2009

Biomechanical evaluation of a 6.5-mm headless, tapered variable pitch screw (Acutrak Plus) in equine and synthetic bone

Andrew James Czak Lewis
Louisiana State University and Agricultural and Mechanical College

BIOMECHANICAL EVALUATION OF A 6.5-MM HEADLESS, TAPERED VARIABLE PITCH SCREW (ACUTRAK PLUS) IN EQUINE AND SYNTHETIC BONE

A Thesis

Submitted to the Graduate Faculty of the
Louisiana State University and
Agricultural and Mechanical College
In partial fulfillment of the
requirements for the degree of
Master in Science

in

The Interdepartmental Program in
Veterinary Medical Sciences
through the
Department of Veterinary Clinical Sciences

by
Andrew J. C. Lewis
D.V.M., Louisiana State University, 2005
May, 2009

DEDICATION

I would like to dedicate this work to two people:

My wife, Crystal, for her continued patience, support, sacrifice, and love throughout both my residency training and graduate studies. She has certainly put up with a lot, and I wish to thank her for that. Sorry for the book in the wall!

And, to the G-Unit…

ACKNOWLEDGEMENTS

The author would like to thank his major professor, mentor, and dear friend, Dr. Gary A. "G-Unit" Sod for all of his guidance throughout his graduate program. The author could have not made it through this program without his help, support, and provision of mental balance.

The author would also like to thank the other members of his committee, Drs. Daniel J. Burba, and Colin F. Mitchell, for guidance and support during the research, as well as critical review of the manuscript and providing recommendations on its improvement.

The author would also take this time to thank the members of the surgery faculty at LSU-SVM, Drs. Daniel J. Burba, Colin F. Mitchell, Charles T. McCauley, Mustajab H. Mirza, and Laura M. Riggs for training, mentoring, and support throughout his residency program. The author would like to express his sincere appreciation to Dr. Jeremy D. Hubert, for his friendship, support, and advice throughout veterinary school and his residency program. The author would also like to thank his brother, Dr. Nelson Lewis, for his support over the years.

Finally, the author would like to thank his parents, John and Sophia Lewis, for all of their support throughout the years and telling him he could do anything he set his mind to which has enabled him to achieve his goals thus far.

TABLE OF CONTENTS

LIST OF TABLES

LIST OF FIGURES

ABSTRACT

Objectives – To compare compression pressure (CP) of 6.5mm Acutrak Plus (AP) and 4.5mm AO cortical screws (AO) when placed in simulated lateral condylar fractures of variable thickness in cadaveric equine third metacarpal (MC3) bones, and to compare pullout force and strength of AP and AO screws when placed in a synthetic bone substitute.

Results – The AO screw configurations generated significantly greater compressive pressure compared to the AP configurations. The ratio of mean CP for AP screws to AO screws at 20, 12, and 8-mm, were 21.6%, 26.2%, and 34.2% respectively. For the pullout study, the AP constructs generated a significantly higher pullout force and strength (60% greater) when compared to the AO constructs.

Conclusion – Mean CP for AP screw fixations are weaker than those for AO screw fixations, most notably with the 20 mm fragments. The 12- and 8-mm groups have comparatively better compression characteristics than the 20-mm group, however they are still significantly weaker than the AO fixations. Pullout characteristics for AP screws were expected however don't correlate with compressions characteristics based mainly on screw thread design.

Clinical Relevance – Given that the primary goals of surgical repair are to achieve rigid fixation, primary bone healing, and good articular alignment, based on these results, it is recommended that caution should be used when choosing the AP screw for repair of lateral condylar fractures, especially complete fractures. Since interfragmentary compression plays a factor in the overall stability of a repair, it is recommended for use only in patients with thin lateral condyle fracture fragments, as the compression tends to

decrease with an increase in thickness. Typically, pullout and compression characteristics are directly proportional for many compression screws, but based on screw thread design, most notably the thread pitch and angulation, greater pullout characteristics are expected with the AP screw without the concurrent increased compressive force and strength. Further works needs to be performed including single cycle to failure and cyclic fatigue testing prior to use in clinical cases.

CHAPTER 1

GENERAL INTRODUCTION

1.1 The Metacarpo(-tarso)phalangeal Joint

The equine metacarpophalangeal and metatarsophalangeal joints (fetlock joints) are made up of the third metacarpal bone, both proximal sesamoid bones, and the first phalanx. It is classified as a ginglymus (hinged) joint, thus it has only one axis of movement in the sagittal plane through flexion and extension. This limitation is due to the presence of the prominent medial and lateral collateral ligaments of the fetlock. It is the one joint in the horse with the highest range of motion ranging from 120° of extension to 120° of flexion. This is best appreciated during athletic activity such as racing or jumping (Bertone, 2004). In the standing horse, the fetlock is supported by the suspensory ligament, intersesamoidean ligament, distal sesamoidean ligaments, deep and superficial flexor tendons, the common digital and lateral digital extensor tendons, and the medial and lateral collateral ligaments.

The locomotor functions of the digit and fetlock include the flexion essential to movement, extension when the foot is off the ground, the diminution of concussion when the hoof contacts the ground, and the recovery from extension (Kainer, 2002). The normal movemements of the digit were elaborated by Rooney (1974). When the body weight of the horse is applied to the leg, the fetlock joint extends and translates downward as a result of distal interphalangeal joint motion. With extension, the cannon bone and the first phalanx (P1) tend to rotate about their long axes in a medial to lateral direction. As the limb is unloaded the fetlock joint begins to flex, allowing the dorsal angle of the fetlock to open, and the pastern elevates because of the distal interphalangeal joint motion in the opposite direction. During this opening and elevating movement, the cannon bone and the first phalanx again rotate, however this time about their long axes in a lateral to medial direction. This movement necessitates that the two bones must move in synchrony.

1.2 The Equine Third Metacarpal Bone

The MC3 is a substantial bone, measuring approximately 26 cm in length from the midsagittal ridge to the carpometacarpal joint. At the distal metaphysis, it is approximately 50 wide from the lateral to medial supracondylar fossae. Cortical and trabecular densities for equine bone have been previously measured and determined to be 1035.25 mg hydroxyapatite/ml and 1048.55 mg hydroxyapatite/ml, respectively (Furst et al, 2008). Mean tensile strengths and modulus elasticity have been reported to be 2137.9 to 2295.7 MPa (El Shorafa et al, 1979) and 16.3 GPa (Batson et al, 2000), respectively.

1.3 Development of Condylar Fractures

It has been theorized that condylar fractures result due to asynchronous movement between the cannon bone and the proximal phalanx. The asynchronous movement leading to lateral condylar fractures was described (Alexander and Rooney 1972, Rooney 1974) as the proximal phalanx remains stationary with the fetlock in a fully dorsiflexed position while the cannon bone rotates in a lateral to medial direction, owing to fracture of the lateral condyle as it strikes the stationary lateral aspect of proximal P1. This asynchrony would be expected near the end of the support phase of the stride when the fetlock joint dorsal angle is opening (Rooney 1974). No work has been performed to support this theory, however more recent efforts have been made to determine predisposing factors associated with condylar fracture formation.

Recent work evaluating the pathophysiology of the development of condylar fractures suggests that they originate from microfractures in the subchondral bone that develop secondary to repetitive loading from exercise related stress remodeling (Fig. 1.1A and 1.1B), and is associated with significant cyclic shear loading of the condyle in dorsopalmar (-plantar) bending which then leads to propogation of a dominant crack proximally (Radtke et al. 2002). Ultrastructural work further revealed that these microfractures develop into clusters and most notably in the palmar aspect of the condyle; these macrofractures propogate along cement lines and interfaces between bone lamellae (Fig. 1.2A and 1.2B) (Stepnik et al. 2004).

Condylar fractures involving the third metacarpal and metatarsal bones have long been recognized as a significant problem among racehorses, and in severe cases can require euthanasia (Rooney 1974, Pool 1990, Wilson et al 1992, Stover 1994, Kane et al 1996, Kane et al 1998, Estberg 1998). Severe condylar fractures account for 20% (Johnson et al, 1994) to 25% (McKee, 1995) of catastrophic injuries in Thoroughbred racehorses in California and the United Kingdom, respectively. Condylar fractures of the left third metacarpal bone are the most commonly noted site in Thoroughbred racehorses, accounting for 39% to 57.1% of all condylar fractures (Rick et al. 1983, Ellis et al. 1994); whereas there appears to be a more even distribution with regards to left vs. right in Standardbred racehorses. This increased prevalence for occurrence in the left forelimb among Thoroughbred racehorses has been theorized to be due to the direction in which racehorses are run, which places the left limb on the inside, thus increasing the load placed on this limb, when compared to the right limb. This same observation has been made in England, however one study reported that the majority of condylar fractures were incurred during training, which involves galloping mainly straight distances (Ellis et al. 1994).

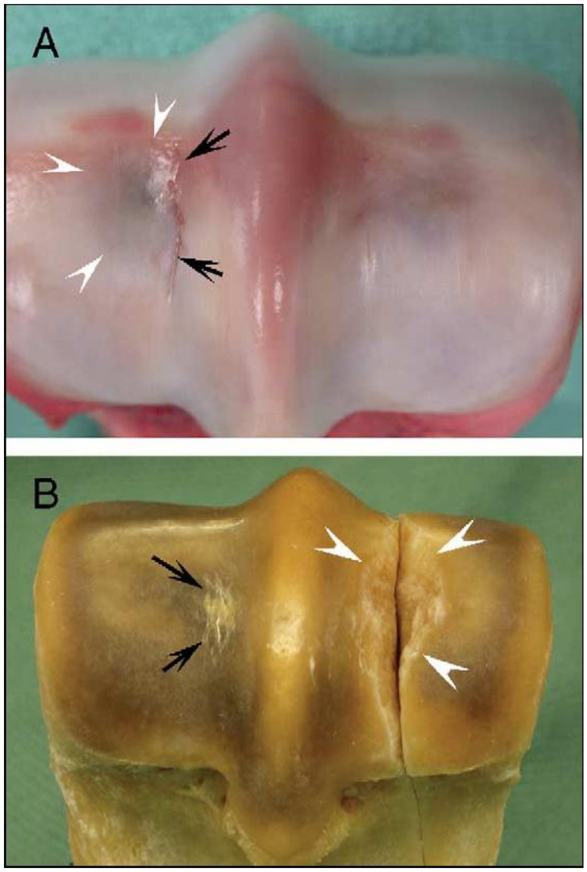


Figure 1.1. Photographic view of the palmar distal region of the distal joint surfaces of the MC3 bones from a 3-year-old male Thoroughbred. (A) In the right MC3 bone, a parasagittal defect can be seen in the articular cartilage of the lateral condylar groove (black arrows). Adjacent to this lesion is circular area of cartilage degeneration in the lateral condylar (white arrow heads). A similar lesion is also present to a lesser extent in the medial condyle. Parasagittal linear wear lines in the articular cartilage are also visible. (B) In the left MC3 bone, a parasagittal condylar fracture is present in the lateral condylar groove (white arrowheads). In the medial condylar groove, a branching array of subchondral cracks can be seen (black arrows). In the lateral condylar groove, comminution of this subchondral bone developed during propagation of the fracture. The articular cartilage was removed by treatment with 0.1 M NaOH to permit the articular surface of the subchondral bone to be examined. (Pictures taken from Radtke et al. 2002)

