

Ex-vivo Evaluation of a Modified Teno Fix® Device Repair Pattern Versus a Three-Loop Pulley for Repair of Equine Flexor Tendons

by

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Abstract

Current techniques for equine tendon laceration repair are not strong enough to support the normal load placed on equine tendons, which can be up to 450 kg at standing rest. Repairs can result in further wound healing complications including compromised blood flow, adhesion formation, development of excessive scar tissue, and serve as a foreign body for infection. Previous research into use of the Teno Fix® stainless steel tendon fixation system for equine tendon laceration repair revealed that repair using four devices was similar in strength of resistance to a standard three-loop pulley suture technique in the prevention of a 2mm gap at the repair site. However, the Teno Fix® repair was weaker than the three-loop pulley in regards to maximum load it could withstand before failure. Development of a different implant configuration using the Teno Fix® system that may provide an option for a stronger repair is warranted due to other benefits of the repair, which include minimal inflammation and development of scar tissue, as well as minimal interference with tendon healing based on previous studies. This project will evaluate the Teno Fix® stainless steel tendon fixation system compared to the three-loop pulley, utilizing the implants applied at staggering distances on either side of the tendon laceration, to evaluate the effect of suture pattern on maximum pull-out strength and gap formation. Our hypothesis was that by distributing strain at staggered distances from the laceration, it would increase the ultimate strength and strength to a 2mm gap of the fixation device repair. The repair configuration also allowed for insertion of an additional implant for evaluation of the effect of the number of stainless steel anchors on repair strength.

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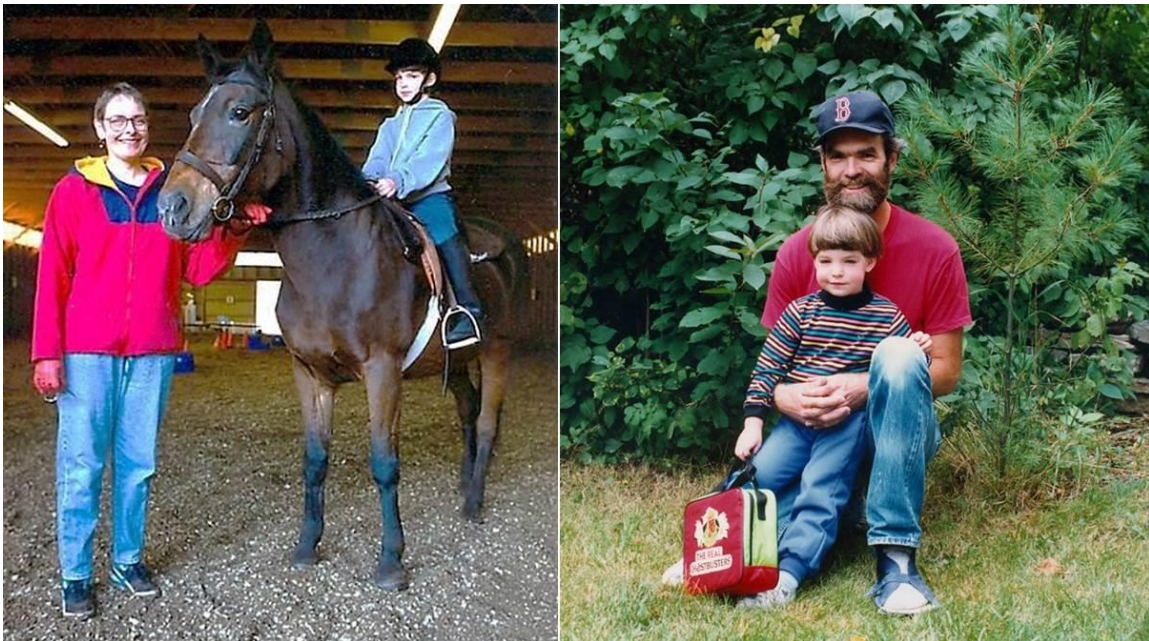


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List of Terms

Abbreviations and Acronyms

3LP	Three-loop pulley
4TF	Four Teno Fix® devices
5TF	Five Teno Fix® devices
10SS	10-strand Savage
ϵ	Strain of a stress-strain curve measured as a percentage, plotted on the x-axis
σ	Stress of a stress-strain curve, plotted on the y-axis
AAEP	American Association of Equine Practitioners
bFGF	Basic fibroblast growth factor
CSA	Cross-sectional area
DASH	Disabilities of the Arm, Shoulder and Hand questionnaire
DDFT	Deep digital flexor tendon
DIP	Distal interphalangeal joint
IGF-1	Insulin-like growth factor
LL	Locking loop
MCP	Metacarpophalangeal
MTP	Metatarsophalangeal
PDGF	Platelet derived growth factor
PIP	Proximal interphalangeal joint
<i>P_{lin}</i>	Yield-point of a tendon repair unit as measured with a load-displacement curve

PMK	Pennington modified Kessler suture
PVC	Polyvinyl chloride
SDFT	Superficial digital flexor tendon
SSS	Six-strand Savage
TGF- β	Transforming growth factor
TF	Teno Fix®
VEGF	Vascular endothelial growth factor

Units

mm	Millimeters
Mpa	Strain energy density as measured with a stress-strain curve
N	Newtons
N/mm	Energy absorbed at failure as measured with a load-displacement curve
N/mm⁻¹	Stiffness
N/mm²	The ultimate stress of a stress-strain curve prior to failure
SE	Standard error

Suture Material

Braided polyethylene coated in silicone	Fiberwire
Bidirectional barbed suture	Quill™
Monofilament nylon	Ethilon
Monofilament polypropylene	Prolene
Multifilament stainless steel	MRSS
Polymerized caprolactan	Supramid

Polybutylate coated braided polyethylene terephthalate	Ethibond
Polydioxanone	PDS
Polyglactin 910	Vicryl
Polyglycolic acid	Dexon
Polyglycolide-trimethylene carbonate	Maxon
Silicone coated braided polyethylene terephthalate	Ti-Cron
Uncoated braided polyethylene terephthalate	Mersilene
Unidirectional barbed suture	V-Loc™

I. Introduction

The SDF and DDF are flexor tendons in the horse that are part of a biomechanical apparatus that maintain the metacarpo/metatarsophalangeal joints suspended above the ground, absorb shock, store elastic energy during movement, and contribute to weight bearing.¹ Flexor tendon lacerations in horses result in a loss of support of the fetlock and biomechanical function of the limb, which can have a negative impact on a patient's future athletic performance and may be life-threatening.^{2,3-5} A recent study investigating the overall outcome after lacerations of the superficial and deep digital flexor tendons, suspensory ligament and/or distal sesamoidean ligaments in horses found that 54-55% of patients returned to an equal level of performance and 24%-27% returned to a lower level of performance.² Potential reasons for poor clinical outcomes of horses that have sustained flexor tendon lacerations that have been repaired include adhesion formation, wound complications, persistent infection, altered limb conformation and lameness impeding performance.⁶ Primary repair of tendon lacerations is recommended to facilitate healing and improve overall outcome by maintaining apposition of tendon ends until the repair site can withstand physiologic tensile loading.⁶ Although several methods of tendon repair have been tested experimentally,^{3,6-13} an ideal suture material or repair technique that is strong enough for equine flexor tendon laceration repair has not been identified and as a result successful primary repair is often not achieved.

Goals of successful tenorrhaphy include creating a strong repair with a smooth juncture and minimal gap formation at the tendon ends, thus restoring gliding function, with minimal bulk

at the repair site, adhesion formation, and disruption of blood supply. To date, there is no repair technique described *in vivo* that prevents gap formation.⁶ Normal equine SDFT support a tensile load of 1845N to 3559N at a walk.⁶ The Savage pattern and bioabsorbable tendon plates are only able to withstand up to 28-42% of this load,^{6, 8} increasing risk of potential gap formation and, or, complete repair failure.

Previous studies have found that a gap of 2mm at a tenorrhaphy site significantly increases the gliding resistance of the tendon; up to 600% more than the gliding resistance of an intact tendon, and up to 100% more than a repair with a 1mm gap due to a disruption in the smooth gliding surface of the tendon.¹⁴ Gap formation and increased gliding resistance increases the risk of adhesion formation and future tendon rupture.¹⁴ To reduce potential gap formation, repairs are often supported by prolonged external coaptation to allow for immediate weight bearing for the horse.¹⁵ However, additional complications can develop as a result of cast application including cast sores, loss of bone density, decreased muscle strength and tendon laxity.^{8, 16, 17} Without early controlled movement of the tendon post-operatively, the risk of adhesion formation further increases. Therefore, increasing the strength of a repair may aid in preventing gap formation and, or, repair failure and possibly allow for decreased time spent in external coaptation.

Several repair techniques have been developed for equine tenorrhaphy including various suture patterns, autologous tendon grafts, bioabsorbable implants and bioabsorbable plates.^{8, 17, 18, 19, 20} There has been only one previous investigation of the use of intratendinous devices for equine tenorrhaphy.¹² One such device is the Teno Fix[®] (Teno Fix[®] Tendon Repair System, Ortheon Medical, Columbus, OH), which is a knotless construct that anchors linear stainless steel suture completely within a tendon by interdigitating with the collagen fibers of the tendon. Use of this device avoids excess exposed suture on the exterior of the tendon and reduces suture bulk, thus

reducing complications such as adhesion formation, and excessive scar tissue.^{10, 11, 21} Previous evaluation of the Teno Fix® (TF) device in a canine model found that at 12 weeks post-repair there was minimal granulation tissue and neovascularization at the repair, and repair tissue was aligned longitudinally and mature. All repairs were enveloped in vascular connective tissue and provided a smooth gliding tendon surface with minimal adhesion formation compared to a normal tendon.²² Also, there was no evidence of focal tissue necrosis, which can occur due to compromised intratendinous vasculature secondary to repair. This is in contrast to other repair techniques, such as the 3LP and six strand Savage (SSS), which have been found to cause a significant reduction in the number of perfused intratendinous vessels, which may have a negative impact on tendon healing.^{23, 24}

Previous *ex vivo* evaluation of four TF devices placed equidistant from a tendon laceration site compared to the 3LP suture for equine SDFT lacerations found that the TF repair was similar in strength to the 3LP in the development of a 2mm gap, however, it was not as strong as the 3LP to maximum load at failure of the suture tendon interface.¹² Given other benefits of the TF device as observed in canine models, further evaluation of the use of this device was warranted to aid in determining a suitable repair method for equine tendon lacerations. Variation in the implant pattern for the TF device has yet to be evaluated for improving strength of repair.

Core purchase length is the exit and, or, entry point of a core suture from the cut ends of a tendon. It has been reported that a longer purchase length resulted in a greater force required to develop a 2mm gap at a repair site, as well as ultimate suture failure in porcine tendons.²⁵ It was speculated that by applying this principle to the TF device, it may strengthen repairs for equine flexor tendon lacerations. The goal of this study was to compare a staggered TF device repair pattern to a 3LP in load to a 2 mm gap (N), mode of failure, ultimate load to failure in Newtons

(N) and gap at failure (mm). Staggered placement of the TF device would be achieved by increasing the core purchase length of some of the implants. It was hypothesized that staggered placement of TF anchors would increase the strength of the repair to ultimate load to failure and load to a 2mm gap. In addition, it would allow for placement of an additional implant, which was hypothesized to further increase the strength of repair to ultimate load to failure and load to a 2mm gap.

II. Literature Review

The History of Flexor Tendon Repair

Tendon laceration repair has been extensively studied in human medicine, with particular focus on flexor tendon laceration repair of the human hand. Tenorrhaphy was not widely performed initially as it was thought to be nervous tissue as reported by Hippocrates and Galen.²⁶ The concept that tendon was not nervous tissue was disputed with two different studies; Meekren in 1682 and Von Haller in 1752.^{26, 27}

In 1767, Hunter performed the first experimental study investigating tendon healing using the canine Achilles tendon.^{26, 28} He noted that a healing tendon formed a callus, similar to healing bone. In subsequent studies, effort was made to define morphologic changes that contribute to tendon healing through extrinsic and intrinsic processes. The majority of these studies were performed on the Achilles tendon. It was not until the 20th century when research began to focus on flexor tendons of the human hand.²⁶

In 1920, research by Saloman on flexor tendon laceration repair at the level of digital sheaths in dogs revealed poor post-operative healing.²⁶ It was hypothesized that there was an inhibitory hormone within the synovial fluid which resulted in poor healing, as well as an inability of cells to migrate through synovial fluid to facilitate tendon repair.²⁶ It was therefore suggested to leave a defect in the digital sheath to permit contact between repairing tendon and subcutaneous tissue to promote extrinsic healing. In contrast, Hueck in 1923 found that there was poor tendon healing whether or not the sheath was left open or sutured closed.²⁶ In addition to this research, Bunnell and Garlock discovered the formation of restrictive adhesions at flexor tendon repair sites

within the human digit in clinical cases.²⁶ Bunnell advised that surgeons should be cautious when repairing tendon lacerations in this region. Furthermore, he outlined conditions that were ideal for the repair of flexor tendon lacerations in the digital sheath, including the use of stainless steel suture, repairing only the human flexor digitorum profundus tendon, and postoperative immobilization of the wrist in flexion to prevent failure of the repair due to excessive muscular strain.^{26, 29} He noted that a flexed wrist position still allowed sufficient motion to stimulate healing and reduce the formation of peritendinous adhesions.^{26, 29}

Despite the development of these guidelines, the predominant opinion during this time period was that primary repair of flexor tendon lacerations in the digital sheath of the hand should be discouraged due to the development of post-operative infections, excessive scarring and flexor tendon contracture.^{26, 30} This concept was widely accepted until 1950, when Siler and colleagues reported a 62% post-operative success rate with primary repair of flexor tendon lacerations within digital sheaths. In 1967, Kleinert and colleagues presented “Primary Repair of Flexor Tendons in No Man’s Land” at the American Society for Surgery of the Hand,^{26, 31} highlighting the success of primary flexor tendon repair versus second intention healing. This was a turning-point in which primary repair of flexor tendon lacerations of the human hand became widely accepted. Following this presentation, a significant amount of research was invested to determine a tendon’s role in healing.²⁶ Once it was established that tendons had an intrinsic capability to heal and did not rely solely on extrinsic healing,^{32, 33-37} several investigators focused on improving repair methods to facilitate mobilization and reduce peripheral adhesion formation.²⁶ This included the use of heavier suture material, multiple suture strands and the use of various core and peripheral suture patterns.²⁶

Current concepts in equine digital flexor tendon laceration repair have been adapted from tenorrhaphy of the human hand and will be discussed in detail later in this literature review. Despite

the various suture methods that have been developed, the amount of force an equine digital flexor tendon experiences in relation to the human hand is not equivocal. Furthermore, patient compliance proves to be a complicating factor in management of flexor tendon laceration repairs post-operatively in horses. As a result, continued research of flexor tendon laceration repair in horses is imperative to improve post-operative outcomes.

Anatomy and Function of Equine Digital Flexor Tendons

i. Tendon Morphology

Tendon is a fibro-elastic, dense connective tissue structure which is defined based on anatomical position, as well as muscular origin (myotendinous junction) and insertion on bone (osteotendinous junction). Mature tendon consists primarily of fibrous protein type I collagen and elastin, but is also composed of proteoglycans, glycoproteins, water and cells.³⁸⁻⁴¹ Type I collagen accounts for approximately 80% of the dry weight of tendon tissue and within a tendon is arranged as a hierarchical structure (Figure 1).³⁹ Collagen molecules, formed by tenocytes, assemble to form filamentous collagen fibrils.⁴² Variable numbers of collagen fibrils grouped together form a collagen fiber, which is the basic unit of a tendon.⁴⁰ Each collagen fiber is surrounded by a sheath of connective tissue known as the endotenon, which binds individual collagen fibers together, and also contains blood vessels, lymphatics and nerves (Figure 2).⁴⁰ A subfascicle, or primary fiber bundle, is formed by a variable number of collagen fibers. A cluster of primary fiber bundles form a fascicle, or secondary fiber bundle, which when grouped form a tertiary bundle. Tertiary bundles form the tendon unit, which is surrounded by a thin connective tissue sheath known as the epitenon (Figure 2). The epitenon contains the vascular, lymphatic and nervous supply to the tendon.⁴¹

An additional connective tissue layer known as the paratenon surrounds the epitenon and tendon, which is replaced by a synovial sheath in regions where a tendon sheath is present (Figure

2). The outer surface of the epitenon is contiguous with the paratenon, forming the peritendon. Combined, these structures reduce friction with adjacent tissues.⁴¹ They also provide new blood vessels and cellular elements for repair.¹⁵ A tendon sheath consists of an outer fibrous wall and an inner synovial membrane, providing an environment for smooth tendon gliding in areas where tendon changes direction over a joint surface. The tendon within a tendon sheath is nourished by synovial fluid through passive diffusion as highlighted through the use of radioactive tracers in a canine model.⁴³ Tendon within a tendon sheath is also nourished by extrinsic vasculature from the mesotenon or mesotendon, which is an elastic membrane found on the friction-free surface of the sheath, and is richly supplied with blood vessels and lymphatics (Figure 3).⁴⁴

The most predominant cell-type present in tendons is the tenocyte, which has been likened to the fibroblast due to similarities in function. Tenocytes, however, differ from fibroblasts based on the proteins they synthesize. Tenocytes primarily synthesize organized type I and type III collagen matrix, which is remodeled during tendon healing.⁴⁵ Tenocytes are not a uniform population of cells because of differences in nuclear morphology that have been identified on light microscopy and when grown in culture.⁴⁵ Currently, four different types of tenocytes have been identified including resting, active, chondrocytic and precursor tenocytes.⁴⁵ Overall, tenocytes comprise approximately 90 to 95% of cellular elements present in a tendon, and are arranged in rows between collagen fiber bundles.^{40, 41} Tenocytes have a large number of cytoplasmic extensions arranged longitudinally and laterally, connecting neighboring cells through gap junctions. This creates an efficient system for mechanotransduction, which allows tenocytes to produce a coordinated response to a variety of stimuli, including mechanical loading.^{45, 46} Other cell types in tendon include chondrocytes at tendon insertion sites, synovial cells within tendon sheaths and on the inner surface of the paratenon, capillary endothelial cells, smooth muscle cells

of arterioles and inflammatory cells when pathologic conditions are present.⁴⁰

Collagen fibrils are arranged along a longitudinal axis within the tendon. When examined under polarized light, tendon fibrils appear to have a crimp. A crimp is the characteristic waveform that can be seen in a tendon histologically (Figure 4). It allows the tendon to elongate when only a low level of stress is applied and is eliminated within the first 2% of tendon elongation. It represents the “toe” region on a stress-strain curve, which is the first non-linear region of the curve. The majority of tendon elongation, however, comes from the sliding of tendon fascicles over one another within the endotenon.¹⁵

ii. Anatomy of the Equine Digital Flexor Tendons

The body of the equine superficial digital flexor (SDF) muscle originates from the medial epicondyle of the humerus in the forelimb.⁴⁷ It continues distally as the superficial digital flexor tendon (SDFT), starting proximal to the radiocarpal joint (Figure 5). The SDFT also originates from the accessory ligament of the superficial digital flexor tendon, which begins on the caudomedial aspect of the distal radius in the forelimb.⁴⁷ The accessory ligament of the SDFT is a strong dense fibrous band that courses distally and fuses with the SDFT just proximal to the radiocarpal joint.⁴⁷ The SDFT passes distally through the carpal canal and becomes flattened and crescent-shaped at the level of the proximal metacarpus.⁴⁷ Just proximal to the metacarpophalangeal (MCP) joint, the SDFT encircles the deep digital flexor tendon, forming the manica flexoria (Figure 6).⁴⁷ The SDFT branches distally and inserts on the palmar surface of the proximal second phalanx and the distal aspect of the first phalanx behind the collateral ligaments of the proximal interphalangeal (PIP) joint (Figure 5).⁴⁷⁻⁴⁹

The deep digital flexor (DDF) muscle has three heads of origin in the forelimb; the medial epicondyle of the humerus, the middle aspect of the medial radius and the caudal aspect of the

proximal ulna.^{1,47} Tendons of all three heads fuse at the level of the carpus to form the deep digital flexor tendon (DDFT), which courses distally, deep to the SDFT (Figure 5). The accessory ligament of the DDFT originates from the palmar carpal ligament and fuses with the DDFT distal to the carpus.⁴⁷ The DDFT inserts on the palmar/plantar aspect of the third phalanx/phalange in the facies flexoria (Figure 5).⁴⁷ As both the SDFT and DDFT pass over the palmar/plantar aspect of the MCP/MTP joint, they are surrounded by the digital flexor tendon sheath, which provides a smooth gliding surface as both tendons change direction along the joint.⁴⁷

In the hindlimb, SDF muscle originates from the supracondyloid fossa of the femur.⁵⁰ It continues distally as the SDFT (Figure 7). The SDFT in the hindlimb inserts on the tuber calcis as it traverses the caudal aspect of the hock (Figure 8).⁵⁰ As in the forelimb, the SDFT is arranged on the plantar aspect of the limb as it courses distally along the metatarsus and fetlock (Figure 7). Its second and third points of insertion include the proximal eminences of the second phalanx and the distal aspect of the first phalanx behind the collateral ligaments of the proximal interphalangeal (PIP) joint.⁵⁰ As in the forelimb, the DDF muscle has three heads of origin, including the long digital flexor, the tibialis posterior and the flexor hallucis.⁵⁰ All three originate from the lateral condyle of the tibia. Tendons of all three heads, which begin proximal to the tarsus, course through the tarsal groove on the plantar medial aspect of the hock within a synovial sheath known as the tarsal sheath. The tendons of all three heads blend at the level of the proximal metatarsus to form the DDFT.^{47, 50} The remainder of the tendon is arranged in the same manner on the distal hindlimb as it is in the forelimb (Figure 7).⁴⁹ The accessory ligament of the DDFT is typically smaller in the hindlimb than the forelimb.⁵⁰

iii. Blood Supply of the Equine Flexor Tendons

Blood supply to tendon originates from muscular origins and osseous insertions, as well as

accessory ligaments, paratenon, and mesotenon within synovial sheaths.^{15, 44, 51} Vasculature at muscular origins and osseous insertions only supply approximately 25% of the proximal and distal portions of tendon.⁴⁴ The paratenon and mesotenon are assumed to play an important role for the remainder of tendon. However, previous studies in which equine digital flexor tendons were stripped of the paratenon failed to result in ischemic damage to the tendon, indicating the importance of an intratendinous blood supply.^{15, 52} Based on the microvascular anatomy determined by microradiographs, this intratendinous blood supply was found to be most abundant around the periphery of the tendon.⁵¹ Further investigation *in vivo* using ¹³³xenon injected intratendinously revealed that the SDFT has a blood supply similar to that of resting skeletal muscle, with a recorded blood flow of 1 to 2mL/min/100g.^{45, 51, 53} Blood flow is capable of increasing during exercise and also following injury by over 300%.^{15, 45, 53}

iv. Function of Tendons of the Equine Limb

Tendons connect muscle to bone, transmitting force from muscle to move and stabilize joints.³⁹ In addition, tendons act to absorb impact during weight-bearing.⁴⁷ Tendons are either spring-like or positional in function, and have a variation in morphological, molecular and mechanical properties based on their action.³⁹ Tendons that function as positional tendons transmit muscle-generated force to bone, resulting in movement around a joint. Based on their function, positional tendons are required to be relatively inextensible under physiologic loads. An example of a positional tendon is the equine common digital extensor tendon (Figure 9).³⁹ Spring-like tendons store and release elastic strain energy, increasing efficiency of locomotion. Examples of spring-like tendons include the equine superficial and deep digital flexor tendons (Figure 9).^{39, 45, 54, 55}

The equine distal limb is unique from other species due to adaptations obtained during

evolution, allowing horses to move faster and more efficiently.^{1, 45, 47, 56} The equine distal forelimb consists of a single digit, the third metacarpal/metatarsal bone, with vestigial second and fourth metacarpal/metatarsal bones.⁴⁵ The digital flexor tendons and the suspensory apparatus are arranged on the palmar/plantar aspect of the limb, providing support to the MCP/MTP joints.^{1, 47} The digital flexor tendons and suspensory ligament are able to withstand high weight-bearing loads.¹ To facilitate their action, digital flexor tendons contain a significant amount of connective tissue and have accessory ligaments, which transfer load directly to bone. This allows the musculotendinous unit to withstand greater loads than associated muscles on their own and prevent overextension of the MCP/MTP joints during exercise due to muscle fatigue.⁴⁷ In addition to supporting the MCP/MTP joints, the SDFT stabilizes the PIP joint by limiting its flexion due to high tension experienced by the tendon during weight bearing (exerts palmar/plantar force on the joint) and preventing lateromedial and rotational displacement of the joint due to insertion of the distal branches on the proximal sides of the middle phalanx and direct contact of the distal branches with the condyles of the proximal phalanx.⁴⁷ The SDFT also aids in preventing hyperextension of the carpus.⁴⁷ In contrast, tension placed on the DDFT facilitates flexion of the PIP joint during weight-bearing, opposing the action of the SDFT.⁴⁷ This further contributes to stabilization of the PIP joint.⁴⁷ Dorsally located digital extensor tendons are responsible for extending the limb during protraction. In comparison to the digital flexor tendons, digital extensor tendons have higher structural stiffness.⁴⁵

Muscle is located proximally on the limb primarily to decrease the weight of the distal limb.⁴⁵ Due to proximally located musculature, associated tendons are long to traverse the joints of which they support.⁴⁵ Although it was traditionally thought that the digital flexor muscles were responsible for limb flexion, recent investigation has suggested that these muscles fix the origin of

the digital flexor tendons. *In vitro*, the superficial digital flexor muscle had a maximal contraction of 2mm.⁵⁵ It is hypothesized that these muscles dampen high-frequency limb vibrations when the foot hits the ground during weight-bearing.^{55, 57} Therefore, loading of the digital flexor tendons can be considered a passive process where energy is stored by elastic properties of the tendon and returned when the limb flexes.⁵⁵ Overall, this allows the horse to reach high speeds with reduced energy required for anterior movement of the limb.

