8-7-2020

# Biomechanical comparison of external fixation and double plating for the stabilization of a canine cadaveric Supracondylar Humeral Fracture Gap Model

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A Thesis
Submitted to the Faculty of
Mississippi State University
in Partial Fulfillment of the Requirements
for the Degree of Master of Science
in Veterinary Medicine
in the Department of Clinical Sciences, College of Veterinary Medicine

Mississippi State, Mississippi

August 2020

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Date of Degree: August 7, 2020

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Title of Study: Biomechanical comparison of external fixation and double plating for the stabilization of a canine cadaveric Supracondylar Humeral Fracture Gap Model

Pages in Study 45

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A 2 cm ostectomy was performed on 10 pairs of canine cadaveric humeri proximal to the supratrochlear foramen. Stabilization was with a double plate construct (DB-PLATE) (n=10) or external skeletal fixator with intramedullary pin tie-in configuration (ESF-IMP) (n=10). Cyclic testing was performed. Axial compressive load to failure testing followed. Data analyzed included dynamic stiffness, stiffness and yield load. No constructs failed during cyclic testing or lost stiffness over time, although mean dynamic stiffness was greater for DB-PLATE compared to ESF-IMP. Mean stiffness of DB-PLATE in load-to-failure testing was not significantly different than ESF-IMP. Yield force of DB-PLATE was significantly higher than ESF-IMP. These results suggest that both DB-PLATE and ESF-IMP would be appropriate fixation techniques for stabilization of comminuted supracondylar humeral fractures in dogs with appropriate exercise restriction. Double plate fixation may be preferable when prolonged healing or inadequate post-operative restraint was anticipated because it was stronger in destructive testing.

DEDICATION

I would like to dedicate this research to my parents, Dorothy and Louis Castaldo for their faithful support in my education as well as Matthew King for his unconditional love, support and patience.

ACKNOWLEDGEMENTS

I would like to thank Dr. Jason Syrcle for giving me the opportunity to pursue a surgical residency and for his time, patience, and contributions to this research project. I would like to thank Dr. Steve Elder for his assistance and expertise, and Dr. Robert Wills for his assistance with statistical analysis. I would also like to thank Drs. Betsy Swanson, Ron McLaughlin, Ryan Butler, and Claudio Correa Natalini for their support throughout this project and my surgical residency. Finally, I would like to thank Tom Thompson, and the MSU-CVM radiology department for their contributions and technical assistance.

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# INTRODUCTION

## Background

Supracondylar humeral fractures comprise approximately 30% of distal humeral fractures in dogs (1,2). Surgical repair of these fractures is challenging due to limited bone stock of the distal humeral fragment(s), proximity to the elbow, presence of the supratrochlear foramen, presence of adjacent neurovascular structures, and inherently complex shape of the humerus (3). Humeral T-Y fractures, which consist of an intracondylar fracture and supracondylar component, share the challenges of supracondylar fracture stabilization. These fractures comprise an additional 20-43% of distal humeral fractures (1,2).

Fixation methods for supracondylar fractures include cross-pin fixation, medial or lateral bone plating, plate-rod fixation, external skeletal fixation, or double plating techniques (3-5). Simple pinning techniques are only recommended in juvenile patients with anatomically reconstructable fractures (3). Rigid internal fixation has been recommended for comminuted supracondylar humeral fractures (1). Clinical studies have described successful stabilization of such fractures with medial or double-plate application of conventional or locking plates (1,6-11). Successful external fixation of supracondylar fractures is also reported and can be achieved with a type I-II hybrid linear fixator and diagonal connecting bar, often in conjunction with an intramedullary pin tied in to the fixator (11-17).

Fixation of supracondylar humeral fractures with a single medial or caudomedial bone plate has been described but may be best reserved for reconstructable fractures or those with a relatively large distal fragment (1,6). Double plating is recommended for fractures with limited distal bone stock (1,8-10). Plates can be applied medially, laterally, caudomedially, caudolaterally, or caudally (3). Double plating is most commonly performed using a combination of caudolateral and medial or caudomedial conventional or locking plates (1,8-10). Double plating is considered standard-of-care for human patients with similar fractures (18).

A type I-II linear fixator with an intramedullary pin tie-in and an acrylic diagonal was used in a case series of six large breed dogs with supracondylar humeral fractures with successful healing and good outcomes reported in five of these patients (14). Similar techniques have resulted in successful healing of supracondylar and humeral T-Y fractures in a feline case series, as well as case reports of both cats and dogs (15-17). Advantages of external skeletal fixation include decreased surgical time, ability to dynamize the fixation, and limited requirement for soft tissue dissection, which may all contribute to faster healing times (13-17). Normograde placement of an intramedullary pin in the humerus starting at the distal metaphysis allows maximal pin purchase within the distal fragment, which is ideal for fixation of distal humeral fractures (19).

The mechanical properties of single versus double plating with String-of-Pearls® locking plates in a canine distal humeral metaphyseal gap model was previously described (20). The double-plate group consisted of caudomedial and caudolateral plates secured with short monocortical screws; this was compared to a medial plate group with bicortical screws, including a transcondylar screw. The double-plate group demonstrated greater stiffness in torsional and axial compression testing compared to the single-plate group. However, the single-plate group ultimately had a higher strength than that of the double-plate group, which was likely due to screw-bone interface failure associated with short monocortical screws placed in the distal humeral metaphysis. The pullout strength of screws with differing lengths placed in various regions of the humerus was investigated, demonstrating that short monocortical screws placed in the distal humeral metaphysis had low resistance to pullout, and the pullout strength of screws placed in the humeral condyle was directly related to the length of the screw (21). Consequently, maximizing screw length in metaphyseal bone should increase the strength of the construct.

Successful stabilization of comminuted supracondylar humeral fractures is challenging, and there are presently a limited number of biomechanical studies. In veterinary medicine, there are no studies looking at the biomechanics of external skeletal fixation versus bilateral plating for supracondylar humeral fractures.

**Canine Humerus & Elbow Anatomy**

Understanding the anatomy of the canine and feline humerus is imperative for successful fracture fixation, as well as restoring appropriate form and function. The position of the humerus makes it an important part of transferring weight and propelling the body forward during locomotion (3,22). The humeral head is on the proximal caudal aspect of the bone and articulates with the glenoid cavity of the scapula to form the shoulder. The humeral condyle articulates with the radius and ulna to form the elbow (3,22). The greater tubercle, which is located on the craniolateral aspect of the proximal humerus, is separated from the lesser tubercle by the intertubucular groove (3,22). The intertubercular groove is also where the tendon of origin of the biceps brachii runs. As the biceps brachii tendon is transmitted across this groove, it is held in place by the transverse humeral ligament (3,22). The site of origin of the lateral head of the triceps brachii muscle is a bony ridge known as the tricipital line. This ridge runs from the humeral head cranially and toward the deltoid tuberosity distally (3,22). The bone just caudal to this tricipital line is cortical which makes it more ideal for implant placement due to its inherent holding power (24). The deltoid tuberosity is the insertion site for the deltoid muscle and is located cranial and distal to the tricipital line (3,22).

The canine humerus has a more pronounced S-shape in the dog than the cat, which is particularly pronounced in chondrodystrophic breeds (3,22). The humerus also tapers from proximal to distal which affects the size of intramedullary implants that can be used for fracture repair (3). The S-shape of the humerus makes implant contouring more challenging than other long bones such as the femur, tibia, radius, and ulna.

The distal part of the humerus is the condyle which is composed of a medial trochlea and lateral capitulum (3,22). Their respective articulations are described below. The medial and lateral sides of the humeral condyle have eminences called epicondyles. These projections serve as the area of attachment for the medial and lateral collateral ligaments, as well as tendons (3,22). The radial fossa is located on the cranial aspect of the bone just proximal to the condyle (22). The comparable structure on the caudal aspect of the bone just proximal to the condyle is the olecranon fossa (22). The radial and olecranon fossae communicate via the supratrochlear foramen (22). No soft tissue structures pass through this foramen in the dog (22). The olecranon fossa receives the anconeal process during extension of the elbow (22). In the cat, a supracondylar foramen is located proximal to the medial epicondyle and serves as a conduit for the median nerve and brachial artery. In contrast to dogs, cats lack a true supratrochlear foramen (3).