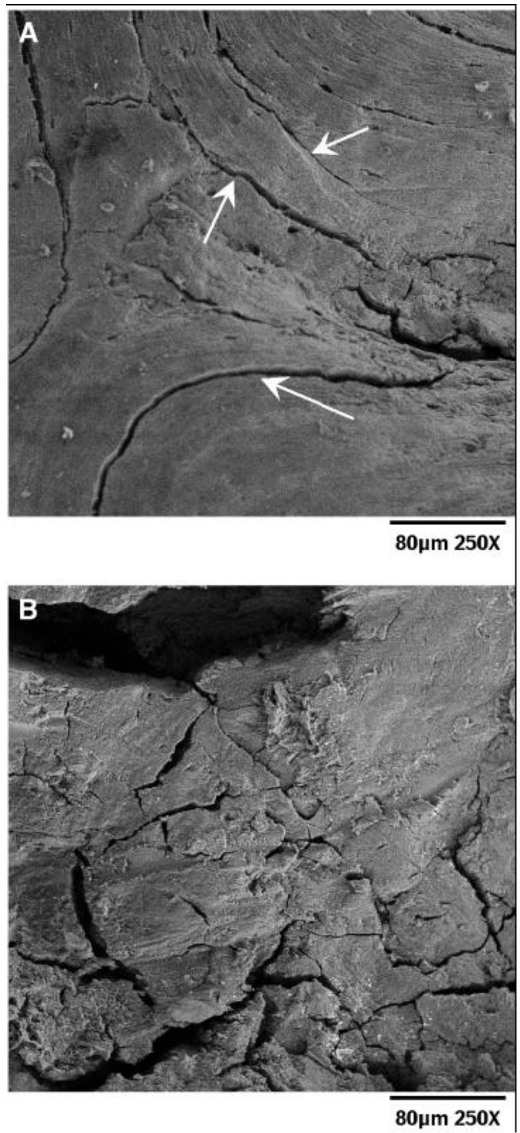


Figure 1.2. Scanning electron microscopic views of the failure surface from a 5-year-old racing Thoroughbred with a catastrophic MC3 lateral condylar fracture. (A) In adapted palmar subchondral bone from the distal end of the MC3 bone, microcracks were often seen propagating along cement lines and the interfaces between bone lamellae. (B) In adapted dorsal subchondral bone from the distal end of the MC3 bone, arrays of branching microcracks were also seen, which were similar to the palmar/plantar region of the condyle. (Pictures taken from Stepnik et al. 2004)

Condylar fractures can occur in either the lateral or medial condyle; however, fractures of the lateral condyle are the most commonly occurring type (Alexander et al 1972, Meagher 1976, Rick et al 1983, Ellis 1994, Johnson et al 1994, Bassage et al 1998, Richardson 1999, Zekas 1999, LeJeune 2003), reported as being approximately 85% in various racehorse populations (Zekas et al 1999; Bassage et al 1998). Condylar fractures are typically described as being one of four types: 1) incomplete – no evidence of joint malalignment or complete extension of the fracture through the proximal cortex; 2) complete-nondisplaced fractures – neither a step at the joint surface nor evidence of separation of the fragment proximally – although the fracture line penetrates through the cortex; 3) complete-displaced fractures – malalignment at the joint surface and abaxial displacement at the proximal cortical surface; 4) special longitudinal diaphyseal fractures – either complete or incomplete – involving the medial condyle and extending various distances up the diaphysis (Figure 1.3) (Rick et al. 1983).

Historically, an onset of severe lameness following intense exercise occurs typically either immediately following injury or within a few hours of exercise. Typical physical examination findings reveal a variable lameness which does not correlate well with severity of the fracture, as a severe non-weightbearing lameness is commonly seen with incomplete, nondisplaced fractures, yet a milder weightbearing lameness is typically appreciated with displaced condylar fractures (Richardson, 2006). Additional findings include variable effusion of the metacarpo- or metatarso-phalangeal joint of the affected limb, elicitation of pain on flexion of the joint, and either the presence or absence of crepitation on palpation of the limb. Radiographic evaluation of the fetlock joint typically confirms the presence of condylar fracture. Traditional radiographic projections including the lateromedial and dorsopalmar (-plantar) are diagnostic, however it is strongly recommended that a horizontal dorsopalmar projection with the fetlock placed in slight flexion be performed highlighting the palmar aspect of MC3/MT3 to rule out the presence of palmar comminution, a common finding associated with condylar fractures (Richardson, 2006).

Treatment goals for type 1, 2, and 3 condylar fractures are aimed at either conservative management consisting of prolonged periods of stall confinement, immobilization via external coaptation by heavy bandaging or distal limb casting (Meagher, 1976, Rick, et al. 1983), or surgical reduction; however, surgical reduction is typically recommended for the best possible prognosis. Surgical repair is most commonly performed through fracture fragment fixation via placement of either 4.5-mm or 5.5-mm AO cortical bone screws (Synthes, Paoli, PA) using the lag screw principle to allow for adequate compression and allows for fracture healing as well as to align the articular surfaces of the fracture to minimize the chances for development of degenerative joint disease (Richardson. 2006). Another recently proposed method of reduction that is being performed in clinical cases in nondisplaced lateral condylar fractures is the use of a headless, tapered, variable pitch screw (Acutrak Plus) bone screw (Acumed, Beaverton, OR).

1.4 The AO/ASIF

The Arbeitsgemeinschaft für Osteosynthesefragen (Association for the Study of Internal Fixation) was formed in 1958 to further research into the concepts of immediate functional rehabilitation after rigid internal fixation. They have done this through research into osteosynthesis, and the development of instruments and implants that

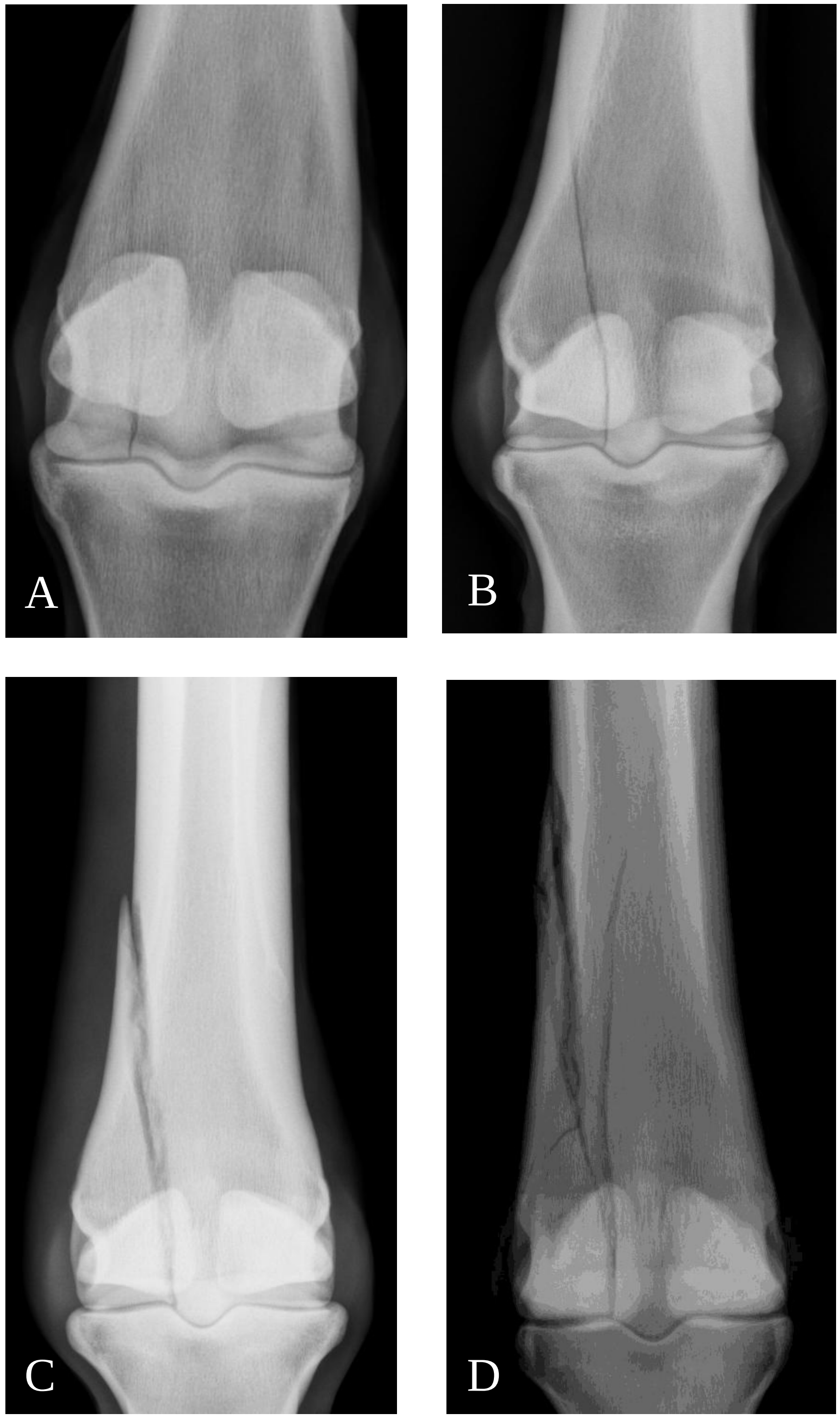


Figure 1.3. Standard dorso-palmar radiographic images of a Type 1 (A), Type 2 (B), Type 3 (C), and Type 4 (D) condylar fractures.

promote rigid internal fixation. This has evolved into a worldwide organization evaluating not only the development of implants in human orthopedics, but as well in veterinary orthopedics (Colton. 1981).

1.4.1 The 4.5-mm AO Cortical Screw

The 4.5-mm AO cortical screw is made of implant quality 316L stainless steel, which contains roughly 62.5% iron, 17.6% chromium, 14.5% nickel, 2.8% molybdenum, and minor alloy additions. It is meant to be resistant to corrosion due to its low carbon content (Texhammar. 1981).

This screw was designed for use in the lag principle or for plate fixation. It is a fully threaded non-self-tapping screw. In cortical bone, the screw has a holding strength of approximately 2500 N.

The dimensions of the screw are as follows:

Head diameter:	8.0-mm
Hexagonal socket width:	3.5-mm
Core diameter:	3.0-mm
Thread diameter:	4.5-mm
Pitch:	1.75-mm
Glide hole diameter:	4.5-mm
Thread hole diameter:	3.2-mm
Tap diameter:	4.5-mm

The 4.5-mm cortical screw also has characteristics similar to that of the other large AO/ASIF screws including first a spherical screw head which ensures optimal screw to plate contact even if a screw is placed at an angle. Second, the AO/ASIF screw

thread which is characterized as a buttress thread profile which allows excellent holding in cortical bone due to the shallow thread and fine pitch leading to a large screw-to-bone contact area. The third common characteristic is the core diameter which is the solid stem of the screw from which the threads protrude.

1.4.2 The Lag Screw Principle

The lag screw was defined by Perren and Buchanan (1981) as *the production of interfragmental compression by compressing the bone under the screw head against the fragment in which the screw threads are anchored.*

The steps for proper lag screw principle for a 4.5-mm cortical screw are as follows:

1. The fracture is reduced and held with reduction forceps.
2. The glide hole is drilled through the near cortex or fragment with a 4.5-mm drill bit protected by the drill sleeve.
3. The 3.2-mm drill sleeve is then inserted into the glide hole until it comes into contact with the far cortex or the parent bone.
4. The thread hole is then drilled in a coaxial direction in the trans-cortex or parent bone with a 3.2-mm drill bit.
5. The cis-cortex is countersunk with the large countersink.
6. Screw length is then measured with the large depth gauge
7. The thread hole is then tapped with a 4.5-mm bone tap. By turning two turns clockwise and one-half turn counterclockwise, the cut bone is directed into the channels of the cutting flutes to be removed.
8. A 4.5-mm cortical screw is inserted with the large hexagonal screwdriver, ensuring engagement of the trans cortex.