Biomechanics of Tendon

Tendons are viscoelastic in nature, having variable stiffness as they stretch during active loading.⁵⁶⁻⁵⁸ When stretched, they return to their original state once stress is removed. It is believed that this is due to interactions of their constituents including collagen, water, surrounding proteins and ground substance.⁵⁹ These properties are time-dependent and change with different rates of loading.⁵⁹

Tendons are typically analyzed *ex vivo*, using mounted specimens to obtain load-elongation and stress-strain curves (Figure 10, 11). This allows evaluation of the mechanical and structural properties of tendons.^{45, 60} A load-elongation curve provides information regarding the tensile capacity of a tendon to the point of failure (Figure 10).⁶⁰ The slope of the curve is considered tendon stiffness (N/mm), which represents the amount of elongation a tendon can sustain before it fails (Figure 10).^{59, 60} The highest load placed on a tendon before failure is the ultimate load (N) (Figure 10).⁵⁹ Ultimate elongation (mm) is the maximum elongation of a tendon at failure (Figure 10).⁵⁹ The energy absorbed at failure (N/mm) is the area under the entire curve, which represents the maximum energy stored by a tendon (Figure 10).^{59, 60}

There are several regions in a load-elongation curve that characterize the behavior of a tendon. The first region is the non-linear “toe” region, in which changes in elongation are believed

to be the result of the change in crimp pattern of a tendon, as a result of elongation of tendon fascicles (Figure 10).¹⁵ In this region, tendon stretches easily without force due to a loss in crimp. As load increases, tendon stiffness increases and elongation changes, represented as the linear region in a load-elongation curve (Figure 10). It is observed as a sudden increase in slope of the curve following the toe region (Figure 10). Elongation is attributed to the sliding of tendon fibrils relative to one another.⁶⁰ When the linear region is surpassed, tendon fiber failure occurs in an unpredictable manner resulting in a load-elongation curve that ends abruptly, or curves downward as a result of permanent tendon damage (Figure 10).⁶⁰ When the load-elongation curve levels off, the load value is designated P_{lin} , or the yield-point of the tendon (Figure 10).

A stress-strain curve is useful in evaluating tensile deformation of a tendon (Figure 11). It is different from a load-displacement curve due to the fact that it accounts for the cross-sectional area (CSA) of the material being tested. The elongation of a tendon is expressed as strain (ϵ) along the x-axis, which is the deformation of a tendon calculated as a percentage of the original length of the tendon. Strain is dependent on cross-sectional area, force applied to the tendon, as well as material properties of the tendon tissue.³⁹ Stress (σ), expressed on the y-axis, is the force per unit area. The tensile strength is the maximum stress achieved (N/mm^2), the ultimate strain (in percentage) is the strain at failure, and the strain energy density (Mpa) is the area under the stress-strain curve.⁶⁰ When tendons are exposed to increased loading rates, the linear portion of the stress-strain curve becomes steeper, indicating greater stiffness of the tissue at higher strain rates. With higher strain rates, tendons store more energy, require more force to rupture and undergo greater elongation.⁶⁰

Ultimate tensile strength for the equine SDFT has been previously reported to be 12,000N. Typically, rupture occurs at the mid-metacarpal region of the tendon. At heel strike during normal

movement, loads rise very quickly in the SDFT due to its role in supporting the MCP/MTP joint above the ground.⁴⁵ In contrast, increase in load in the DDFT is slower.⁴⁵ This may explain why the SDFT is more prone to injury.⁴⁵ Equine flexor tendons can extend 10 to 12% of their original length before failure. However, this is the ultimate tensile strain, reflecting final strain prior to rupture. Recent evaluation of normal strains in flexor tendons *in vivo* in ponies using force plate analysis and kinematic motion analysis systems with markers revealed a normal strain of 2 to 4% at a walk and 4 to 6% at a trot. In a racing Thoroughbred, normal strain can reach 16% in the SDFT at a gallop.⁵⁶

Physiology of Tendon Repair

Restoration of normal tendon function following a laceration requires reestablishment of tendon fiber pattern, as well as gliding function. Overall, healing after primary tendon repair is a slow process⁶¹ and can take up to 240 days post-injury before a repair site reaches strength similar to adjacent normal tendon.⁴⁴ Once injury occurs, the body initiates a sequence of repair events, characterized by tissue inflammation, cell proliferation and remodeling. Following acute tendon injury, blood vessels rupture and signaling molecules released by intrinsic cells, such as cells from the epitenon and endotenon, will initiate a coagulation cascade, resulting in blood clot formation.⁶² The resulting blood clot stimulates enzymatic release of various chemotactic factors including transforming growth factor- β (TGF- β), insulin-like growth factor-1 (IGF-1), other pro-inflammatory molecules and vasodilators.⁶² Erythrocytes, platelets and inflammatory cells (neutrophils, monocytes and macrophages) migrate to the wound site and remove necrotic debris through phagocytosis. These cells, along with cells of the surrounding intact endotenon and epitenon, stimulate release of vasoactive and chemotactic factors, such as platelet derived growth factor (PDGF), vascular endothelial growth factor (VEGF), basic fibroblast growth factor (bFGF)

and IGF-1, which recruit tenocytes to begin collagen synthesis and deposition, initiate the proliferative phase and stimulate the formation of an extensive blood vessel network at the site of injury.⁶² The resultant repair tissue is highly cellular and contains relatively large amounts of water and extracellular matrix components compared to normal tendon. This tissue establishes continuity and partial stability early on at the site of injury.⁶²

Collagen synthesized by tenocytes is initially arranged in a random manner, and consists primarily of type III collagen.⁶³ Continued recruitment and proliferation of tenocytes and fibroblasts at the wound site are responsible for the synthesis of collagens, proteoglycans and other components of the extracellular matrix.⁶² Remodeling occurs 14 to 21 days after initial injury and lasts for several months.⁶² During this stage, there is a decrease in cellularity, synthesis of matrix components and type III collagen, and increased synthesis of type I collagen.⁶⁴ Type I collagen fibers are arranged longitudinally along the long axis of the tendon and contribute to the strength of the repair tissue.⁶⁴ As remodeling progresses, interaction between collagen structural units increases tendon stiffness and tensile strength. Despite this process, the repair tissue never achieves the characteristics of normal uninjured tendon.⁶² Resultant scar tissue has a higher ratio of type III collagen (up to 50% in injured tendon; 10% in normal tendon) to type I collagen in contrast to normal tendon.⁶³ It also has a higher hydration status and higher levels of glycosaminoglycans when compared to normal tendon.^{44, 63}

It is believed that the mechanisms for tendon healing occur through both extrinsic and intrinsic processes. Extrinsic repair consists of cellular infiltration from the paratenon or tendon sheath, resulting in angiogenesis and proliferation of fibroblastic cells. Intrinsic repair occurs through migration and proliferation of cells from the endotenon and epitenon into the site of injury. In cases of severe tendon disruption and, or, transection, extrinsic repair may predominate.⁴⁴ This

can result in increased peritendinous adhesions, which have a negative impact on gliding function of a tendon.⁴⁴ In addition, these cases are usually managed with prolonged external coaptation, which further promotes extrinsic repair.⁴⁴

Flexor Tendon Lacerations

Due to a lack of soft-tissue protection in the equine distal limb, any laceration of the dorsal or palmar/plantar surface of the distal limb in a horse may involve a tendon (Figure 12).^{44, 65} Possible causes include direct trauma from sharp objects, such as barbed wire, or blunt force trauma as result of a kick from another horse.⁴⁴ Lacerations can range from involving only the paratenon to full transection of a tendon. If a tendon is not completely transected, digital palpation of the lacerated tendon while passively moving the distal joints can aid in determining the integrity of a lacerated tendon.^{44, 65} Identification of a partially lacerated tendon is critical because, if left untreated, it may progress to complete tendon disruption.⁴⁴ Ultrasound can be used to determine the percentage of tendon still intact if a partial laceration is present.⁶⁵

Complete transection of a tendon will usually result in a change in posture or gait that will be noticeable on initial examination of the horse.⁶⁵ Complete transection of the SDFT results in mild hyperextension of the MCP/MTP joint from partial loss of support.^{44, 65} Complete transection of the SDFT and the DDFT results in hyperextension of the MCP/MTP joint and distal interphalangeal joint (DIP), causing the toe to flip upwards (Figure 13).^{44, 65} Complete transection of both digital flexor tendons and the suspensory ligaments results in complete loss of support of the MCP/MTP joint.⁴⁴ As a result, the fetlock will rest on the ground, and the toe will be flipped upwards.

Lacerated tendons exhibit diminished mechanical properties compared to normal tendons. *In vitro*, lacerated tendons have been shown to have decreased strength, decreased stiffness and

increased compliance (increased strain under a given load).^{58, 66-70} As the size of a laceration increases from partial to complete, there is increased mechanical compromise of the tendon. In a study by Kondratko and colleagues in 2013, the effects of different laceration sizes on elastic and viscoelastic mechanical properties of tendons were investigated in digital flexor tendons harvested from porcine lower limbs.⁵⁸ It was found that as laceration size increased across a tendon, the mechanical properties of the tendon, including stiffness and viscoelastic properties, became progressively more compromised.⁵⁸ In a clinical setting, this may increase the risk of further tendon damage or complete tendon rupture.⁵⁸ Also, as the size of a laceration increases, it can have a negative impact on a tendon's ability to absorb, store and transfer energy like a normal tendon.⁵⁸

Initial Treatment of a Tendon Laceration

During evaluation of a potential tendon laceration, the patient should be assessed for blood-loss due to the fact that major arteries and veins of the distal limb are in close proximity to the digital flexor tendons and can easily become damaged.⁶⁵ If blood-loss is evident, the patient should be evaluated for hypovolemia prior to use of chemical restraint or general anesthesia.⁶⁵ Chemical restraint and general anesthetics reduce myocardial contractility, cardiac output and vascular responsiveness. In addition, they cause vasodilation.⁷¹ Treatment consists of intravenous volume replacement and pressure bandaging or ligation to control active bleeding. Due to the potential of significant soft tissue injury, contamination and, or, active infection at the site of a laceration, broad spectrum antibiotics are typically administered. Common systemic antibiotic combinations include penicillin (22,000IU/kg administered intravenously (IV) once every six hours for Penicillin G Potassium or 22,000IU/kg administered intramuscularly (IM) once every 12 hours for Procaine Penicillin G) and gentamicin (6.6mg/kg administered IV once every 24 hours) or penicillin and amikacin (21mg/kg administered IV once every 24 hours). Duration of

administration is dependent on the patient's response to the medications. Regional limb perfusion with antimicrobials can also be performed using one third of the systemic dose of a selected antimicrobial, diluted to 30mL with sterile saline if the wound is located within the distal limb and the tourniquet is placed distal to the carpus and tarsus.⁷² If the tourniquet is placed proximal to the carpus or tarsus, a selected antimicrobial should be diluted to 60mL with sterile saline.⁷² The tourniquet is typically left in place for 30 minutes post-injection (Figure 14).⁷² Esmarch and pneumatic tourniquet systems can be used for regional limb perfusion.⁷³ For the pneumatic tourniquet system, an inflatable pneumatic cuff is placed 10 to 15cm proximal to the site of regional limb perfusion.⁷³ If the tourniquet is placed proximal to the carpus or tarsus, rolled gauze is applied over blood vessels on either side of the limb prior to tourniquet application to facilitate blood vessel occlusion.⁷³ In a study by Alkabes et al. in 2011, it was determined that an Esmarch tourniquet was more effective at maintaining antibiotic regionally than a pneumatic tourniquet when tourniquet was placed distal to the carpus in a standing horse.⁷⁴ This should be taken into account when selecting a tourniquet for regional limb perfusion.⁷⁴ Determining the patient's tetanus vaccination history is important and vaccination should be considered if the patient is not up-to-date or the vaccination history is unknown.⁶⁵

The injured limb should be properly immobilized prior to transport to a surgical facility to prevent further damage to injured tendons and preserve neurovascular structures of the distal limb.⁴⁴ Immobilization can be achieved through placement of a well-padded splint.^{44, 65} Commercially available splints, such as the Kimzey Leg Saver (Figure 15) (Kimzey, Woodland, CA) can be used, or a splint can be constructed out of polyvinyl chloride (PVC).^{44, 75} Other materials that can be used include metal rods, broom handles, and wooden boards.⁷⁵ If a PVC splint is used, it should be applied to the dorsal surface of the forelimb, and plantar surface of the

hindlimb, extending from the proximal metacarpus/metatarsus to the ground.⁴⁴ For the forelimb, the fetlock joint is pulled dorsally into the splint; this keeps the limb in slight flexion.^{44, 76} For the hindlimb, the fetlock joint is pulled in a plantar direction into the splint.^{44, 76} The splint is placed on the plantar aspect of the hindlimb as opposed to the dorsal aspect, due to the fact that splints tend to break more readily when placed on the dorsal aspect of the distal hindlimb.^{77, 78} It is important to ensure with application of a splint, the limb is properly padded to prevent the splint from causing further soft tissue damage to the limb.⁷⁵ This is achieved with the use of layered padding with each layer tightened with non-adhesive gauze.⁷⁵

Prior to tenorrhaphy, the wound should be debrided to aid in reducing bacterial load and contamination, which may have a negative impact on future healing.^{44, 79} All contaminated, avascular and necrotic tissue should be removed from the wound site, which can be achieved through sharp dissection. Instruments that can be used for sharp dissection include a scalpel, scissors and laser, with the scalpel being the least traumatic.⁷⁹ In addition, the wound can be irrigated to further reduce contamination. A large volume of a non-cytotoxic solution delivered at 10-15 PSI is an ideal irrigation solution (Figure 16).^{44, 79} High-pressure lavage is discouraged as this can push debris into deeper tissue layers in the wound.⁷⁹ Once the wound is adequately debrided, tenorrhaphy can be performed if possible.⁴⁴

For tendon lacerations within the digital flexor tendon sheath, the injury sustained by the tendon sheath should be treated as if the tendon sheath was septic, especially if the injury has been present for greater than three hours, or if the wound is heavily contaminated.⁷⁶ During surgical debridement, the tendon sheath should be lavaged with several liters of lactated Ringer's solution.⁷⁶ Regional limb perfusion can be performed at the time of surgery and a fenestrated polyethylene tube can be inserted into the tendon sheath to facilitate post-operative flushing, as

well as intrathecal antibiotic administration.⁷⁶ To maintain patency of the tubing post-operatively, the tube can be flushed with heparinized saline.⁷⁶ As an alternative to copious lavage of the tendon sheath and placement of polyethylene tubing, McNally et al. reported performing a tenosynoviotomy of the digital flexor tendon sheath with concurrent tenotomy of the distal lateral branch of the SDFT for treatment of tendon sheath sepsis.⁸⁰ When performed on clinical cases of chronic, refractory sepsis of the digital flexor tendon sheath, 71% of patients (5/7) treated were serviceable for their intended use as described by their owner.⁸⁰ One patient in this study had sustained a laceration to the SDFT within the tendon sheath, which involved approximately 90% of the tendon. It was not specified in this study if the tendon was sutured as part of the treatment. This particular patient was one of the five horses that was considered serviceable for its intended use by the owner.⁸⁰

Post-Operative Management of Flexor Tendon Lacerations in Horses

Once tenorrhaphy is complete, the external wound can be sutured closed. One critical layer to consider with closure is the paratenon.⁷⁶ By closing the paratenon, a highly vascular layer is covering the repair site, which provides cellular elements for healing, as well as provides a protective layer to aid in reducing post-operative infection.⁷⁶ Subcutaneous tissue and skin should be closed when possible to further protect the area and prevent the formation of exuberant granulation tissue.

Due to the fact that current repairs of equine digital flexor tendons cannot support early post-operative tensile loads, it is recommended that all horses are maintained in prolonged external coaptation post-operatively. This consists of placement of a fiber-glass cast on the distal limb, which incorporates the hoof.⁷⁶ The distal limb is typically casted in slight flexion to reduce strain on the flexor tendons post-operatively.⁸¹ It has been recommended to leave the cast in place for

six to eight weeks, or until the repair can withstand physiologic tensile loading.⁸¹ One author argued that external coaptation should only be maintained for four weeks post-operatively as it has been found in one *in vivo* equine model that tendon strength following maintenance of a cast for four weeks was similar to tendons that were maintained in external coaptation longer.^{3, 76} The cast should be removed with the patient sedated and restrained to avoid catastrophic failure of the repair.⁷⁶

Following cast removal, it is recommended transitioning horses to splints and bandages, as needed, until sufficient tensile strength has been gained at the site of injury.^{65, 82} If the DDFT was involved in the initial injury, an extended heel shoe should be placed on the hoof of the involved limb (Figure 17).⁷⁶ The shoe is usually maintained in place for several months following removal of the cast.⁷⁶ This prevents compromise of the repair site by preventing hyperextension of the distal interphalangeal joint.⁷⁶ An elevated heel shoe could also be applied to reduce tension on the deep digital flexor tendon.⁶⁵ If the SDFT is involved, placement of a wedge shoe alone should be avoided as placement of the wedge encourages extension of the SDFT by a compensatory drop in the fetlock joint.⁴⁷

The most common complication associated with application of a fiber-glass cast in horses is the development of cast sores, which has a reported incidence of up to 81%.^{82, 83} Cast sores most commonly occur on the dorsal aspect of the proximal metacarpus/metatarsus, as well as the palmar/plantar aspect of the proximal sesamoid bones.⁸³ As an alternative to a fiber-glass cast, Whitfield-Cargile et al. investigated the use of a fetlock support brace in 15 clinical cases with flexor tendon lacerations.⁸² The fetlock support brace consists of a 120cm x 1.5cm stainless steel rod, bent to act as a palmar/plantar support for the limb, welded to a flat steel shoe with 2.5cm-3.5cm extended heels.⁸² Eyelets are welded to the bent stainless steel rod to allow for placement

of nylon straps and a PCV splint on the dorsal aspect of the limb. A stainless steel pipe is placed below the extended heels of the shoe to mimic a patent wedge shoe (Figure 18).⁸² The fetlock support brace is applied to the limb following placement of a heavy Robert Jones-type bandage.⁸² Injuries ranged from partial laceration of only the DDFT to complete transection of the SDFT and DDFT. All horses were treated with surgical debridement of the wound, and the wound was sutured closed if possible. Tenorrhaphy was not performed in any of the cases in this study. Patients presented at a mean of 6.9 days from the initial injury, which the authors concluded was too long of a duration from initial injury to consider primary repair of the tendons. Duration of fetlock support brace application ranged from 63-143 days. Complications encountered with placement of the fetlock support brace included mild superficial abrasions on the dorsal aspect of the cannon bone in 46% of cases. This was resolved by applying more padding with subsequent bandage changes. Nine out of the 15 horses (60%) were sound and returned to their previous level of work following treatment and application of this brace.⁸² Mean time from the initial injury to return to work was 14.5 months.⁸² Three out of the 15 horses were reported to be sound by the owners, but at a lower level of work.⁸² The remaining three horses remained persistently lame.⁸²

Exercise during the first six to eight weeks post-operatively consists of strict stall rest.⁸⁴ Following that time, ultrasonography should be performed on the affected tendon(s) to determine further exercise instructions. Exercise can transition from strict stall-rest to stall-rest with hand-grazing or brief periods of hand-walking.⁸⁴ Repeat ultrasound can be performed every 60 to 90 days to aid in tailoring a specific controlled exercise program for the patient.⁷⁶ Typically, a total convalescence time of six months to a year should be expected depending on the degree of the injury.^{76,84}

Characteristics of an Ideal Tendon Laceration Repair

i. Sutured versus Un-sutured Tendons

Suturing lacerated tendons can produce a mechanically stronger and histologically superior repair tissue as opposed to leaving a tendon laceration un-sutured.^{3,5} In humans, primary repair of flexor tendon lacerations is advocated in order to restore manual dexterity of the hand. The impact of suturing flexor tendons in horses has been previously investigated. Bertone et al. used an *in vivo* model to compare primary repair to second intention healing following removal of 3cm sections from the mid-point of equine SDFT.⁵ The first group was left un-sutured, the second group was repaired with a double locking-loop (LL) suture pattern (Figure 19) using carbon fiber, and the third group was repaired with a double LL pattern using nylon suture. Horses were euthanized at six, 12 and 24 weeks post repair. The breaking stress, which is defined as the breaking load divided by the cross-sectional area of the tendon, was greatest for nylon sutured tendons at 24 weeks post-repair. This means that the repair site was able to support larger loads prior to failing while having a smaller cross-sectional area than the repairs made with carbon fiber suture and tendons left un-sutured. Scar tissue present at the site of repair was mature, arranged longitudinally and had minimal inflammation.⁵ In contrast, the carbon fiber repairs had a marked inflammatory response and significant tissue necrosis.⁵ Scar tissue in un-sutured tendons was moderately organized and comparatively had a bigger callous than the tendons sutured with nylon suture. Based on these findings, sutured repairs healed with a more mature scar tissue that was as strong as the larger callous formed in un-sutured tendons.

In an *in vivo* study, Jann et al. compared un-sutured tendon lacerations to those repaired with a three-loop-pulley technique (3LP) (Figure 20) using #1 polyglyconate (Maxon).³ Common, lateral and long digital extensor tendons were tested, as well as the SDFT and DDFT. Load to

failure was significantly greater for sutured versus un-sutured repairs at five (average of 2427N for sutured, versus 923N unsutured) and nine weeks (average of 6363N for sutured, versus 5364N unsutured) post-repair. Histologically, scar tissue at the repair site for sutured tendons was more mature than un-sutured tendons.³ These findings agree with findings made in the previous study conducted by Bertone et al.^{3,5}

ii. Time to Laceration Repair

There has been little investigation on the optimal time for tendon laceration repair. In humans, gross contamination of a wound may preclude early repair of flexor tendon lacerations.⁸⁴ In these particular cases, the laceration may be managed by delayed repair several days after the initial injury once contamination and infection have been successfully managed.⁸⁴ However, this could potentially have a negative impact on the gliding function of the tendon, especially in the human hand where manual dexterity is critical. To determine the most optimal period for flexor tendon laceration repair, Gelberman and colleagues investigated the effect of delayed flexor tendon laceration repair using a canine model.⁸⁵ Animals were divided randomly into three groups, and laceration repair was performed on the flexor digitorum profundus tendon immediately, at seven days or 21 days post-laceration.⁸⁵ A significant benefit to early repair performed immediately after injury was observed, resulting in improved gliding function post-operatively.⁸⁵ There was no significant impact on the total concentration of collagen or strength of the tendon if the repair was delayed. Based on the improved functional characteristics of early repaired tendons, it is recommended that tendon repair is ideally performed within the first week of injury.⁸⁵

In horses, due to the fact that the digital flexor tendons contribute to weight-bearing, delayed repair of contaminated wounds is not recommended.⁸⁴ Therefore, surgical debridement of the wound is critical to reduce contamination and the risk of post-operative infection. If gross

contamination and, or, active infection at the site of a tendon laceration precludes immediate repair, one author recommended immobilizing the limb in a Kimzey Leg Saver splint (Figure 15) with frequent bandage changes until delayed closure could be performed.⁷⁶

iii. The Ideal Suture Pattern

The development of an ideal suture pattern has been investigated in great detail in human medicine for flexor tendon laceration repair of the hand. Goals of a successful repair including restoring gliding function of the tendon, reducing peritendinous adhesion formation and preserving the overall function of the tendon.⁸⁶⁻⁹¹ This is critical in the hand to ensure maintenance of manual dexterity. Several studies have investigated the effects of post-operative immobilization on healing at tenorrhaphy sites; specifically in human, canine, porcine and avian models.⁶¹ Using a canine model, Gelberman and colleagues investigated the effects of immobilization versus early post-operative mobilization on tendon laceration repairs. Immobilization versus early post-operative mobilization was controlled with application of shoulder spica casts.⁶¹ Repaired tendons were studied by biomechanical, microangiographic, biochemical and histologic techniques at intervals through 12 weeks. Early mobilized tendons showed improved gliding function, higher tensile strength and increased stiffness when compared to immobilized tendons.⁶¹ In addition, early mobilization improved peritendinous vessel density and configuration, as well as increased total DNA content at the repair site.⁶¹ Finally, immobilization resulted in the formation of peritendinous adhesions, which were speculated to be due to a healing process dominated by extrinsic repair.⁶¹ In contrast, mobilized tendons developed a smooth gliding surface of the repair site, consisting of cells from the epitenon, and the repair was maintained free of adhesions through 42 days post-repair.⁶¹

Further human studies support that immobilization results in decreased repair strength,

especially during the initial post-operative period. Early passive and active motion have been shown to prevent the initial weakening of a repair, leading to progressively increasing repair strength starting from the time of repair.⁹² A major disadvantage of early active mobilization is that it generates high cyclical forces across the repair site, which often exceed the mechanical properties of commonly used repair patterns.⁸⁸ Schuind et al. reported forces of up to 34.3N during active flexion of the human finger.⁹³ This was in a normal tendon and did not take into account the presence of post-surgical edema or possible adhesions, which would further increase the work of flexion.²¹ To facilitate early post-operative mobilization, repairs have to be strong, and result in minimal gap formation with early tensile loading, which can be as early as five days post-repair.⁹⁴ Important variables of tendon repair that help achieve these goals include core purchase length of suture, suture strand number, type and size, and grasping versus locking attributes of the suture.^{8, 88, 94-97} In horses, post-operative immobilization is warranted because early post-operative passive mobilization would be difficult and impractical due to poor patient compliance.