One of the most challenging aspects of distal humeral fracture repair is the associated complex neurovascular anatomy. The nerves that are most commonly encountered during surgical approaches are the median and ulnar nerves medially and the radial nerve laterally (3). All of these nerves originate from the brachial plexus which is formed by the ventral branches of the 6$^{th}$, 7$^{th}$, and 8$^{th}$ cervical and 1$^{st}$ and 2$^{nd}$ thoracic spinal nerves (3). The median nerve in addition to the brachial artery and vein pass cranial to the medial epicondyle before continuing distally to enter the antebrachium (3,22). The median nerve innervates the pronator teres, pronator quadratus, flexor carpi radialis, superficial digital flexor, the radial head and parts of the humeral and ulnar heads of the deep digital flexor (22). It is also responsible for supplying sensory innervation to the palmar surface of the manus (22). The ulnar nerve runs caudal to the medial epicondyle and innervates the flexor carpi ulnaris and parts of the ulnar and humeral heads of the deep digital flexor (22). The ulnar nerve is responsible for sensory innervation to the palmar aspect of the manus and motor innervation to the intrinsic muscles of the manus (22). The radial nerve travels a short distance with the median and ulnar nerves before entering the triceps distal to the teres major (22). The radial nerve is responsible for motor function to all of the extensor muscles of the elbow, carpal, and phalangeal joints (22). Three clinically important muscles that receive radial nerve innervation include the triceps brachii, tensor fasciae antebrachii, and anconeus. As the radial nerve courses distally, it coils around the humerus, first on the caudal and then on the lateral surface of the brachialis muscle (22). The radial nerve is at risk for iatrogenic damage when approaching the craniolateral distal diaphysis of the humerus due to its lateral position in this region (3,22). The radial nerve terminates into deep and superficial branches on the lateral side of the distal third of the thoracic limb (22). On the lateral aspect of the distal humerus, the vasculature encountered includes the cephalic, omobrachial, and axillobrachial veins (22,23). On the medial distal surface, the most common vessels encountered are the brachial artery and vein which were described above (22,23).

The canine elbow is a complex, synovial hinge (ginglymus) joint that is composed of three smaller joints. The principle bones that form the elbow articulation are the humerus, ulna, and radius (22). The main weight-bearing bone of the antebrachium is the radius (22); however, the proximal articular surface of the ulna and radius contribute almost equally to load-sharing through the canine elbow joint (25). The main articular regions of these bones are the radial head (radius); ulnar trochlear notch, anconeal process, medial coronoid process, and lateral coronoid process (ulna); and the trochlea, capitulum, and supratrochlear foramen (humerus) (3). The three smaller joints are the humeroulnar joint (humeral trochlea and ulnar trochlear notch from the anconeal process to the radial incisure [radial notch], including the medial coronoid process), humeroradial joint (capitulum and radial head), and the proximal radioulnar joint (3,22).

The ligaments supporting these osseous structures include the medial and lateral collateral ligaments, annular ligament, and interosseous ligament which are all extrasynovial (3,22). The joint capsule is comprised of cranial and caudal compartments and surrounds the entire joint, including the supratrochlear foramen cranially but not caudally (3,22). The joint capsule has both an inner synovial layer and an outer fibrous membrane (3,22).

There are few muscles that contribute to flexion and extension of the elbow joint. The main extensor muscle is the triceps brachii with contributions from both the tensor fascia antibrachii and anconeus muscles which are all innervated by the radial nerve (22). The main flexors are the biceps brachii and brachialis muscles (22). During the swing phase of locomotion, the extensor carpi radialis also contributes to elbow flexion (3). The biceps brachii and brachialis muscles are innervated by the musculocutaneous nerve, while the extensor carpi radialis is innervated by the radial nerve (3,22).

The normal standing angle of the canine elbow is approximately 130° with normal range of motion of approximately 36° in flexion (range, 34° to 38°) and 165° in extension (range, 164° to 167°) (26). The collateral ligaments are responsible for rotational stability of the elbow joint when both the elbow and the carpus are held at 90° (3). A Campbell's test is used to evaluate the integrity of the collateral ligaments (27). A Campbell's test is performed with the elbow joint and carpus at 90 degrees of flexion to position the anconeal process caudal to the olecranon fossa, so that rotational stability of the elbow joint relies primarily on collateral ligaments. Transection of the medial collateral ligament increases pronation to 60 to 100 degrees, whereas transection of the lateral collateral ligament increases supination to 70 to 140 degrees (27). Although this test has been validated, the rotational ability of the elbow joint varies greatly (27). The normal angles of rotation of the elbow are 17-50° laterally (supination) and 31-70° medially (pronation) (23,27,28).

**Humeral Fractures**

Humeral fractures are the least common long bone fracture in small animals (29). Fractures of the canine humerus account for 8-10% of fractures and are more likely to affect the distal part of the bone (condylar, supracondylar) (30-32). Humeral fractures in the cat account for 5-13% of fractures and usually occur at the mid-diaphysis (30-33). Proximal humeral fractures are least common for both species, as most fractures occur at the middle and distal one-third of the bone (3,6). The cause of diaphyseal fractures tends to be road traffic accidents, gunshots, and falls as compared to distal humeral fractures which are primarily caused by jumping or falling (1,34). In one retrospective study, vehicular trauma was responsible for approximately 70% of humeral fractures (1). Vannini et al. reported that 90% of unicondylar humeral fractures were caused by minor trauma, whereas 83% of the supracondylar and distal diaphyseal fractures were caused by severe trauma (2).

It is crucial to do a thorough physical examination as part of the clinical workup for humeral fractures due to nearby vital structures and body compartments, including the thorax, head, and neck. This is especially important in cases of vehicular trauma. Cardiovascular, respiratory, and neurologic examinations should be performed prior to administration of pain medications and anesthetic induction for surgical repair. Thoracic radiographs are recommended to help look for the presence of pneumothorax, pleural effusion secondary to hemothorax, diaphragmatic hernia, rib fractures, scapular fractures, and vertebral fractures or luxations (35). Selcer et al. diagnosed concurrent thoracic morbidities in 57 of 100 dogs presenting with musculoskeletal injuries (36). In that study population, 77% had abnormal thoracic radiographs, 44% had low $PaO_2$ (hypoxemia), and 30% had cardiac arrythmias (36). Electrocardiogram should be used to help rule out arrythmias secondary to traumatic myocarditis (35). Patients presenting with a humeral fracture often have a dropped elbow with the manus resting on its dorsal surface. This mimics the appearance of nerve injury which makes differentiation crucial prior to determining the course of treatment (3). Neurologic examination is necessary to determine the extent of damage and offer prognostic information to clients (29). Injuries that may present concurrently with humeral fractures include spinal trauma (fracture/luxation), brachial plexus or spinal nerve root avulsion, and radial nerve injury (29,35). Horner's syndrome or loss of panniculus reflex, in conjunction with thoracic limb deficits, may be indicative of brachial plexus injury (3). The brachial plexus originates from the 5-8$^{th}$ cervical and 1$^{st}$ and 2$^{nd}$ thoracic spinal nerves and provides sensory and motor innervation to the thoracic limb (37,38). Brachial plexus avulsion occurs when there is traction on the thoracic limb or severe abduction of the scapula. The nerve roots are more likely to be damaged than the plexus itself due to their lower capacity to stretch (39-42). Radial nerve paresis will typically resolve with fracture fixation and time (35). Assessment for cutaneous sensation is crucial in patients presenting for trauma. If cutaneous sensation is present, motor function to the limb is usually expected to return in 1-6 weeks (35).

Orthogonal (mediolateral and craniocaudal) radiographic views should be taken of both thoracic limbs preoperatively in order to appropriately plan surgical correction (3). Radiographs of the contralateral limb can serve as an important point-of-reference for surgical planning, particularly in cases where the injured limb has a severely comminuted or displaced humeral fracture. Bandaging the limb is not necessary if surgery is scheduled imminently. However, if several days are expected between initial stabilization and surgical repair of the humerus, the only appropriate bandage is a spica splint (3). Adequate analgesia should be provided both pre-operatively as well as postoperatively (3).

Fractures of the humerus are classified according to their anatomic location: proximal, midshaft, and distal (1). Subclassifications exist within each of these categories. Proximal fractures can be divided into greater tubercle fractures (Salter-Harris type I and type II), neck fractures, and proximal metaphyseal fractures (1). Fractures of the midshaft are classified as transverse, oblique, spiral, comminuted, and segmental (1). Finally, distal fractures include distal shaft, supracondylar, and lateral/medial humeral condylar fractures (1). Dicondylar fractures are also called “T” or “Y” fractures (1).