1.5 The Acutrak Screw Series

The Acutrak Plus (AP) screw is a titanium alloy 6.5-mm headless, variable-pitch, self-tapping, tapered screw (Galuppo et al. 2001; Galuppo et al. 2002; Carpenter et al. 2006; Galuppo et al. 2006). The Acutrak series of screws was first introduced in the late 1980's in human orthopedics and has been utilized for repair of scaphoid waist fractures of the wrist (Haddad et al. 1998; Dao and Shin. 2003, Slade and Merrell 2003; Geissler 2006), fifth metatarsal metaphyseal-diaphyseal stress fractures (Casillas and Strom 2006), distal interphalangeal joint arthrodesis (Rehak 2007), and first metatarsal basal osteotomy fixation (Fadel et al. 2007). Reported applications for the Acutrak screws in veterinary medicine are limited at this time but do include repair of equine nondisplaced lateral condylar fractures (Galuppo et al. 2006), equine frontal plane third carpal bone slab fractures (Hirsch et al. 2007), and bovine distal interphalangeal joint arthrodesis (Lewis et al. 2008).

1.5.1 Geometry of the Acutrak Plus, 4.5mm AO Cortical and 5.5mm AO Cortical Screws

The thread surface profile of a 45 mm Acutrak plus screw in comparison to a 4.5mm AO cortical screw and 5.5mm AO cortical screw are described below:

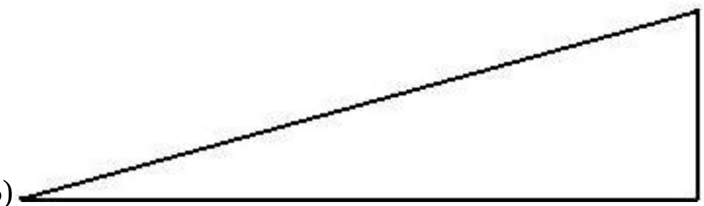
Acutrak Plus 45 mm Length Screw

Taper 5.0 mm Diameter → 6.5 mm Diameter or 2.5 mm Radius → 3.25 mm Radius

Broad Pitch 2.15 mm with 15 threads

Narrow Pitch 1.65 mm with 7 threads

Thread Depth 1.25 mm



Slope = 0.75 / 45 = 0.01667

Y_{outer} – 2.5 / X = 0.01667 → Y_{outer} = 0.1667X + 2.5

Y_{inner} – 1.25 / X = 0.01667 → Y_{inner} = 0.1667X + 1.25

Area of Annulus = Π (Y^2_{outer} – Y^2_{inner}) = Π [(0.01667X + 2.5)^2 – (0.01667X 1.25)^2]

$(a+b)^2 - (a+c)^2 = a^2 + 2ab + b^2 - (a^2 + 2ac + c^2)$

$= 2a (b - c) + b^2 - c^2$

→ Area of Annulus = Π[2X 0.01667 x (2.5 – 1.25) + (2.5^2 – 1.25^2)]

= Π[0.0417X + 4.6875]

= 0.131X + 14.73

Course Pitch

$$\Delta_c = \sum_{l=1}^{15}(0.131x_1 + 14.73) = 0.131\sum_{l=1}^{15} x_i + 220.95$$

$$x_i = 0.5 + (i-1)\times pitch = 2.15(i-1) + 0.5$$

$$\Delta_c = 0.131\sum_{l=1}^{15}(2.15(i-1) + 0.5) + 220.95$$

$$= 0.282\sum_{l=1}^{15} l - 1 + 0.49 + 220.95$$

$$= 0.282(105) + 221.44 = 251.05mm^2$$

Fine Pitch

$$\Delta_F = \sum_{l=16}^{22}(0.131_{i'} + 14.73) = 0.131\sum_{l=16}^{22} x_{i'} + 103.11$$

$$x_i \leq x_{16} + i' \times pitch = 32.75 + 1.65i'$$

$$\Delta_F = 0.131\sum_{l=1}^{7}(1.65\boldsymbol{i} + 32.75) + 103.11$$

$$= 0.131 \times 1.65(28) + 0.131 \times 32.75 + 103.11$$

$$= 6.052 + 4.29 + 103.11$$

$$\rightarrow \Delta_F = 113.45$$

Total Surface Area of Threads = Δ_C + Δ_F = 364.5 mm^2

AO Cortical Screw 4.5mm 46 mm long

Pitch = 1.75 mm
Core Radius = 1.5mm
Thread Radius = 2.25 mm

Area of Annulus = Π[2.25^2 – 1.5^2] = 8.836 mm^2

$$\#Threads = \frac{Length}{Pitch} = \frac{46mm}{1.75mm} = 26$$

Total Surface Area of Threads = 26 x 8.836 mm^2 = 229.7 mm^2

*****AO 4.5mm has 63% of thread surface area of Acutrak Plus 45mm screw*****

AO Cortical Screw 5.5 mm   46mm Long

Pitch = 1.75mm
Core Radius = 1.95mm
Thread Radius = 2.75mm

Area of Annulus = Π[2.75^2 – 1.75^2] = 11.81mm^2

$$\#Threads = \frac{Length}{Pitch} = \frac{46mm}{1.75mm} = 26$$

Total Surface Area of Threads = 26 x 11.81mm^2 = 307.1 mm^2

*****AO 5.5 mm has 84% of thread surface area of Acutrak Plus 45mm screw*****

1.5.2 Biomechanical Studies on the Acutrak Screw

Interfragmentary compression is considered to be one of the most important properties involved in determining the stability of a fracture fixation. Compression promotes bone healing, provides stability, and resists rotation of the components (Cosio MQ et al. 1986; Inoue G et al. 1991; Wozasek GE et al. 1991; Whipple TL 1992; Ledoux P et al. 1995; Whipple TL 1995; Faran et al. 1999; Slade JF et al. 2002; Adla et al. 2004). Previous biomechanical studies have been performed evaluating interfragmentary compression of the Acutrak series of screws both in human and veterinary orthopedics (Faran et al. 1999; Galuppo et al. 2002; Adla et al. 2004; Eddy et al. 2004; Hausmann et al. 2006; Bailey et al. 2006). Hausmann et al. (2006) concluded in an interfragmentary compression test in synthetic bone material that the Acutrak screw had a significantly higher mean compression force than the 3.0 mm AO cannulated cancellous screw and the Headless Bone Screw (HBS; Heintl/Martin, Tuttlingen/D). In another study comparing compression forces of the Acutrak Mini screw, the Herbert/Whipple screw, the 3.0 mm AO cannulated cancellous screw, and the AO 2.0 mm cortical screw, the Acutrak generated higher interfragmentary compression forces compared to the 2.0 mm cortical screw, but generated 70% of the interfragmentary compression force of a cancellous screw (Adla et al. 2004). Galuppo et al. (2002) reported in an interfragmentary compression study of the Acutrak Plus and the 4.5mm AO cortical screw in simulated equine third metacarpal bone lateral condylar fractures that the Acutrak Plus screw generated 65% and 44% the compressive pressure and force, respectively, to that of the AO screw.

Reported advantages of this screw over the traditional stainless AO cortical screw include reported faster screw preparation and placement time (Galuppo et al. 2001), increased stability by providing compression and greater bone to screw contact due to its tapered and variable pitch design (Galuppo et al. 2001; Galuppo et al. 2006), and decreased irritation and impingement of the lateral collateral ligament and metacarpophalangeal joint capsule due to its headless design (Galuppo et al. 2001; Galuppo et al. 2006). The latter was recognized early on in human orthopedics with the development of the Herbert screw (Herbert 1986). However, there are also disadvantages of this screw design, including less compression as the screw cannot be placed in true lag fashion compared to an AO cortical screw (Galuppo et al. 2001), and difficulty in placing the screw as this can be a technically challenging technique to learn (Galuppo et al. 2006). If this lack of compression compared an AO screw is significant, this could lead to fixation failure during continuous cyclic loading. This coupled with the difficulty of removal of the Acutrak screw, may lead to an unfavorable end-result for return to soundness in horses. An unknown point to consider is regarding to fracture fragment size and the ability for the Acutrak screw to have enough purchase within that fragment to reach its maximal compression. Since the screw is not placed in lag fashion, there is question as to whether or not distraction of the fracture fragment can occur while tightening the screw, as may be seen with an AO cortical screw not placed in lag fashion through a fracture. These questions led to the development of this author's Masters' project. Thus, the specific objectives of this study were to: 1) evaluate and compare *in vitro* the interfragmentary compression pressure with regards to fracture fragment thickness following placement of a single 6.5-mm AP screw versus a single 4.5-mm AO cortical screw when placed in artificially created lateral MC3 condylar fractures of equine cadaver limbs (Chapter 2); 2) evaluate and compare *in vitro* the pullout force and strength of 6.5-mm AP and 4.5-mm AO cortical bone screws in a synthetic bone substitute (Chapter 3).

CHAPTER 2

COMPRESSION OF CONDYLAR FRACTURES OF VARYING FRAGMENT THICKNESS WITH ACUTRAK PLUS VS 4.5-MM AO CORTICAL SCREWS IN EQUINE THIRD METACARPUS

*Reprinted by permission of Veterinary Surgery

2.1 Introduction

Condylar fractures involving the third metacarpal and metatarsal bones have long been recognized as a significant problem among racehorses (Pool et al, 1990; Wilson et al, 1993; Stover et al, 1994; Kane et al, 1996; Kane et al, 1998). Severe condylar fractures account for 20% (Johnson et al, 1994) to 25% (McKee, 1995) of catastrophic injuries in Thoroughbred racehorses in California and the United Kingdom, respectively. Condylar fractures of the left third metacarpal bone are the most commonly noted site in Thoroughbred racehorses, accounting for 39% to 57.1% of all condylar fractures (Rick et al, 1983; Ellis, 1994).

Treatment goals for incomplete, complete nondisplaced, and complete displaced condylar fractures are aimed at either conservative management consisting of prolonged periods of stall confinement, immobilization via external coaptation by heavy bandaging or distal limb casting (Meagher, 1976; Rick et al, 1983), or surgical reduction; however, surgical reduction is typically recommended for the best possible prognosis. Surgical repair is most commonly performed by fracture fragment fixation via placement of either 4.5-mm or 5.5-mm AO cortical bone screws (Synthes, Paoli, PA) using the lag screw principle to facilitate primary bone repair as well as to align the articular surfaces of the fracture to minimize the development of degenerative joint disease (Richardson, 2006). Reported success rates for lag screw fixation of nondisplaced condylar fractures are good with 70 to 80% of horses returning to full function (Rick et al, 1983; Bassage et al, 1998; Zekas et al, 1999).

Another recently proposed method of reduction that has been performed in clinical cases in nondisplaced lateral condylar fractures is the substitution of the AO cortical screw with the Acutrak Equine bone screw (Acumed, Beaverton, OR), a non-cannulated version of the Acutrak Plus (AP) bone screw (Galuppo et al, 2006). Suggested advantages of this screw over the traditional stainless AO cortical screw include decreased irritation and impingement of the lateral collateral ligament and metacarpophalangeal joint capsule due to its headless design (Galuppo et al, 2001; Galuppo et al, 2006), reported faster screw preparation and placement time (Galuppo et al, 2001), and increased stability by providing compression and greater bone to screw contact due to its tapered and variable pitch design (Galuppo et al, 2001; Galuppo et al, 2006). In a recent retrospective study evaluating the use of Acutrak Equine screw in repair of nondisplaced condylar fractures in Thoroughbred racehorses, 73% successfully returned to racing (Galuppo et al, 2006).