Gliding resistance is a measure of the force which resists relative motion of a tendon. In uninjured tendons, gliding resistance is low due to the smooth gliding surface of the tendon, as well as the presence of lubricating synovial fluid in tendon sheaths. Following tendon injury, gliding resistance has been reported to increase by at least 50% in humans.⁹² Increased gliding resistance after tendon injury and repair can impair the motion of the tendon, increasing the work of flexion as well as the risk of post-operative adhesion formation. Contributing factors of increased gliding resistance following tendon injury and repair include edema of surrounding tissues, suture technique, gap formation and wound healing. Specific factors involved in suture technique include the number of exposed suture loops and knots on the outside of the tendon surface, suture caliber, and suture material. In a study by Zhao et al. on human cadaver hands,

gliding resistance was greatly influenced by the choice of repair method used for partially lacerated tendons. In this study, gliding resistance following repair was four times greater than that of an intact tendon. As the number of suture strands, knots located outside of the tendon and locking loops on the surface of the tendon increased, so did gliding resistance.¹⁴ Multiple suture strands and loops increase suture strength. Therefore, this raises the question of what is more important for tendon laceration repair: gliding resistance or repair strength?

Low gliding resistance suture techniques should not be overemphasized, as their decreased strength can result in gap formation. Gap formation at a tenorrhaphy site post-operatively is a common complication, and currently, there is no repair technique described *in vivo* that prevents gap formation in horses.⁶ Typically gap formation occurs within the first few weeks following tendon repair. In humans, it is attributed to applied tension during either active motion or passive therapy, of which exceeds the tolerance of suture repair.¹⁴ Gaps as small as 1 to 2 mm at a tenorrhaphy site for human flexor tendon laceration repair have been associated with poor functional performance, resulting in increased gliding resistance, peritendinous adhesion formation, reduced repair strength and possible tendon rupture.^{14, 98} In some cases, a second surgical procedure is required for lysis of peritendinous adhesions.⁹⁹ In a study by Zhao and colleagues in 2004 using a human cadaver model, gliding resistance increased by 100% from a 1mm to 2mm gap at a tenorrhaphy site and nearly 600% from an intact tendon to a tenorrhaphy site with a 2mm gap.¹⁴ In the same study, a threshold effect was observed with a gap between 2mm and 3mm. Beyond this point, catastrophic failure of the repair was likely. Using a canine model, Gelberman and colleagues found that a gap greater than 3mm at a tenorrhaphy site impaired healing and increased the risk of rupture within the first six weeks of rehabilitation through passive flexion by decreasing the tensile properties of the repair site.⁹⁸ Therefore, gap formation can have

a detrimental effect on overall tendon healing and possibly result in poor clinical outcomes.¹⁴ As a result, using a repair technique that has reduced gliding resistance, but is strong enough to reduce potential formation of a gap post-operatively should be considered. In horses, the digital flexor tendons have a shorter amplitude of motion and do not change direction as substantially as the flexor tendons of the human hand.¹⁰ Therefore, gliding function may not be as important a variable to maintain post-operatively as the strength of repair for equine flexor tendon lacerations.¹⁰

1. Suture Type

For flexor tendon lacerations of the human hand, suture material is typically chosen based on characteristics including high tensile strength, ease with handling, good knot security, minimal tissue reaction, prolonged retention of strength post-operatively, as well as low extensibility.¹⁰⁰⁻¹⁰³ Currently, there is no clear consensus on the best type of suture material to use for tenorrhaphy and there are several studies that have investigated the use of different suture materials for flexor tendon laceration repair. As a result, selection of a particular suture material is usually dependent on surgeon preference.¹⁰⁴

In previous studies, the use of stainless steel monofilament suture was advocated due to its high tensile strength, good knot security, low extensibility and low tissue reactivity.^{100, 105} In a study by Ketchum et al., six different suture materials were evaluated for flexor tendon laceration repair, including monofilament stainless steel, polymerized caprolactan (Supramid; S. Jackson Inc., Alexandria, VA), monofilament nylon (Ethilon; Ethicon, Somerville, NJ), monofilament polypropylene (Prolene; Ethicon, Somerville, NJ), and braided polyester (Ethibond; Ethicon, Somerville, NJ).¹⁰⁵ Canine flexor tendons were repaired with a long mattress suture technique and strength characteristics were evaluated either immediately or at three weeks post-repair. It was

found that immediately after repair and at three weeks post-repair, monofilament stainless steel was most resistant to gap formation and rupture and had good knot security.^{102, 105} Although monofilament stainless steel is strong and has good knot security, it is a very difficult suture to handle, and often kinks.¹⁰⁰ Also, due to its high tensile strength, it is typically stronger than surrounding tissue, and will often pull through tissue with applied strain to a tenorrhaphy site. In a study by Mangus et al., microscopic examination of monofilament stainless steel revealed even with gentle manipulation of the wire with needle drivers, marked roughening and flaking developed on the surface of the wire. The presence of these changes in the wire could contribute to the saw-like effect of the material.¹⁰⁶ For these reasons, monofilament stainless steel wire is not used for flexor tendon tenorrhaphy.

Suture is either absorbable or non-absorbable. Absorbable suture initially has equivalent or superior strength to non-absorbable suture; however, the strength of the material decreases over time as the suture dissolves by hydrolysis.^{107, 108} Braided absorbable sutures such as polyglycolic acid (Dexon; Covidien, New Haven, CT) and polyglactin 910 (Vicryl; Ethicon, Somerville, NJ) are commonly used in a variety of surgeries, however, they have a short half-life tensile strength. In an *in vivo* lagomorph study by Bourne et al., it was found that the half-life of tensile strength of polyglycolic acid suture and polyglactin 910 suture was 14 days after being placed in subcutaneous tissues.¹⁰⁷ Due to their short half-life, it precludes their use in situations in which prolonged retention of suture strength would be needed, such as tendon laceration repair. Monofilament absorbable sutures such as polydioxanone (PDS; Ethicon, Somerville, NJ) and polyglycolide-trimethylene carbonate (Maxon; Davis & Geck, Danbury, CT), have a smaller surface area available for hydrolysis, making it more difficult for the suture to dissolve.¹⁰⁷ As a result, these sutures retain their strength for a greater period of time.¹⁰⁷ In the same study by Bourne et al., it

was found that the half-life tensile strength for polyglycolide-trimethylene carbonate suture was three weeks after placement, and six weeks for polydioxanone suture.¹⁰⁷ The use of absorbable suture material has not been recommended clinically for flexor tendon laceration repair in humans as the half-life tensile strength of absorbable suture may prevent it from being able sustain post-operative mobilization over a prolonged period. There may also be a risk of increased tissue reaction and adhesion formation with absorbable suture material as it dissolves.¹⁰⁹ However, repairs in which absorbable suture is used may have improved gliding function post-repair, as excessive fibrosis and stitch granulomas may occur when non-absorbable sutures are placed.¹⁰⁸ In addition, non-absorbable suture could act as a nidus for infection, especially in a contaminated wound.¹¹⁰ In *in vitro* tensile tests performed by Bourne et al. on four different size 4-0 absorbable sutures, it was found that polyglycolide-trimethylene carbonate was the strongest, with an ultimate tensile strength of 106N. Of the sutures tested (polyglycolic acid, polyglactin 910, polyglycolide-trimethylene carbonate, and polydioxanone), polydioxanone was the weakest, with an ultimate tensile strength of 79N.¹⁰⁷ When knotted with a square knot, polyglycolide-trimethylene carbonate had the best knot security, while polyglycolic acid and polyglactin 910 both had poor knot security.¹⁰⁷ This quality could also preclude the use of polyglycolic acid and polyglactin 910 in tendon laceration repair as knot slippage could compromise the strength of a repair. Although polyglycolic acid had the highest tensile strength, a half-life of three weeks would not be ideal for tendon laceration repair.

Wada et al. compared 4-0 polydioxanone to 4-0 non-absorbable braided polyester suture (Ethibond) in an *in vivo* canine model in which post-operative mobilization was used. Polydioxanone was chosen due to its longer half-life tensile strength than other absorbable suture materials, as well as the fact that it has been found to be the least reactive in tissues compared to

other absorbable sutures.¹¹¹ Flexor digitorum profundus tendons were repaired with a four-strand, double-modified, locking Kessler core suture of the two different suture materials. A circumferential suture of 6-0 monofilament polypropylene was placed along the tendon ends for each repair. Reasoning for the use of a circumferential suture in addition to a core suture was not specified; it can only be assumed that this repair was chosen to reflect current clinical standards for human flexor tendon laceration repair. Repaired tendons were harvested seven, 14, 28, and 42 days post-operatively for mechanical testing and histologic evaluation. All repairs, whether polydioxanone or braided polyester suture was used, healed without ruptures or gap formation.¹⁰⁸ It was found that there was a reduction in strength of the polydioxanone suture repairs in the first 14 days, with ultimate strength decreasing from $48.1\text{N} \pm 4.3\text{N}$ to $40.1\text{N} \pm 3.1\text{N}$.¹⁰⁸ In contrast, the ultimate strength of the braided polyester suture repair increased from $48.2\text{N} \pm 8.3\text{N}$ to $52.5\text{N} \pm 5.8\text{N}$.¹⁰⁸ The decrease in strength of the polydioxanone repair over the first 14 days was suspected to be due to suture resorption, rather than weakening of the tendon ends with inflammation, as this same decrease in strength was not observed with the braided polyester suture repair.¹⁰⁸ Overall, both repairs were stronger than the maximum force applied to human flexor profundus digitorum tendons during active flexion, of 34.3N .⁹³ Repairs were also evaluated histologically at 14 and 42 days post-repair to assess inflammatory response. Inflammatory response was divided into four grades; no inflammatory response, slight inflammatory response (slight leukocyte infiltration within epitenon and endotenon), mild inflammatory response (leukocytes surrounding suture material within epitenon and endotenon), and severe inflammatory response (diffuse inflammation within tendon).¹⁰⁸ When evaluated histologically, there was either none or a very mild inflammatory response noted for the braided polyester suture repairs at 42 days post-repair. A slight inflammatory response surrounding the suture material of the polydioxanone repairs,

however, was observed by day 14. This progressed to moderate to severe in some of the repairs by day 42. The inflammatory response did not result in large peri-tendinous adhesion formation or large callous formation at the repair site.¹⁰⁸ Based on these findings, the authors concluded that absorbable suture could potentially have some use in flexor tendon laceration repair in humans, even when post-operative mobilization is used. One downfall of this study was the use of a non-absorbable circumferential suture (monofilament polypropylene) in combination with an absorbable core suture. Further investigation of the use of an absorbable circumferential suture combined with an absorbable core suture for flexor tendon laceration repair is warranted.

Commonly used non-absorbable synthetic suture materials for human flexor tendon laceration repair include uncoated (Mersilene; Ethicon, Somerville, NJ) and coated (Ethibond or Ti-Cron; Covidien, New Haven, CT) braided polyethylene terephthalate, monofilament nylon (Ethilon; Ethicon, Somerville, NJ) and monofilament polypropylene (Prolene).^{100, 104, 112} Coated braided polyethylene terephthalate differs from uncoated braided polyethylene terephthalate due to the presence of an external coating, which improves its handling characteristics compared to the uncoated braided polyethylene terephthalate.¹¹³ Ti-Cron differs from Ethibond due to the presence of a silicone coating on the exterior of the suture as opposed to polybutylate. As with polybutylate, the silicone coating aids in improving the handling characteristics and pliability of the suture.¹¹³ Lawrence et al. evaluated the mechanical properties of various non-absorbable suture materials, including monofilament nylon, monofilament polypropylene and coated braided polyethylene terephthalate.¹⁰⁴ Using an *ex vivo* porcine model, flexor tendon lacerations were repaired with a four-strand single cross suture technique and tested immediately following repair. It was found that coated braided polyethylene terephthalate created a stiffer ($10.2\text{N/mm} \pm 1.4\text{N/mm}$) for coated braided polyethylene terephthalate, $8.0\text{N/mm} \pm 0.9\text{N/mm}$ for monofilament

polypropylene, $6.0\text{N/mm} \pm 1.1\text{N/mm}$ for monofilament nylon) and stronger repair than monofilament polypropylene and monofilament nylon ($65.6\text{N} \pm 4.7\text{N}$ for coated braided polyethylene terephthalate, $63.4\text{N} \pm 10.7\text{N}$ for monofilament polypropylene, $46.7\text{N} \pm 5.9\text{N}$).¹⁰⁴ Coated braided polyethylene terephthalate was more resistant to gap formation than monofilament nylon ($52.3\text{N} \pm 4.8\text{N}$ for coated braided polyethylene terephthalate, $36.3\text{N} \pm 2.8\text{N}$ for monofilament nylon); there was no difference in resistance to gap formation between coated braided polyethylene terephthalate and monofilament polypropylene ($52.3\text{N} \pm 4.8\text{N}$ for monofilament polypropylene).

Trail et al. compared the mechanical properties of various 4-0 absorbable and non-absorbable suture materials used for flexor tendon laceration repair. When individual sutures were tested *ex vivo* by linear load to failure as unknotted strands, the three strongest suture materials were polyglycolide-trimethylene carbonate, monofilament stainless steel, and silicone coated braided polyethylene terephthalate ($23.95\text{N} \pm 3.78\text{N}$ for polyglycolide-trimethylene carbonate, $21.12\text{N} \pm 2.15\text{N}$ for monofilament stainless steel, $19.16\text{N} \pm 1.09\text{N}$ for silicone coated braided polyethylene terephthalate).¹⁰⁰ When knotted and tested by linear load to failure, silicone coated braided polyethylene terephthalate lost almost 50% of its strength.¹⁰⁰ This was in contrast to monofilament stainless steel, which lost only 10% of its initial strength, and polyglycolide-trimethylene carbonate, which lost 30% of its initial strength.¹⁰⁰ It was found that silicone coated braided polyethylene terephthalate required three to five throws of the knot to make it secure. Silicone coated braided polyethylene terephthalate was found to be the least extensible, extending $33.00\text{mm} \pm 10.43\text{mm}$ prior to failure. This was in contrast to monofilament stainless steel, which extended $37.82\text{mm} \pm 26.62\text{mm}$ prior to failure and polyglycolide-trimethylene carbonate, which extended $98.54\text{mm} \pm 23.48$ prior to failure.¹⁰⁰ The author concluded that based on its mechanical

properties, silicone coated braided polyethylene terephthalate may be a suitable suture to use for flexor tendon laceration repair.¹⁰⁰ One downfall is its poor knot security; more throws of the knot are required to keep it secure.

Although selection of a suture material with a high tensile strength is important, the viscoelastic properties of a suture material are also critical.¹¹² This includes stress relaxation and creep. Stress relaxation is a measure of a material's ability to relieve stress under constant strain. Creep is a measure of how much a material deforms over time with applied strain.¹¹² Vizesi et al. investigated the viscoelastic properties of three common suture materials used for flexor tendon laceration repair including 4-0 monofilament polypropylene, 4-0 monofilament nylon and 4-0 silicone coated braided polyethylene terephthalate.¹¹² Sutures were tested using a micromechanical tester with each suture material submerged in 0.9% phosphate-buffered saline maintained at room temperature and body temperature.¹¹² Changes in temperature were incorporated into the study to mimic physiologic conditions. It was found that changes in temperature had a significant impact on the stiffness of monofilament polypropylene and monofilament nylon, but not silicone coated braided polyethylene terephthalate. As the temperature increased, the stiffness of both monofilament polypropylene and monofilament nylon decreased.¹¹² Only monofilament polypropylene had significantly increased stress relaxation and creep ratios with changes in temperature. This means that over time and with an increase in temperature, monofilament polypropylene deforms irreversibly more than monofilament nylon and silicone coated braided polyethylene terephthalate when a constant strain is applied.¹¹² Based on these findings, the author recommended the use of silicone coated braided polyethylene terephthalate for flexor tendon laceration repairs.¹¹²

As with absorbable suture, inflammatory response to non-absorbable suture materials is

a concern for flexor tendon laceration repair.¹¹³ Using a rabbit model, Esenyel et al. placed various non-absorbable sutures in various tissues, including tendon.¹¹⁴ It was found that polybutylate coated braided polyethylene terephthalate was the least reactive in tendon tissue compared to other non-absorbable sutures at six weeks following placement of the sutures.¹¹⁴ In contrast, monofilament polypropylene caused a significant inflammatory response in tendon tissue.¹¹⁴

A similar investigation was performed by Carr et al., also using a lagomorphous model. Of eight common orthopedic sutures implanted into dorsal fascia of New Zealand white rabbits, silicone coated braided polyethylene terephthalate caused an intense foreign body inflammatory response.¹¹³ The reason for this degree of reaction is not known; it is speculated that it is due to silicone coated braided polyethylene terephthalate's unique composition of high molecular-weight polyethylene with recurrent polyethylene terephthalate aromatic rings.¹¹³ In a separate study evaluating 3-0 polyglactin 910 versus 3-0 silicone coated braided polyethylene terephthalate for tracheal anastomosis using a canine model, it was found that silicone coated braided polyethylene terephthalate resulted in grossly visible moderate to severe granuloma formation and inflammation (graded no visible inflammation or granuloma formation, visible inflammation, but no granuloma formation, moderate granuloma formation and severe granuloma formation) at the anastomosis site at 60 days following surgery.¹¹⁵ This is in contrast to anastomoses using 3-0 polyglactin 910 in which there was no visible granuloma formation or inflammatory response.¹¹⁵

Although silicone coated braided polyethylene terephthalate has good viscoelastic properties and is a relatively strong suture material, the fact that it can result in a significant foreign body inflammatory response as well as the fact it is prone to knot slippage may limit its use in flexor tendon laceration repairs. Monofilament nylon is typically avoided because it may

result in early gapping at a repair site and fails at low loads. Because monofilament polypropylene has been found to deform easily with applied strain, is weaker than other currently available suture materials, and can cause an intense inflammatory response when implanted in tendon tissue, it may not be the best suture material for flexor tendon laceration repairs. Currently, polybutylate coated braided polyethylene terephthalate tends to be favored for use clinically in humans as it provides a strong repair that is relatively inextensible, it has good handling characteristics and has been found to cause only mild inflammation of tissues.¹⁰⁴

Newer suture materials that have been tested for flexor tendon laceration repair include Fiberwire (Arthrex, Naples, FL) and multifilament stainless steel (MFSS) (Core Essences Orthopedics, Fort Washington, PA).^{103, 104} Fiberwire is a polyethylene-based braided suture coated in silicone. It has been previously reported to be superior in strength to other suture materials.¹⁰⁴ It consists of a core of several small biocompatible strands of polyethylene covered with braided polyester.¹¹⁰ When compared to braided stainless steel, monofilament polypropylene and monofilament nylon in an *ex vivo* study, Fiberwire and braided stainless steel were significantly stronger. When used to repair porcine deep flexor tendons *ex vivo*, gap formation and ultimate failure occurred at a significantly higher force for both Fiberwire and braided stainless steel compared to the other suture materials tested.¹⁰⁴ McDonald and colleagues investigated the use of MFSS (Core Essences Orthopedics, Fort Washington, PA) for flexor tendon tenorrhaphy.¹⁰³ The suture is composed of 49 stainless steel filaments arranged in seven bundles, with each bundle containing seven filaments.¹⁰³ This particular arrangement prevents to the formation of kinks in the material when handled. When compared to other suture materials used for flexor tendon tenorrhaphy, it was found that the MFSS suture had increased stiffness (47.1N/mm± 6.9N/mm), ultimate tensile strength (121±3.2N) and decreased elongation.¹⁰³ The next strongest suture

material, Fiberwire, had an ultimate tensile strength of $53\text{N} \pm 15.8\text{N}$ and a stiffness of $8.7\text{N/mm} \pm 0.8\text{N/mm}$. In addition, the MFSS suture had better knot security than Fiberwire.¹⁰³ One major downfall of this study was that MFSS was not tested in tendons, *ex vivo* or *in vivo*. Although Fiberwire and MFSS have improved stiffness and ultimate tensile strength to other commonly used suture materials, they have not been accepted for general use in tendon repair due to lack of *in vivo* testing.¹⁰³ In addition, it has been found that Fiberwire causes significant inflammation in tendon tissue compared to other types of non-absorbable suture, such as polybutylate coated braided polyethylene terephthalate and monofilament polypropylene.¹¹⁴

The ideal type of suture material for repair of equine digital flexor tendon lacerations has not been investigated to the same degree as in human medicine and the best suture material for repair of equine flexor tendon lacerations has yet to be determined.⁴ Suture is often selected based on a surgeon's preference.⁷⁶ The use of braided synthetic materials is generally avoided due to their potential for harboring infection between the filaments of the suture.¹¹⁶ This is critical due to the fact that there is a high risk of contamination associated with trauma of the distal limb of a horse. As a result, synthetic monofilament suture is considered the suture of choice.¹¹⁶ Synthetic monofilament suture that has been used for equine tenorrhaphy includes monofilament nylon, monofilament polypropylene, polydioxanone, polyglycolide-trimethylene carbonate and polymerized caprolactam (Supramid). Adair et al. reported that polydioxanone may be an ideal suture to use for equine flexor tendon tenorrhaphy due to the fact that it is an absorbable suture, and there was no significant difference found between the tensile strength of #1 polydioxanone and #1 monofilament polypropylene when tested by linear load to failure in 3LP and LL suture patterns.⁹ One author recommended using either monofilament absorbable polydioxanone or polyglycolide-trimethylene carbonate as they have a longer half-life tensile strength than other

absorbable sutures, and the risk of suture sinus development is reduced because the sutures are absorbable.⁷⁶ Based on previous studies, it may be beneficial to use polydioxanone over polyglycolide-trimethylene carbonate, as it will retain its tensile strength longer at the repair site than polyglycolide-trimethylene carbonate.¹⁰⁷

One synthetic implant that has been investigated in great detail for tendon injuries in horses is carbon fiber. Carbon fiber displays considerable strength under tension.¹¹⁶ A carbon fiber suture unit consists of a bundle of 10,000 carbon fiber filaments; this composition is known as a tow.¹¹⁶ Based on previous evaluation of the use of carbon fiber implants, it has been found that fibroblasts grow along the individual carbon fibrils. Subsequent collagen that is produced is arranged longitudinally, similar to normal tendon.¹¹⁶ In horses, carbon fiber implants were typically used as a treatment of tendinitis, in which the implant was surgically placed within diseased tendon.¹¹⁶ Valdez and colleagues investigated the use of carbon fiber implants for repair of transected equine digital flexor tendons.¹¹⁷ It was found that as early as 30 days post-repair, fibrous tissue developing around the carbon fiber implant closely resembled mature tendon tissue.¹¹⁷

Nixon and colleagues investigated the use of polylactic acid coated carbon fiber suture versus nylon suture for repair of SDFT lacerations *in vitro*.¹¹⁸ Ten horses of various breeds were included in the study. Under general anesthesia, sharp transection of the SDFT was performed on both forelimbs at the level of the mid metacarpus. One tendon was repaired using 1-0 monofilament nylon in a LL suture pattern (Figure 19), while the other tendon was repaired with polylactic acid coated carbon fiber suture in the same suture pattern. Both forelimbs were maintained in a neutral angle distal limb cast post-operatively. Horses were euthanized at six, eight, 12, 20 and 24 weeks post-operatively. Tendons were tested by linear load to failure using a materials testing machine as well as evaluated histologically. At the time of necropsy, a gap of 2

to 4 cm was noted in all repairs, despite the suture used. Although the carbon fiber suture had good handling characteristics, and did not display any fraying or knot slippage, it did result in a significant amount of fibrous tissue reaction within the tendon, especially at 8 to 12 weeks post repair.¹¹⁸ Cosmetically, this appeared as gross thickening of the repaired tendon, as well as a significant amount of swelling of the limb surrounding the repaired tendon (Figure 21).¹¹⁸ At 24 weeks post-repair, in tendons repaired with the carbon fiber suture, there was a layer of immature, cell-dense tissue in close association with the suture. This was in contrast to the nylon suture repairs, in which mature tendon tissue was observed adjacent to the suture. When harvested tendons were tested by linear load to failure, tendons repaired with nylon were stronger.¹¹⁸ This became more apparent as the duration of post-operative healing increased (breaking strength of 11,234N for nylon repair versus 7425N for carbon fiber repair at 24 weeks post repair). Based on these findings, the authors concluded that carbon fiber suture was not a suitable suture material to use for repair of equine flexor tendon lacerations.¹¹⁸ Other reports have found that carbon fiber fragments over time and fibers can migrate to regional lymph nodes.⁸⁴ Overall, the use of carbon fiber for repair of equine digital flexor lacerations has fallen out of favor.

2. *Suture Size*

It has been found that the strength of a suture is related to its cross-sectional area.¹¹⁹ In a human cadaver study by Taras and colleagues, use of 3-0 polybutylate coated braided polyethylene terephthalate (Ethicon, Somerville, NJ) resulted in a repair that was 52% stronger than a repair using 4-0 suture of the same composition.¹²⁰ Similar findings were made by Osei and colleagues in which there was a 49% increase in polyfilament caprolactam (Supramid; S. Jackson Inc, Alexandria, VA) suture strength when caliber was increased from 4-0 to 3-0.¹¹⁹ However, increasing suture caliber may have a negative impact on tendon function.¹²¹ In an *ex vivo* study by Momose et al. using a

canine model, it was found that increased suture size from 5-0 to 4-0 increased the gliding resistance of the repair.¹²¹ In the same study by Osei et al., an eight-strand repair using 4-0 polyfilament caprolactam produced a greater load to failure when compared to a four-strand repair using 3-0 suture material.¹¹⁹ Therefore, using a multi-strand repair may be more beneficial than increasing suture caliber to improve the strength of a repair.¹¹⁹ However, a multi-strand repair could also have a negative impact on tendon function by increasing gliding resistance through increased bulk at the repair site.

In a separate study by Alavanja et al., the impact of suture size on the strength of a flexor tendon repair was investigated in a human cadaveric study. Transected flexor profundus digitorum tendons were repaired with a four-strand locking cruciate pattern using 2-0, 3-0 and 4-0 polybutylate coated braided polyethylene terephthalate. Tendons were maintained within an intact hand during testing. Repairs were tested with cyclic loading over a total of 1000 cycles in a tensometer that mimicked digital motion. It was found that repairs made with 2-0 suture were significantly stronger than repairs made with 4-0 suture (80N± 12N mean maximum tensile load for 2-0 repair, 66N± 12N mean maximum tensile load for 4-0 repair). No significant difference in strength was found between 2-0 and 3-0 suture (74N±18N mean maximum tensile load for 3-0 repair), and between 3-0 and 4-0 suture.¹²² Gap formation at the site of repair was not significant for all three repairs. This indicates other factors, such as repair configuration, may have a more significant impact on gap formation than suture caliber.¹²² The 2-0 repair created greater work of flexion than the 3-0 and 4-0 repairs by increasing the gliding resistance of the tendon, resulting in more force needed for tendon excursion.¹²² There was no significant difference in the work of flexion between the 3-0 and 4-0 repairs,¹²² All three repairs had a mean maximum tensile strength that was greater than the force exerted on flexor tendons of the human hand during active flexion.