Due to the large cross-sectional area of the bone and its close proximity to the body, proximal humeral factures are uncommon; however, when they do occur, they are often present in skeletally immature animals and are physeal in origin (3,4). When mature animals present with proximal humeral fractures, they are usually associated with traumatic, comminuted diaphyseal fractures or develop as a result of gunshot injuries (3,4). It is extremely uncommon for proximal humerus fractures in mature animals to be isolated, simple transverse or oblique fractures (3,4). The proximal metaphysis is predisposed to developing primary bone tumors with osteosarcoma being the most common. Although neoplasia is the primary cause of pathologic fracture in this region, fungal and systemic diseases should also be considered (3). Iatrogenic fracture can also occur in the proximal humeral metaphysis, as the greater tubercle is a common site for harvesting cancellous autograft (43). As previously mentioned, proximal humeral fractures can be divided into greater tubercle fractures (Salter-Harris type I and type II), neck fractures, and proximal metaphyseal fractures (1, 29).

The developmental anatomy of the proximal humerus is complex. The capital center of ossification will appear first between 14 and 16 days (44,45). It will then begin to slowly invade the cranial epiphysis to form the greater tubercle at 4 months (44,45). Fusion of the epiphysis with the metaphysis occurs between 7.5 and 12 months in the dog and 19 and 26 months in the cat (44-46). The epiphysis is formed by the fusion of the humeral head and greater tubercle with the metaphysis (3,47). The angle of fusion of the humeral head and greater tubercle in the dog is approximately 102° which forms a "cap" that has inherent stability once fracture reduction occurs (47).

Proximal humeral physeal fractures are most commonly repaired with wires, wire and tension band fixation, lag screws, or a combination of these methods (3,4). In young animals with continued growth potential, it is preferable to place wires in parallel fashion (3,4). This will allow for continued physeal growth by avoiding compression of the growth plate (3,4). In older animals, it is more acceptable to use lag screws and tension band wires in order to provide additional stability (3,4). Complicated fractures of the proximal humerus can be stabilized with lag screws and a plate (4). Due to the abundance of cancellous bone in the proximal humerus, healing is usually rapid (4).

Fractures of the diaphysis occur most commonly due to blunt-force trauma such as falls, kicks, or vehicular impact (1). They can be simple transverse, oblique, spiral, or comminuted (3). Various fixation methods can be used to stabilize this region of the humerus: external skeletal fixation, bone plates, screws, cerclage wire, closed or open intramedullary pinning, interlocking nail, or a combination of these methods (3,4). Fractures of the diaphysis can be further subdivided into the region affected (proximal, middle, and distal), as well as the degree of comminution (3).

Distal humeral fractures include the supracondylar region, the condyle, and the distal physis (3). A supracondylar fracture is defined as a fracture that communicates with the supratrochlear foramen but not the articular surface (3). Fracture lines can be transverse, oblique, or comminuted with comminuted being the most common in dogs and cats (14, 48). As previously mentioned, these fractures can be challenging because of their close proximity to the elbow, limited distal bone stock, and regional anatomy (3,48). Briefly, acceptable fixation methods include cross-pin fixation, medial or lateral bone plating, plate-rod fixation, external skeletal fixation, and double-plating techniques) (3-5).

The humeral condyle develops from two centers of ossification that fuse at approximately 83 days (3 months) with the medial center appearing between 14 and 22 days of life, and the lateral center appearing at 21 to 43 days (44,49). The condyle then fuses with the metaphysis by 5.5 to 6 months (44,49). Fractures of the humeral condyle extend through the articular surface or one or both epicondyles or epicondylar crests or into the distal diaphysis (3). Humeral condylar fractures are primarily diagnosed in skeletally immature dogs under 1 year of age or adult dogs with underlying humeral intracondylar fissures (50). In one survey, which included 107 humeral fractures in the dog, 43 (41%) were condylar (25/107 dicondylar, 15/107 lateral condylar, and 3/107 medial condylar) (1). The lateral condyle is fractured more frequently than the medial condyle due to biomechanical and anatomic differences and accounts for 34-67% of humeral condyle fractures and 37% of distal humeral fractures (1,51,52). More specifically, the lateral condyle is smaller and less robust than the medial condyle (1,51,52). Fractures of the medial condyle occur in approximately 6.9-11% of condyle fracture cases, and T-Y fractures occur in 25.9-35% of fractures affecting the humeral condyle (1,51,52). French Bulldogs and Spaniels (English Springer Spaniel, Cocker Spaniel, and Cavalier King Charles Spaniel) are predisposed to humeral condylar fractures with medial being more common in French Bulldogs (50,52,53). Medial humeral condylar fractures occur because of the interaction between the ulna and humerus, while radial loading is the cause of lateral fractures (54). It is thought that medial humeral condylar fractures are more common in chondrodystrophic breeds because of the difference in elbow anatomy and subsequent loading patterns (53). Cats have a much lower incidence of condylar fractures than dogs. This is likely due to the fact that their lateral and medial epicondyles are wider and subsequently stronger, as well as the lack of a supratrochlear foramen (1). Methods of fracture repair also vary for the humeral condyle and can include Kirschner wires, position or lag screws, bone plates and screws, intramedullary pinning, or a combination of these methods (3).

Prognosis after humeral fracture repair is usually good with appropriate stabilization, activity restriction, and adequate healing (3). When the articular surface is involved, the prognosis is more guarded because elbow function may be compromised due to the development of decreased range of motion, subsequent residual lameness, and osteoarthritis (3,8). In one study, 13 of 13 dogs with humeral condylar fractures developed posttraumatic osteoarthritis following surgical correction (55). Outcome after repair of lateral humeral condylar fractures was reviewed with owner interpretation being excellent in 79%, good in 14%, and poor in 7% of cases (56). T-Y fractures carry a more guarded prognosis with 41% classified as excellent, 52% considered good, and 10% considered fair (8). Animals with incomplete ossification of the humeral condyle carry the worst prognosis with nonunion and implant failure commonly occurring (3). It is vital to achieve anatomic reduction and absolute stability when repairing the articular surface of the humeral condyle. This, along with postoperative rehabilitation, is typically expected to lead to a satisfactory outcome (3).

## External Skeletal Fixation

External skeletal fixation employs a more biologic approach to fixation as compared to internal fixation (4,57). It preserves local soft tissues, avoids vascular compromise, requires minimal exposure, achieves anatomic alignment without excessive manipulation or disruption of the fracture fragments, allows for minimal contact with the periosteal surface, and preserves the fracture hematoma (4,57). The majority of the fixation remains outside the skin surface and uses percutaneously applied transosseous pins or wires secured to a connecting bar with clamps (4,57). Because the fixation remains external, it satisfies “open but do not touch” or closed reduction technique requirements which results in a more biologic approach to osteosynthesis (58-62).

Fractures that have been repaired with external skeletal fixation undergo indirect healing (secondary bone healing) (57, 63). Fractures that are allowed to heal in nature without surgical correction similarly undergo indirect bone healing (63). The formation of callus is the hallmark of indirect bone healing, as it requires the organized response of the periosteum and surrounding soft tissue envelope (63,64). Indirect bone healing decreases interfragmentary strain in two ways (64). First, osteoclasts remove dead bone from the margins of the fracture resulting in an initial widening of the fracture gap (63,64). The resorption of dead bone at the fracture gap increases interfragmentary strain and distance which, in turn, allows granulation to form and survive within the gap (63, 64). Secondly, external callus begins to form on the abaxial surface of the bone (63). Stability is correlated with radial distance of the callus – the greater the distance, the greater the stability due to the increase in the area moment of inertia (63). The formation of various types of tissue within a fracture gap is dictated by the degree of interfragmentary strain; granulation tissue can survive in conditions of 100% strain, fibrocartilage can withstand 10% to 15%, and bone can tolerate 2% (63). Five overlapping phases can be used to describe secondary bone healing: inflammation, intramembranous ossification, soft callus formation (chondrogenesis), hard callus formation (endochondral ossification), and bone remodeling (65). Fracture healing times with external skeletal fixation are decreased as compared to direct bone healing and more invasive open techniques due to the biological approach to osteosynthesis (59,60,62).