However, there are disadvantages to this screw design, including less fracture fragment compression as the screw cannot be placed in true lag fashion compared to an AO cortical screw (Galuppo et al, 2001), and difficulty in placing the screw as it can be a technically challenging technique to learn (Galuppo et al, 2006). If this lack of compression compared to an AO screw is significant, this could lead to prolonged fracture healing, increased risk for the development of osteoarthritis, and at worse case cause repair failure during continuous cyclic loading, which may lead to an unfavorable end-result for return to soundness. In addition, removal of an Acutrak screw in the event of stripping during placement, or if required, post-operativey, is could be difficult (Eddy et al, 2004).

The objectives of this study are to evaluate and compare *in vitro* the interfragmentary compression pressure with regards to fracture fragment thickness following placement of a single 6.5-mm Acutrak Plus screw versus a single 4.5-mm cortical screw when placed in artificially created lateral MC3 condylar fractures of equine cadaver limbs.

We hypothesize that due to the variable pitch nature of the AP screw, differing fragment thicknesses will lead to variable compression and that the thicker fragments will yield lesser interfragmentary compression due to the location of the most drastic thread pitch change in the screw remaining within the fragment when compared to a 4.5 mm AO cortical screw fixation of condylar fractures in horses.

2.2 Materials and Methods

2.2.1 Acutrak Plus Screw Description

The Acutrak Plus screw is a titanium alloy, headless, cannulated, self-tapping, variable pitched, tapered screw (Galuppo et al, 2001; Galuppo et al, 2002; Galuppo et al, 2006). At its base, it is at its widest diameter of 6.5-mm. This tapers down to 5.0-mm at its apex. The change in pitch occurs gradually throughout the length of the screw, with the widest pitch at the apex, and narrowing towards the base. Close examination of the screw reveals that the most abrupt change in pitch occurs at a point 12-mm from the base of the screw (Fig 2.1). In a 45-mm long AP screw, the most narrow pitched threads encompass approximately 1/3 of the screw with the wider pitched threads covering the remaining 2/3 of the screw. The screw is cannulated to allow pre-placement of a guide wire for positioning purposes, if needed. A modification of the AP screw, called the Acutrak Equine, was of a similar design, however was non-cannulated and had a reinforced screwdriver insert (Galuppo et al, 2006). This screw has since been taken off the market, thus was not utilized in this study.

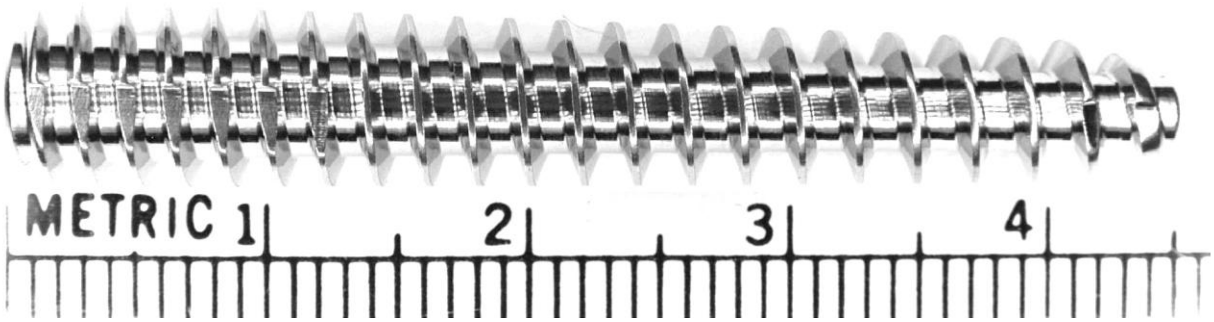


Figure 2.1 Magnified view of the 45-mm long AP screw to show the variable pitch nature of the screw. The most abrupt change in pitch occurs 12-mm from the base of the screw. Also note the cutting flutes at both the base and apex of the screw.

2.2.2 Fragment Compression Testing

Twelve pairs (left and right) of 3rd metacarpal bones (MC3) including the 2nd and 4th metacarpus were collected from adult (2-6 years) Thoroughbred horses euthanatized for reasons unrelated to orthopedic disease. The MC3 were wrapped in saline (0.9% NaCl) soaked towels and stored in pairs at -20°C. The MC3 bones were thawed for 24 hours at room temperature prior to testing. The paired MC3 were randomly assigned to one of 3 groups (n=4) identified by fracture thickness. A complete parasagittal osteotomy was created at 20, 12, and 8-mm axial to the epicondylar fossa on different specimens and grouped accordingly. In each fracture thickness group, one MC3 was randomly selected from each pair for AP screw fixation with the contralateral MC3 receiving the AO screw fixation.

2.2.3 Fracture Preparation

The MC3 was secured in a positioning jig, and a lapidary table saw (Covington Slab Saws) with a blade thickness of 0.8 mm was used first to create a flat surface on the lateral condyle at the level of the epicondylar fossa. Next, the complete lateral condylar osteotomies were created at one of the three fragment thicknesses (Figure 2.2). An additional two osteotomies were created 50- and 60-mm proximal to the distal aspect of the lateral condyle perpendicular to the level of the parasagittal osteotomy. This 10mm section of bone was then removed and discarded (Figure 2.2). The bone was then secured in a positioning jig attached to a drill press, and the condylar fragment was then secured in position to the parent bone using pointed reduction forceps. For the AP screw fixation, a 50-mm deep hole was drilled using a cannulated tapered dense bone bit (Acumed, Inc., Beaverton, OR) at the center of the epicondylar fossa of the lateral condyle, perpendicular to the saw cut, and parallel to the articular surface. Bits were discarded following each use. The hole was then tapped with a 45-mm long, tapered bone tap (Acumed, Inc., Beaverton, OR) (Galuppo et al, 2002).

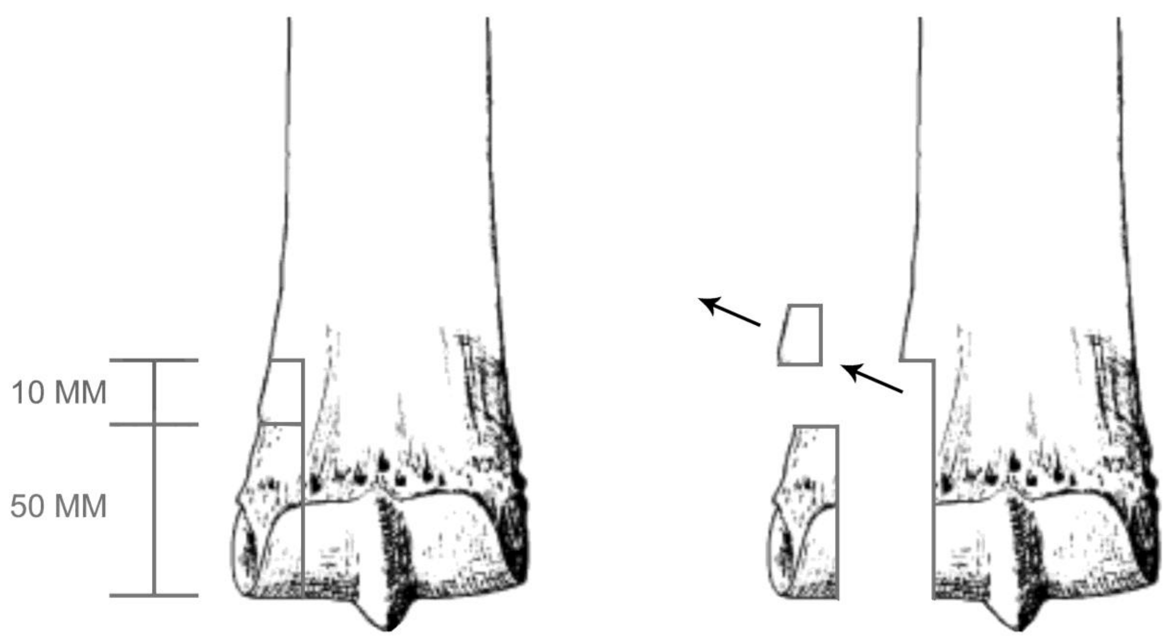


Figure 2.2. Exploded diagram of condylar osteotomy creation. Note that a 10-mm section has been removed to prevent sensor damage.

The construct was secured in a bench vise. The reduction forceps were then removed to allow placement of a circular 0.10-mm thick pressure sensor with a center hole to accommodate screws (I-Scan 6220 Pressure Sensor, Tekscan Inc, Boston MA) connected to a computer. The sensor was then positioned in the osteotomy plane centered around the screw hole. Reduction forceps were then replaced to once again secure the construct. A 45-mm long 6.5-mm diameter Acutrak Plus screw was inserted in the screw hole and tightened until the base of the screw was tightened until the base of the screw was flush with the surface of the supracondylar fossa, providing a consistent reference point.

For the AO fixation groups, the MC3 was secured in the drill press and the osteotomy plane was reduced using pointed reduction forceps. Glide and thread holes were then created and tapped using standard AO/ASIF principles for a 4.5mm cortical screw. The reduction forceps were then removed to allow placement of the pressure sensor in the same manner as for the AP fixation group. A single 46-mm long 4.5-mm cortical bone screw was then placed in lag fashion (Nunamaker, 2000; Perren et al, 2000). All screws were tightened with a digital torque wrench (CDI Torque Products, City of Industry, CA). A final insertional torque of 4.6 N m for AP and 4.4 N m for AO screws was achieved (Galuppo et al, 2002).

2.2.4 Interfragmentary Compression Testing

Prior to testing, all sensors underwent two-point calibration in a materials testing machine. A single sensor was used for each pair of MC3, labeled, and stored post-testing. Final compression pressure was acquired by the I-Scan system and stored in a computer file.

2.2.5 Statistical Methods:

Mean + SD were calculated for screw compression pressure for each fracture distance and fixation configuration (AP and AO). Paired samples for each construct were evaluated using *t*-tests for paired sample means within each fracture fragment testing group. Statistical significance was set at $P<0.05$.

2.3 Results

There were significant differences between the AO and AP constructs for compression pressure for the 20-mm (P = 0.03), 12-mm (P = 0.04), and 8-mm (P = 0.04) groups (Table 2.1).

The ratio of mean compressive pressures for AP screws to AO screws at 20, 12, and 8-mm, respectively were 21.6%, 26.2%, and 34.2% (Fig. 2.3).

Table 2.1 Mean + SD Compressive Pressures for AO and AP screw types at fracture thicknesses 20-, 12-, and 8-mm. Statistical significance ($p \leq 0.05$) was achieved for all configurations.