Although the 2-0 repair was stronger than the 4-0 repair, the fact that the suture increased the work of flexion may not make it an ideal suture caliber to use clinically in human flexor tendon lacerations.

For equine flexor tendon tenorrhaphy, it has been previously reported that the largest monofilament suture that can be buried should be used.¹¹⁶ Similar recommendations have been made previously for flexor tendon laceration repairs in small animals.¹²³ It may be difficult for a surgeon to obtain and stock their facility with a monofilament absorbable and non-absorbable suture with sizes larger than #2 suture caliber.¹⁰ This has prompted surgeons to use multi-stranded suture repairs as opposed to suture repairs that may rely on increased size of suture caliber to contribute to the strength of repair.¹⁰ Currently, the most common suture calibers used for equine flexor tendon tenorrhaphy include #1 and #2.

3. Number of Suture Strands

Previous human studies evaluating multi-strand tenorrhaphies have found that the number of suture strands crossing a tendon laceration site is directly proportional to the strength of repair; increased suture strands increases the strength of repair.^{96, 124-127} One such multi-strand technique is the Savage suture pattern, first reported by Savage in 1985 (Figure 22).^{124, 128} When tested by linear load to failure using 3-0 polybutylate coated braided polyethylene terephthalate in porcine extensor digiti proprius tendons, the six-strand Savage suture pattern had enhanced gap resistance and ultimate tensile strength when compared to other commonly used repair techniques with fewer suture strands, including the Bunnel, Kessler, Kleinert, modified Kessler (Figure 23) and Becker repairs.¹²⁴

In a study by Barrie et al. in 2000, four-strand suture repairs were compared to two-strand

and six-strand repairs to determine if four-strand repairs had improved work of flexion over six-strand repairs. Human cadaver profundus tendons were repaired with three different four-strand repairs, including the four-strand Kessler, the cruciate (Figure 24) and the locking cruciate, as well as a two-strand Kessler and a six-strand Savage. All repairs were performed using 3-0 polybutylate coated braided polyethylene terephthalate. Work of flexion of each repair was determined by assessing tendon excursion within a 20mm testing range. In addition, ultimate tensile strength and force at 2mm gap formation were measured. It was found that the six-strand repair did have improved ultimate tensile strength compared to the two and four-strand repairs; however, there was no significant difference in resistance to 2mm gap formation between the four-strand and six-strand repairs.¹²⁹ The author could not correlate a relationship between an increase in the number of strands for a multi-strand repair and work of flexion; the author attributed this to the fact that it was an *ex vivo* and not an *in vivo* study.¹²⁹ As the number of suture strands increases in a repair, the complexity of the repair increases. In addition, with an increased amount of suture material at the site of repair, the work of flexion could potentially increase due to increased resistance to gliding from increased bulk at the repair site.¹²⁹ A more complex repair may require more surgical manipulation, potentially increasing edema in tissues post-operatively, increasing the risk of further tendon damage, and compromising tendon nutrition by compromising intratendinous blood supply.^{6, 129-131} Further tendon damage and edema in tissues post-operatively could subsequently increase work of flexion post-operatively. In addition, it is important that equal tension is placed on all strands of the repair. If the strands of the repair do not have equal tension, it may result in uneven loading with strain on the repair, potentially weakening the repair.²⁰

As an alternative to increasing the number of suture strands at a repair site, making core sutures double-stranded as opposed to single-stranded has been proposed. The benefit of double-

stranded core sutures is that they require fewer needle passes within the tendon tissue, reducing the amount of tissue handling and possible further tendon damage during surgery. The most common technique in which double stranded sutures are used is the Tsuge (Figure 25). A four-stranded Tsuge repair was compared to five other four-strand techniques in a study by Angeles and colleagues using human cadaver profundus tendons *ex vivo*. It was found that the repair had improved gliding, ease of placement of the suture strands and sufficient ultimate tensile strength to withstand early post-operative mobilization.¹³⁰

4. Placement of Suture Knots

Previous studies investigating biomechanical properties of flexor tendon repairs have found repairs often fail at the knot and suture material is weakest at its knot.^{128, 132} The strength of a repair is influenced by the location and number of knots in a repair, as determined in *ex vivo* studies.^{128, 133} In a study by Aoki et al. on cadaver canine flexor digitorum profundus tendons, placement of knots as far away from the repair as possible, as well as decreasing the number of knots of the repair increased the ultimate tensile strength of the repair.¹³³ Pruitt et al. further investigated the effect of knot placement on the interior versus the exterior of a tendon using an *in vivo* canine model.¹³⁴ Using a four-strand Savage technique, knots were placed between the transected ends of tendon, or as far away from the transected ends of the tendon as possible on the exterior of the tendon. Tensile testing performed immediately after repair revealed that the repair in which the knots were located on the exterior of the tendon away from the tendon ends was stronger than the repair in which knots were located at the transected ends of tendon.¹³⁴ It was speculated that this was due to the knots bearing the entire tensile load at the repair site, which is the weakest area of the construct immediately following repair.¹³⁴ At one week post-repair, both techniques showed a decrease in the tensile strength by 85%.¹³⁴ This was attributed

to weakening of the tendon ends associated with the inflammatory stage of repair. At three weeks post-repair, there was an increase in tensile strength of both repairs of 120%, while at six weeks post-repair, the group with knots placed between the tendon ends had increased tensile strength to 167% of the initial value, while the repair in which the knot was placed on the outside of the tendon did not further increase in strength.¹³⁴ These findings demonstrate that over time, repairs in which knots are located near the transected tendon ends have tensile strengths comparable to those in which the knots are located away from the repair site. Therefore, the presence of knots at the repair site does not inhibit tendon healing. This is different than what was observed in Aoki's previous study in which knots located near the repair site weakened the repair, highlighting the importance of *in vivo* testing.^{133, 134}

Placement of the knots on the surface of a tendon has been found to increase the gliding resistance significantly by inhibiting smooth tendon excursion through increased friction.^{121, 135} Increased gliding resistance impairs the function of the tendon, potentially resulting in peritendinous adhesion formation during healing.¹²¹ In an *ex vivo* study by Momose et al. using a canine model, it was found that when knots were located on the volar aspect of a tendon, or when there was more than one knot present, the gliding resistance of the repair increased.¹²¹ When knots were placed laterally along a tendon or when the number of knots was reduced, gliding resistance decreased.¹²¹ Therefore, the location and number of knots present in a repair is an important consideration for future function of the tendon.

As an alternative to using suture material in which a knot is required to secure the repair, the use of barbed suture has been investigated. Currently, there are two types of barbed suture; unidirectional and bidirectional.¹³⁶ Unidirectional barbed suture (V-Loc™; Covidien, Mansfield, MA) is composed of glycolic acid and trimethylene carbonate with unidirectional barbs that are

arranged circumferentially around the thread. A curved needle is located on one end of the suture, while a small loop is located at the opposite end of the suture for locking the suture with initial placement.¹³⁶ With this configuration, the barbs prevent pull-out in one direction, while the locking loop prevents pull-out in the opposite direction.¹³⁶ Bidirectional barbed suture is composed of monofilament polypropylene (Quill™; Angiotech Pharmaceuticals, Inc., Vancouver, BC). The suture is arranged with curved needles on each end and for one half of the suture, barbs point one direction, while for the other half of the suture, barbs point the opposite direction. This allows the suture to resist pull-out in opposite directions.¹³⁶

Zeplin investigated the use of a unidirectional barbed suture for flexor tendon laceration repair *ex vivo*.¹³⁷ V-Loc was tested in harvested human flexor digitorum tendons. Size 3-0 V-Loc was compared to 3-0 polydioxanone in transected tendons using both a modified two-strand and four-strand Kirchmayr-Kessler suture technique. It was found that the two-strand technique using the barbed suture device was significantly lower in tensile strength when compared to the two-strand technique using 3-0 PDS.¹³⁷ There was no significant difference in tensile strength between the four-strand barbed suture device and the four-strand repair using 3-0 polydioxanone.¹³⁷ Based on these findings, the author concluded that the device may have potential in use for flexor tendon laceration repair; however, it did not provide any advantage in tensile strength over a four-strand polydioxanone suture repair.¹³⁷ Further testing *in vivo* may be warranted.

In a separate study, Lin et al. investigated the use of V-Loc suture of substantially larger diameter for repair of flexor digitorum profundus tendons *ex vivo*.¹³⁶ A four-strand Kirchmayr-Kessler suture technique was performed comparing 3-0 polybutylate coated braided polyethylene terephthalate to the V-Loc wound closure device with a diameter comparable to 0 sized suture. Repairs were tested by linear load to failure using a materials testing machine. It was found that

the load to failure for the knotless repair using V-Loc was higher than that of the repair using 3-0 polybutylate coated braided polyethylene terephthalate ($52.3\text{N} \pm 12.5\text{N}$ for V-Loc repair, $42.3\text{N} \pm 12.7\text{N}$ for 3-0 polybutylate coated braided polyethylene terephthalate repair).¹³⁶ There was no significant difference in load to 2mm gap formation between the two repair patterns ($10.9\text{N} \pm 3.8\text{N}$ for V-Loc repair, $11.7\text{N} \pm 4.5\text{N}$ for polybutylate coated braided polyethylene terephthalate repair). Based on these results, large diameter knotless suture produces a repair that is stronger to failure and equivalent in strength to resisting gap formation to that of a traditional repair pattern.¹³⁶ Despite creating a strong repair, the larger diameter unidirectional barbed suture was more difficult to handle than 3-0 polybutylate coated braided polyethylene terephthalate and the initial locking loop of the suture was difficult to keep in low profile on the exterior of the tendon. It is not known if the exposed locking loop, as well as exposed barbs of the suture would increase the work of flexion at the repair site post-repair. Based on the repair's strength characteristics, further investigation of the use of barbed suture to repair flexor tendon lacerations *in vivo* may be warranted.¹³⁶ At this time, bidirectional barbed suture has not been tested for repair of flexor tendon lacerations.

In order to facilitate early post-operative mobilization in human flexor tendon laceration repair, several studies have investigated increasing repair strength by increasing the number of suture strands bridging the repair site and adding circumferential sutures. Use of additional suture strands and circumferential sutures can be labor intensive, technically demanding, and may result in extensive dissection of the surgery site in order to achieve repair.¹⁷ In addition, multi-strand techniques that require locking components result in focal areas of tenocyte death due to the fact that these repairs distribute strain in only focal areas of tendon.^{17, 138} With additional suture present at the repair site, it may impair tendon gliding, resulting in increased work of flexion and possible

peritendinous adhesion formation.¹⁷ Due to these disadvantages, several variations of knotless intratendinous implants have been developed for tendon laceration repair. Specific aims of producing intratendinous devices were to minimize dissection of the paratenon surrounding tendon in order to achieve repair, minimize exposed suture outside the tendon, and hopefully create a repair that was as strong as or stronger than current repairs.^{17, 139} Specific types of intratendinous devices that have been investigated include wire coils, barbed devices and stainless steel anchor systems.

One of the first reported intratendinous devices was developed by Gordon and colleagues.¹⁴⁰ The device consisted of a low-profile stainless steel anchor that was 20mm in length, 3mm in width and 1mm in thickness (Figure 26). The anchor was secured in place by introducing it through the cut ends of the tendon and by placing stainless steel sutures perpendicular to the long axis of the anchor. With tension, the suture secured tendon substance around the triangular projections of the anchor. The device was tested by linear load to failure in harvested human cadaver flexor digitorum profundus tendons and was compared to three other repairs: the Kessler repair using 4-0 monofilament nylon, the Becker repair using 6-0 monofilament polypropylene suture and the Savage repair using 4-0 monofilament nylon. Reasoning behind the use of each particular size and type of suture for each sutured repair was not specified; it can only be assumed that these materials were used as a clinical standard. The maximal linear tensile load (74N) and the ultimate tensile strength (75N) of the stainless steel anchor were significantly greater than all three sutured repairs.¹⁴⁰ One drawback to the use of this implant in human flexor tendon repair was the lack of flexibility of the implant, not lending to bending during tendon excursion across a joint. Further research of the use of an intratendinous anchor was warranted based on its higher level of strength compared to other sutured repairs. The use of an absorbable material, such as poly-L-

lactic acid, was proposed to improve motion of tendon over joints.

Erol et al. developed a stainless steel wire in two different spiral configurations to act as an intratendinous repair device. Stainless steel was chosen as a repair material, due to its low level of elasticity. Model 1 was composed of 0.8mm stainless steel wire formed into a five-turn coil with a portion of the wire passing through the middle of the coil.¹³⁹ Model 2 was composed of a seven-turn stainless steel coil made out of 0.8mm stainless steel wire, with a separate 1.5mm Kirschner T-shaped wire within the proximal portion of the device (Figure 27).¹³⁹ The diameter and distance between spirals for both models was not specified. A coil configuration was chosen for the implants as it was believed this configuration increased the contact surface area between the tendon and device. This study compared both models to the LL and Bunnell suture repairs using polybutylate coated braided polyethylene terephthalate in lacerated cadaveric sheep Achilles tendons. When tested by linear load to failure, it was found that Model 2 withstood a higher tensile load to ultimate failure (274N) than the LL technique (216N) and the Bunnell technique (198N). Model 1, which withstood a tensile load of 222N, was prone to implant failure at the point of wire bending in the middle of the coil. It was felt that the T configuration of the middle wire in the Model 2 distributed tensile load more evenly through the implant. In addition, there was not a weak point in the second implant created from wire bending.

Hirpara et al. developed a barbed implant composed of nitinol shape memory alloy for tendon laceration repair (Figure 28).¹⁷ Nitinol is composed of approximately 56.2% nickel and 43.2% titanium.¹⁷ This product was chosen as it is a remarkably flexible, and biocompatible.^{17, 141} A barbed configuration was chosen as barbs have been shown *in vivo* to gain purchase in other tissues, including bone and menisci.¹⁷ Hirpara compared the strength of the barbed nitinol device in porcine profundus flexor tendons to three different sutured repairs including the six-strand

Savage, the Pennington modified Kessler suture (PMK) and a four-strand cruciate repair using 6-0 monofilament nylon. Each repair was tested with and without a supplemental circumferential suture at the repair site. For the nitinol barbed device, repairs using one or two devices were tested. When one device was used for repair, the force needed to produce a 3mm gap was not statistically different than the two-strand PMK. When two devices were used for repair, the repair was significantly more resistant to gapping than the PMK. When a circumferential suture was added to the two-device repair, there was no statistical difference in the amount of force required to create a 3mm gap between this repair, and the Savage technique. With the circumferential suture, it was significantly more resistant to gapping than the PMK and cruciate repairs. Although the barbed device was as strong as a two-strand repair technique, placement of a circumferential suture at the repair site was required to make it as strong as the six-strand Savage technique.

The Teno Fix® (TF) Tendon Repair System (Ortheon Medical, Columbus, OH) is a commercially available product used for repair of flexor tendon lacerations in human orthopedic surgery. The device was designed to take advantage of the strength characteristics of stainless steel, while having knotless tendon anchoring.²¹ It consists of two intratendinous stainless steel anchors connected with a single multifilament 2-0 stainless steel suture (Figure 29).²¹ The anchors are 4mm in length with an outer diameter of 2.2 mm. Anchors consist of an inner core surrounded by a coil and are introduced into a tendon through an obturator after a stab incision has been made into the epitenon (Figure 30).²¹ The stainless steel suture is fed through the anchors and the tendon ends are opposed (Figure 31). A stainless steel crimp bead is placed on the end of the suture to lock the implant in place (Figure 32). Incisions made in the epitenon to place the anchors are sutured closed and a circumferential suture is placed around the laceration.²¹

Su and colleagues first examined the device in 2005 *ex vivo* in 60 harvested cadaveric

human flexor digitorum profundus tendons.²¹ Tendons harvested were randomized into the following repair groups: cruciate pattern with 3-0 polybutylate coated braided polyethylene terephthalate, cruciate pattern with 4-0 polybutylate coated braided polyethylene terephthalate, and the Teno Fix® repair. Repairs were tested as core repairs alone, as well as with a running locked circumferential epitendinous suture using 5-0 monofilament polypropylene.²¹ When tested by linear load to failure, it was found that the force to a 2mm gap for the TF repair significantly greater than that of the 4-0 cruciate repair (38.9N for the TF repair, 28.59N for the 4-0 cruciate repair, $p \leq 0.05$).²¹ There was no significant difference in force at a 2mm gap between the TF repair and the 3-0 cruciate repair (32.03N for the 3-0 cruciate repair).²¹ There was also no significant difference in the peak force between the TF repair and both the cruciate repairs.²¹ Addition of a circumferential suture increased the strength of all the repairs. Specifically, it increased force to a 2mm gap by 40% and peak force by 48% for the TF repair.²¹ Due to the importance of gap formation clinically for human digital flexor tendon lacerations, the author concluded that the TF implant had promise because of its comparable and superior strength to 2mm gap formation when compared to a 3-0 and 4-0 cruciate repair.²¹

Following *ex vivo* testing, Su et al. examined the device *in vivo* using a canine model.²² Common flexor digitorum superficialis tendons were repaired with the TF device and a continuous circumferential suture using 5-0 monofilament polypropylene in 16 dogs.²² All dogs were maintained in external coaptation for four weeks following surgery using a forelimb spica cast that extended from the shoulder to the elbow for four weeks. With this configuration, the cast allowed active flexion and extension of the carpus and interphalangeal joints, but did not allow the dogs to fully weight-bear on the limb.²² In addition, this configuration allowed for passive flexion and extension exercises. In nine of the 16 dogs, the dogs removed the spica casts prior to the completion

of the four-week period. Of these dogs, only three of the repairs healed primarily without gapping; the remaining repairs ruptured with full weight-bearing on the limb.²² Of 16 dogs, seven dogs healed without gap formation or rupture at the site of repair; three of which removed the spica cast prematurely and four of which the spica cast remained in place for four weeks.²² The remaining three dogs had failure of the repair, either through repair rupture or gap formation of greater than 2mm.²² Of the successful repairs examined, only one of the dogs evaluated developed peritendinous adhesions. The remaining repairs healed with a smooth epitendon or with few adhesions present. Histologic examination at 12 weeks post-repair revealed that the TF devices were surrounded by well vascularized connective tissue with no discernable inflammatory response present.²² The devices did not cause necrosis of surrounding tendon substance and there were small areas of fibroblastic proliferation in the region of the repair.²² Fibrous tissue at the repair site was arranged longitudinally.²² Therefore, the TF device did not appear to impair tendon healing.

In a third study by Su et al., a multicenter, randomized, blinded clinical trial was performed in which the TF device was tested in clinical cases of human flexor tendon lacerations.¹⁴² Sixty-seven patients with 85 injured digits consisting of flexor tendon injuries were randomized and treated with either the TF implant or a control repair comprised of a four-strand cruciate suture repair using either 3-0 or 4-0 monofilament polypropylene.¹⁴² A running circumferential suture using 6-0 monofilament nylon was also used in each repair. Patients were eligible for the study if there was a laceration of the flexor digitorum profundus tendon, with or without concurrent injury of the flexor digitorum superficialis muscle. Initially, 41 digits were randomized into the TF group and 44 digits in the control group; however, due to limited surgical exposure at the time of surgery, seven digits were moved to the control group. Post-operatively, all patients began a controlled

rehabilitation protocol consisting of passive flexion and active extension starting on the first day following surgery. Mobility of the proximal and distal interphalangeal joints was evaluated at 12 and 24 weeks post-operatively, as well as grip and pinch strength of the affected hand. Within six weeks of repair, 18% (9/51) of tendons that were repaired with the control ruptured.¹⁴² Causes for repair rupture included a lack of patient compliance, persistent wound infection and rupture during active flexion in patients that were compliant with post-operative rehabilitation.¹⁴² The mode of repair failure in these cases was either suture pull-out or suture rupture.¹⁴² No ruptures of the repair site were observed with the TF repair.¹⁴² Grip and pinch strength steady increased over time for both repairs, and at 24 weeks post-operatively, was approximately 80% of that of uninjured fingers.¹⁴² Other factors used to determine success of treatment included the Strickland mobility scale as developed by Strickland in 1985¹⁴³ and the Disabilities of the Arm, Shoulder and Hand (DASH) questionnaire.¹⁴⁴ The Strickland mobility scale is calculated as follows:^{142, 143}

$$\frac{((\text{proximal interphalangeal} + \text{distal interphalangeal flexion}) - (\text{proximal interphalangeal} + \text{distal interphalangeal extension deficit}))/175^\circ}{\text{}} \times 100$$

Based on the calculation, repairs were classified as excellent (75-100%), good (50-74%), fair (25-49%) or poor (<25%). At 24 weeks post-operatively, 51 patients were available for post-operative follow-up.¹⁴² For the TF repair, 29% (7/24) had an excellent outcome, 38% (9/24) had a good outcome, 21% (5/24) had a fair outcome and 13% (3/24) had a poor outcome.¹⁴² For the control repair, 30% (8/27) had an excellent outcome, 41% (11/27) had a good outcome, 19% (5/27) had a fair outcome and 11% (3/27) had a poor outcome.¹⁴² There was no significant difference in patients with a good or excellent outcome between the two groups.¹⁴²

The DASH questionnaire consists of 30 questions on daily activities, as well as pain. Graded out of 100, as the DASH score increases, the greater the disability of the patient. At six

weeks post-operatively, the DASH score for the TF repairs was 27.2 ± 22.7 . At 12 weeks post-operatively, the DASH score for the TF repairs was 7.4 ± 8.6 and at 24 weeks was 2.5 ± 3.8 .¹⁴² For the control group, the DASH score at six weeks was 22.7 ± 19.8 , at 12 weeks it was 8.1 ± 13.8 and at 24 weeks it was 2.0 ± 4.2 .¹⁴² There was no significant difference in DASH scores between both groups at six, 12 and 24 weeks post-operatively.¹⁴²

Based on the results obtained from the study, the author concluded that the TF device showed promise for repair of flexor tendon lacerations as the repair had a lower rupture rate and had similar results to a conventional cruciate repair. One downfall of the device was that it could not be placed in all circumstances due to limited surgical exposure.¹⁴²

Lewis et al. examined the mechanical characteristics of three TF devices versus a modified Kessler suture repair (Figure 23) using #2 polybutylate coated braided polyethylene terephthalate in cadaveric human Achilles tendons.²⁰ The modified Kessler repair is similar to a LL suture pattern, with knots of the repair tied at the tendon ends. When tested by linear load to failure, it was found that there was no significant difference in the peak tensile stress between the two repairs ($1.03 \text{N/mm}^2 \pm 0.51 \text{N/mm}^2$ for the modified Kessler repair, $1.19 \text{N/mm}^2 \pm 0.12 \text{N/mm}^2$ for the TF repair).²⁰ Peak stress was calculated using the following formula:

$$\text{Peak stress (N/mm}^2\text{)} = \text{Final cross-sectional area} / \text{Peak force at repair separation}$$

The gap formation stress for the TF was repair $0.8 \text{N/mm}^2 \pm 0.46 \text{N/mm}^2$, while the gap formation stress for the modified Kessler repair was $0.3 \text{N/mm}^2 \pm 0.15 \text{N/mm}^2$.²⁰ The gap formation stress was calculated using the following formula:

$$\text{Gap formation stress (N/mm}^2\text{)} = \text{Cross-sectional area closest to gap} / \text{Force at gap formation}$$

The gap formation stress was higher for the TF repair than the modified Kessler repair,

most likely due to the fact that the #2 polybutylate coated braided polyethylene terephthalate was more elastic (less stiff) than the stainless steel suture of the TF system.²⁰ Having a higher gap formation stress correlates with greater force needed to create a gap at the repair site.²⁰ This is beneficial as a reduction in the propensity of gap formation will result in a more uniform repair.²⁰ There was a higher standard deviation for the TF repair than the modified Kessler repair, possibly due to uneven loading of the TF devices at the repair site.²⁰ Uneven loading could occur because there is no way of knowing how much tension is placed on the TF device when it is initially placed. As a result, tension between each device is variable. This is one downfall of using multiple devices for one repair as uneven loading could potentially weaken the repair.²⁰ One limitation of this study is that it does not report the cross-sectional area used to calculate the peak stress and the gap formation stress. Therefore, peak force cannot be determined for both repairs.

Kubat et al. reported the use of a TF device in a clinical case of a patient that had sustained multiple injuries to the left hand due to an accident with a table saw.¹⁴⁵ Prior to this injury, the patient had sustained a significant neurological injury, resulting in diffuse muscle spasms.¹⁴⁵ At initial presentation for the injury to the hand, tendon lacerations of the index finger were repaired with a four-strand core suture technique using 3-0 polybutylate coated braided polyethylene terephthalate, with a continuous circumferential suture using 6-0 monofilament polypropylene.¹⁴⁵ Rehabilitation consisted of an early active range of motion program.¹⁴⁵ Post-operatively, uncontrollable muscle spasms of the left hand resulted in rupture of the superficialis and profundus tendons of the left index finger causing failed repair of the tendons twice. Following the second rupture, the profundus tendon was repaired with one TF device.¹⁴⁵ The repair healed well, and at two years post-repair, the patient had normal range of motion of the left index finger.¹⁴⁵

5. Core Purchase Length of Suture

Core purchase length is the exit and, or, entry point of a core suture from the cut ends of a tendon. Previous studies using porcine tendon have found that increasing the core purchase length for both locking and grasping suture patterns increased the force required for 2mm gap formation and ultimate failure, as well increased overall repair stiffness.^{25, 146} It is speculated that increasing the core purchase length of suture increases the strength of repair by increasing tendon-suture interaction. This is critical in a clinical setting, as cut ends of a tendon have been shown to soften within the first week following a laceration due to collagen degradation from proteolytic enzyme release as part of the inflammatory stage of tendon repair.²⁵ This would most likely result in decreased holding power of the repair if it was closer to the tendon ends due to weakened tendon tissue. In these studies, a threshold point of 0.7mm to 1.0cm from the cut ends of the tendon was identified. Beyond this point, the repair did not become stronger.^{25, 146} Core purchase length has not been investigated in the same manner for equine tendon laceration repair. Therefore, a threshold point for core purchase length for equine tenorrhaphy is not known. Further investigation of core purchase length as a variable for equine tenorrhaphy may be beneficial in increasing repair strength for current repair patterns.