Advantages of external skeletal fixation go beyond the aforementioned biological approach. External skeletal frames are readily accessible and can therefore be adjusted throughout healing (57). Components can be added or removed from the frame which is particularly advantageous in cases of angular limb deformity (57). Disadvantages of external skeletal fixation can be both biological and mechanical in origin. Because the fixation pins are percutaneous, the risk for infection is greater as compared to internal fixation devices (57).

Pin tract infections are the most common significant complication, and surgeons can expect all pins to produce some degree of drainage and inflammation (66). Because the implants are placed externally, large bending moments act on the fixation pins (57). As the fracture heals, the risks of pin tract infection, fixation failure, and premature pin loosening increase, so careful selection of tolerant patients and compliant owners is essential (57).

A linear external skeletal fixator uses pins, clamps, and connecting bars. Pins can be smooth or threaded; however, the use of smooth pins is contraindicated due to their poor bone-holding power (67-71). Pins can be classified as negative- or positive-profile and by the depth of penetration (half or full). Half pins only penetrate the soft tissues on one side but span both cortices. Full pins penetrate both soft tissue surfaces and both cortices (57). Positive-profile pins have a single diameter shaft with threads that are rolled onto the end or center making the threaded portion larger than the rest of the shaft. Positive-profile pins have increased pin stiffness, greater axial pull-out strength, and greater fatigue life compared to smooth pins (67,69,71). Negative-profile pins have threads cut into the diameter so that the outer diameter of the threaded portion is the same as the shaft diameter. The core of the threaded portion is smaller than the shaft diameter. Older negative-profile pins were characterized by an abrupt transition from the non-threaded portion to the threaded portion, which, in turn, created a stress riser and ultimately led to failure at this pin junction (72). Newer negative-profile pins have been designed with a taper transition between the threaded and non-threaded portions, which maximizes both the diameter of the threaded portion within the bone and the shaft diameter outside the bone. This eliminates the stress riser effect that the old negative-profile pins carried. The Duraface® pin is one example of a negative-profile threaded pin with a tapered thread-run-out (TRO). In an *in vitro* mechanical study, Duraface® pins were compared to currently available positive-profile pins (73). Under static loading, TRO pins were significantly stiffer (55%) and had a higher maximum load (54%) compared to positive-profile pins (73). In cyclic fatigue testing, TRO pins lasted 2.3- to 4.9-fold more cycles than positive-profile pins (73).

During the process of pin selection, it is important to consistently choose the largest appropriate pin diameter for the patient's size. This is because large pin diameters have greater resistance to bending or failure with cyclic loading. The diameter of the pin chosen should be no greater than 25% of the diameter of the bone. Additionally, larger diameter clamp-pin interfaces result in more stable constructs that are less likely to undergo slippage or rotation (57). A minimum of two pins must purchase each bone fragment; however, the stress at the pin-bone interface is decreased if more pins are utilized, so it is preferable to use three to four pins per segment. In situations where this not possible, pins can be angled to one another in order to increase construct stability. Pins should be placed an appropriate distance from the fragment ends, which is typically defined as two pin diameters away from the edge (4).

Linear external skeletal fixation clamps rely on friction from compression of components within the clamps, which results in stabilization of the construct; therefore, the bone segments (57). Traditional external fixation clamps, such as the Kirschner-Ehmer® system, had many limitations, including 1) the fact that all clamps had to be applied to the connect bar prior to placing transfixation pins in the bone; 2) the clamps could not be removed without removing the entire connecting bar from the frame; 3) acute structural deformity was common after single-use tightening of the clamp; 4) clamps preplaced on the connecting bar would not accommodate the thread diameter of positive-profile pins (74-77). These disadvantages prompted the development of modern-day clamps. Contemporary clamp systems include IMEX® SK clamps and Securos® TITAN and U-clamps. These newer systems limit slippage along the connecting bar and slippage of the pin through the clamp, as well as rotation of the pin. Modern clamps can be disassembled, which allows for removal or addition of clamps in the middle of the frame, thereby preventing loss of fracture reduction. Newer clamps also accommodate multiple pin sizes and are available in single- or double-clamp models (57). Pin-gripping clamps should be placed so that the bolt locking the pin is as close to the bone as possible while still avoiding skin contact. Clamps should generally be placed 1 cm from the skin surface which shortens the pin length and makes the construct stiffer (4,57).

Connecting bars externally connect transfixation pins or wires of the linear system to the fixator frame through pin-connecting fixator clamps (57). Although they may be composed of a variety of materials, contemporary, lightweight, large-diameter connecting rods consist of an aluminum, titanium, acrylic, and carbon fiber composite. This has led to a substantial increase in strength and stiffness of the frame construct (78,79). Stainless steel is another material used in the production of connecting bars (57).

Construct stiffness may be increased by using a large connecting bar and smaller fixation pin. Utilizing smaller fixation pins decreases the risk of compromising bone integrity with large-diameter holes (57). Using a large bar decreases the load and stress on individual pins and evenly distributes the load between the pins, which in turn, may protect the pin-bone interface and reduce the chance of pin loosening (57,79).

Classification of linear external fixators has evolved from describing device names to a more descriptive system based on the number and planar geometry of connecting bars (80). Frame configurations can be broken down into unilateral (one side of the limb/half pins) or bilateral (both sides of the limb/full pins) and either uniplanar (all pins in one plane), biplanar (pins in two planes), or multiplanar (pins in multiple planes) (57). As configuration complexity increases, strength, stiffness, and resistance to shear forces, torsional loads, and axial loads likewise increase (81,82). The following table describes common nomenclature of external skeletal fixation frames (57):

Table 1 Nomenclature of External Skeletal Fixation Frames

| **TYPE** | **PINS** <br> **(Half vs. Full)** | **CONNECTING BAR NUMBER** | **PIN GEOMETRY** |
|---|---|---|---|
| Ia | Half | 1 | Unilateral uniplanar |
| Ib | Half | 2 | Unilateral biplanar |
| I-II | Half with 1 full | 2 | Bilateral uniplanar |
| II modified | Half with 2 full | 2 | Bilateral uniplanar |
| II | Full | 2 | Bilateral uniplanar |
| III modified | Half and full | 3 | Bilateral biplanar |

One of the downsides of mechanically superior configurations (i.e., type II and III frames) is that they are biologically more compromising than simple frames. Augmentation techniques can be applied to increase frame stiffness without the need for more complex frames (57). The use of interconnections for bilateral or multiplanar frames can add additional rigidity. Articulations are interconnecting bars that do not cross the fracture gap, while diagonals are interconnecting bars that do cross the fracture gap (57). Diagonals add more stability than articulations to the frame (83). An intramedullary pin can also be incorporated to add rigidity and should be no greater than 40% of the diameter of the medullary cavity (84,85).

## Bone Plating

Open reduction and internal fixation (ORIF) with bone plates requires a surgical approach to the bone, as well as manipulation of the fracture fragments, fracture hematoma, and the surrounding musculature. Although ORIF allows for rigid fixation and promotes weight bearing on the affected limb, its inherently invasive nature may delay fracture healing (86-87). Fractures that may be considered for open anatomic reconstruction include transverse, short oblique, long oblique, segmental, minimally comminuted (butterfly fragments) and articular fractures (86). It is especially important to achieve anatomic reconstruction of joint surfaces. In contrast, this may not be necessary for diaphyseal fractures (86). Fractures that cannot be anatomically reconstructed can also be repaired using bridging fixation (86).