Screw Type / Fragment Size	Mean ± C.P. (KPa)
AO / 20-mm	1085.34 ± 463.05
AP / 20-mm	234.51 ± 195.32
AO / 12-mm	3140.68 ± 1312.71
AP / 12-mm	822.62 ± 195.98 KPa
AO / 8-mm	3259.13 ± 1301.64
AP / 8-mm	1113.60 ± 262.03

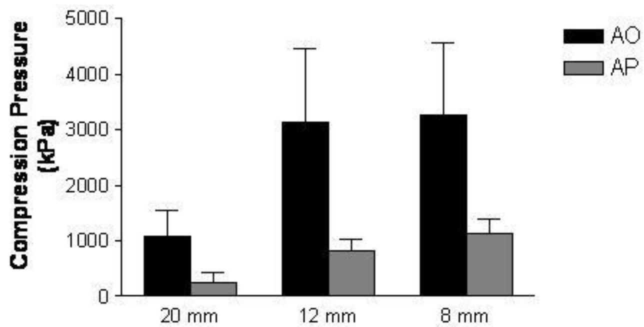


Figure 2.3 Compressive pressure measurements for AO and AP constructs at fracture thicknesses 20-, 12-, and 8-mm.

2.4 Discussion

Previous work on interfragmentary compression claimed compression pressure and force of the Acutrak Plus to be 65% and 44%, respectively, to that of a 4.5mm AO cortical screw (Galuppo et al, 2002). Our findings do not agree with this. One possible reason for this discrepancy is that the fragment thickness played an important role in the ability to obtain interfragmentary compression with the Acutrak screw. In regard to pitch change, the screw can be divided into three distinct sections. Beginning at the screw apex, the first seven threads have a pitch of 2.5 mm. The second set of seven threads have a pitch of 2.0 mm. The third set of seven threads has a pitch of 1.75 mm. This third set of seven threads corresponds to the beginning of the cutting flutes near the screw base which is at a length of approximately 12 mm from the base. Therefore, there is essentially a coarse section of threads occupying approximately 2/3 of the screw beginning at apex and extending towards the base, and a narrow section of threads occupying the remaining 1/3 of the screw nearest the base or headless end of the screw. While the screw is being tightened, the fracture line is being gradually compressed, however the majority of this compression occurs when the more narrow pitched threads reach the fracture plane. Once the fine pitch threads engage the fragment, the interfragmentary compression begins. A problem may arise when the fragment thickness is significantly different from the length of the portion of the screw that has the finer pitch threads.

Mean compressive forces and pressures for AP screw fixations are significantly less than those for AO screw fixations. The most profound differences were noted with the 20-mm fragments in which interfragmentary compressive force and compressive pressure for AO screws were approximately 5 times greater and 9 times greater than the AP screws, respectively. Additionally, the interfragmentary compressive force and pressure for the 12-mm AO groups were 4 times and 8 times greater than AP screws, and 3 times and 4 times greater for the 8-mm group, respectively. These findings suggest that the Acutrak Plus screw may only be able to be utilized within a very narrow window of fracture fragment thickness. If the fragment is too thick, these very narrow pitched threads and the most proximal coarse threads will stay in the fragment and never purchase the parent bone, leading to intrafragmentary compression, creating minimal interfragmentary compression.

The 8-mm group resulted in the greatest amount of compression. Although the most abrupt change in pitch is within the parent bone in this situation, there is still compression generated across the fracture plane because continued compression is occurring between both the fragment and parent bone. Furthermore, with the tapered configuration of the screw, the fine pitch threads in the narrower section of the screw engage less in the parent bone than the fine pitch threads in the part of the screw with the greatest diameter, hence the greater thread surface area, which engage more in the fragment. This results in a differential compression in the fine pitch thread region.

Another possible reason for lesser interfragmentary compression is the interference of the sensor which was placed in the osteotomy plane. Given that the thickness of the sensor is 0.10-mm, we find it highly unlikely that this affected the compression to any great degree. Since this is the same sensor used in a previous study (Galuppo et al, 2002), it was not mentioned in that study that the sensor posed any problem. Although a complete fracture model was created in that particular study, by placing the bone segments in a positioning jig attached to a load cell to stabilize the fragment and parent bone during screw placement, a stable,incomplete fracture was essentially repaired. In our study, the use of reduction forceps to secure the fragment to the parent bone simulated more closely the clinical setting for repair of a complete condylar fracture. This may also lend to the discrepancy in compression pressure when compared to previous work. Another possible explanation is the final insertion torque used. In our study, all AP screws were tightened to a final torque of 4.6 Nm, and the AO screws were tightened to a final torque of 4.4 Nm. This is in contrast with another study (Galuppo et al, 2002), in which all screws were tightened to a perceived maximal torque which was determined by assessing the maximal torque required to strip the recessed screwdriver insert of the AP screw and cause screw head failure of the AO screw performed in a preliminary test.

For this study, we chose to drill the AP group with the dense bone bit in order to maintain a consistent protocol throughout the study. Previously, it was recommended in the clinical setting to drill initially with the standard solid core tapered bit, and, after tapping, if screw placement was difficult in the dense epiphyseal bone it was recommended to either re-tap the hole or over drill with the dense bone bit and then re-tap in order to place the screw (Galuppo et al, 2006). The main difference between the dense bone bit and the standard bit is that the dense bone bit is 0.5-mm diameter larger throughout the tapered length compared to the standard drill bit. This allows for easier placement of the screw due to less thread engagement with the bone. We felt that if we had re-tapped this may lead to cross-threading of the threads thus hindering screw placement and optimal compression. This, in turn, could likely lead to less compression.

Previous biomechanical work (Galuppo et al, 2001; Galuppo et al, 2002) was followed up by a clinical evaluation of AP fixation in sixteen lateral condylar fractures (Galuppo et al, 2006) which reported a 73% success rate for return to racing following repair. The mean fracture fragment thicknesses were 13.6 + 2.1 mm (forelimb) and 18.7 + 2.2mm (hindlimb). Our test thicknesses ranged between 8- and 20-mm, encompassing almost all of their reported range. Based on our work, the compressive pressure is what we consider inadequate for use in a clinical situation.

The Acutrak Equine bone screw, a non-cannulated version of the AP screw, would have been the ideal screw to test, given that this screw was used in the clinical retrospective study (Galuppo et al, 2006). Unfortunately, this screw has been taken off the market and was therefore unavailable for evaluation.

The clinical report (Galuppo et al, 2006) suggests that this fixation does not rely predominantly on compression for stability. However, the concern remains that articular surface integrity is compromised in a situation in which compression at this site is not optimal which would lead to a construct not rigid enough to stimulate primary bone healing or possible shifting of the fracture fragment. This is of the most importance along the articular surface where compression is required to eliminate any gaps or defects which, if not dealt with properly, may lead to the development of degenerative joint disease, thus decreasing an athlete's chances of competing successfully. This poor interfragmentary compression could also lead to excessive movement at the fracture line, thus leading to excessive cycling of the construct which could either result in failure of the repair or degenerative changes along the articular surface. Wheeler et al (1998) compared the interfragmentary compression and cyclic fatigue testing of an Acutrak screw, Herbert screw, and a 3.5-mm AO cancellous screw. He discovered that the Acutrak screw did not lose compression compared to the AO screw by cycling up to 500 cycles. This may be considered adequate for biomechanical testing in the human field, however no conclusions can be made with regards to cyclic fatigue in the horse. McDuffee et al (2000) found that the mean number of steps per hour within a box stall taken by a horse was 190 $\pm$ 180 steps (McDuffee et al, 2000). Further work performing cyclic fatigue testing of a construct would obviously be required to confirm this.

A minimum amount of compression of a nondisplaced condylar fracture may be adequate to achieve a satisfactory clinical outcome (Rick et al, 1983; Galuppo et al, 2006). The amount of compression generated by the Acutrak screw certainly is concerning from the standpoint of fracture stability and healing in the case of a complete condylar fracture. Further work is now needed to determine what degree of interfragmentary compression is necessary to facilitate healing in complete condylar fractures in general, and then to determine how this relates to our findings as well as previous work.

CHAPTER 3

COMPARISON OF PULLOUT STRENGTH OF 6.5-MM HEADLESS TAPERED VARIABLE PITCH AND 4.5-MM AO CORTICAL BONE SCREWS IN A SYNTHETIC BONE SUBSTITUTE

3.1 Introduction

The pullout strength is used as a standard for evaluating the inherent holding strength of orthopedic screws in bone. There are multiple variables that are involved in affecting the pullout strength of a screw, including thread depth, major diameter, bone quality, cortical purchase, depth of screw penetration, thread angulation, pitch diameter, screw placement within the bone, physical changes to the screw or bone between insertion time and withdrawal time, and screw withdrawal speed (Philips et al, 1989; Uhl, 1989; DeCoster et al, 1990; Maiman et al, 1990; Daftari et al, 1994; Halvorson et al, 1994; Kohn et al, 1994; Schwimmer et al, 1994; Asnis et al, 1996; Ferrera et al, 2000).

Previous work evaluating screw pullout for AO cortical and cancellous screws in both foal and adult equine bone has been performed (Sedrish et al, 1998; Sedrish et al, 1998; Andrea et al, 2002; Johnson et al, 2004, Sod et al, 2004). This has been evaluated extensively because adequate interfragmentary compression between bone fragments to produce stable osteosynthesis is reliant on pullout strength (Sod et al, 2004). There have been few studies evaluating the pullout strength of the Acutrak screw, a headless, tapered variable pitch compression screw in the human literature (Sides et al, 2006). Other studies have been performed in both human and veterinary literature comparing pushout strength of Acutrak screws to AO screws (Wheeler et al, 1998; Carpenter et al, 2006). Pushout testing was performed as an alternative to pullout testing due to the headless design of the Acutrak screw (Carpenter et al, 2006).

A concern of the authors in a previously written paper (Carpenter et al, 2006) on pushout strength of the Acutrak Plus screw in equine bone was the mode of failure. The Acutrak Plus screws failed via screw deformation in pushout, whereas the AO screws failed at the bone-screw interface. They speculate that the Acutrak Plus would most likely be stronger in pullout than the 4.5 mm AO cortical screw.

The purpose of this study was to evaluate and compare the pullout strengths of an 80-mm Acutrak Plus (AP) screw with that of a 4.5-mm AO cortical screw (AO) in a synthetic bone substitute. Our hypothesis was that the Acutrak Plus screw would be significantly stronger in pullout when compared to a 4.5 mm AO cortical screw.

3.2 Materials and Methods

3.2.1 Synthetic Bone Description and Preparation

Twelve blocks of synthetic bone (Sawbones, Pacific Research Labs, Vashon, WA) measuring 170mm x 120mm x 63mm (length x width x thickness) were obtained. Cross-sectional evaluation of the synthetic bone exposed a 6-mm cis-cortex, 51-mm medullary cavity, and 6-mm trans-cortex (See Figure 3.1). Cortical and medullary densities were 1.70 g/ml and 0.160 g/ml, respectively. Cortical and medullary ultimate tensile strengths were 90 MPa and 2.20 MPa. 3/4-inch holes were drilled through both cortices at the corners of the block approximately 63mm x 37mm apart. These were drilled out to allow for mounting in the pullout jig for the materials testing machine.

3.2.2 Acutrak Screw Preparation and Placement

For the AP screw fixation, the bone substitute was secured in a drill press vice, and a hole was drilled to a depth of 65-mm drill press using a standard bone bit in the center of the bone substitute perpendicular to the cortical surface. The hole was then cleared of any debris and then tapped using a 45-mm long, tapered bone tap by rotating the tap until it was snug. The tap was then rotated clockwise an additional ¼ turn, followed by four counterclockwise turns, repeating the process each time. Care was made to ensure that the tap was not backed out beyond four turns, as this increased the risk of accidentally double-tapping the hole. The hole was tapped to a final depth of 50-mm. After tapping, the hole was again cleared of all debris.

