,19,20,23,24,26,28-30,32,43,45,47,48] All external skeletal fixators consist of percutaneously inserted fixation elements which transfix bone segments and which are incorporated into an extra-corporeal framework that serves to stabilize the involved limb segments.[1] Types of external skeletal fixation include linear fixation which uses linear rods as frame elements and pins as fixation elements, circular fixation which utilizes rings as supporting elements and small diameter wires as fixation elements, and the more recently developed linear-circular hybrid fixation which incorporates components of both linear fixation and circular fixation into a highly customizable fixation system.[20,22,26,33,37]

Chapter 1 provided a brief overview of the history of external skeletal fixation in dogs and cats including the development of linear fixators, circular fixators, and finally culminating in the development of hybrid fixators. Hybrid fixation is purported to combine the beneficial aspects of both linear fixation and circular fixation by allowing controlled axial micromotion of stabilized bone segments, which is the hallmark of circular fixation, but incorporating linear hybrid rods and fixation pins which allow hybrid constructs to be used in anatomic locations where application of circular fixators is not practical. We reviewed hybrid fixator nomenclature, biomechanical properties of the components which are typically incorporated in hybrid fixator constructs, as well as the current literature describing the clinical application of hybrid fixators. We concluded that current knowledge regarding hybrid fixator biomechanical properties has been an extrapolation from extensive studies performed on linear and circular fixator

components. Clinical outcomes with hybrid fixators have been very good overall and the most commonly utilized hybrid fixator would be classified as a type Ib hybrid fixator.

Incomplete rings are commonly incorporated in hybrid fixator constructs because the open section of the ring can be positioned to avoid impingement of the regional anatomic structures, particularly when the ring is positioned near joints. In Chapter 2 we evaluated the ring deformation induced by wire tensioning in single ring constructs utilizing both complete and incomplete rings. We tested rings of four diameters and three wire crossing angles by sequentially applying wire tension from 0 kg to 30 kg, to 60 kg and to 90 kg. Fixation wire tensioning did not cause significant ring deformation in complete ring constructs; however, significant ring deformation was observed in incomplete ring constructs and the magnitude of the ring deformation increased with increasing fixation wire tension. Wire crossing angle had an effect on the magnitude of ring deformation induced by fixation wire tensioning but the effect was not consistent between ring diameters or between wire tensions.

The effect of ring deformation on incomplete ring construct performance was not clear, so in Chapter 3 we tested single ring constructs of 50 mm, 66 mm, 84 mm, and 118 mm diameter utilizing complete and incomplete rings at the same four wire tensions utilized in Chapter 2 but using a single wire crossing angle which allowed wires to cross in the center of the incomplete ring constructs. We applied axial compression to the single ring constructs and obtained load-displacement curves from which we calculated construct stiffness. Incomplete ring constructs displayed a more gradual increase in load-displacement curve slope compared to complete ring constructs. We found that constructs utilizing complete rings were stiffer than constructs which utilized incomplete rings. The stiffness of incomplete ring constructs could be increased with application of fixation wire tension. Axial displacement in both complete and

incomplete ring constructs decreased with fixation wire tensioning. We also measured ring deformation associated with wire tensioning and construct loading. Complete ring constructs did not deform during testing while incomplete ring constructs underwent deformation during wire tensioning and construct loading. We attributed the decreased stiffness of incomplete ring constructs when compared to complete ring constructs to ring deformation occurring in the incomplete ring constructs. Based on construct performance and observed plastic ring deformation we recommended that fixation wires on incomplete rings be left untensioned in 50 mm diameter rings, tensioned to 30 kg or less in 66 mm diameter rings and be tensioned to 60 kg in 84 mm and 118 mm diameter rings.

In Chapter 4 we evaluated three hybrid fixator construct configurations: A type Ia, a type Ia with an additional diagonal strut, and a type Ib. The hybrid constructs were tested in axial compression, craniocaudal bending, mediolateral bending, and axial torsion. Construct load-displacement curves were linear rather than displaying the exponentially increasing slope typical of circular fixators. We found that the addition of a second hybrid rod as a diagonal strut (construct Ia_d) increased construct stiffness in axial compression while the addition of a secondary hybrid rod and bilateral insertion of fixation pins (construct Ib) increased construct stiffness in all modes of loading over the stiffness of the type Ia construct. Bone model motion occurring during construct loading was quantified and determined to be relatively similar between construct types. Reversible ring deformation was induced with fixation wire tensioning and construct loading.

Based on the results of our study, we conclude that incomplete ring deformation decreases ring construct stiffness but is not detrimental to construct performance as long as the deformation is in the elastic range. Hybrid constructs may behave more like linear fixators rather

than circular fixators based on load-displacement curve shapes and thus may not allow fracture site axial micromotion, as was previously thought. Orthogonal insertion of fixation pins and addition of more than one hybrid rod are valid strategies to increase hybrid fixator stiffness. Additional studies are warranted to further characterize the biomechanical performance of hybrid fixator constructs.

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BIOGRAPHICAL SKETCH

Caleb C. Hudson was born in Lafayette, Louisiana. Caleb grew up in the Ozarks of southern Missouri as the oldest of five children. He was homeschooled from kindergarten through high school. After graduation he enrolled in the University of Missouri at Columbia as an Animal Science major. In fall 2003 Caleb was accepted into the College of Veterinary Medicine at the University of Missouri and he graduated Summa Cum Laude with a DVM degree in May 2007. After graduating from Vet school he traveled to Gainesville Florida and completed a small animal rotating internship at the University of Florida's Small Animal Hospital. When the internship year was completed Caleb stayed in Gainesville and entered into the small animal surgical residency program at the University of Florida's Small Animal Hospital. Now starting the final year of the surgical residency program, Caleb is looking forward to the new challenges and opportunities that await as he takes on the role of chief resident.