Bone plates are typically applied to the tension surface of the bone and can be applied in compression, neutralization, buttress, or bridging (86). Dynamic compression can only be utilized if the fracture can be anatomically reconstructed such as with simple transverse and short oblique fractures (86). Bone plates can be loaded with screws placed eccentrically in order to achieve compression in specially designed dynamic compression plate holes (4). When a plate is applied in dynamic compression, the loads are transferred primarily through the reconstructed diaphysis, which spares the plate from cyclic bending stresses (86). Interfragmentary repair and reconstruction of the diaphysis can also be achieved in cases of long oblique, spiral, or butterfly fractures (86). The use of lag screws and cerclage wire can achieve effective interfragmentary compression, but in order to counter the forces applied during weight bearing, the bone plate is applied in neutralization mode (86). When a plate is applied in neutralization, the loads are carried primarily by the plate and, to a lesser extent, by the reconstructed diaphysis (86). When the diaphysis cannot be anatomically reconstructed, a bone plate applied assumes a bridging function (86). The bone plate alone prevents fracture gap collapse, resists the load applied, and withstands all forces applied at the fracture gap until clinical union is achieved (86). Although an "open" surgical approach to the bone is required when applying a plate in bridging fashion (with the exception of minimally invasive plate osteosynthesis [MIPO]), the principles of biological osteosynthesis, with preservation of the blood supply and fracture hematoma, are emphasized (4,86). Bridge plates are typically applied using an "open but do not touch" or MIPO approach (86). When choosing a plate to fix a fracture that cannot be anatomically reconstructed, a long plate affixed to the bone extremities (metaphyses or epiphyses) with fewer bone screws is utilized (86). The fracture site is spanned with the bridging plate, which acts like an extramedullary splint (4,87,88). Bridging plates are characterized by a high plate bridging ratio (plate-to-bone length ratio), low plate screw density (number of screw to number of plate holes ratio), and low plate span ratio (plate-to-fracture length ratio) (86). Buttress plating is an older term applied to plates used to fix cortical defects within metaphyseal regions. Bridge plating is now used for this type of repair (4,86).

When choosing the appropriate length for an internal fixator, the means of two values should be considered – the plate span width and the plate screw density (89). The plate span width is the quotient of plate length and overall fracture length. Gautier et al. found that the plate span width should be > 2 to 3 in comminuted fractures and > 8 to 10 in simple fractures. The plate screw density is the quotient formed by the number of screws inserted and the number of plate holes (89,90). The recommended value is < 0.4-0.5, which suggests that less than half of the plate holes should be occupied by screws (90). AO recommends that the diameter of the screw not exceed 40% of the diameter of the bone given that bone strength decreases as screw size increases (4). During bone plate application, screws should be placed at each end of the plate first, close to the fracture secondarily, and in the remaining holes last in order to maintain axial alignment (4). A minimum of 2 bicortical screws per fracture fragment is recommended when screws are inserted correctly in good quality bone (90). In all other cases, when possible, a minimum of three bicortical screws per fragment is recommended for improved security (90).

Complications associated with bone plating include delayed union, nonunion, malunion, need for reoperation due to implant failure or implant removal, osteomyelitis, sequestration, and mechanical failure (86,87,89). Rozbruch et al. retrospectively reviewed conventional plate osteosynthesis in humans over three decades and identified several factors associated with the development of the aforementioned complications, including extensive soft tissue dissection, disruption of the fracture hematoma, multifocal periosteal necrosis secondary to plate compression, and iatrogenic trauma associated with interfragmentary implants such as lag screws and cerclage wires. One of the best predictors of success in this study was the use of a longer bridging plate with fewer screws (89). As plating techniques shifted toward biological osteosynthesis during the course of this study, time to union decreased from 20 to 13 weeks, nonunion rates decreased from 10% to 4%, revision surgery rates decreased from 43% to 13%, and overall success rate increased from 62% to 87% (89).

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CHAPTER II

# BIOMECHANICAL COMPARISON OF EXTERNAL FIXATION AND DOUBLE PLATING FOR THE STABILIZATION OF A CANINE CADAVERIC SUPRACONDYLAR HUMERAL FRACTURE GAP MODEL

## Objectives

The authors hypothesized the double plate construct would be stiffer, stronger and more resistant to repeated loading than the external fixator construct when evaluated by simulated load bearing in a cadaveric model. These hypotheses will be addressed in the study with the following specific aims. The aim of this study was to compare two fixation methods, double plate fixation (DB-PLATE) and a hybrid Type I-II linear external fixator with an intramedullary pin tie-in (ESF-IMP), for the stabilization of a cadaveric supracondylar humeral gap model. The DB-PLATE construct was designed based on results of previous biomechanical studies and clinical experience in fixation of similar fractures (1,2). The ESF-IMP construct combines the advantages demonstrated in both biomechanical and clinical studies of a Type I-II external fixator with an intramedullary pin tie-in, with the intramedullary pin placed normograde from the distal metaphysis for maximal distal bone purchase (3-9).

## Materials and Methods

### Specimen Preparation

Twenty forelimbs previously harvested immediately following euthanasia for reasons unrelated to this study from ten young adult purpose-bred dogs weighing 27-35 kg were frozen at -20°C until needed for this study. Forelimbs were thawed at room temperature over a 24-hour period. The scapulae were disarticulated from the humerus and discarded. Prior to construct formation, a midshaft radioulnar osteotomy was performed and the distal limb discarded. All soft tissues other than the joint capsule and collateral ligaments of the elbow were removed.

Each pair of humeri were treated with both fixation methods, DB-PLATE and ESF-IMP. A coin flip determined which fixation method was applied to the right humerus of the first pair tested. The other fixation method was applied to the left humerus. Fixation methods were alternated between the left and right humerus of each subsequently tested pair. The margins of a proposed 2 cm ostectomy were marked on each humerus prior to implant placement. The distal margin of the ostectomy was the proximal aspect of the supratrochlear foramen and perpendicular to the humeral shaft. The proximal margin of the ostectomy was 2 cm proximal and parallel to the distal margin.

### Double-Plate Fixation

DB-PLATE fixation **(Figure 1)** consisted of two eight-hole plates applied to the distal humerus. All screw holes were drilled with a standard 2.5 mm drill bit, and 3.5 mm cortical screws were used.[I] The medial plate, a broad 3.5 mm compression plate[II] was placed first, contoured and applied to the caudomedial surface of the humerus. The plate was positioned with four holes proximal to the proposed ostectomy, two empty holes over the ostectomy and two holes over the distal segment. The plate was secured with 6 cortical screws, including four proximal bicortical screws. The two distal screws were monocortical, but angled to maximize screw purchase. The most distal screw was angled cranially, and the second most distal screw was angled craniodistally. Each of the 2 distal holes were drilled until the articular surface was penetrated, and a screw 2 mm shorter than measured was placed. The humeral ostectomy was performed with an oscillating saw, being careful not to damage the medial plate.

An 8-hole 3.5 mm String of Pearls plate[III] was contoured and applied to the caudolateral surface of the humerus, using 3.5 mm screws.[IV] Contouring was minimal for this plate, a gap of up to 2 mm under the plate at any screw hole was tolerated. The plate was positioned with four holes proximal to the ostectomy, two empty screw holes over the ostectomy site and two holes over the distal segment. The proximal four screws were 16 mm long and placed in monocortical fashion. An additional 16 mm screw was placed into the lateral supracondylar crest. For the most distal screw, a hole was drilled until the articular surface was penetrated, and a screw 2 mm shorter than the measurement was placed.

**External Fixator with Intramedullary Pin Tie-In Fixation**

ESF-IMP fixation **(Figure 2)** consisted of a hybrid Type I-II linear external fixator with an intramedullary pin tie-in. All fixator pins had a 4.8 mm thread diameter and were pre-drilled with a 3.9 mm bit. All fixator pins were threaded. Full pins were positive profile and half-pins were negative profile. A centrally threaded full condylar pin[V] was placed transversely first, just distal and cranial to the humeral epicondyles. The first half-pin[VI] was placed 2 cm proximal to the deltoid tuberosity, started just caudal to the tricipital line and oriented caudally. A 9.5 mm diameter, 200 mm carbon-fiber connecting bar[VII] was secured to the pins with fixation clamps[VIII] and two more clamps were pre-loaded onto the bar. The second half-pin was placed 1.5 cm distal from the first and the third 1.5 cm distal from the second. These two pins were inserted in the lateral cortex of the humerus in a lateral to medial direction, angled slightly cranially to allow for the passage of the intramedullary pin. A clamp, built from two standard fixation clamps, modified to grip two pins, was placed on the medial end of the condylar full pin. A 4.8 mm Steinman pin[IX] (cross bar) was contoured with a cranial bend to simulate the clinical situation and attached to the medial condylar clamp and to a fixation clamp connecting it to the carbon fiber connecting bar proximally. All clamps were secured tightly with a 10 mm wrench. For the condylar pin clamps, a gap of 2 cm was left between the bone and the clamp. For all other clamps, a gap of 2.5 cm was left, to simulate the clinical situation with soft tissues in place. The humeral ostectomy was completed as described previously. A 3.2 mm Steinman pin was placed in a normograde fashion from distal to proximal, starting in the fossa of origin of the deep and superficial flexor muscles, caudal and distal to the medial epicondyle (9). The pin was driven with a surgical drill until it exited the proximal humerus in the region of the greater tubercle. The distal point was cut from the pin and the proximal end was grasped with the drill and withdrawn until 2 mm of the pin remained protruding distally. The proximal aspect of the pin was connected with a single clamp and a 100 mm carbon-fiber connecting bar to the 200 mm connecting bar with a double clamp[X].