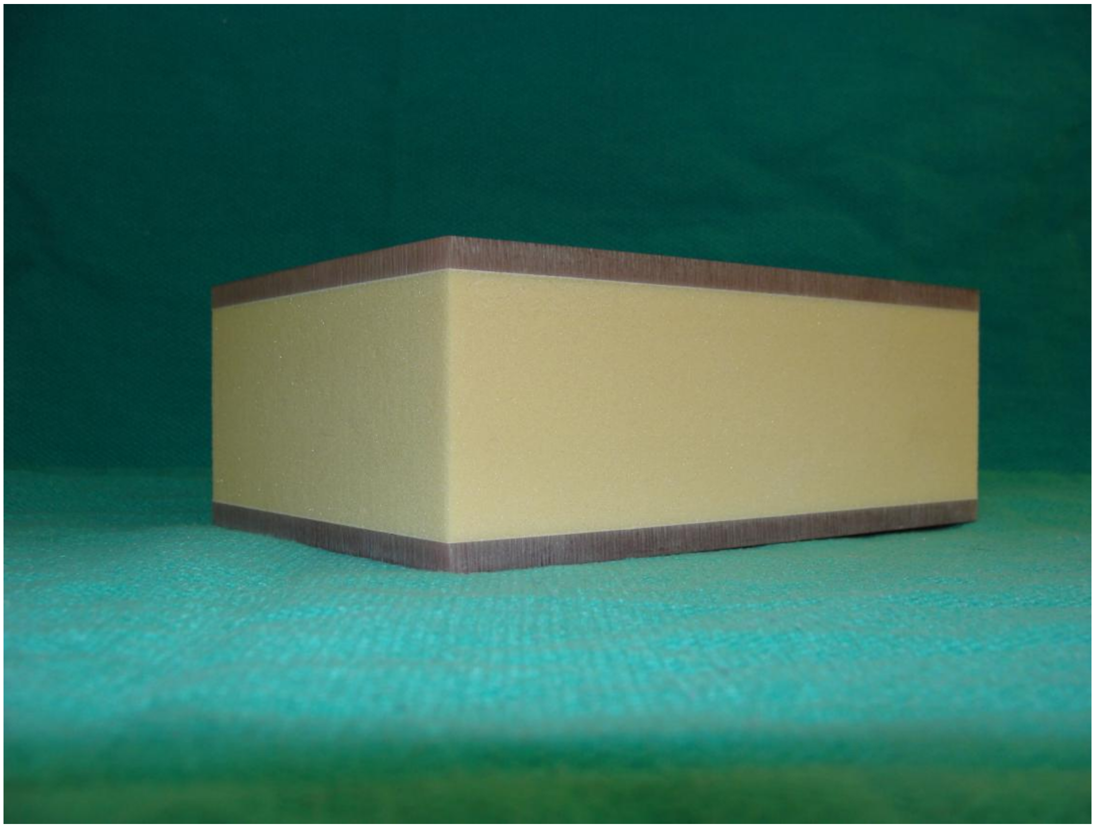


Figure 3.1 Cross-sectional photo of synthetic bone substitute. Cortical sections are brown, and cancellous section is yellow.

An 80-mm Acutrak Plus screw (60-mm threaded length, 20-mm shank length) was used in the pullout test. Prior to testing, the first 4 mm of non-threaded shank (shank diameter 5 mm) was threaded using a tap and die set to facilitate pullout testing. After creating the pullout threads, the screw then placed in the bone substitute and tightened to the point that all manufacturer threads were embedded in the bone substitute, leaving the non-threaded shank exposed. Pullout testing was then performed as described below.

3.2.3 AO Screw Placement

For the AO screw fixation, the bone substitute was secured in a drill press vice, and a 70-mm deep thread hole was drilled with a drill press using a 3.2-mm drill bit. The hole was then lavaged and subsequently tapped with a 4.5-mm cortical bone tap by rotating the tap two turns clockwise and one-half turn counterclockwise until the tap could be seen exiting the trans-cortex by a distance of at least 4-mm, ensuring that the entire hole was tapped. A 110-mm long 4.5-mm cortical screw was then placed through a three-hole 4.5-mm narrow dynamic compression plate and advanced through the bone substitute until 4 mm of screw was exposed from the trans cortex leaving approximately 35-mm of the screw closest to the screw head exposed to allow enough room for the pullout testing jig to engage the plate and head without coming in contact with the bone substitute.

3.2.4 Pullout Testing

For the AO group, the synthetic bone-screw constructs were mounted in a bone holding frame that was attached to the hydraulic cylinder of a servo-hydraulic biaxial material testing system (MTS) (Sod et al. 2004). The screw and section of the round hole plate, located under the screw head, were connected by a metal jig to the testing system load cell. The screw-metal jig allowed sufficient lateral motion to ensure that the uniaxial tensile force was applied along the longitudinal axis of the screw (See Figure 3.2).

For the AP group, the synthetic bone-screw constructs were mounted in a bone holding frame that was attached to the hydraulic cylinder of the MTS. The threaded section of shank was then threaded on to a 15 cm length of steel rod which had a hole bored in its core measuring 4.5 mm diameter which was tapped to match the threaded shank of the screw. The rod was then threaded onto the screw to facilitate pullout testing. This was a modification of a previously described pullout jig (Sides et al, 2006). The proximal portion of the rod was then attached to a universal joint connected to the MTS. The screw-metal jig allowed sufficient lateral motion to ensure that the uniaxial tensile force was applied along the longitudinal axis of the screw (See Figure 3.3).

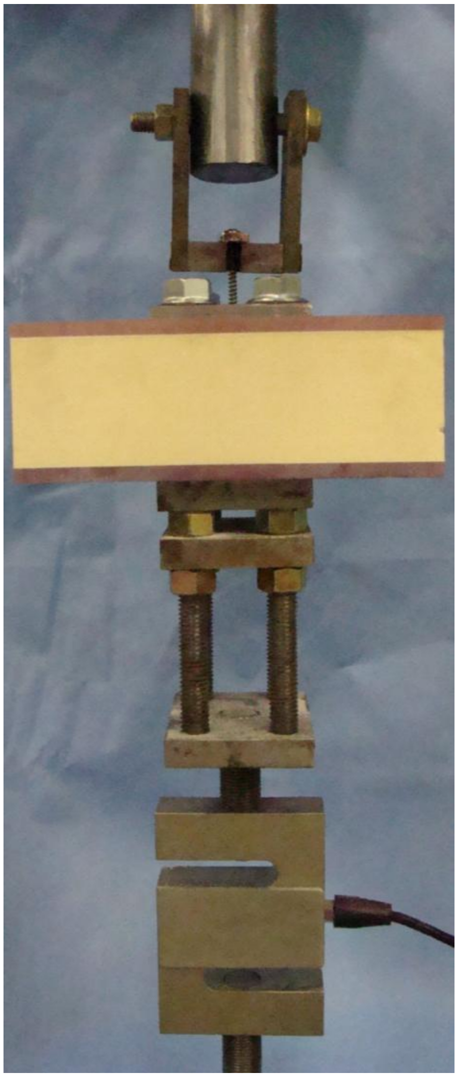


Figure 3.2. Photograph showing the AO pullout test construct in the MTS.

Figure 3.3. Photograph showing the AP pullout test construct in the MTS.

The synthetic bone-screw constructs were subjected to a tensile loading at 19.0mm/sec to mimic a catastrophic failure. The tensile force (N) as a function of displacement was recorded for each test and the maximal tensile forces needed to pull the screws from the synthetic bones were recorded. The point of failure was defined as the peak force obtained immediately before screw or bone failure. The tests were stopped when either the screw broke or pulled out of the bone.

After testing, the bone substitutes were transected through the screw holes and the cortical and medullary cavity thicknesses were measured with a dial caliper to confirm thicknesses. Because total bone thickness affects both tensile strength and mode of bone-screw failure, the pullout strength in N/mm (force required for screw extraction divided by the total cortical bone thickness) was calculated (Koryani et al. 1972).

3.2.5 Materials Testing Machine Description

A modified MTEST Windows material testing system (ADMET Inc, Norwood, MA) controlling a servo-hydraulic material testing machine (designed and built by **Sod**, LSU-SVM, Baton Rouge, LA) equipped with a 250kN biaxial load cell and a 50 kN fatigue rated load cell was used for all mechanical tests. The MTEST Windows system provided closed loop servo control, managed calibration, and provided analog/digital conversion of the data acquired and storage in a computer data file. Calibration was verified by an external contractor every 6 months, when the MTS was moved or when a new load cell was installed. MTEST Windows was used to perform load cell calibration, using the shunt calibration technique, and position calibration, before each cyclic fatigue test and before each group of single cycle to failure tests.

3.2.6 Statistical Analysis

Mean $\pm$ standard deviations (SD) for AO and AP were calculated for the maximum pull-out force, pullout strength per mm of screw purchase (total cortical bone thickness or total cortical bone thickness plus medullary cavity thickness). For both the

4.5mm AO screw and the AP screw, comparisons between groups were made using *t*-tests for paired samples. Values of $p \leq 0.05$ were considered significant.

3.3 Results

The mean ± SD pullout force and pullout strength for the AO group were 916.72±303.13 N and 76.39±25.26 N/mm, respectively (See Figure 3.4). The mean ± SD pullout force and pullout strength for the AP group were 2296.76±917.44 N and 191.40±76.45 N/mm, respectively (See Figure 3.5). The AP group had a significantly greater pullout force (p=0.025) and pullout strength (p=0.025) when compared to the AO group. All screws failed at the bone-thread interface. A crater-shaped section of cis-cortex (approximately 15mm x 2.5mm) failed on pullout of the AO screws and remained with the screw (See Figure 3.6A). A similar phenomenon was noted with the AP screws (See Figure 3.6B), with the exception that the failed portion of cortex was thicker (approximately 5mm).

Table 3.1 Mean + SD Pullout Force and Strength for AO and AP screw constructs. Statistical significance ($p \leq 0.05$) was achieved for all configurations.

Screw Type	Mean ± S.D. P.F. (N)	Mean ± S.D. P.S. (N/mm)
AO	916.72±303.13	76.39±25.26
AP	2296.76±917.44	191.40±76.45

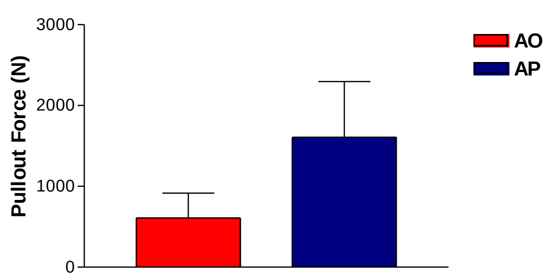


Figure 3.4. Pullout force measurements for AO and AP screw constructs.

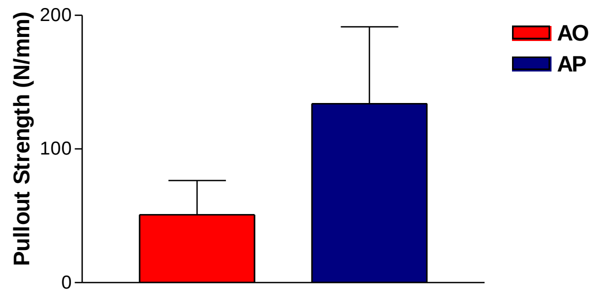


Figure 3.5. Pullout strength measurements for AO and AP screw constructs.