6. Grasping vs. Locking Attributes of Suture Patterns

Grasping and locking attributes of a repair refer to the configuration of the suture and how the repair gains purchase in a tendon. A suture pattern is considered a locking pattern if the suture loop formed by the repair tightens around tendon fiber bundles when tension is applied to suture. A suture pattern is considered a grasping pattern if the suture loop does not tighten around tendon fibers and as a result pulls through the tendon fibers when tension is applied to the suture (Figure 33). These configurations were first described by Pennigton in 1979.^{86, 147} The locking pattern has

been further described as circle-locks, cross-locks or Pennington locks.¹⁴⁸ Circle-locks are either looped or double circled. An example of a suture pattern that employs the use of circle-locks is the locking cruciate repair. Cross-locks are either embedded or exposed. An example of a suture pattern that consists of cross-locks is the Savage suture pattern. An example of a repair that uses a Pennington lock configuration is the Kessler repair.¹⁴⁸ It has been found that there is no significant difference in strength between a cross-lock or circle-lock configuration.¹⁴⁸ However, it has been found previously that a repair in which a Pennington lock configuration is used is weaker than use of a cross-lock or circle-lock.¹⁴⁸

The efficacy of a grasping or locking configuration is dependent on the size of suture material used, the distance from which the grasping or locking loop is away from the laceration site, as well as the amount of tendon substance within the grasping or locking loop.¹⁴⁸ Suture material should be larger than 4-0, especially with locking patterns, as suture in a locking pattern has a tendency to fail at the locking loop.⁸⁶ This is due to the fact that the kink created by the crossing of the transverse and longitudinal throws of the suture creates a point of weakness in the suture due to shear stress. With a smaller caliber suture, the likelihood of suture failure at this point is much higher. Locking and grasping loops should be at least 7mm to 10mm away from the tendon ends, as tendon tissue in the region of the wound is weaker, especially within the first week following the initial injury due to inflammation of the injured tissue. The size of the grasp or locking loop should be at least 15% of the cross-sectional area of the tendon.¹⁴⁸

The mechanical properties of grasping versus locking patterns have been investigated in great detail. Hotokezaka et al. compared five different suture configurations using 3-0 Ethibond in cadaver human flexor digitorum profundus tendons; two-strand suture without grasping or locking loops, two-strand suture with one locking loop per suture, two-strand suture with two grasping

loops per suture, two-strand suture with one locking loop per suture and two-strand suture with two locking loops per suture (Figure 34).⁸⁶ When tested by linear load to failure, it was found that the mode of failure for the grasping configurations was suture pull-out and for the locking loop configurations it was suture rupture at the level of the locking loop.⁸⁶ The suture configuration in which a grasping or locking loop were not used was the weakest configuration, while the one locking loop, two locking loop and two grasping loops were the strongest configurations. There was no significant difference in the tensile strength of the one locking loop, two locking loop or two grasping loop configurations.⁸⁶ The stiffness of both locking loop patterns was greater than that of the grasping configurations.⁸⁶ Based on these findings, the author recommended that a single locking loop configuration would be most ideal for repair as there was no additional benefit observed in tensile strength with additional locking loops. In addition, the locking loop configuration had greater stiffness than the grasping configuration, potentially making it less prone to gap formation at a repair site.⁸⁶

Hatanaka et al. investigated the effect of the size of locking loops on the strength of a tendon laceration repair. Using human cadaver flexor digitorum profundus tendons, two locking loops were applied in one side of a tendon using 3-0 polybutylate coated braided polyethylene terephthalate in a Pennington repair pattern. The cross-sectional area of the locking loops was increased from 10-50% of the total cross-sectional area of the tendon. It was found that with an increase in the cross-sectional area of tendon fibers included in the locking-loop, the greater the strength of the repair.¹⁴⁹ When the locking-loops were overlapped with no further increase in cross-sectional area, the configuration had no significant impact on the tensile strength of the repair, but it did increase the propensity for gap formation.¹⁴⁹ Therefore, the author concluded that increasing the cross-sectional area of a locking-loop increases the tensile strength of a repair, but overlapping

may result in gap formation.

Dona et al. investigated the effect of cross-sectional area of locking loops on a four-strand cruciate repair. Using deep flexor tendons harvested from sheep, a four-strand cruciate repair was placed, increasing the cross-sectional area of the locking-loops from 10% to 25%, 33% and 50% of the total width of the tendon (Figure 35). When tested by linear load to failure, it was found that locking-loops incorporating 25% of the width of the tendon were the greatest ultimate tensile strength, force to 2mm gap formation and stiffness.¹⁵⁰ Past 25%, the tensile strength and stiffness of the repair began to decrease, and the repair was more prone to 2mm gap formation. Therefore, the author concluded that a locking-loop incorporating at least 25% of the width of a tendon should be used to optimize the mechanical properties of a four-strand cruciate repair.¹⁵⁰

Xie et al. evaluated the area within a locking loop configuration using an *ex vivo* porcine model for 4-0 monofilament nylon two-strand and four-strand suture techniques.¹⁵¹ Three different diameters of locking-loops were tested; 1mm, 2mm and 3mm. When tested by linear load to failure, repairs using 2mm and 3mm locking-loops had greater force at failure and greater force to gap formation than the 1mm locking-loops.¹⁵¹ There was no significant difference found between the 2mm and 3mm locking-loop repairs.¹⁵¹

Not only does the diameter of the locking-loops influence the strength of repair, but the orientation of locking loops does as well. Tan et al. investigated differences in the mechanical properties of locking-loops placed perpendicular to the longitudinal orientation of the tendon versus placement parallel to the tendon using an *ex vivo* porcine model (Figure 36). It was found that when placed perpendicular to the longitudinal axis of the tendon, the repair was stronger than if the locking-loops were placed parallel to the longitudinal axis of the tendon.¹⁵² When placed

parallel to the longitudinal axis of the tendon, strength of the locking loop is dependent on collagen cross-links. When placed perpendicular to the longitudinal axis of the tendon, strength of the locking loop is dependent on compression of the collagen fibrils, which results in a stronger repair.¹⁵²

When the appropriate caliber of suture material is used, the locking and grasping loops are placed a sufficient distance away from the tendon ends and the size of the grasp or locking loop is at least 15% of the cross-sectional area of the tendon, grasping and locking configurations are a reliable way of engaging a suture material in tendon, however, the locking loop is a somewhat stronger configuration than the grasping configuration. It should be noted that the differences in strength are negligible.¹⁴⁸ Other factors which may contribute to the strength of the repair, such as the caliber of the suture material, or the number of strands crossing the repair site may have a more significant impact on the overall strength of the repair.¹⁴⁸ At this time, the impact of grasping versus locking attributes of suture have not been tested for equine flexor tendon laceration repair.

7. Preservation of Intra-Tendinous Blood Supply

Crowson, et al. evaluated the effect of the 3LP versus the LL on intratendinous vasculature of the SDFT in anesthetized horses.²³ SDFT were transected and repaired with either a 3LP, or LL using #2 PDS. Circumferential ligatures were also placed around a single tendon to evaluate their effect on intratendinous vasculature. After placement of sutures, horses were administered heparin intravenously and euthanized. Flexor tendons were removed and examined histologically in the region of each repair. It was found that both patterns resulted in decreased intratendinous vessel density when compared to normal tendon, but that the 3LP had significantly more vessels in cross section when compared to the LL. The author concluded that both tenorrhaphy techniques reduced

intratendinous blood flow, however, a suture pattern that consisted of locking components, such as the LL had a more profound effect on intratendinous vasculature than grasping patterns, such as the 3LP.²³ This is most likely due to compression of intratendinous vessels within the locking loops.²³

Freeman et al. evaluated the effect of the 3LP versus the SSS on intratendinous vasculature of the SDFT in anesthetized horses.²⁴ Under general anesthesia, forelimb SDFT were transected and repaired with either a SSS or 3LP using #2 PDS. Following repair, horses were heparinized, euthanized, and forelimbs were perfused with barium sulfate solution then fixed with formalin. The tendons were transected every 5mm and microangiographic images were obtained using a Faxitron X-ray cabinet with computed radiography imaging. Microvascular analysis of sections proximal to the tenorrhaphy, throughout the tenorrhaphy and distal to the tenorrhaphy was completed. A significant reduction in the number of perfused intratendinous vessels was observed in the SSS repair compared to the 3LP.²⁴ The author concluded that this may limit clinical use of the SSS.²⁴

Core and Peripheral Sutures

Core sutures are sutures placed within a tendon. Configuration of core sutures varies greatly and many different materials and suture caliber sizes have been tested to optimize flexor tendon laceration repair in humans. Two main factors that have a strong impact on the strength of core sutures are core purchase length, as well as placement of secure locking and grasping sutures.¹⁴⁸

Peripheral sutures are sutures that are placed around the periphery of the tendon ends at the site of a tendon laceration in addition to placement of core sutures. Peripheral suture patterns may be interrupted, running or locking.¹⁴⁸ Peripheral sutures were initially used to approximate tendon

ends and improve the gliding surface of the tendon post-operatively.¹⁵³ It has been found however, that peripheral sutures can contribute to increasing the strength of a tendon laceration repair.¹⁴⁸ Wade et al. compared the strength characteristics of simple continuous, interrupted and vertical mattress peripheral sutures in a human cadaver model and found that the vertical mattress repair increased load to 2mm gap formation by 93%.¹⁵⁴

The peripheral suture pattern that is used most often is the simple continuous peripheral pattern. This method is chosen as it is relatively easy to perform. The strength of a simple continuous peripheral pattern can be increased by taking deeper bites into the tendon.¹⁵⁵ Using cadaver flexor digitorum superficialis tendons, Diao et al. investigated the effect of penetrating depth of peripheral suture repairs on the overall strength of repair. Tendons were repaired with a core Kessler suture technique, and two different peripheral repairs; placement of superficial peripheral sutures and placement of deep peripheral sutures that were half the depth to the center of the tendon. When tested by linear load to failure, it was found that the repairs containing deep peripheral sutures were almost 80% stronger than repairs with superficial peripheral sutures (38.96N ultimate load to failure for the deep peripheral suture repair, 21.68N ultimate load to failure for the superficial peripheral suture repair). In addition, repairs in which deep peripheral sutures were used were 90% stiffer than repairs using superficial peripheral sutures.¹⁵⁵

A second factor that increases the strength of a peripheral suture repair is increasing the purchase length of peripheral sutures. Using a *ex vivo* flexor digitorum profundus model, Merrell et al. found that when the purchase length of peripheral sutures was increased from 1mm to 2mm, the ultimate tensile strength of the repair significantly increased (50.8N for the 2mm peripheral suture repair, 37.1N for the 1mm peripheral suture repair).¹⁵⁶

A third factor which increases the strength of peripheral suture repair is the number of bites

made to place the peripheral suture. Using cadaver human flexor profundus tendons, Kubota et al. found that when the number of throws of a peripheral suture increased, the force to gap formation and ultimate tensile strength of the repair increased.¹⁵⁷ As the number of suture passes was increased, there was no significant impact on the gliding function of the tendon.¹⁵⁷ Repairs tested included the simple, simple-locking, Lambert, Halsted, cross-stitch, and Lin-locking suture repairs.

Equine Digital Flexor Tendon Tenorrhaphy

i. Traditional Suture Patterns

Traditional suture patterns that have been previously described for repair of equine flexor tendon lacerations include the Bunnell and the Bunnell-Mayer suture patterns (Figure 37).^{84, 116} The Bunnell suture pattern fell out of favor for equine flexor tendon laceration repair with adoption of the Locking-Loop (LL) (Figure 19) pattern from human medicine. When compared to the Bunnell suture pattern, the LL required less surgical exposure of tendon ends and was easier to place than the Bunnell pattern.⁸⁴ However, in previous studies it was found that a single LL repair pattern was not strong enough to prevent gap formation in repaired equine tendon lacerations, even with external coaptation.¹¹⁸ One suture pattern that was adopted from human medicine that proved to be even stronger than the LL was the three-loop pulley (3LP) (Figure 20). Adair et al. investigated the LL and 3LP for repair of equine SDFT and DDFT lacerations *ex vivo* using #1 polydioxanone and polypropylene.⁹ When tested by linear load to failure, it was found that the 3LP withstood a greater load to failure than the LL regardless of the material used (92.10N± 10.91N for LL using monofilament polypropylene, 112.13N± 9.59N for LL using polydioxanone, 268.56N± 52.59N for 3LP using monofilament polypropylene, 245.76N± 69.58N for 3LP using polydioxanone).⁹ There was no significant difference in tensile strength of the repairs between the

SDFT and the DDFT.⁹ The 3LP formed a larger gap (2.97cm) prior to failure than the LL (2.13cm). The 3LP consistently failed by suture pull-out, while the mode of failure for the LL was suture breakage. The author concluded that the 3LP was a superior repair pattern to the LL due to its strength characteristics.⁹

In a separate *ex vivo* study, Easley et al. compared a single, double and triple LL to the 3LP suture pattern on cadaver equine SDFT. Repairs were made with #2 monofilament nylon. When tested by linear load to failure, the 3LP and the triple LL were the strongest of the repairs (308.5N± 39.5N load at failure for the 3LP, 323.2N± 70.1N load at failure for the triple LL).¹⁰ The single LL was the weakest repair tested, and also had the largest gap at the point of failure (145.1N± 40.5N load at failure, 26.66mm± 6.12mm gap at failure).¹⁰ The mode of failure for the 3LP was suture pull-out while the mode of failure for the triple LL was inconsistent with either suture pull-out or suture breakage.¹⁰ The author noted that the triple LL was the most technically difficult repair to perform and resulted in a significant amount of suture material within the tendon at the repair site. Based on the amount of suture at the repair site, the author expressed concern that the triple LL may have a significant impact on intratendinous vasculature.¹⁰

In a study by Jann and colleagues, a single LL was compared to the 3LP using 3-0 polydioxanone of varying strand numbers, including a single strand of 3-0 PDS, a single braid of three strands of 3-0 polydioxanone, or a double braid of six strands of 3-0 polydioxanone.¹¹ The single strand 3LP was stronger than the single strand LL when tested by linear load to failure (104.0N± 2.8N load at failure for single strand 3LP, 46.1N± 2.9N load at failure for single strand LL).¹¹ The 3LP repair with a single braid of 3-0 polydioxanone was similar in strength to the LL with a double braid of 3-0 polydioxanone (129.5N± 26.5N load at failure for the single braid 3LP, 128.5N± 17.7N load at failure for the double braid LL). The strongest repair was the 3LP repaired

with double braid 3-0 polydioxanone ($155.0\text{N} \pm 28.5\text{N}$ load at failure). There was greater gap formation with the LL repairs at the point of failure, and less force was required to create a 2mm and 10mm gap for the LL repairs compared to the 3LP repairs. The mode of failure for the LL repairs was primarily suture breakage, while the mode of failure for the 3LP repairs was primarily suture pull-out. Overall, the author found the 3LP provided more support, less distraction, and less tendon matrix constriction than the LL.¹¹

The most widely accepted repair pattern used for equine digital flexor tendon lacerations clinically is the 3LP using #2 polydioxanone.^{8,76} This may be due to the fact that this repair pattern is stronger than the LL and there is less tendon matrix constriction and distortion than the LL repair.^{11,23} It is difficult to create a repair that is strong enough due to the fact that normal equine flexor tendons need to be able to support up to 450kg at rest, with the superficial digital flexor tendon alone supporting an estimated tensile load of 1845N to 3559N at the walk.⁶ Based on previous studies, the 3LP is only able withstand up to one third of the normal weight-bearing load.^{6,7,9,10,12} This is further complicated by the fact that post-operative tenomalacia may develop at the suture-tendon junction, decreasing initial repair strength.¹⁴⁶ As a result, initial strength of the repair depends on properties of the repair technique, requiring external coaptation for prolonged periods to reduce strain on the repair until it can withstand physiologic tensile loading.¹⁵ Unfortunately, prolonged immobilization through external coaptation can have many deleterious effects on tendon laceration repair, namely, the strength of the repair, as well as post-operative functional characteristics. An additional criticism cited by authors evaluating the 3LP repair includes the presence of a significant amount of suture material on the exterior of the tendon. This could result in peritendinous adhesion formation due to impaired gliding function of the tendon.¹⁰ The amount of exposed suture could also serve as a nidus for infection.¹⁰ Finally, although the pattern lacks

locking attributes, the 3LP has been shown to compromise intrinsic vasculature of the SDFT, which could result in impaired tendon healing.²³ These disadvantages have prompted investigation of various other suture techniques, as well as intratendinous devices in an attempt to improve the quality of repair of equine digital flexor tendon lacerations.

ii. Novel Suture Patterns

1. Tendon Plates

The use of plates to repair equine digital flexor tendon lacerations was first investigated by Roush and colleagues. Stainless steel tendon plates were used to repair equine DDFT in an *ex vivo* model.⁹⁷ Application of the plates was compared to a single LL and 3LP using #2 monofilament polypropylene. The stainless steel plates were secured in place with 1-5 single LL sutures on either side of the plate. When tested by linear load to failure, the stainless steel plate technique was found to be stronger than the 3LP and a single LL (406.1N± 69.6N load to failure for the stainless steel plate, 310.9N± 55.8N load to failure for the 3LP, 170.2N± 8.8N load to failure for the single LL).⁹⁷ Mode of failure of the stainless steel plate was suture breakage on one side of the plate.⁹⁷ Use of four or five sutures to secure the stainless steel plate in place created the strongest repairs.⁹⁷ Although the stainless steel plate repair was stronger than the 3LP and the single LL, *in vivo* evaluation of the stainless steel plate application would need to be performed before it could be recommended clinically.

As an alternative to the non-absorbable stainless steel plate, Jenson et al. developed a bioabsorbable tendon plate composed of poly-L-lactic acid (Figure 38) and evaluated its strength in comparison to the 3LP. Poly-L-lactic acid is a synthetic bioabsorbable polymer that has been used for a variety of maxillofacial and orthopedic procedures in human medicine.¹⁵⁸ Evaluation of poly-L-lactic acid in tissues of mice has found that it does not cause local tissue necrosis, acute

inflammation or abscess formation at the area of implant placement.¹⁵⁸ Complete resorption of a poly-L-lactic acid implant has been estimated to be approximately 501 days from initial placement.¹⁵⁸ Resorption occurs through lactate/pyruvate metabolism pathways and is eliminated from the body as carbon dioxide.¹⁵⁸ In the study by Jenson et al., equine DDFT were harvested and repaired with either the poly-L-lactic acid tendon plates or the 3LP with #2 polydioxanone.⁸ The poly-L-lactic acid plates were shaped in a semi-cylindrical hour-glass shape which conformed to the palmar surface of the DDFT. Plates were 11.5cm x 2.5cm x 3mm in size and each plate had sixteen holes on either end to place five simple interrupted full-thickness sutures and two figure-of-eight full-thickness sutures using #2 PDS. The peak force of failure of plated tendons was 1507.8N, which was significantly higher than those repaired with the 3LP (460.9N). The plates failed by sutures pulling out of the tendon.⁸

Although repairs in which tendon plates are used appear to be stronger than the standard 3LP, authors investigating their use have expressed concerns over how they would be accepted in tissue. With plate placement, it can be assumed that post-operative gliding function would be greatly impaired, potentially inducing adhesion formation.⁸ In addition, they could limit tendon healing by vascular compromise, and harbor sepsis.⁸ To date, there have been no published reports of the *in vivo* use of stainless steel or poly-L-lactic acid tendon plates.

2. Bridging Implants

Eliashar et al. investigated the use of a poly-L-lactic acid implant to bridge equine flexor tendon lacerations that could not be sutured in four clinical cases.¹³ Two of the horses had complete transection of the SDFT, one horse had complete transection of the SDFT and DDFT and one horse had complete transection of the DDFT.¹³ In two horses, the injury was within the digital flexor tendon sheath. Three of the horses had sustained the injury within 24 hours prior of initially

being examined, however, one horse sustained the injury approximately four to five weeks prior to being examined.¹³ All tendons involved were in the hindlimbs and all horses were AAEP grade 4/5 lame at presentation. Following evaluation with digital palpation, ultrasonography and radiography, all horses were placed under general anesthesia and the wounds were revised. Tendon ends of three out of four horses were debrided, leaving a gap between tendon ends of approximately 4 to 5cm. In the horse in which the injury was present for a prolonged period prior to presentation, the tendon ends had reflected on themselves and were fixed in fibrous tissue. Tendon debridement was more extensive in this case, leaving a 10cm gap at the repair site.¹³ The implant was 25cm in length and 9mm in width, consisting of 1080 braided filaments of poly-L-lactic acid. The implant was placed through longitudinal stab incisions in the tendon ends and secured in place by suturing the ends of the implant together with #3-#5 polydioxanone. In the horse in which both flexor tendons were transected, two implants were placed. All horses were placed in a distal limb cast with the metatarsophalangeal joint kept in neutral position and a wooden wedge was placed on the bottom of the hoof and incorporated into the cast.¹³ External coaptation was maintained in place for up to 16 weeks post-operatively. All four horses experienced no complications associated with the implant, which appeared to become well incorporated in scar tissue present at the tendon gap.¹³ Two of the four horses (50%) returned to work within a year. The horse which sustained lacerations of the SDFT and DDFT, as well as the horse which sustained the initial injury four to five weeks prior to presentation were still lame after one year and remained on pasture rest. There was no control group as part of this study. As a result, the authors were not able to determine if the implant contributed to improved tendon healing compared to other forms of tenorrhaphy.¹³ To date, there has been no further reported use of this implant *ex vivo* or *in vivo*.

3. *Savage Suture Pattern*

The Savage suture pattern was first described by Savage in 1985. The suture pattern was developed for repair of human digital flexor tendons in order to increase repair strength, allowing for earlier post-operative mobilization.^{124, 128} The six-stranded repair uses a grasping cruciate-like technique to anchor suture within the tendon substance in three equally spaced locations around the circumference of the tendon (Figure 22). This grasping pattern has been found to be superior in both strength and resistance to gap formation experimentally in human studies.^{124, 128} To increase the strength of the repair, epitenon at the tendon ends can be sutured in a Lin-locking epitenon suture (Figure 39). The Lin-locking suture pattern is a continuous locking suture pattern that out of six commonly used circumferential epitenon sutures in human flexor tendon laceration repair provided the highest additional strength to a primary core tendon suture repair.¹⁵⁷

The Savage suture pattern was first evaluated in equine tenorrhaphy by Smith and colleagues in 2011. The suture pattern was modified to create a 10-strand technique in which sutures were placed in five locations equally spaced around the circumference of the tendon. The repair was tested with and without a Lin-locking epitenon suture in an *ex vivo* study using equine superficial digital flexor tendons.⁶ Suture material used for repairs included #2 polydioxanone or #2 polyglactin 910. The 10-strand Savage was compared to the 3LP through linear load to failure. All Savage repairs failed by suture breakage at the knot-suture interface while the 3LP failed by suture pull-out from the tendon. The 10-strand Savage with or without the Lin-locking suture was significantly stronger than the 3LP regardless of suture material used. The Lin-locking suture made the 10-strand Savage significantly stronger (998 N) than without the epitenon suture (872 N). Use of #2 polyglactin 910 significantly increased the strength of any of the Savage repairs over the use of #2 polydioxanone in the same repair (978 N vs. 765 N without epitenon suture or 1105 N vs 890 N with an epitenon suture). The opposite was true for the 3LP where repairs made using #2

polydioxanone were significantly stronger than 3LP repairs performed with #2 polyglactin 910 (382 N vs 292 N). Smith did not evaluate the load required to produce a 2 mm gap but found a gap at failure of 17.6 mm for the Savage repair without the epitenon suture and 18.9 mm with the epitenon suture. The gap at failure for the 3LP was 31.7 mm which was significantly larger than the Savage repairs. Smith concluded that although the 10-strand Savage was three times as strong (978 N-1105 N) as the 3LP (292-382N) there could be difficulties using the pattern in clinical cases. The pattern requires 360-degree access to the tendon to place it appropriately and it is technically challenging to perform. Smith also noted that the circumferential nature of the pattern and its locking attributes may theoretically affect blood flow at the repair site because of disruption of the peritendinous tissues in placement, as well as compression of intratendinous vasculature.⁶

A six-strand Savage (Figure 22) has been evaluated by Everett, et al. in comparison to the 3LP in cadaveric equine SDFT.⁷ Repairs were performed with #2 polydioxanone without an epitenon suture. This study found the six-strand Savage had a higher ultimate load to failure (421 N) than the 3LP (193 N). However, the two suture patterns had similar resistance to 3 mm gap formation of 102N for the Savage pattern and 109N for the 3LP. Similar to the previous study the authors concluded that there may be some limitations to the use of the suture pattern in *in-vivo* trials because the pattern may affect blood flow and that the overall strength of the pattern was not strong enough to resist gap formation in standing horses.⁷ Impairment of intratendinous vasculature was confirmed by Freeman et al. when it was found that the Six-Strand Savage significantly reduced intratendinous vessel density when compared to the 3LP in equine SDFT.²⁴

4. Tendon Implants

Barrett et al. investigated the use of the TF for repair of equine digital flexor tendons *ex vivo*.¹² SDFT were harvested from forelimbs of horses euthanized for reasons un-related to the

study. Tendons were transected at mid-point and repaired with either four TF devices placed in line (Figure 40) or a standard 3LP using #2 polydioxanone. All repairs were tested by linear load to failure. It was found that the TF repair was similar in strength to the standard 3LP to 2mm gap formation ($114.5\text{N} \pm 21.5\text{N}$ load at 2mm gap for TF, $164.9\text{N} \pm 67.7\text{N}$ load at 2mm gap formation for 3LP), however, the TF was significantly weaker than the 3LP to ultimate load at failure ($132.4\text{N} \pm 26.8\text{N}$ ultimate load at failure for the TF, $363.5\text{N} \pm 83.7\text{N}$ ultimate load at failure for the 3LP).¹² Mode of failure was suture pull-out for the 3LP and device pull-out for the TF repairs.¹² The gap at failure for the 3LP was significantly larger for the 3LP than the TF repair. Given other benefits of the device as observed in previous canine models, further evaluation of this device was warranted to aid in determining a suitable repair method for equine digital flexor tendon lacerations.