**Biomechanical Testing of Constructs**

The osteotomized radius and ulna at the distal end of each construct was potted in a steel fixture with polymethylmethacrylate (PMMA),[XI] leaving 2 cm between the elbow joint surface and the proximal aspect of the steel fixture. The steel fixture was secured to the load cell of a servohydraulic testing device[XII] **(Figure 3)**, with the humerus hanging freely from the fixture. Another steel fixture consisting of a 5 cm long and 3 cm diameter steel pipe welded to a plate was secured to the base plate of the testing device. The pipe was filled with PMMA in the dough phase and the humeral head was pressed into the PMMA, creating a custom mould for load application and simulating the glenoid cavity of the scapula. The elbow was placed at full extension and 20 N of preload applied to the construct.

A sinusoidal load between 20 and 200 N at 2 Hertz for a total of 63,000 cycles was applied to each construct. This simulated the load and number of cycles applied by a 30.5 kg dog (mean weight of test sample dogs) walking a total of 30 minutes per day (four 5-10 minute walks/day) for four weeks. The 200 N was derived from peak vertical force of 65% of body weight of a 30.5 kg dog, the approximate force applied at the walk to the thoracic limb (10,11). The number of cycles was calculated from a reported forelimb stride frequency of 1.25 strides/second at the walk (12). Samples were kept moist during testing by periodic spraying with 0.9% saline solution. After completing non-destructive cyclic testing, samples were maintained in the servohydraulic loading system and compressive load was applied at 6 mm/min until catastrophic implant failure occurred. Construct stiffness was calculated from the first linear portion of the load versus displacement curve. Yield load was defined as the point at which the load displacement curve deviated from the linear portion of the curve, determined by the intersection of a line with 2 mm offset from the construct dataset. Any obvious bone or implant deformation, or displacement of 2 cm was considered to be failure and the testing was stopped. Failure mode was documented with digital images, radiographs and video of each specimen.

**Statistical Analysis**

The effect of construct and time on dynamic stiffness was assessed by linear mixed models using PROC MIXED in SAS for Windows v9.4[XIII]. The initial model included construct, cycle number, and the construct-cycle number interaction as fixed effects. Dog was designated as a random effect. Type3 method option and Kenward-Roger approximation for the degrees of freedom was specified. In the case the construct-cycle number interaction was not significant, it was removed and the model refit. Similarly, in the case the cycle number variable was not significant, it was also removed from the model and separate models were fit for the middle and final times. Conditional residual plots were assessed to ensure the statistical models had met the assumptions of normality and homoscedasticity. The effect of construct on stiffness and yield force was also initially assessed with linear mixed models but the models did not meet the assumptions of normality and homoscedasticity. This appeared to be due to increased variation in the DB-PLATE construct compared to the ESF-IMP construct. It also may have been due to the relatively small sample size. Consequently, Wilcoxon Signed Rank tests were used to assess the effect of construct on stiffness and yield force. An alpha level of 0.05 was used to determine statistical significance.

## Results

None of the ten paired constructs failed during cyclic testing. Representative values for dynamic stiffness are reported in **Table 2**. A significant effect on dynamic stiffness due to cycle number ($p=0.9614$) or construct-cycle number interaction ($p=0.8304$) was not detected; however, the least squares mean of dynamic stiffness of the ESF-IMP construct (229.19 N/mm, std error=17.302) was significantly less ($p<0.0001$) than that of the DB-PLATE construct (277.46 N/mm, std error=17.302). A significant difference between median stiffness of DB-PLATE in compression load to failure and that of ESF-IMP **(Table 3)** was not detected. Yield load of DB-PLATE was higher than that of ESF-IMP **(Table 3)**.

Failure mode for all ESF-IMP specimens included collapse of the ostectomy gap with bending of the cross bar (n=10) **(figure 4).** Additional findings included a bent intramedullary pin (n=3) **(figure 4)**, fracture of the caudal cortex of the proximal segment created by the intramedullary pin (n=2), distal intramedullary pin migration (n=1) and ulnar fracture (n=1). Mode of failure for the DB-PLATE constructs included failure of the distal screws by bending (n=3) **(figure 4)**, bending and breaking (n=1) and screw cut-out (n=1). In the three specimens with screw bending, plate screws of both the medial and lateral plates were affected. Screw breakage and cut-out were only seen in screws engaged in lateral String of Pearls plates. Another mode was fracture of the proximal segment through the holes created by the two distal screws of the lateral String of Pearls plate (n=1). The connection between the bone-implant construct and the testing machine broke down prior to construct failure in 4 specimens, including ulnar fracture (n=3) and displacement of the proximal humerus from the PMMA bed (n=1). Testing of these four constructs produced the three lowest yield load values.

## Discussion

Consistent with the hypothesis, DB-PLATE had greater dynamic stiffness than ESF-IMP in cyclic testing and was stronger as measured by compression yield load. However, no advantage was noted for DB-PLATE over ESF-IMP during cyclic testing regarding construct failure or maintaining stiffness. In compressive load-to-failure testing, no difference in stiffness was found between the constructs.

All test constructs withstood the cyclic loading protocol, designed to simulate 30 minutes of walking per day during a one-month convalescent period. Additionally, no constructs lost stiffness over the course of 63,000 cycles. This suggests that either construct type would be adequate for stabilization of similar fractures in patients with appropriate exercise restriction for at least one month.

As hypothesized, the DB-PLATE construct was stronger than the ESF-IMP construct as measured by yield force in compressive load to failure. Premature failure of the connection between four of the constructs and the testing machine was considered to be a failure of the model, not of the fixation construct. This likely caused an underestimation of the median yield force for the DB-PLATE construct. Even with this limitation, DB-PLATE constructs reached a median yield force of 845.7 N before beginning to fail. This load is more than four times the estimated load applied to the thoracic limb at the walk.

Amongst the remaining six DB-PLATE specimens exhibiting true construct failure, distal screw failure in five was probably due to the limited available bone for screw purchase in this segment. Mode of failure for all ESF-IMP constructs included collapse of the ostectomy gap and bending of the cross bar. Median yield load of the ESF-IMP constructs (501.4 N) is more than twice the estimated load applied to the thoracic limb at the walk (10,11).

The model used was modified from a previous study (1). Maintaining the radio-ulnar articulation with the humerus was advantageous due to the distal placement of implants in the humerus, which makes stabilization of the distal humerus for testing otherwise challenging. Also, load transfer across the elbow is likely more physiologic and less constrained when compared to models that immobilize the distal humerus. The previous model was modified by maintaining the proximal humerus and using PMMA to create a simulated glenoid cavity. Preserving the proximal humerus was necessary for testing of the ESF-IMP construct. This modification eliminated rigid constraint of the proximal humerus, although it also allowed failure of the model in one instance during destructive testing of the DB-PLATE construct. However, the model was adequate for cyclic testing of both constructs and destructive testing of the ESF-IMP construct. It should be noted that this model only provides an approximation of normal ground reaction forces acting on the humerus. In our estimation, forces acting on the tested humeri were primarily craniocaudal bending and axial compression.

Inherent limitations exist for any biomechanical study attempting to simulate *in vivo* clinical conditions. Increasing constraint of the proximal humerus could be considered for future studies, either by preserving the shoulder joint or by rigid fixation of the proximal humerus. Increasing the number of specimens may have allowed detection of further differences between test constructs. A post hoc power analysis indicated that 23 specimens per group would be needed to achieve a power of 0.80 and detect a difference of 33 N/mm in stiffness between groups. Additionally, it should be noted that the results from this study only apply to the tested DB-PLATE and ESF-IMP constructs and should not be extrapolated to other similar constructs. Ideally, cyclic testing would have performed over more cycles to more accurately represent activity during the convalescent period. In this study, this was limited by total available testing time per construct and concerns that increasing testing above 2 Hz would be non-physiologic.