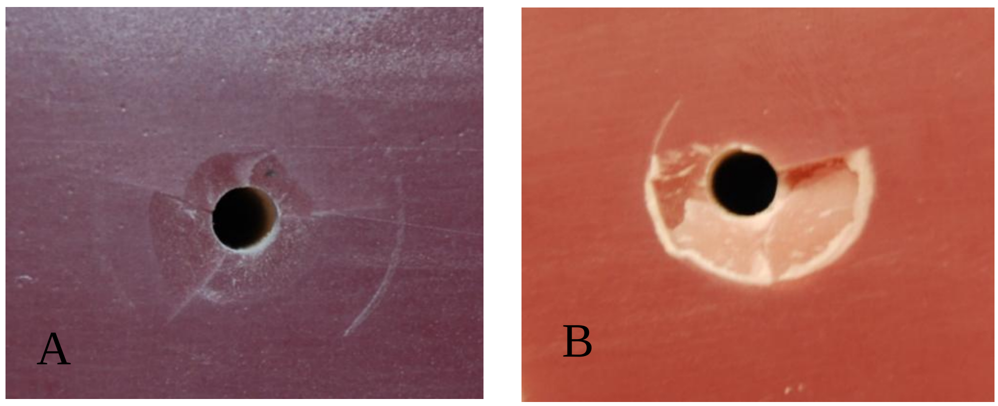


Figure 3.6. Close-up image of the defect remaining following pullout of the AO (A) and AP (B) screw constructs. Note that the defect following pullout of the AP constructs is larger in size and deeper.

3.4 Discussion

The purpose of this study was to assess the pullout forces and strengths of an 80-mm, 6.5-mm Acutrak Plus screw and a 4.5-mm AO cortical bone screw in a synthetic bone substitute. Similar work had been previously been performed by Carpenter et al (2006) in an equine model evaluating pushout characteristics of both a 45-mm long, 6.5-mm AP screw and a 4.5-mm AO cortical screw. Their work showed that mean pushout strength to failure was similar between the two screws. One point brought up in that paper was how the failure occurred. The AO screws failed at the bone-thread interface, whereas the AP screws typically failed by screw deformation. The theory was that had screw deformation not occurred in the experiment, the AP screw may actually have a greater pushout strength to failure. By performing pullout testing, we were able to avoid screw deformation and confirmed this theory, as all screws failed at the bone-thread interface.

One point of objection with the previous pushout study (Carpenter et al, 2006) that we have is the statement made that as long as the pushout force is directed along a longitudinal axis, it should equal pullout force. We disagree with this statement based on our findings in comparison with their findings. The mode in which failure occurred in pushout for the majority of the AP constructs was that the screw failed by deformation. This resulted in a dissipation of energy that would not occur in a pullout test. Failure during pullout with the AP construct in our test revealed that the screw failed at the bone-thread interface. The major force of pullout occurred by failure in the cis-cortex, as evidenced by the deep saucer fracture seen at the end of each test. This encompassed two to three threads of the screw. This implies that in order for pullout to occur, the major force must be overcome at the cis cortex, followed by a series of smaller failures at each bone-thread interface. With regards to the pushout study (Carpenter et al, 2006), by having failure first occur at the screw apex and distal shaft, the screw would then have to still overcome the forces of the of the dense cis-cortex. Failure at the cis-cortex occurred only once in that study. The pushout tests were stopped following a displacement of 10mm. This gives an inaccurate assessment of how the screw would fail, and essentially an inaccurate assessment of the inherent strength of the screw in bone. Had the pushout test continued beyond this point, failure may have occurred at the cis-cortex, then smaller failures would have occurred beyond this.

The amount of cortex lost with each pullout test is an interesting point. Both AO and AP constructs had similarities in that failure both resulted in the development of a “saucer-like” fracture. The major difference was the thickness of the fragment. A much larger fragment was removed during pullout testing of the AP screw when compared to the AO construct. Although the diameter was similar, we believe this was associated with the hole in the plate which held the composite material in place during the test. The failure of the AP construct is much more violent compared to AO failure. This was observed in a preliminary pullout test in which the jig plate was removed from the cis-cortex of the bone. A large section of cis-cortex was essentially ripped away from the cancellous substitute foam, causing multiple fractures throughout the foam. This was not the case with the AO constructs in which failure was similar to that of the test constructs for the study.

Pullout testing was performed in a synthetic bone substitute to ensure pullout occurred. In planning this project, we did test pullout in the distal radial metaphysis, but due to the density of both cortical and cancellous equine bone, AO screws failed at the screw-head junction and AP screws failed at the pullout jig. Previous work showed that pullout of 4.5-mm AO cortical screws placed in adult third metacarpal bones led to failure of the screw at the junction of the screw head and threads (Andrea et al, 2002). The synthetic bone substitute has a cortical density of 1700 mg/ml and which is similar to equine cortical bone which has a mean cortical density of 1035.25 mg hydroxyapatite/ml (Fürst et al, 2008). Although the density of the synthetic bone is greater than cortical bone, the inherent strength of the bone is affected more by its tensile strength and modulus of elasticity. The synthetic substitute has a tensile strength and modulus of elasticity of 90.0 MPa and 12.4 GPa, respectively (Sawbones Third-Generation Simulated Cortical Bone Specifications, www.matweb.com), whereas equine cortical bone has a mean tensile strength and modulus of elasticity of 2137.9 to 2295.7 MPa (El Shorafa et al, 1979) and 16.3 GPa (Batson et al, 2000), respectively. This decrease in both the tensile strength and elastic modulus allows us to evaluate the behavior of how bone fails under pullout, whereas had we used equine bone, the screws would most likely have all failed by breaking.

In conclusion, the pullout force and strength were significantly greater for the Acutrak Plus screws compared to the 4.5-mm AO screws. Our results suggest that the Acutrak Plus may provide a more stable construct for repair of fractures where screw fixation alone is appropriate. Further work including single cycle to failure and cyclic fatigue testing may be of benefit.

CHAPTER 4

FINAL DISCUSSION AND CONCLUSIONS

4.1 Summary

In summary, we have compared the Acutrak Plus and 4.5-mm AO cortical screws measuring the interfragmentary compression for 3 different simulated fracture fragment thicknesses in adult equine third metacarpal bones, and the pullout strength of the same screws in a composite bone substitute.

We have determined that the AO screw is superior in compression at all fragment thicknesses compared to the AP screw. This certainly lends concern to the surgeon when interfragmentary compression is directly related to construct stability and articular alignment. Further work needs to be performed to define what ideal interfragmentary compression is with regards to its role in primary bone healing and construct stability.

The AP screw is significantly stronger than a 4.5-mm AO cortical screw in pullout strength. This plays a significant role in construct stability as well. However, given the density of equine bone and the potential difficulty of insertion of the AP screw in equine bone, one must continue to use caution when choosing this screw, especially if basing the repair on pullout strength alone.

In the clinical retrospective study by Galuppo et al.(2006), evaluating the outcomes following repair lateral condylar fractures with the AE screw, they reported that 73% of cases successfully returned to racing. However, 12 of 16 of these cases were classified as type 1 fractures. In comparison to Rick et al. (1983) which evaluated outcomes of return to race career by fracture classification and type of repair, 21 fractures were classified as type 1 fractures, 11 were repaired surgically using standard AO techniques, and 10 were managed conservatively. Of the 11 managed surgically 9 successfully returned to racing. Of the 10 managed surgically, 9 successfully returned to racing, leading one to question the necessity for repair of type 1 fractures. Galuppo et al.'s (2006) did not compare to any cases managed conservatively, but one could assume that those cases repaired with the AE screw may have returned to racing successfully without repair. Until repairs can be assessed in more complicated fractures (types 2 and 3), no conclusions can be made to their efficacy in displaced condylar fractures. Caution should be used when choosing this screw for repair of condylar fractures.

BIBLIOGRAPHY

Adla DN, Kitsis C, Miles AW: Compression forces generated by Mini bone screws – a comparative study done on bone model. *Int J Care Injured* 2005;36:65-70.

Alexander JT, Rooney JR: The biomechanics, surgery, and prognosis of equine fractures 1967-1971. *Proc Am Assoc Eq Pract* 1972;19:219-236.

Andrea CR, Stover SM, Galuppo LD, et al: Comparison of insertion time and pullout strength between self-tapping and non-self-tapping AO 4.5-mm cortical bone screws in adult equine third metacarpal bone. *Vet Surg* 2002; 31:189-194.

Asnis SE, Ernberg JJ, Bostrom MP, et al: Cancellous bone screw thread design and holding power. *J Orthop Trauma* 1996; 10(7):462-469.

Bailey CA, Kuiper JH, Kelly CP: Biomechanical evaluation of a new composite bioresorbable screw. *J Hand Surg* 2006;31:208-212.

Bassage LH, Richardson DW: Longitudinal fractures of the condyles of the third metacarpal and metatarsal bones in racehorses: 224 cases (1986-1995). *J Am Vet Med Assoc* 1998; 212:1757-1764.

Batson EL, Reilly GC, Currey JD, et al: Postexercise and positional variation in mechanical properties of the radius in young horses. *Equine Vet J* 2000; 32: 95-100.

Bertone AL: Distal limb: fetlock and pastern. In Hinchcliff KW, Kaneps AJ, Geor RJ (eds): *Equine Sports Medicine and Surgery: Basic and Clinical Sciences of the Equine Athlete*. Edinburgh: Saunders, 2004:289-318.

Bueno ACD, Galuppo LD, Taylor KT, et al: A biomechanical comparison of headless tapered variable pitch and AO cortical bone screws for fixation of a simulated slab fracture in equine third carpal bones. *Vet Surg* 2003; 32:167-177.

Carpenter RS, Galuppo LD, Stover SM: Effect of diameter of the drill hole on torque of screw insertion and pushout strength for headless tapered compression screws in simulated fractures of the lateral condyle of the equine third metacarpal bone. *Am J Vet Res* 2006; 67:895-900.

Casillas MM, Strom AC: Internal fixation of fifth metatarsal metaphyseal-diaphyseal stress fractures with an intramedullary screw. *Oper Tech Sports Med* 2006;14:232-238

Colton C: History of Osteosynthesis. In Texhammar R, Colton C (eds): *AO/ASIF Instruments and Implants: A Technical Manual*. New York: Springer-Verlag, 1981: 3-4.

Cosio MQ, Camp RA. Percutaneous pinning of symptomatic scaphoid nonunions. *J Hand Surg [Am]* 1986;11:350-355.

Daftari TK, Horton WC, Norman JT, et al: Correlations between screw hole preparation, torque of insertion, and pullout strength for spinal screws. *J Spinal Disord* 1994; 7(2):139-145.

Dao KD, Shin AY: Percutaneous cannulated screw fixation of acute nondisplaced scaphoid waist fractures. *Oper Tech in Orthop* 2003; 13:11-20.

DeCoster TA, Heetderks DB, Downey DJ, et al: Optimizing bone screw pullout force. *J Orthop Trauma* 1990; 4(2):169-174.

Eddy AL, Galuppo LD, Stover SM, et al: A biomechanical comparison of headless tapered variable pitch compression and AO cortical bone screws for fixation of a simulated midbody transverse fracture of the proximal sesamoid bone in horses. *Vet Surg* 2004;33:253-262.

El Shorafa WM, Feaster JP, Ott EA: Horse metacarpal bone: age, ash content, cortical area and failure stress interrelationships. *J Anim Sci* 1979; 49: 979-982.

Ellis DR: Sir Frederick Hobday Memorial Lecture: Some observations on condylar fractures of the third metacarpus and third metatarsus in young thoroughbreds. *Equine Vet J* 1994; 26:178-183.