Testing of Repair Methods

Biomechanical testing of repaired tendons is typically performed *ex vivo* using a linear materials testing machine where the tendon and repairs are distracted by a single linear load to failure.¹⁵⁹ This is also known as static tensile testing.⁵⁸ Tendons are placed between two grips and pulled apart at varying displacement rates. The materials testing machine can be coupled with load cell data collection software and a high speed camera to aid in obtaining data during testing. This allows one to determine the force required to produce a gap at the repair site, force at failure of the repair, as well as the mode of failure. However, physiologic conditions, such as cycling of a repair, are not reflected with this method. As an alternative, dynamic linear testing can be performed. Dynamic linear testing consists of cyclic loading of a tendon, providing a more physiologic evaluation of a repair.^{88, 160, 161, 162, 163} Cyclic loading, also known as conditioning, has resulted in gap formation at significantly smaller loads than with single linear load to failure

testing.^{161, 164} Dynamic testing investigates the remaining strength of a tendon repair after a known number of cycles have been applied to the repair. The number of cycles applied to a repair varies from study to study, ranging from 100 cycles per tendon¹⁶⁵ to as many as 3000 cycles per tendon.¹⁶⁶ One disadvantage of using this testing protocol is that it can be time-consuming, especially if a large number of tendons are to be tested with a high number of cycles per tendon. Also, this method does not allow for easy comparison of the biomechanical behavior of repairs, such as their strength and mode of failure, which can be easily obtained with static tensile testing.^{94, 159}

Another limitation of static tensile testing is that it usually focuses on the failure region of the load deformation curve. As a result, the ultimate force to failure is considered the strength of the repair.⁵⁸ Typically at this point, there is a gap in the repair of several millimeters. In a clinical setting, a gap this large would be considered catastrophic. A better measure of the strength of a repair using static tensile testing may be the yield point of the repair.⁵⁸ The yield point is the point at which irreversible damage to the repair begins to occur prior to complete failure. When applied to a clinical setting, it could be assumed that in order for a repair to remain intact during post-operative rehabilitation, the forces subjected to the repair should not exceed its yield force.⁵⁸

A third limitation of static tensile testing is that stiffness of the repair and tendon is reported based on grip-to-grip displacement and whole tendon load. However, displacement and load within a repaired tendon will vary considerably due to focal changes in stiffness. In current studies, this factor has not been considered; it may be of benefit to investigate this parameter in future studies when determining the strength and stiffness of a tendon repair.⁵⁸

Prognosis

Based on previous retrospective studies of flexor tendon lacerations in horses, prognosis

for future athletic performance is variable, but generally poor (Table 1).^{2, 4, 167} Foland et al. investigated the prognosis in 35 horses that sustained lacerations of the SDFT and DDFT. Of the 35 cases, 11 horses were euthanized shortly after presentation without pursuing any treatment. The reason(s) for immediate euthanasia were not specified.⁴ The remaining horses were treated. Of the 24 horses treated, 22 had follow-up; one horse developed fatal enterocolitis and was euthanized prior to discharge and the other horse was lost to follow-up. Of the 22 horses treated, three horses had complete transection of only the SDFT, one horse had partial transection of only the SDFT, three horses had complete transection of only the DDFT, one horse had only partial transection of the DDFT and 14 horses had involvement of both flexor tendons, either partial or complete.⁴ Tendons were sutured in 13 of the 22 cases; the remaining cases were left to heal by second intention.⁴ Suturing consisted of either a single (9/13 cases) or double (4/13 cases) LL suture pattern using #2 monofilament nylon. Reasoning for the use of this suture pattern and material was not specified. External coaptation was used in 21 of the 22 cases for 17 to 66 days following initial treatment, with the exception of two horses, which were euthanized prior to completion of treatment.⁴ Of the patients that survived, 18% (4/19) returned to their previous level of athletic activity, 41% (9/19) returned to a reduced level of activity and 32% (7/19) were pasture/breeding sound.⁴ Of the cases in which tenorrhaphy was performed, 62% returned to their previous level of activity or reduced level of activity, while the remaining 31% were pasture/breeding sound.⁴ Of the surviving cases in which tenorrhaphy was not performed, 41% returned to their previous level of performance, while 33% returned to a reduced level of performance.⁴

Taylor et al. investigated the long-term prognosis of flexor tendon lacerations in 50 cases. Case records were obtained for 90 horses that had sustained flexor tendon lacerations.² Twenty-two horses were euthanized at presentation without pursuing further treatment.² The remaining 68

horses were treated; however, follow-up information was only available for 50 cases.² The 22 horses that were euthanized and the 18 cases that were lost to follow-up were not included in the study. Of the 50 cases, 16 presented with complete transection of one tendon (either SDFT or DDFT), 16 presented with both flexor tendons completely transected and 18 presented with one or both flexor tendons only partially transected. Of the 50 cases treated, tendons were sutured in 16 cases. In horses in which there was partial disruption of a flexor tendon and tenorrhaphy was pursued (3/16), simple interrupted or horizontal mattress sutures were used for repair, with the exception of one case in which an unspecified tenorrhaphy pattern was used. In cases of complete tendon transection and tenorrhaphy was pursued (13/16), Bunnell, single or double LL or 3LP suture patterns were used for repair. Suture material used included carbon fiber (6/16), monofilament polypropylene (5/16), polydioxanone (3/16) and polyglactin 910 (1/16). At one year post-treatment, of the 16 horses in which there was complete transection of only one flexor tendon, 75% (12/16) were still alive and 63% (10/16) were sound for riding.² Of the 16 cases in which there was complete transection of both flexor tendons, 81% (13/16) were still alive and 56% (9/16) were still sound for riding.² Of the 18 cases in which there was partial transection of one or both flexor tendons, 78% (14/18) were still alive and 67% (12/18) were sound for riding.² Overall, 78% (39/50) of all of the 50 cases were still alive one year following surgery and 54% (27/39) had returned to their previous level of athletic activity.² For cases in which tenorrhaphy was performed, 94% of cases were still alive one year following initial treatment (15/16) and 56% (9/16) returned to their previous level of performance.² For cases that were allowed to heal by second intention, 71% were still alive one year following initial treatment (24/34) and 53% (18/34) returned to their previous level of performance.² An association could not be found between patient outcome and mode of treatment (sutured versus un-sutured).² Also, differences in prognosis between the

different tenorrhaphy patterns in this study was not specified.²

Jordana et al. investigated the prognosis of horses that had sustained lacerations of the SDFT, DDFT, suspensory ligament and/or distal sesamoidean ligaments in 106 horses. Of these horses, 60 presented with partial laceration of one or more tendons and/or ligaments, while the remaining 46 presented with complete transection of one or more tendons and/or ligaments.¹⁶⁷ Tenorrhaphy was performed in 58 cases; the most common repair patterns consisted of the 3LP or double LL with large diameter polydioxanone.¹⁶⁷ The exact number of each repair pattern used was not specified, nor was the prognosis based on individual tenorrhaphy patterns. External coaptation was used for 84 horses for duration of 0.5 to 14 weeks. Following removal of the cast, limbs were supported with bandages, as well as extended heel shoes when needed.¹⁶⁷ At one year following initial treatment, 55% (58/106) of horses returned to their previous level of performance, 27% (29/106) returned to a lower level of performance and 18% (19/106) were euthanized due to the severity of the injury sustained (3/106) and complications of the injury (16/106), including persistent lameness, or persistent infection of the digital flexor tendon sheath.¹⁶⁷ Other complications encountered post-operatively included persistent thickening of the involved tendons, subcutaneous tissues and skin, permanent hyperextension of the metacarpo/metatarsophalangeal joints, deep cast sores, wound dehiscence, and persistent digital flexor tendon sheath tenosynovitis.¹⁶⁷ Hyperextension of the metacarpo/metatarsophalangeal joints was significantly associated with an unsuccessful outcome, in which horses returned to a lower level of performance.¹⁶⁷ When more than one structure was involved in the initial injury, there was a higher likelihood of an unsuccessful outcome.¹⁶⁷

III. Objectives

Flexor tendons in horses are part of a biomechanical apparatus that maintain the metacarpo/metatarsophalangeal joints suspended above the ground, absorb shock, store elastic energy during movement, and contribute to weight bearing.¹ Flexor tendon lacerations in horses result in a loss of biomechanical function of the limb, which is often career and life threatening.⁷ In a recent study, 55% of horses surgically treated for flexor tendon lacerations returned to athletic function, with 27% returning to limited athletic function.¹⁶⁷ Although several methods of tendon repair have been proposed, an ideal suture material or repair technique has not been identified and successful repair is often challenging. With a guarded prognosis for return to athletic function, research in development of an improved tenorrhaphy technique is warranted.

Goals of successful tenorrhaphy include creating a strong repair and restoring gliding function to the tendon with minimal gap formation, negligible adhesion formation, and insignificant disruption of blood supply. It is difficult to create a repair that is strong enough due to the fact that normal equine flexor tendons need to be able to support loads of up to 3559N at a walk.⁶ Current repair methods are only able to withstand up to one third of the normal weight-bearing load.^{6, 8} As a result, repairs are often supported by external coaptation to reduce strain on the repair until it can withstand physiologic tensile loading.¹⁵ Increasing the strength of repair would allow for decreased time spent in external coaptation and decreased morbidity associated with cast sores, loss of bone density, decreased muscle strength, tendon laxity and failure of tendon repair.^{8, 15, 16}

It can be difficult to restore gliding function and reduce adhesion formation for several reasons. Tendon injury from percutaneous trauma is often associated with blunt trauma to tendons, which can prevent direct apposition of tendon ends. Even with debridement, external coaptation, and application of an implant, second intention healing must occur with a gap present, which results in a larger area of inferior collagen or scar tissue present within the tendon.⁸¹ This reduces the gliding function of the tendon and increases the risk for adhesions. In addition, some repair techniques, including the three-loop pulley (3LP), result in an excess amount of exposed suture material, risking the formation of excessive scar tissue and adhesions. Excess suture can also serve as a nidus for infection, which can easily occur due to contamination from percutaneous trauma.^{10,11}

Another challenge is that current repair techniques often compromise blood flow and tissue healing. Of suture patterns assessed for potential equine flexor tendon laceration repair, the 3LP is superior in terms of strength and failure characteristics and prevents distraction of tendon ends under loading.^{9, 10, 11, 15} As a result, it is currently considered the clinical standard for equine flexor tendon laceration repair. However, the 3LP can have deleterious effects on intrinsic vasculature of the superficial digital flexor tendon, resulting in decreased density of perfused intra-tendinous vasculature.²³ This may be due to the circumferential nature of the suture pattern, causing strangulation of blood supply to tissues.

Current research is focused on repair with bioabsorbable implants and autologous tendon grafts with little research on devices that can be placed within tendons.^{8, 17-20} One such device is the Teno Fix® (TF) device, which is a knotless suture construct that anchors linear stainless steel suture completely within severed tendon, so no portion of the suture is located outside the tendon. This design utilizes the high tensile strength of stainless steel while avoiding the poor knot-tying

qualities of the material. Threads present on stainless steel anchors of the system interdigitate with collagen fibers of normal tendon beyond the laceration site. Multifilament suture of the system does not weave throughout the site of the laceration, reducing overall bulk of suture at the site.²¹ Evaluation in canine tendon repair found that the device produced minimal inflammation and scarring, and did not interfere with tendon healing.²² A previous *ex vivo* study that evaluated the use of four TF devices placed in line in comparison to the three-loop pulley suture pattern for equine superficial digital flexor tendon lacerations found that the TF repair was similar in strength to the three-loop pulley technique in the development of a 2mm gap.¹² Variation in the suture pattern for the TF device has yet to be evaluated in terms of improving strength of the material for repair. Given the other benefits of using the device for tendon repair, further evaluation of the use of this device is warranted to aid in determining a suitable repair method for equine flexor tendon lacerations.

Core purchase length is the exit and, or, entry point of a core suture from the cut ends of a tendon. It has been reported that a longer purchase length resulted in a greater force required to develop a 2mm gap at a repair site, as well as ultimate failure in porcine tendons.²⁵ It was speculated that by applying this principle to the TF device, it may strengthen repairs for equine flexor tendon lacerations. The goal of this study was to compare a staggered TF device repair pattern to a 3LP in ultimate failure in Newtons (N), load to a 2 mm gap (N), mode of failure, and gap at failure (mm). Staggered placement of the TF device would be achieved by increasing the core purchase length of some of the implants. It was hypothesized that staggered placement of TF anchors would increase the strength of the repair to ultimate load to failure and load to a 2mm gap. In addition, it would allow for placement of an additional implant, which was hypothesized to further increase the strength of repair to ultimate load to failure and load to a 2mm gap.

IV. Materials and Methods

i. Collection of Samples

Forelimb SDFT were collected from 20 adult horses euthanized for reasons unrelated to this study (age 4 to 28 years; weighing 450 to 670kg, 16 geldings, 4 mares; breeds represented included seven American Quarter Horses, two Appaloosas, one Arabian, one Belgian, two Hanoverians, two Tennessee Walking Horses, two Thoroughbreds and three Welsh Cobs). Pre- and Post-mortem use of animals and tissues was approved by the Auburn University Animal Care and Use Committee. All horses were evaluated at a walk and trot prior to euthanasia for the absence of lameness using the AAEP lameness scale. Any horse with an observable lameness was excluded from the study. Both distal forelimbs were evaluated ultrasonographically to ensure flexor tendons were structurally normal. Cross sectional area of the SDFT within zone 2B¹⁶⁸ was recorded for all tendons and ranged in size from 123.2 mm² to 137.2 mm² (mean of 128.4 mm² ± 0.79 mm²). Immediately after euthanasia, SDFT were collected by transecting the tendons proximally at the level of the carpometacarpal joint and distally at the level of the proximal sesamoid bones. Tendons were rinsed and wrapped in gauze soaked with saline (0.9% NaCl, Baxter Healthcare Corp, Deerfield, IL) and sealed in plastic for freezing. Tendon specimens were preserved at -70°C until testing.

ii. Experimental Repair

Tendons were thawed on the day of testing to room temperature and transected at midpoint using a #10 scalpel blade. Each of the 20 tendon pairs (one pair consisted of two tendons from the

same horse) were selected at random; each pair was repaired with either a 3LP or one of two TF device patterns. This ensured that each horse was its own control. TF sutures were evenly distributed and staggered either 2 cm or 3 cm away from the point of transection. Four TF devices (4TF) were placed in a staggered pattern for 10 tendons and five TF devices (5TF) were placed in a staggered pattern for 10 tendons (Figure 41). A standard 3LP pattern was placed in a total of 20 tendons. Core purchase length of all implants, as well as suture placement, was determined using a metric ruler placed next to each tendon during repair.

The TF device was placed as described by Su *et al* in 2005.²¹ Briefly, stab incisions were made through the paratenon at 2 cm or 3 cm from the point of transection using a #15 scalpel blade. Stainless steel anchors contained within an obturator were placed into tendons through stab incisions with the obturator facing towards the transected ends of tendon (Figure 30). Multifilament 2-0 stainless steel suture loaded on a 3.8 cm long straight taper needle was passed through the anchor, crossing the transected ends of tendon, and through a second anchor placed in a manner similar to the first anchor (Figure 31). A stop bead placed at the end of the suture engaged the first anchor, preventing the suture from being completely pulled through the anchor (Figure 32). The remaining suture was placed in a crimper and tightened until the transected ends of the tendon were tightly apposed. Excess stainless steel suture was cut and the paratenon closed routinely with #4-0 USP polydioxanone (PDS, Ethicon Inc, Somerville, NJ) in a simple continuous suture pattern.

The 3LP repair was placed as previously described with #2 USP polydioxanone (PDS, Ethicon Inc, Somerville, NJ) (Figure 20).¹⁰ Suture bites 1 and 6 were placed 2cm from the site of transection, bites 3 and 4 were placed 3cm from the site of transection and bites 2 and 5 were placed 4cm from the site of transection.¹⁰ Suture was tied with a surgeon's knot followed by three

square knots. Suture was tightened such that tendon ends were tightly apposed and there was no slack in suture. Immediately after repair, all tendons were placed in phosphate buffered saline (PBS, Baxter Healthcare Corp, Deerfield, IL).

iii. Mechanical Tensile Testing

Ultimate failure in Newtons (N), load at a 2 mm gap (N), mode of failure and gap at failure (mm) were measured for all repairs. Failure was defined as a sudden drop in tension as recorded by the materials testing machine. Failure mode of the 3LP was defined as suture pull-out or suture breakage. Failure mode of the TF device was defined as anchor pull-out of the tendon or anchor failure due to the stop bead coming off the stainless steel wire. An Instron 5565 Universal Testing Machine (Instron, Norwood, MA) was used to perform all tensile tests. Custom-designed steel clamps were used to grip tendons for testing with the distance between clamps set at 150mm (Figure 42). A metric measuring tape was suspended from the top clamp in the same plane as the tendon to act as a frame of reference for acquiring measurements from videos recorded with the high-speed camera. Tendons were placed under a preload of 1N prior to testing and tests were performed at a displacement rate of 8.5 mm/sec until failure. A displacement rate of 8.5mm/sec was chosen to maintain a similar project design to that of Barrett et al. 2014¹² to allow discussion of results obtained between both studies. Load data was recorded every 10ms during testing by load cell data collection software (BlueHill 2, Instron Inc, Norwood, MA, USA). Each test was recorded using a high speed camera recording at 240 frames per second (Casio® Exilin, Casio America Inc., Dover, NJ). Videographic analysis software (Dartfish Pro 7.0, Dartfish, Alpharetta, GA) was used to observe gap formation and determine tensile load at a 2mm gap for each repair. Gap at failure and mode of failure was also determined using this software. Non-linear load-elongation curves were obtained for each specimen through the load cell data collection software.

Stiffness (rigidity of the repair measured in N/nm) was determined using the slope of the linear region of the load displacement curve.

iv. Data Analysis

Data was collated using Microsoft Excel (Microsoft Corporation, Redmond, WA) and analysed using JMP® Pro 11.0.0 (SAS Institute Inc., Cary, NC). The load to failure, load to 2mm gap, gap at failure and stiffness were tested for normality using the Anderson Darling test. Paired t-tests were performed to compare load to failure, load to a 2mm gap, gap at failure and stiffness between the 3LP and both TF repairs. Unpaired t-tests were performed to compare load to failure, load to a 2mm gap, gap at failure and stiffness between the 4TF and the 5TF. Significance for all analyses was set at $P \leq .05$. The frequency of mode of failure was described.

V. Results

i. Load at Construct Failure, Mode of Failure, Gap at Failure, Stiffness (Table 2)

All tendons completed testing immediately after repair; no grip failure or slippage occurred during testing. The mean ultimate load at failure for the 3LP was significantly greater than both TF repairs ($p<0.001$) (Figure 43, Appendix 3). Use of an additional TF implant did significantly increase the strength of repair when comparing the 4TF to the 5TF ($p=0.004$). Mode of failure was suture pull out or suture breakage for the 3LP and anchor pull out or anchor failure for the TF. Failure for the 3LP was defined as the point at which the first loop of the 3LP pulled through tendon. At this point, the gap at failure was significantly larger in the 3LP than in both TF repair patterns ($p<0.001$). The stiffness (the slope of the load displacement curve) of both TF repairs was significantly higher than the stiffness of the 3LP ($p<0.001$). There was no significant difference in the stiffness between both TF repairs ($p=0.91$).

ii. Load at 2 mm Gap Formation (Table 2)

The mean maximum load to create a 2mm gap in the 4TF and 5TF repair was significantly less than that of the 3LP repair ($p<0.001$). There was no significant difference in load to a 2mm gap, or gap at failure between both TF repairs ($P=0.11$; $P=0.15$, respectively).

IV. Discussion

This *ex vivo* study compared the biomechanical properties of two TF implant patterns to the standard 3LP in equine SDFT tenorrhaphy. Failure in this study was defined as a sudden drop in tension as recorded by the materials testing machine. When evaluating the load-displacement curve, it should be noted that after the initial drop in tension for the 3LP repair, there is an additional increase in force. This is associated with tension being placed on the second and third loops of the repair pattern, which are still intact. Based on the definition of failure for this study, failure for the 3LP occurred after the first drop in tension, which is associated with the first loop of the 3LP repair pulling through the tendon. It could be argued that ultimate failure occurs following pull-out of the third loop of the 3LP. The mean gap at the point at which the third loop of the 3LP pulled through the tendon was $90.3\text{mm}\pm 24.0\text{mm}$. This is in contrast to the mean gap when the first loop of the 3LP pulled through the tendon, which was $38.1\pm 5.2\text{mm}$. From a clinical stand-point, a gap as large as when the third loop of the 3LP pulled through the tendon would be catastrophic for tendon healing, as well as the future function of a patient. Therefore, failure for the 3LP was considered to be the point at which the first loop of the 3LP pulled out of the tendon.

Staggered insertion of the TF implant pattern as well as increasing the number of implants increased the ultimate strength of repair in relation to the results obtained from a previous study by Barrett in 2014 (mean, 95% confidence interval of 132.4, 113.8-151.0 N for 4TF in the previous study versus 193.6, 177.5-209.6 for the 4TF and 222.8, 207.8-237.7 N for the 5TF in the present study).¹² Increased ultimate strength of the two TF patterns in this study may, in part, be due to the

increased core purchase length of some of the implants in both patterns. Previous studies using porcine tendon have found that increasing core purchase length for both locking and grasping suture patterns increased the force required for 2mm gap formation and ultimate failure, as well increased overall repair stiffness.^{25, 146} It is believed that increasing core purchase length increases repair strength by increasing tendon-suture interaction. This is critical in a clinical setting, as cut ends of tendon soften within the first week following a laceration due to collagen degradation from proteolytic enzyme release during the inflammatory stage of repair.¹⁴⁶ This could result in decreased strength of the repair if it was closer to the tendon ends in weakened tendon tissue. In these porcine studies, a threshold point of 0.7mm to 1.0cm from the cut ends of tendon was identified. Beyond this point, the repair did not become stronger.^{25, 146} In this study, core purchase length for the TF repairs varied between 2cm and 3cm. This is much greater than the standard core purchase length used in human digital flexor tendon laceration repair of 0.7 to 1.0cm.^{25, 146} Core purchase length has not been investigated in the same manner for equine tendon laceration repair. Therefore, a threshold point for core purchase length for equine tenorrhaphy is not known. Further investigation of core purchase length as a variable for equine tenorrhaphy may be beneficial in increasing repair strength for current repair patterns.

Previous human studies evaluating multi-strand tenorrhaphies have found that the number of suture strands crossing a tendon laceration site is directly proportional to the strength of repair.^{92, 96, 124-127} In this study, addition of a fifth TF implant did significantly increase the strength of the TF repair to the point of failure, but not to a 2mm gap. At this time, it is not possible to determine

the tension placed on each device in the tendon. There could be unequal tension across the devices within the tendon²⁰ and underloading, unequal tension across the repair site could weaken the repair.²⁰ This might explain the failure to see an increase in strength of the 5TF repair to a 2mm gap over the 4TF repair. Placement of the fifth TF implant was difficult, time consuming, increased the manipulation of the tendon and created suture bulk at the repair site. Its suitability in clinical cases where excessive manipulation and suture bulk at the repair site could impair tendon healing and future gliding function, in the face of no obvious gain in strength, is questioned.

Both TF patterns were not as strong as the 3LP in ultimate load at failure and load to a 2mm gap. When taking into account that the SDFT has been reported to support a tensile load of 1845N to 3559N at a walk,⁶ the 4TF could only withstand approximately 5% of this normal weight-bearing load, while the 5TF could withstand 7% of the load. The 3LP could resist 10% of the normal weight-bearing load. In contrast, the 10-strand-Savage (10SS), as reported by Smith and colleagues in 2011, could withstand approximately 25% of the normal weight-bearing load.⁶ When a Lin-Locking-loop suture was applied to the 10SS, the repair could withstand approximately 28% of this load.⁶ In a clinical setting, all three repairs tested in this study would be at high risk of repair failure, especially the TF repairs. This is critical within the first few weeks following a tenorrhaphy, as it has been reported that a weakening of tissue occurs at the repair site.^{95, 126, 127, 132} All three repairs are also weaker than Savage pattern previously described.⁶ Although the 10SS pattern is a stronger repair comparatively,⁶ it may have a significant negative impact on intratendinous vasculature. Based on previous findings of decreased perfused intratendinous

vasculature with both the 3LP and six-strand Savage (SSS),^{23, 24} it can be presumed that with an increase in the number of suture strands of the 10SS, perfusion would be further decreased.