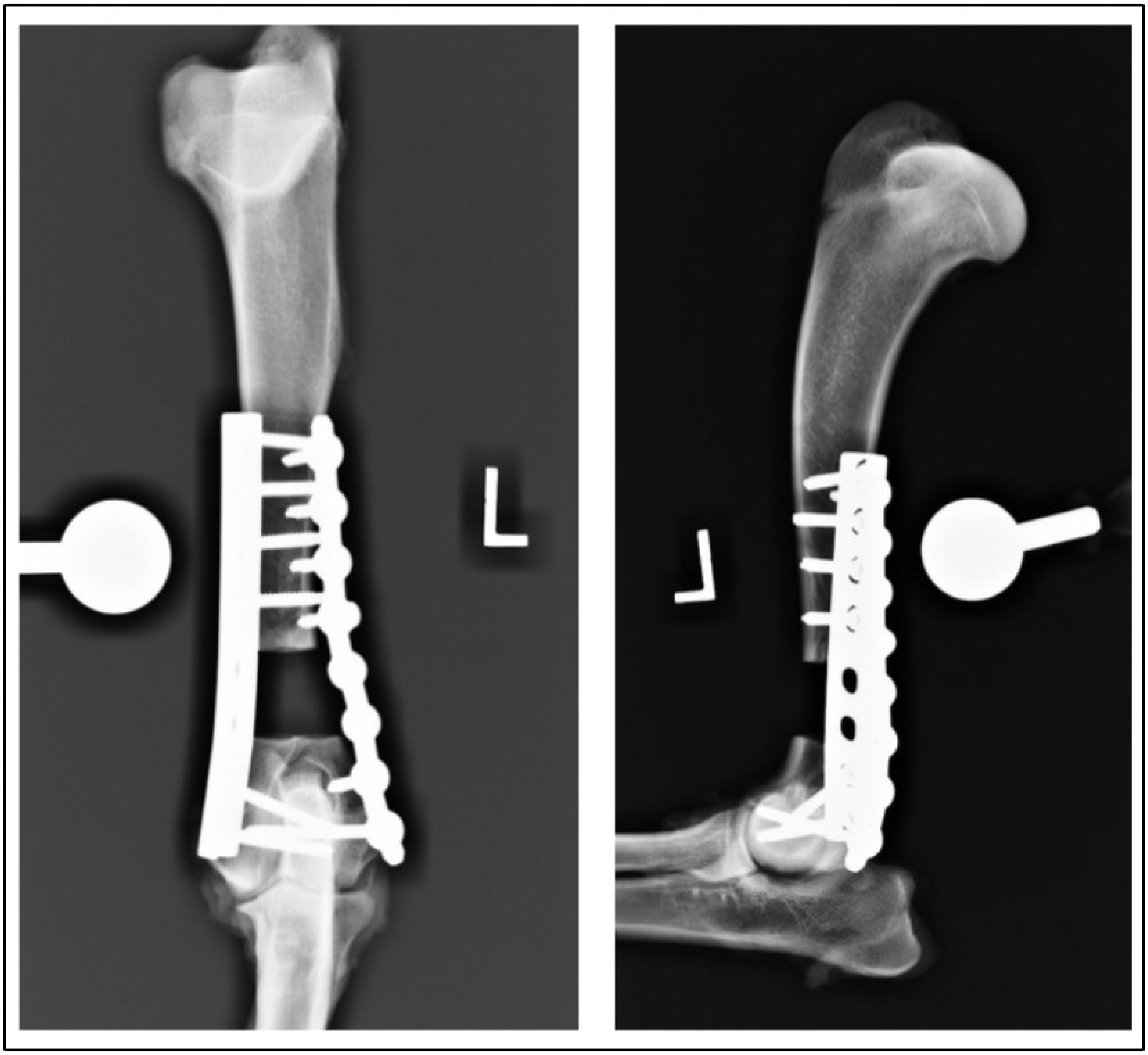


Figure 1 Lateral and craniocaudal radiographic projections of the DB-PLATE construct. The construct consists of a 2 cm cadaveric humeral supracondylar gap model stabilized with a medially placed 8-hole broad compression plate and a laterally placed 8-hole String of Pearls (SOP) plate.

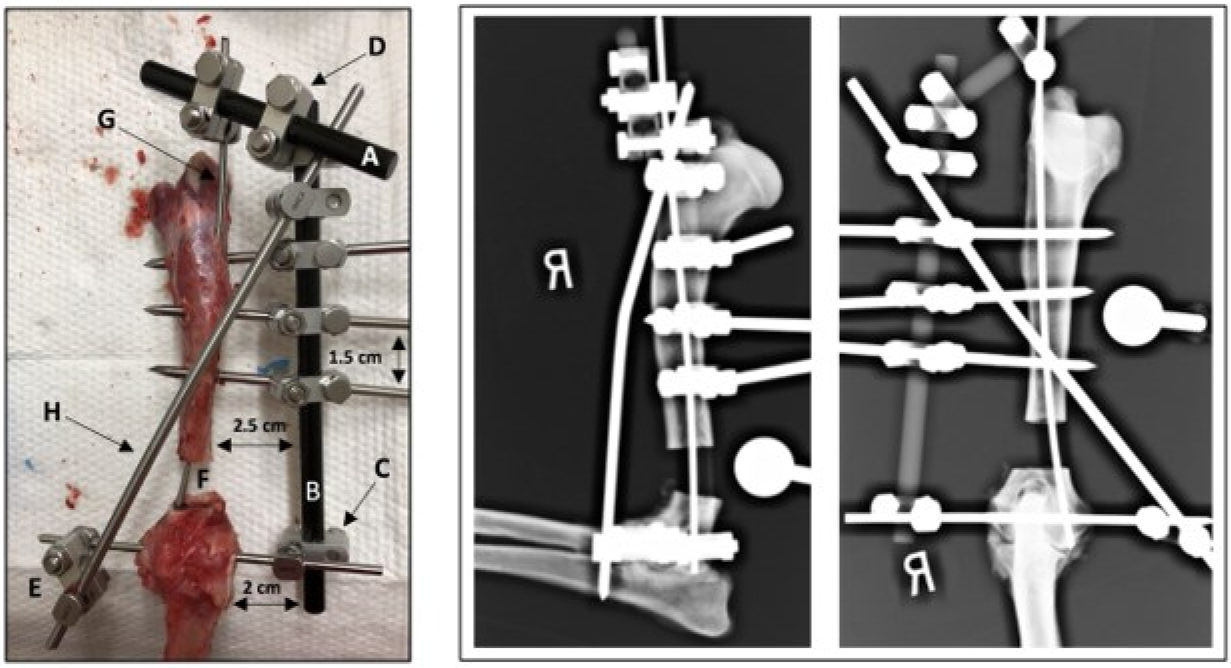


Figure 2 Lateral and craniocaudal radiographs, and a digital image of the ESF-IMP construct. The construct consists of a 2 cm cadaveric humeral supracondylar gap model stabilized with a Type I-II linear external fixator and a tied-in intramedullary pin. A: Carbon-fiber connecting bar (100 mm length, 9.5 mm diameter); B: Carbon-fiber connecting bar (200 mm length, 9.5 mm diameter); C: Single clamp; D: Double clamp; E: Modified clamp made from two single clamps; F: 2 cm ostectomy; G: 3.2 mm intramedullary pin; H: 4.8 mm cross bar