Fadel GE, Hussain SM, Sripada S, et al: Fixation of first metatarsal basal osteotomy using Acutrak screw. *Foot and Ankle Surg* 2008;14:21-25

Faran KJ, Ichioka N, Trzeciak MA, et al: Effect of bone quality on the forces generated by compression screws. *J Biomech* 1999;32:861-864.

Ferrera LA, Ryken TC: Screw Pullout Testing. In An YH, Draughn RA (eds): *Mechanical Testing of Bone and the Bone-Implant Interface*. London: CRC Press, 2000:551-565.

Fürst A, Meier D, Michel S, et al: Effect of age on bone mineral density and micro architecture in the radius and tibia of horses: An Xtreme computed tomographic study. *BMC Vet Res* 2008; 4:3.

Galuppo LD, Stover SM, Jensen DG, et al: A biomechanical comparison of headless tapered variable pitch and AO cortical bone screws for fixation of a simulated lateral condylar fracture in equine third metacarpal bones. *Vet Surg* 2001; 30:332-340.

Galuppo LD, Stover, SM, Jensen DG: A biomechanical comparison of equine third metacarpal condylar bone fragment compression and screw pushout strength between headless tapered variable pitch and AO cortical bone screws. *Vet Surg* 2002; 31:201-210.

Galuppo LD, Simpson EL, Greenman SL, et al: A clinical evaluation of a headless, titanium, variable-pitched, tapered, compression screw for repair of nondisplaced lateral condylar fractures in thoroughbred racehorses. *Vet Surg* 2006; 35:423-430.

Geissler WB: Management of scaphoid fractures in the athlete. *Oper Tech Sports Med* 2006;14:101-107.

Haddad FS, Goddard NJ: Acute percutaneous scaphoid fixation using a cannulated screw. *Ann Chir Main* 1998; 17:119-126.

Halvorson TL, Kelley LA, Thomas KA, et al: Effects of bone mineral density on pedicle screw fixation. *Spine* 1994; 19(21): 2415-2420.

Hausmann JT, Mayr W, Unger E, et al: Interfragmentary compression forces of scaphoid screws in a sawbone cylinder model. *Int J Care Injured* 2007;38:763-768.

Herbert TJ: Management of the fractured scaphoid using a new bone screw. *J of Bone and Joint Surg.* 1986; 66B:114-123.

Hirsch JE, Galuppo LD, Graham LE, et al: Clinical evaluation of a titanium, headless variable-pitched tapered cannulated compression screw for repair of frontal plane slab fractures of the third carpal bone in Thoroughbred racehorses. *Vet Surg* 2007; 36:178-184.

Inoue G, Tamura Y. Closed technique for the Herbert screw insertion in an undisplaced fracture of the scaphoid. *J Orthop Surg Tech* 1991;6:1-7.

Johnson BJ, Stover SM, Daft BM, et al: Causes of death in racehorses over a 2 year period. *Equine Vet J* 1994;26:327-330.

Johnson NL, Galuppo LD, Stover SM, et al: An in vitro biomechanical comparison of the insertion variables and pullout mechanical properties of AO 6.5-mm standard cancellous and 7.3-mm self-tapping, cannulated bone screws in foal femoral bone. *Vet Surg* 2004; 33:681-690.

Kane AJ, Stover SM, Gardner IA, et al: Horseshoe characteristics as possible risk factors for fatal musculoskeletal injury of Thoroughbred racehorses. *Am J Vet Res* 1996;57:1147-1152.

Kane AJ, Stover SM, Gardner IA, et al: Hoof size, shape, and balance as possible risk factors for catastrophic musculoskeletal injury of Thoroughbred racehorses. *Am J Vet Res* 1998;59:1545-1552.

Kainer RA: Functional Anatomy of Equine Locomotor Organs. In Stashak TS (ed): *Adams Lameness in Horses*. Philadelphia: Lippincott, Williams, and Wilkins, 2002:1-72.

Kohn D, Rose C: Primary stability of interference screw fixation. Influence of screw diameter and insertion torque. *Am J Sports Med* 1994; 22(3):334-338.

Koryani E, Bowman CE, Knecht D, Janssen M: Holding power of orthopaedic screws in bone. *Clin Orthop* 1972;72:283-286.

Ledoux P, Chahidi N, Moermans JP, Kinnen L. Percutaneous Herbert screw Osteosynthesis of the scaphoid bone. *Acta Orthop Belg* 1995; 61:43-47.

LeJeune SS, MacDonald MH, Stover SM, et al: Biomechanical Investigation of the Association Between Suspensory Ligament Injury and Lateral Condylar Fracture in Thoroughbred Racehorses. *Vet Surg* 2003;32:585-597.

Lewis AJ, Sod GA, Gill MS, Mitchell CF: Bovine distal interphalangeal joint arthrodesis using the Acutrak Plus screw: a report of seven cases. *Vet Surg* Accepted for publication 2009.

Maiman DJ, Pintar FA, Yoganandan N, et al: Pullout strength of Caspar cervical screws. *Neurosurgery* 1992; 31(6):1097-1101.

McDuffee LA, Stover SM, Coleman K: Limb loading activity of adult horses confined to box stalls in an equine hospital barn. *Am J Vet Res* 2000; 61:234-237.

McKee SL: An update on racing fatalities in the UK. *Equine Vet Educ* 1995;7:202-204.

Meagher DM: Lateral condylar fractures of the metacarpus and metatarsus in horses. *Proc Am Assoc Eq Pract* 1976;22:147-154.

Nunamaker DM: General Techniques and Biomechanics. In Fackelman GE, Auer JA, Nunamaker DM(eds): *AO Principles of Equine Osteosynthesis.* Stuttgart: Thieme, 2000:11-15.

Ozkaya N, Nordin M: *Fundamentals of Biomechanics, Equilibrium, motion, and Deformation*. New York: Van Nostrand Reinhold, 1991:277–278.

Perren SM, Buchanan J: Basic AO/ASIF Technique: Aims and Principles. In Texhammar R, Colton C (eds): *AO/ASIF Instruments and Implants: A Technical Manual*. New York: Springer-Verlag, 1981: 26.

Perren SM, Frigg R, Hehli M, et al: Lag Screw. In Rüedi TP, Murphy WM (eds): *AO Principles of Fracture Management*. Stuttgart: Thieme, 2000:157-167.

Philips JH, Rahn BA: Comparison of compression and torque measurements of self-tapping and pretapped screw. *Plast Reconstr Surg* 1989; 83(3):447-458.

Pool RR, Meagher DM: Pathologic findings and pathogenesis of racetrack injuries. *Vet Clin North Am Equine Pract* 1990;6:1-30.

Radtke CL, Danova NA, Scollay MC, et al: Macroscopic changes in the distal ends of the third metacarpal and metatarsal bones of Thoroughbred racehorses with condylar fractures. *Am J Vet Res* 2003;64:1110-1116.

Rehak DC: Arthrodesis of the distal interphalangeal joint using a headless screw. *Oper Tech Orthop*. 2007;17:140-147.

Richardson DW: The Metacarpal and Metatarsal Bones. In Auer JA, Stick JA (eds): *Equine Surgery*. Philadelphia: Elsevier, 2006:1238–1253.

Rick MC, O’Brien TR, Pool RR, et al: Condylar fractures of the third metacarpal bone and third metatarsal bone in 75 horses: Radiographic features, treatments, and outcome. *J Am Vet Med Assoc* 1983; 183(3):287-296.

Rooney JR: Distal condylar fractures of the cannon bone in the horse. *Mod Vet Pract* 1974:113-114.

Schwimmer A, Greenberg AM, Kummer F, et al: The effect of screw size and insertion technique on the stability of the mandibular sagittal split osteotomy.   *J Oral Maxillofac Surg* 1994; 52(1):45-48.

Sedrish SA, Moore RM, Kelly K, et al: Pullout strength of screws in foal third metacarpal bone after overdrilling a 4.5-mm hole. *Vet Comp Orthop Traumatol* 1998; 11:200-204.

Sedrish SA, Moore MR, Kelly K, et al: *In vitro* pullout strength of screws inserted in adult equine third metacarpal bone after overdrilling a 4.5-mm threaded insertion hole. *Vet Surg* 1998; 27:143-149.

Sides SD, Fetter NL, Glisson R, et al: Bending stiffness and pullout strength of tapered variable pitch screws, and 6.5-mm cancellous screws in acute Jones fractures. *Foot Ankle Int* 2006; 27:821-825.

Slade JF, Gutow AP, Geissler WB. Percutaneous internal fixation of scaphoid fractures via an arthroscopically assisted dorsal approach. *J Bone Joint Surg [Am]* 2002;84A(12):21-36.

Slade JF, Merrell GA: Minimally invasive management of scaphoid fractures. *Oper Tech Plastic Recon Surg* 2003;9(4):143-150.

Sod GA, Martin GS, Gill MS: Pull-out strength of orthopedic screws in polymethylmethacrylate filled medullary cavities of foal third metacarpal bones. *Vet Comp Orthop Traumatol* 2004;17:136-140.

Stepnik MW, Radtke CL, Scollay MC, et al: Scanning electron microscopic examination of third metacarpal/third metatarsal bone failure surfaces in thoroughbred racehorses with condylar fracture. *Vet Surg* 2004;33:2-10.

Stover SM, Read DH, Johnson BJ: Lateral condylar fracture histomorphology in racehorses. *Proc Am Assoc Equine Pract* 1994;40:173-174.

Texhammar R: AO/ASIF Instrumentation. In Texhammar R, Colton C (eds): *AO/ASIF Instruments and Implants: A Technical Manual*. New York: Springer-Verlag, 1981: 73-78.

Uhl RL: The biomechanics of screws. *Orthop Rev* 1989; 18(12):1302-1307.

Wheeler DL, McLoughlin SW: Biomechanical assessment of compression screws. *Clin Orthop Relat Res* 1998; 350:237-245.

Whipple TL. The role of arthroscopy in the treatment of wrist injuries in the athlete. *Clin Sports Med* 1992;11:227-238.

Whipple TL. Stabilization of the fractured scaphoid under arthroscopic control. *Orthop Clin North Am* 1995; 26(4):749-754.

Wilson JH, Howe SB, Jensen RC: Injuries sustained during racing at racetracks in the U.S. *Proc Am Assoc Equine Pract* 1993;39:267-268.

Wozasek GE, Moser KD. Percutaneous screw fixation for fractures of the scaphoid. *J Bone Joint Surg [Br]* 1991;73B:138-142.

Zekas LJ, Bramlage LR, Embertson RM, et al: Characterization of the type and location of fractures of the third metacarpal/metatarsal condyles in 135 horses in central Kentucky (1986-1994). *Equine Vet J* 1999; 31:304-308.

VITA

Andrew Lewis was born in Fort Smith, Arkansas, in 1980. At the age of twelve, his family moved back to his father's childhood home near Powhatan, Louisiana, where Andrew learned what free farm labor was all about, given that he was the labor and he worked for free for his father and uncle on the family farm. He matured to his maximum capacity on this commercial cow-calf operation, and came to the full appreciation for hard work and the benefits that he reaped. He went on to attend Louisiana State University for both his undergraduate and veterinary studies. His goal throughout veterinary school was to enter practice as an equine veterinarian, specializing as a surgeon. Upon graduation, he married Crystal, his veterinary school classmate, and then entered a large animal surgery and medicine internship at the University of Georgia College of Veterinary Medicine, where he cemented his goal of obtaining a residency in equine surgery. He was then accepted into an equine surgical residency back at Louisiana State University. During his residency, he made a close friendship with his mentor, Dr. Gary Sod, who supported him both educationally and psychologically.