Force required to create a 2mm gap for the 3LP repair in this study was comparable to results obtained in a previous study (mean, 95% confidence interval 164.9, 118.1-211.7 N for 3LP in the previous study and 184.8, 165.9-203.7 in the present study).¹² In contrast to previous reports, significantly less force was needed to create a 2mm gap for both TF repairs compared to the 3LP, although the force was not different between TF repairs. A previous study reported a tensile force of 114.5, 99.6-129.4 N for a 4TF pattern¹² compared to 56.9, 25.3-88.5 N in the present study. These findings do not support our hypothesis in which increased core purchase length as well as an additional implant would increase the force required to create a 2mm gap at a tenorrhaphy site for the TF device. From a clinical standpoint, this is significant as gap formation as small as one to two millimeters at a tenorrhaphy site has been associated with poor functional performance, resulting in increased gliding resistance, peritendinous adhesion formation, reduced repair strength and possible tendon rupture in humans.^{9, 14, 98} Although gap formation has not been studied as extensively in equine flexor tendon tenorrhaphy, it is possible that it would have the same impact as observed in human flexor tendon tenorrhaphy. If altering the implant pattern does not make the tenorrhaphy stronger than the 3LP in terms of resisting 2mm gap formation, the repair may not be ideal for equine flexor tendon laceration repairs.

The 3LP in this study failed by either suture pulling through the tendon (18), or suture breakage at the level of the knot (2). In a previous study by Barrett et al., all 3LP repairs failed by

suture pulling through the tendon.¹² It can be presumed that this type of failure occurs due to the fact that the 3LP is a grasping repair. A suture pattern is considered a grasping pattern if the suture loop does not tighten around tendon fibers and as a result pulls through the tendon fibers when tension is applied to the suture because the suture is stronger than surrounding tendon matrix.¹¹ This mode of failure predominates for the 3LP as seen in previous studies.^{7, 11} Suture breakage at or near the knot for the 3LP has been previously described.¹¹ Suture breakage at the knot could be due to increased shear force at the knot with applied tension, favoring knot failure.^{169, 170}

The TF repairs in this study failed by either the device pulling through the tendon or through anchor failure (anchor pull-out in 9 repairs for the 4TF and 7 repairs for the 5TF; anchor failure in 1 repair for the 4TF and 3 repairs for the 5TF). Anchor failure was caused by the stop bead sliding off the stainless steel suture. In a previous study by Barrett et al., failure of the TF repair was due to the devices pulling through the tendon only.¹² To secure the TF device in tendon, a stop bead is applied to the stainless steel suture using a crimper with the suture under tension. Failure of some of the devices in this study could have been due to operator error in which the stop beads were not secured on the stainless steel suture correctly, resulting in failure.

For clinical use, all repairs would have to be supported by prolonged external coaptation.⁸¹ Cast immobilization, however, can delay generation of tendon matrix and remodeling of normal tendon architecture, resulting in inefficient fiber alignment and scar formation.^{8, 81, 171-173} In addition, it is unknown at this time how much strain external coaptation reduces on the flexor tendons to facilitate healing. Therefore, even with external coaptation, these repairs may be at high

risk of significant gap formation, construct failure, longer healing time and poor quality repair tissue.

One limitation of this study was the use of frozen instead of fresh tendons. Frozen tendons were used for this study to allow results obtained to be compared to those of the study by Barrett in 2014 without variability in project design. Previous research of human tendons has revealed that freezing has a negative impact on tendon structure.¹⁷⁴ Freezing and subsequent thawing of human posterior tibial tendon resulted in decreased collagen fibril density.¹⁷⁴ It was believed that this was due to the formation of ice crystals within tendon tissue during freezing.¹⁷⁴ Frozen-thawed tendons had decreased strength at failure and increased stiffness when compared to fresh tendon.¹⁷⁴ The TF device is dependent on its interaction with collagen fibrils for its holding strength. If there is decreased collagen fibril density due to freezing, this may have a negative impact on the repair by reducing its ability to grasp tendon fibrils. However, it is important to note that this study evaluated intact tendon samples. In a study by Hipara in 2008, tendons were harvested from porcine forelimbs and subjected to different forms of preservation, including fresh samples, refrigeration for 24 hours, frozen for 3 months or frozen for 6 months.¹⁷⁵ All tendons were transected and repaired with a Silfverskiold peripheral cross-stitch using 6-0 monofilament nylon and a Pennington modified core suture using 4-0 polybutylate coated braided polyethylene terephthalate.¹⁷⁵ It was found that freezing had no impact on force required to produce a 3mm gap at the repair site or ultimate strength to failure when compared to fresh tendons.¹⁷⁵ Therefore, the impact on freezing of tendon samples to test tenorrhaphies may not be substantial. Further investigation may be warranted to determine if freezing has an impact on the ultimate strength of

the TF repair.

A second limitation of this study was the use of a low strain of 0.5 to 1%, with a distraction rate of 8.5mm/sec to test tendons. Strain of a tendon is the change in length of a tested tendon over the original length of the tendon. This strain and distraction rate were chosen to maintain the same project design as Barrett in 2014 to allow comparison of results obtained between both studies. The strain rate used in this study is much lower than what is experienced by the SDFT naturally, which has been reported to be 2.2% to 4.68% at a walk.⁵⁶ In a previous study evaluating the effect of strain rate on intact chicken flexor digitorum profundus tendons, it was found that increased strain rate increased the overall stiffness and tensile strength of tendon prior to failure.¹⁷⁶ It was hypothesized that this was due to decreased interfragmentary shear, resulting in tissue deforming within a very short period prior to failure and load being distributed evenly among collagen fibers within the tendon.¹⁷⁶ A lower strain may have allowed uneven load distribution due to increased shear, decreasing the strength of the tendon.¹⁷⁶ This study, however, evaluated only intact tendon. In a study by Belvins in 1994, strain had no impact on the strength or stiffness of ACL or patellar tendon grafts in humans.¹⁷⁷ The distraction rate in this study is lower than that which has been used in some previous equine tenorrhaphy studies, ranging from 12-25mm/sec.^{7, 11} When comparing the ultimate strength at failure of a 3LP repair in this study to previous studies that have used a higher distraction rate, all found similar strengths for the 3LP repair.^{7, 11} Therefore, although a lower strain does not mimic natural conditions, it may not have a great impact on testing repair strength.

Although changing the implant pattern as well as number of TF implants increased the overall strength of the TF repair compared to results obtained from a previous study, the device was still not as strong as the 3LP in ultimate load before failure and load to a 2mm gap. It is therefore speculated that all repairs tested would have to be supported by prolonged external coaptation. However, even with additional support, these repairs may still be at high risk of significant gap formation, construct failure, longer healing time and poor quality repair tissue. Based on these findings, use of the Teno Fix® device in the repair patterns described in this study may not be ideal for equine flexor tendon tenorrhaphy. Increased strength of TF repairs in this study compared to a previous study could in part be due to increased core purchase length of the repairs. Previous evaluation of core purchase length using porcine models found that increased core purchase length increased the strength of repair, most likely due to increasing tendon-suture interaction. Core purchase length has not been investigated as a contributing factor to increasing repair strength for equine flexor tendon tenorrhaphy. Further investigation of this repair characteristic is warranted to aid in developing an optimal repair method for equine flexor tendon lacerations.

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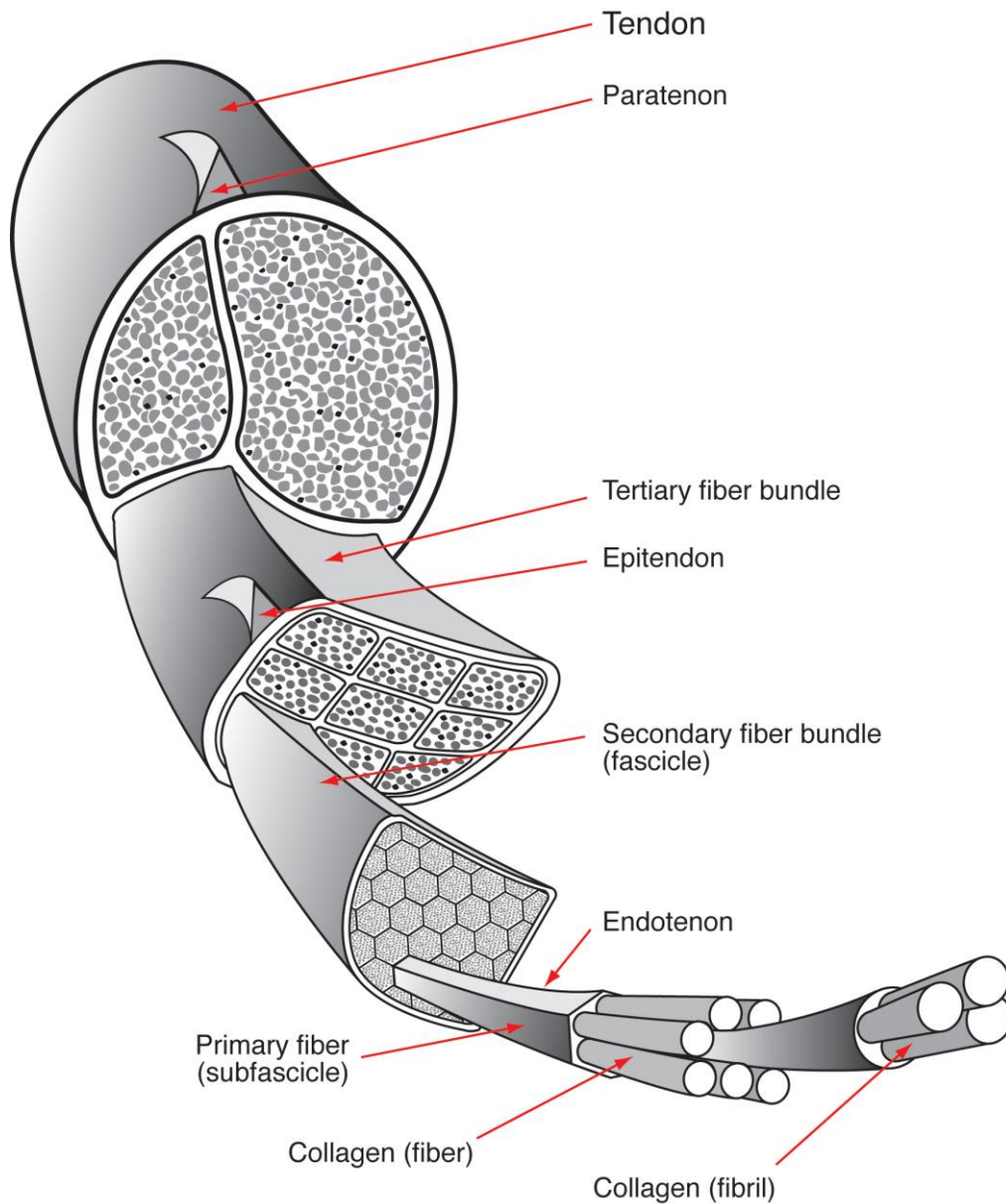
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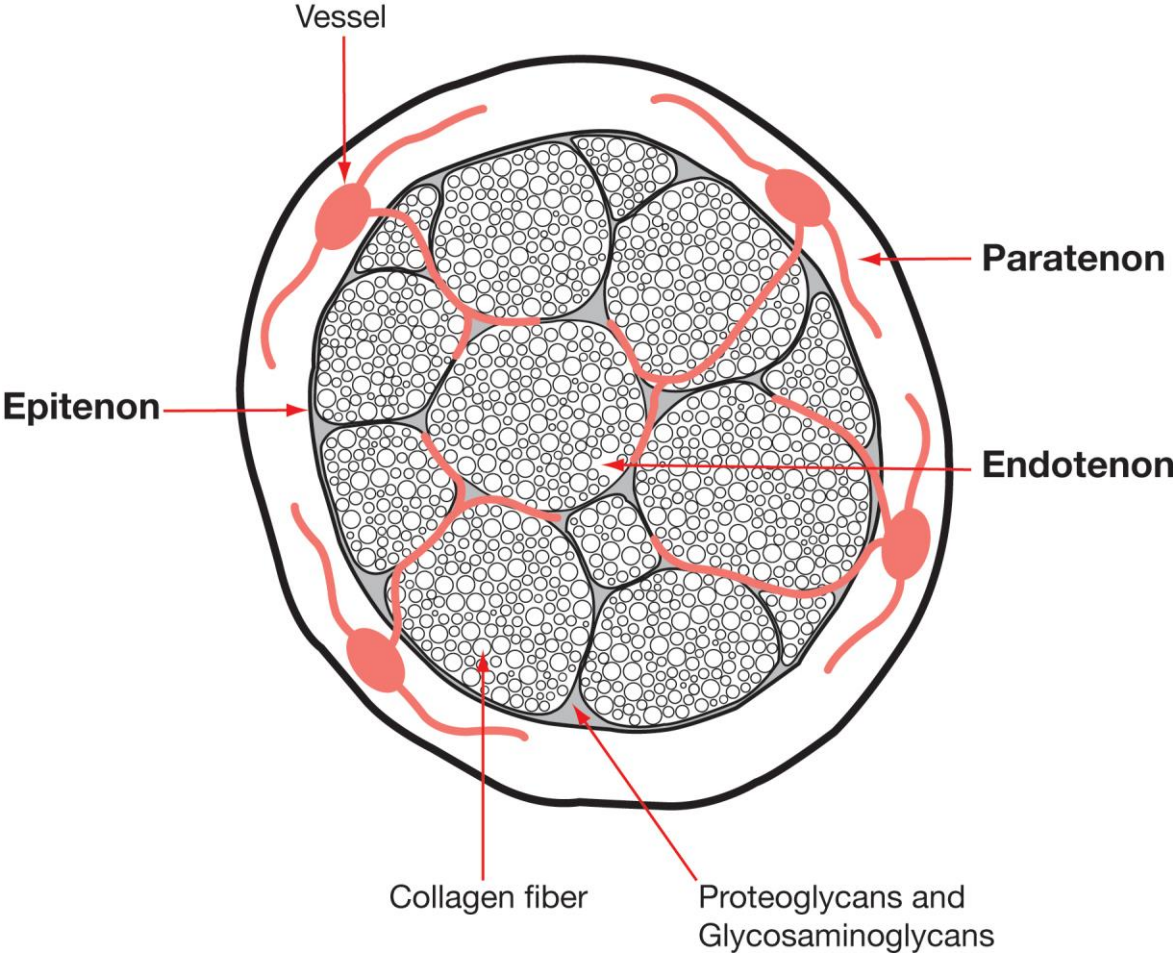
Appendix 1: Figures

Figure 1: Hierarchical structure of a tendon. Original illustration by J. McMaster.



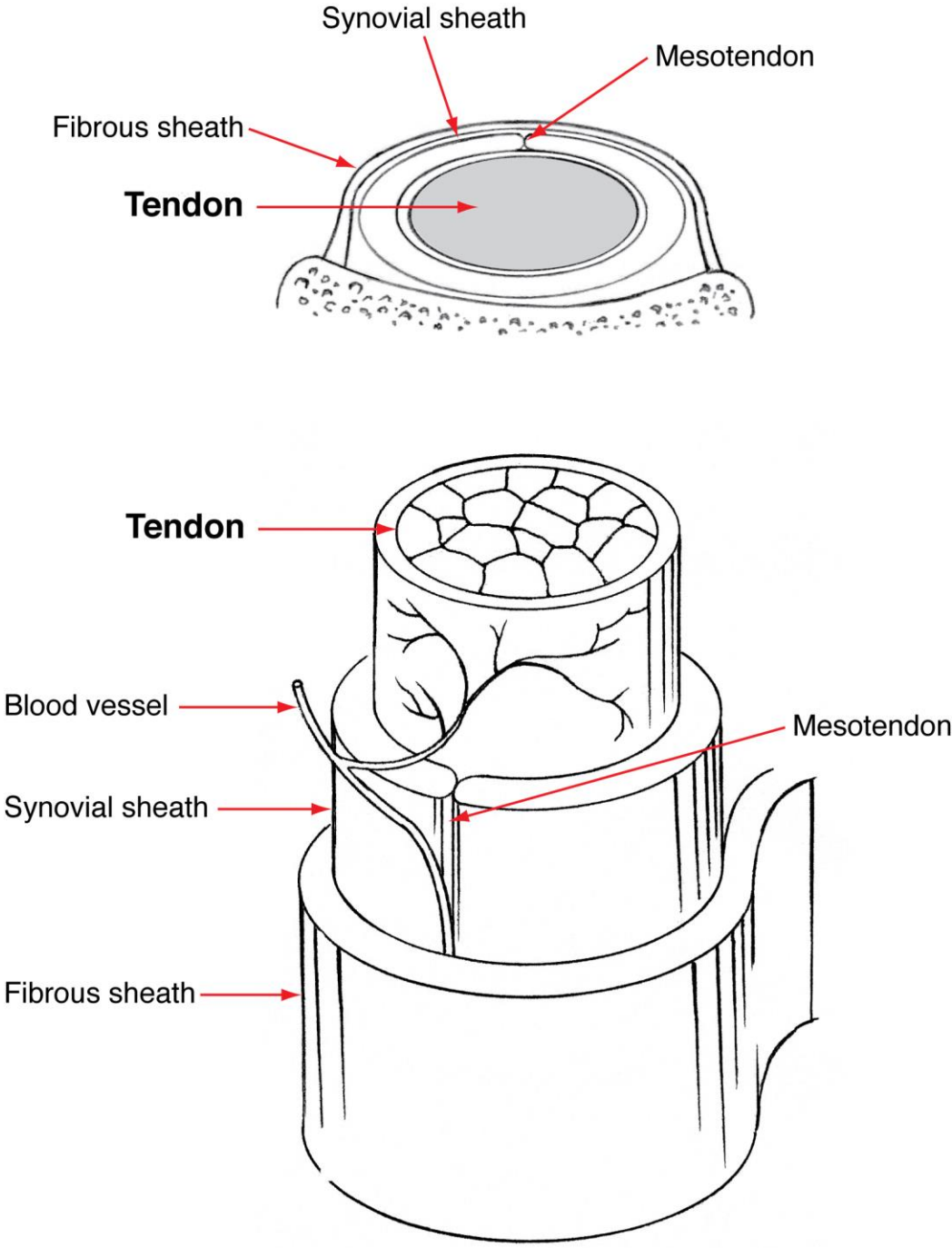
J McMaster

Figure 2: Schematic of the epitenon, endotenon and paratenon of a tendon. Original illustration by J. McMaster



J McMaster

Figure 3: Schematic of a tendon within a tendon sheath. Original illustration by J. McMaster



J McMaster

Figure 4: Histologic appearance of the normal “crimp” pattern of tendon (Yasuda 2004).¹⁷⁸ Used under Fair Use guidelines

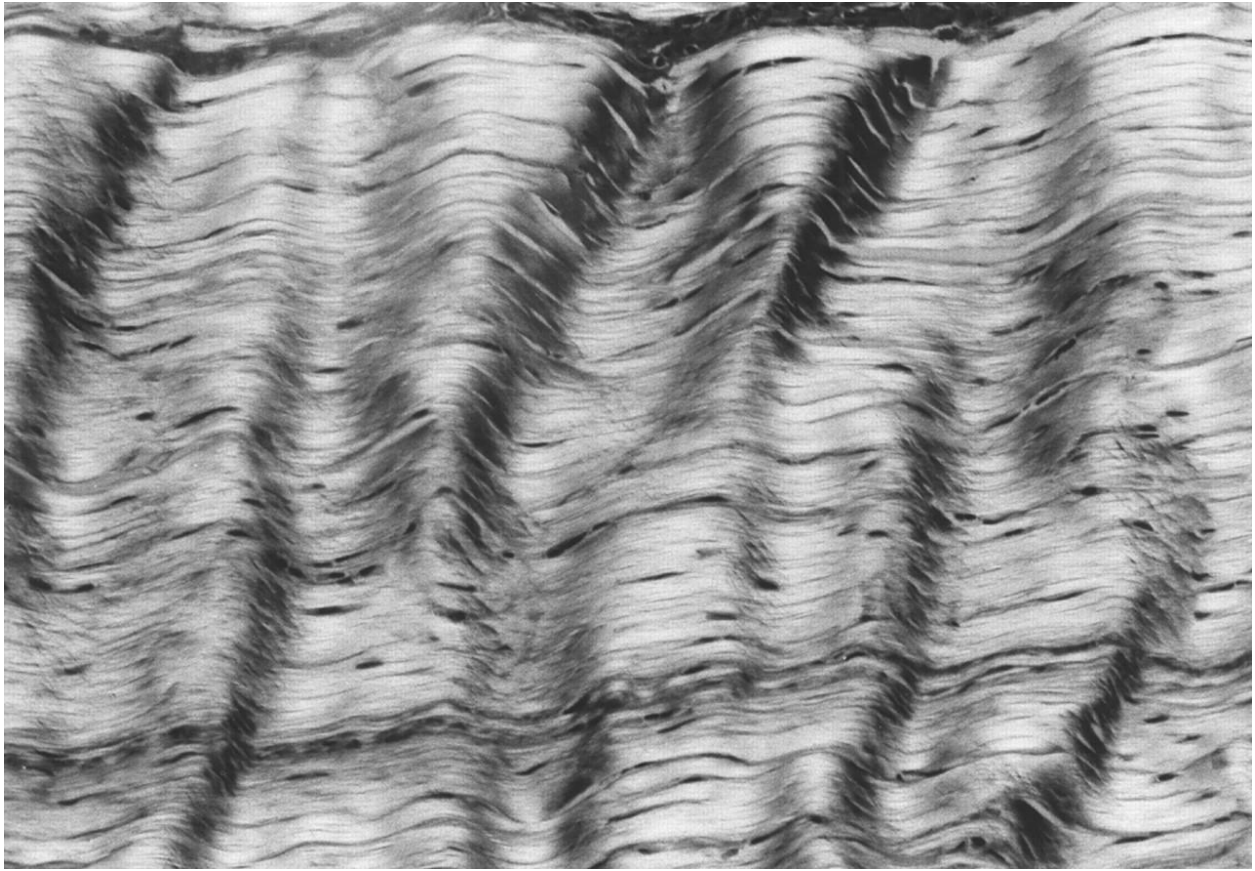


Figure 5: Schematic of the muscles and tendons of the forelimb of the horse. Image provided by R. Wilhite

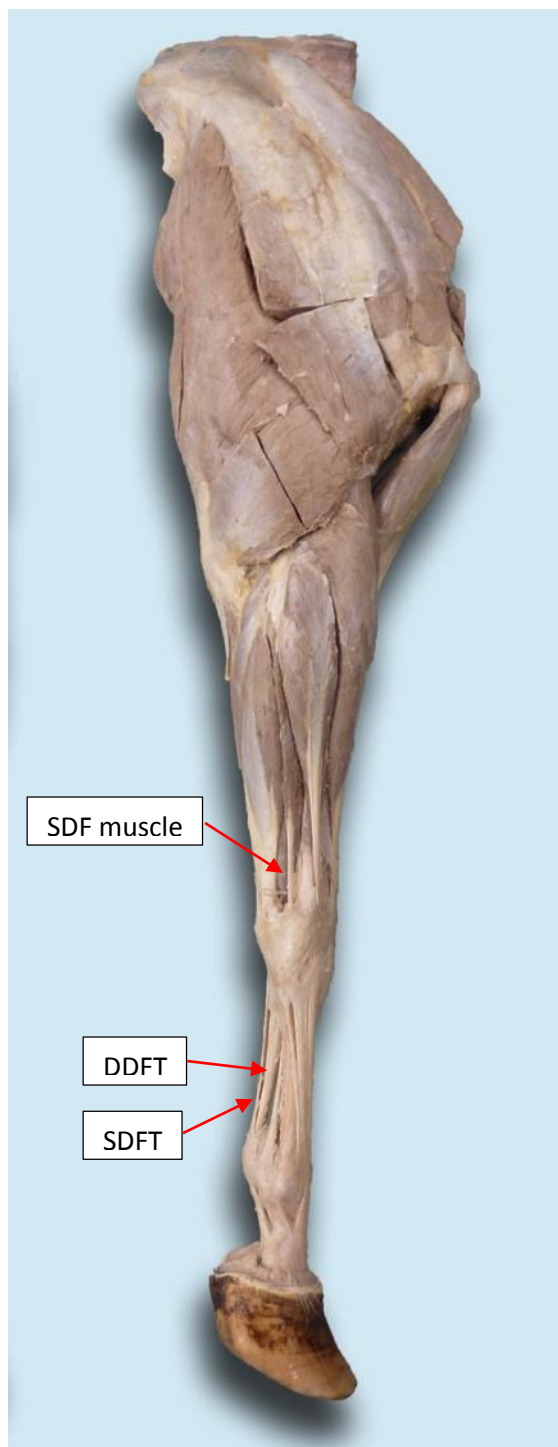


Figure 6: Schematic of the SDFT and DDFT just proximal to the fetlock joint where the SDFT forms the manica flexoria. Image provided by R. Wilhite.