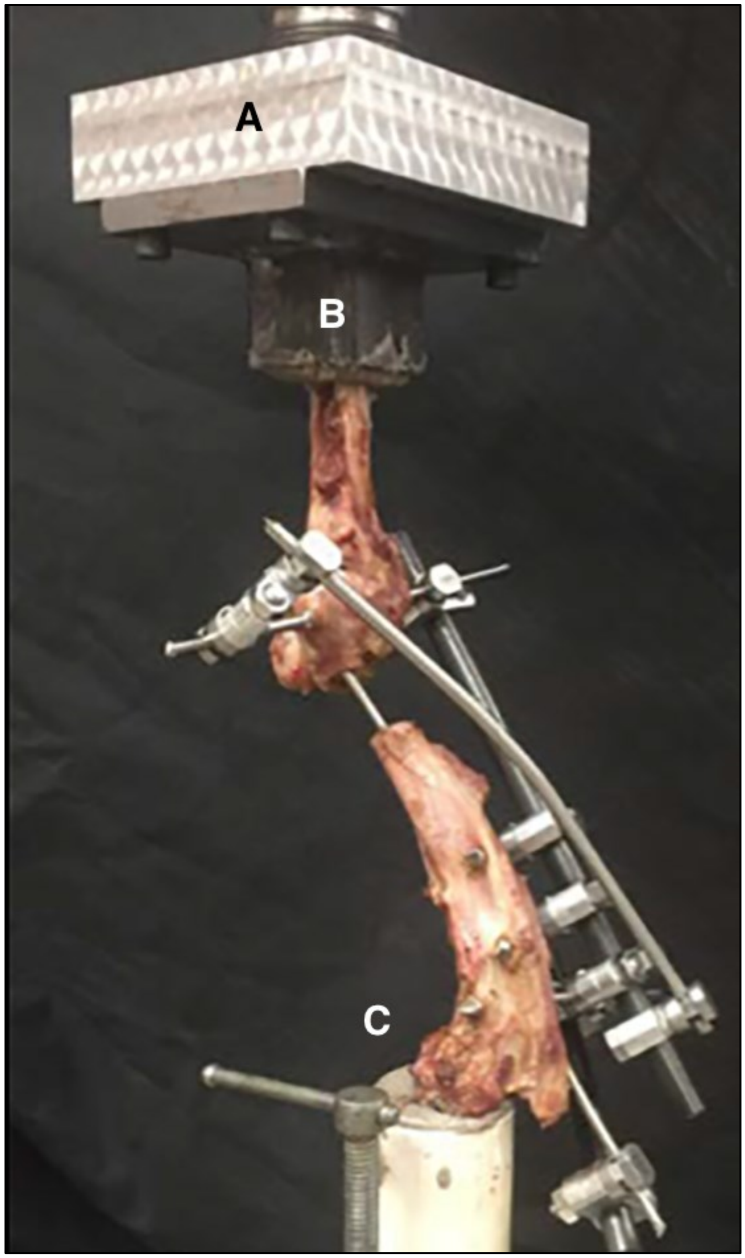


Figure 3 Biomechanical testing setup with test sample in the servohydraulic testing machine. A: Load cell of testing machine; B: Steel fixture with polymethylmethacrylate (PMMA) potted sample; C: PMMA- humeral head interface

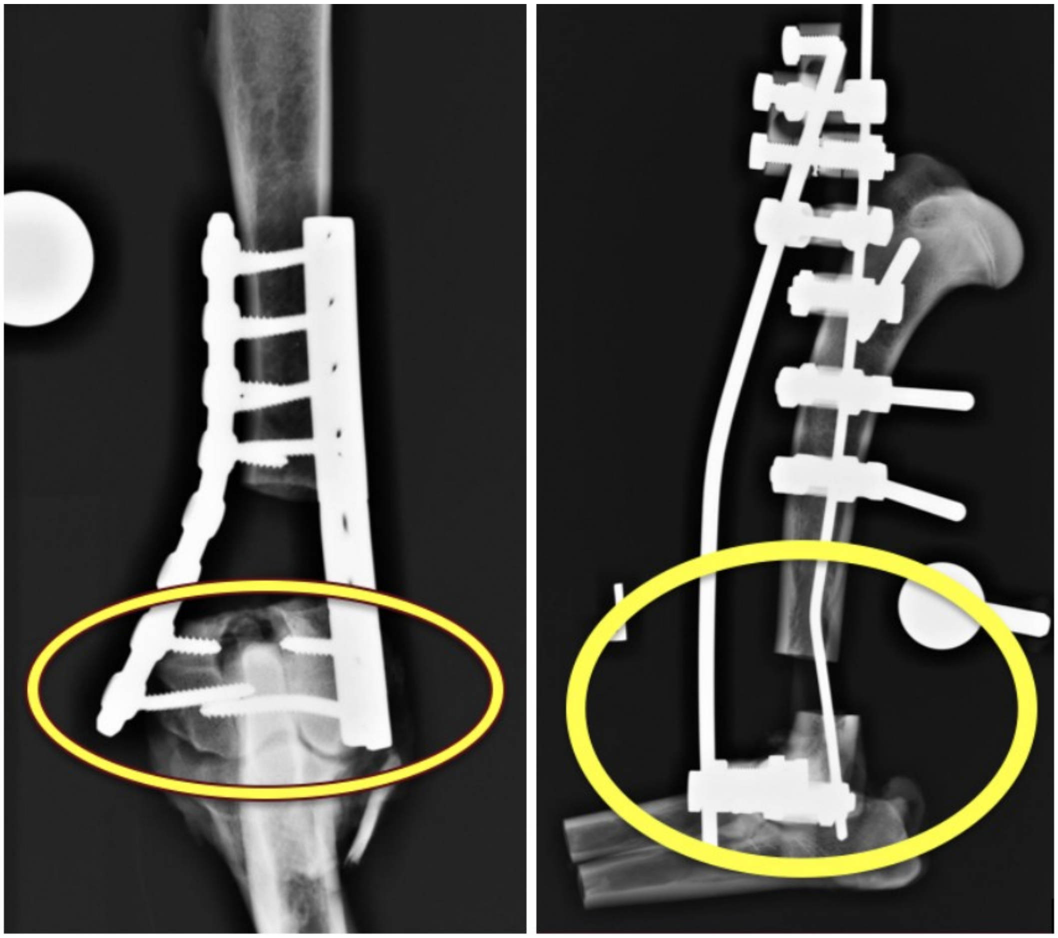

Figure 4 Modes of failure for the DB-PLATE and ESF-IMP constructs. A: DB-PLATE showing distal monocortical screw bending. B: ESF-IMP showing intramedullary pin bending, bending of the cross bar and collapse of the ostectomy gap.

Table 2 Dynamic stiffness of the middle and final 100 cycles of DB-PLATE and ESF-IMP

| | **Middle 100 cycles (N/mm)*** | **Final 100 cycles (N/mm)*** |
|---|---|---|
| **DB-PLATE** | 278.45 +/- 56.76 | 276.47 +/- 58.99 |
| **ESF-IMP** | 228.56 +/- 55.99 | 229.81 +/- 57.03 |
| *** Reported as mean +/- standard deviation** | | |

Table 3 Stiffness and yield load of DB-PLATE and ESF-IMP subjected to destructive compression testing

| | **Stiffness (N/mm)*** | **Yield load (N)*** |
|---|---|---|
| **DB-PLATE** | 180.50 (82.80) | 845.70 (402.60) |
| **ESF-IMP** | 158.95 (45.80) | 501.40 (104.30) |
| **p-value** | 0.1602 | 0.0020 |
| *** Reported as median (interquartile range)** | | |

## Footnotes

I. 3.5 mm screws; Veterinary Orthopedic Implants, St. Augustine, FL.

II. 3.5 mm broad compression plate; Veterinary Orthopedic Implants, St. Augustine, FL.

III. String of pearls (SOP) 3.5 mm interlocking plate; Orthomed(UK) Ltd., West Yorkshire, Halifax, UK.

IV. 3.5 mm screws; Orthomed(UK) Ltd., West Yorkshire, Halifax, UK.

V. Centerface fixation full-pin; IMEX Veterinary, Inc., Longview, TX.

VI. Duraface fixation half-pin; IMEX Veterinary, Inc., Longview, TX.

VII. Large SK carbon fiber connecting rod; IMEX Veterinary, Inc., Longview, TX.

VIII. Large SK single clamp; IMEX Veterinary, Inc., Longview, TX.

IX. Trocar/trocar smooth pin; IMEX Veterinary, Inc., Longview, TX.

X. Large SK double clamp; IMEX Veterinary, Inc., Longview, TX.

XI. Technovit; Jorgenson Laboratories, Loveland, CO.

XII. Bionix 858: MTS Corporation, Eden Prairie, MN, USA

SAS Version 9.4; SAS Institute, Inc., Cary, NC.

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# CHAPTER III

# CONCLUSION

In this biomechanical study utilizing a cadaveric supracondylar humeral gap model, all DB-PLATE and ESF-IMP constructs produced no implant loosening or failure in a cyclic loading protocol simulating limb use during a simulated 4-week convalescent period. In addition, median yield force for each construct was more than twice the calculated force applied at a walk for a 30 kg dog. These results suggest that both DB-PLATE and ESF-IMP would be appropriate fixation techniques for stabilization of comminuted supracondylar humeral fractures in dogs with appropriate exercise restriction. Double plate fixation may be preferable when prolonged healing or inadequate post-operative restraint was anticipated because it was stronger in destructive testing. However, although not evaluated by this study, the more invasive approaches necessary for medial and lateral plate placement may have an adverse effect on blood supply and bone healing.