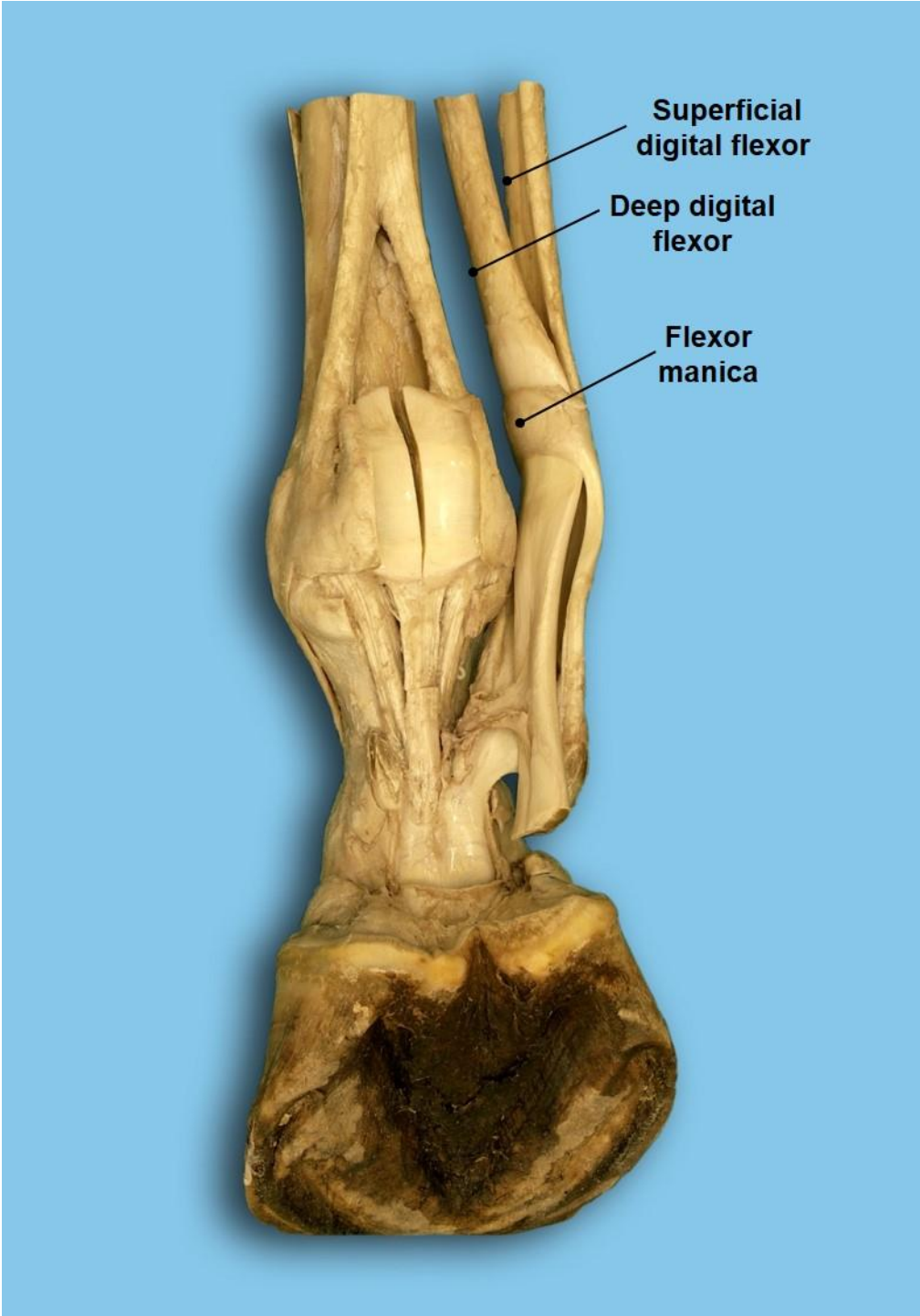


Figure 7: Schematic of the muscles and tendons of the hindlimb of the horse. Image provided by R. Wilhite.



Figure 8: Schematic of the SDFT and DDFT as they course distally on the hindlimb. Image provided by R. Wilhite.

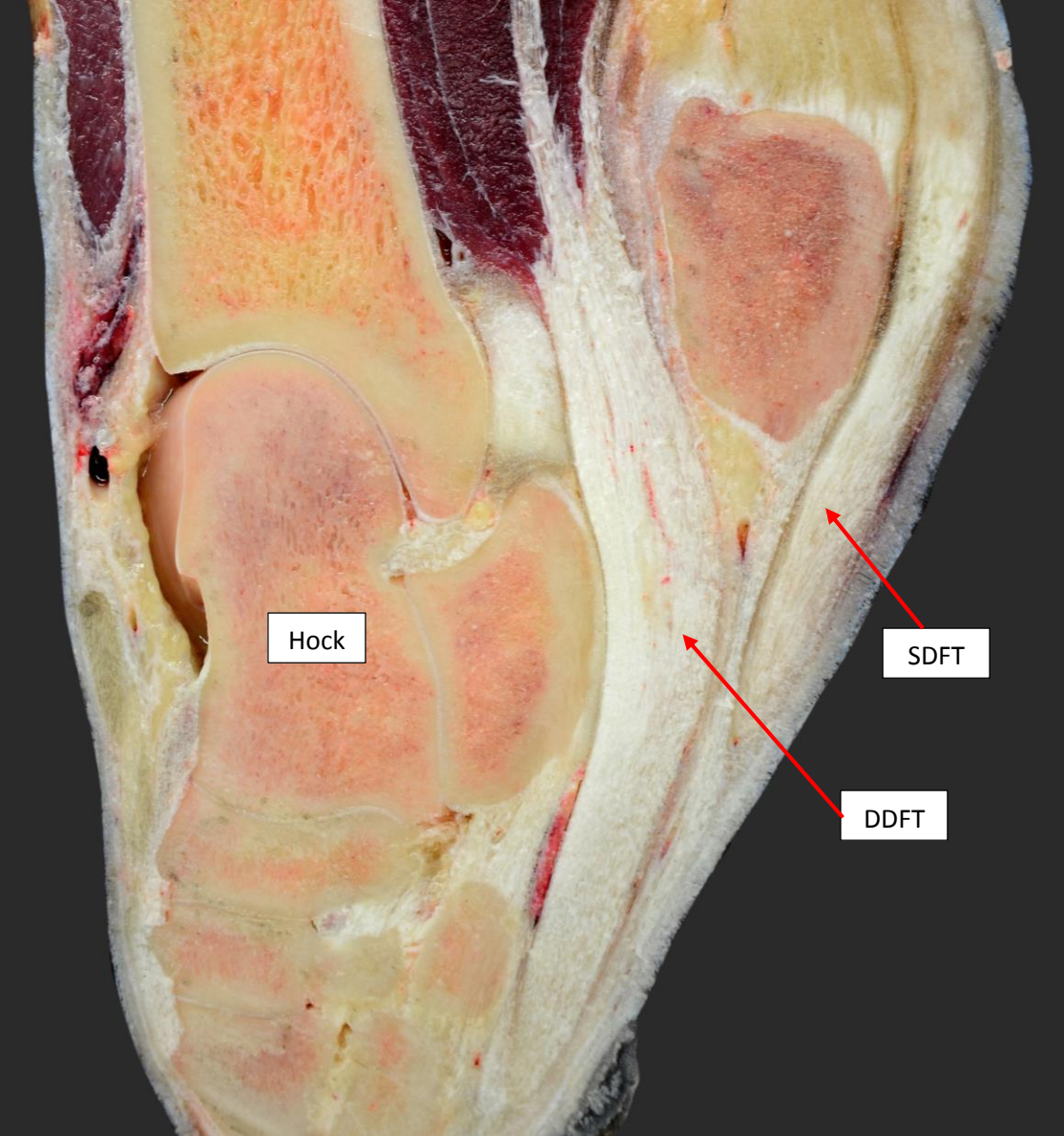


Figure 9: Schematic illustrating the location of the SDFT and DDFT of the equine limb (a), as well as the spring-like function of the SDFT during weight-bearing (b) and the positional action of the common digital extensor tendon during the flight phase of stride (c).

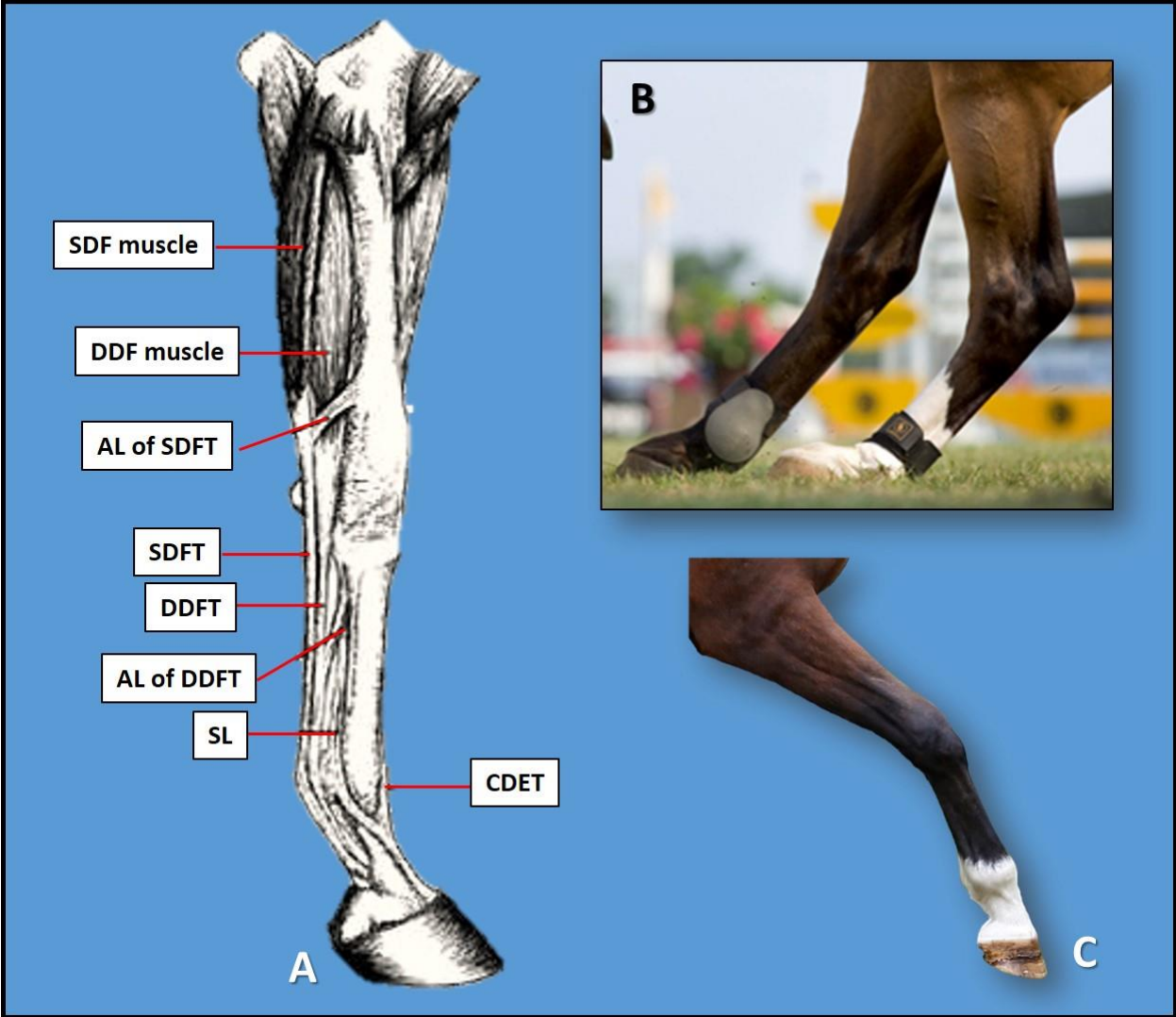


Figure 10: Example of a load-elongation curve of a tendon.

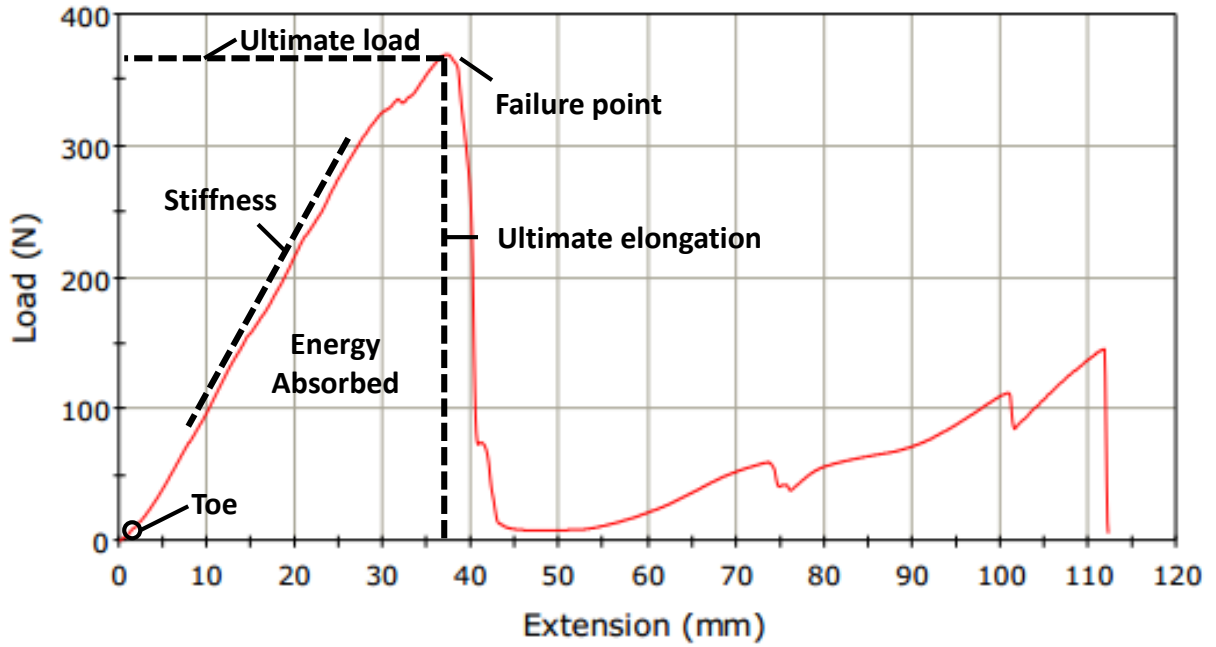


Figure 11: Example of a stress-strain curve of a tendon.

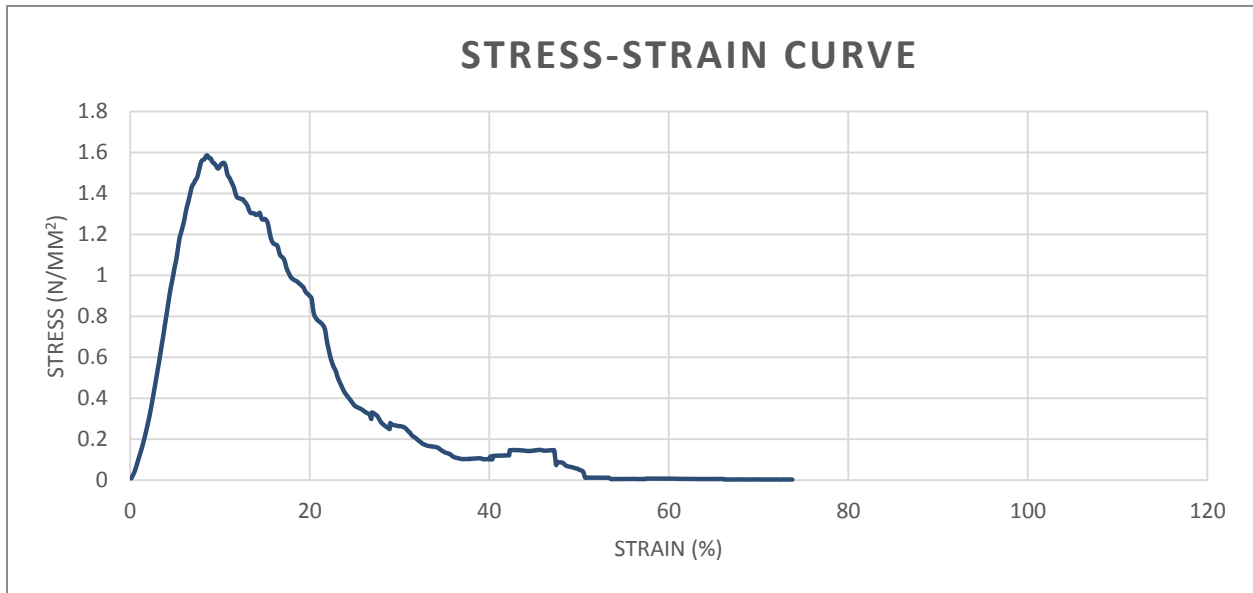


Figure 12: Example of a laceration involving the SDFT and a portion of the DDFT. Image provided by R. Hanson.

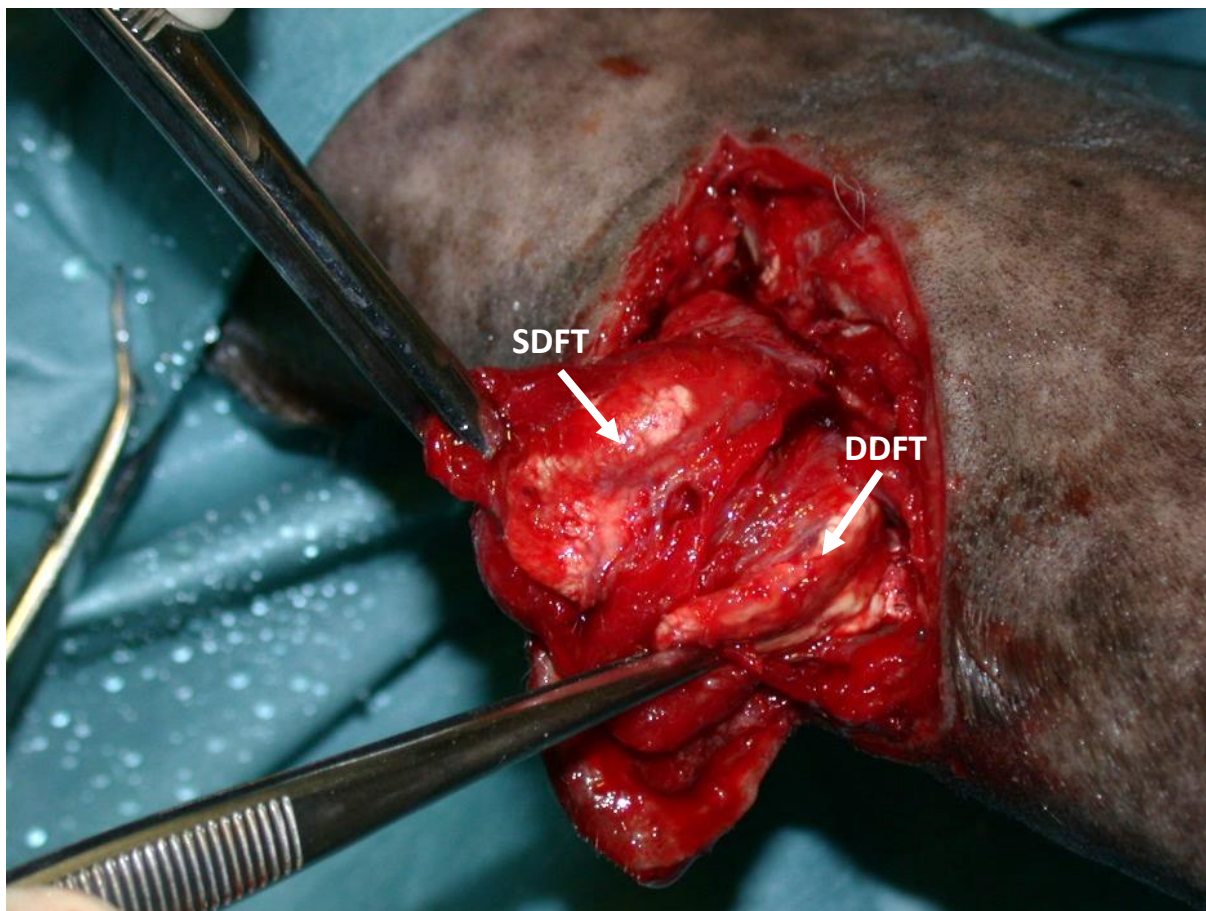


Figure 13: Characteristic limb position following transection of the SDFT and DDFT in a horse.
Image provided by R. Hanson.



Figure 14: Regional limb perfusion performed on a distal limb with a flexor tendon laceration.
Image provided by R. Hanson.

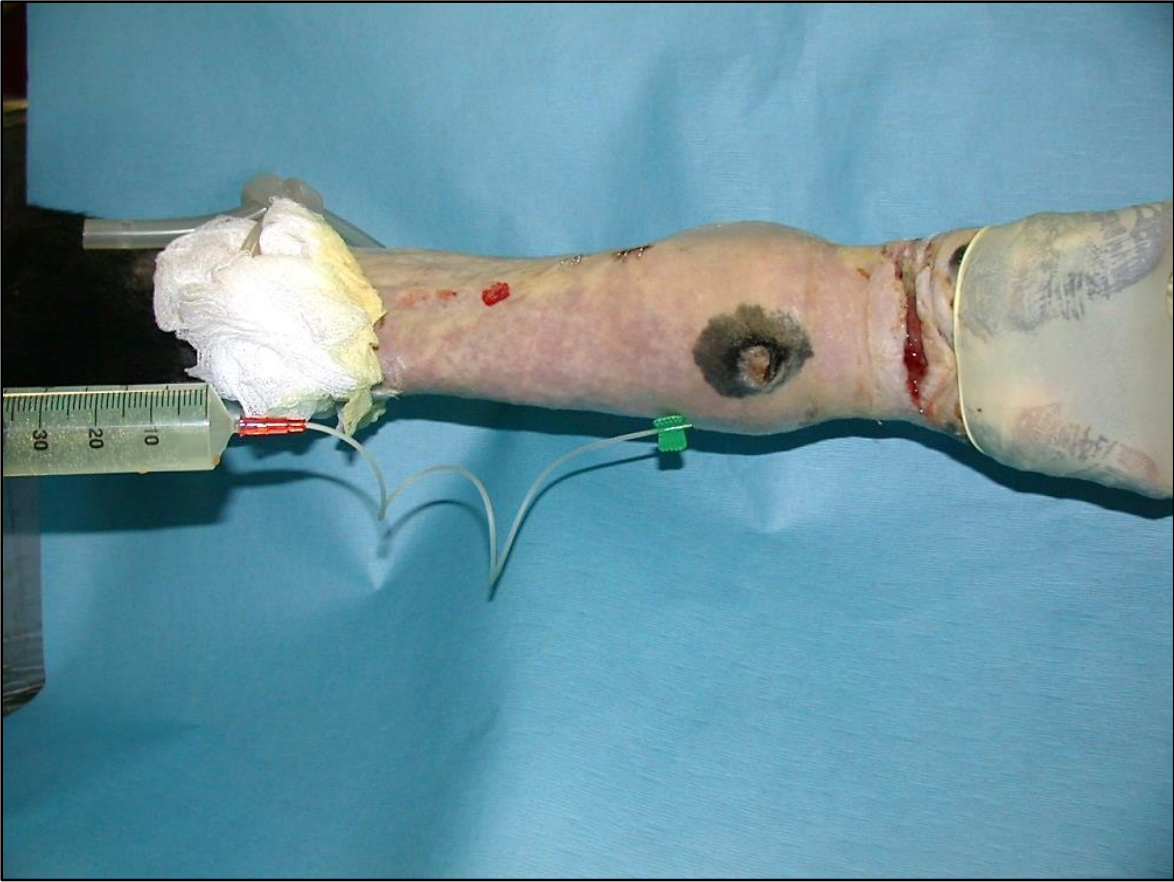


Figure 15: Kimzey Leg Saver splint for the forelimb.



Figure 16: Mechanical wound debridement using lavage. Image provided by R. Hanson.

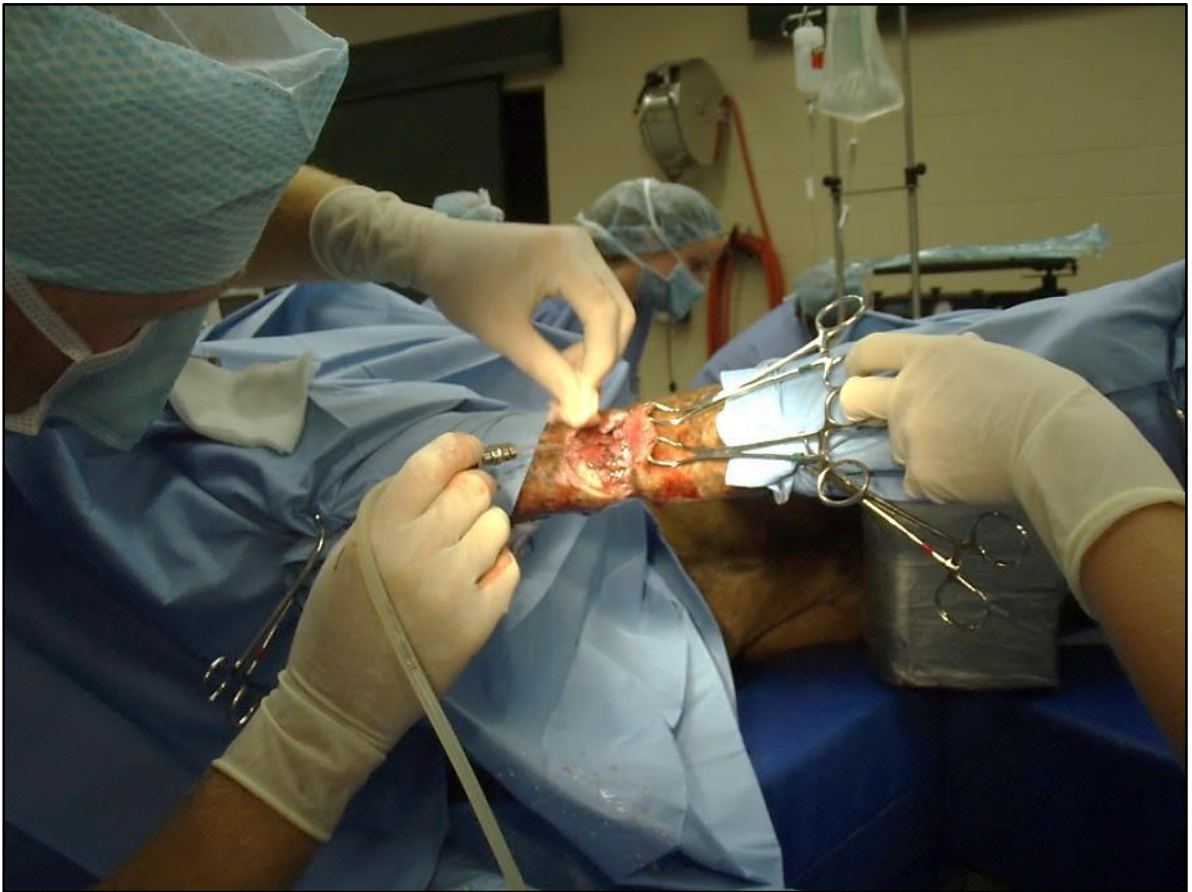


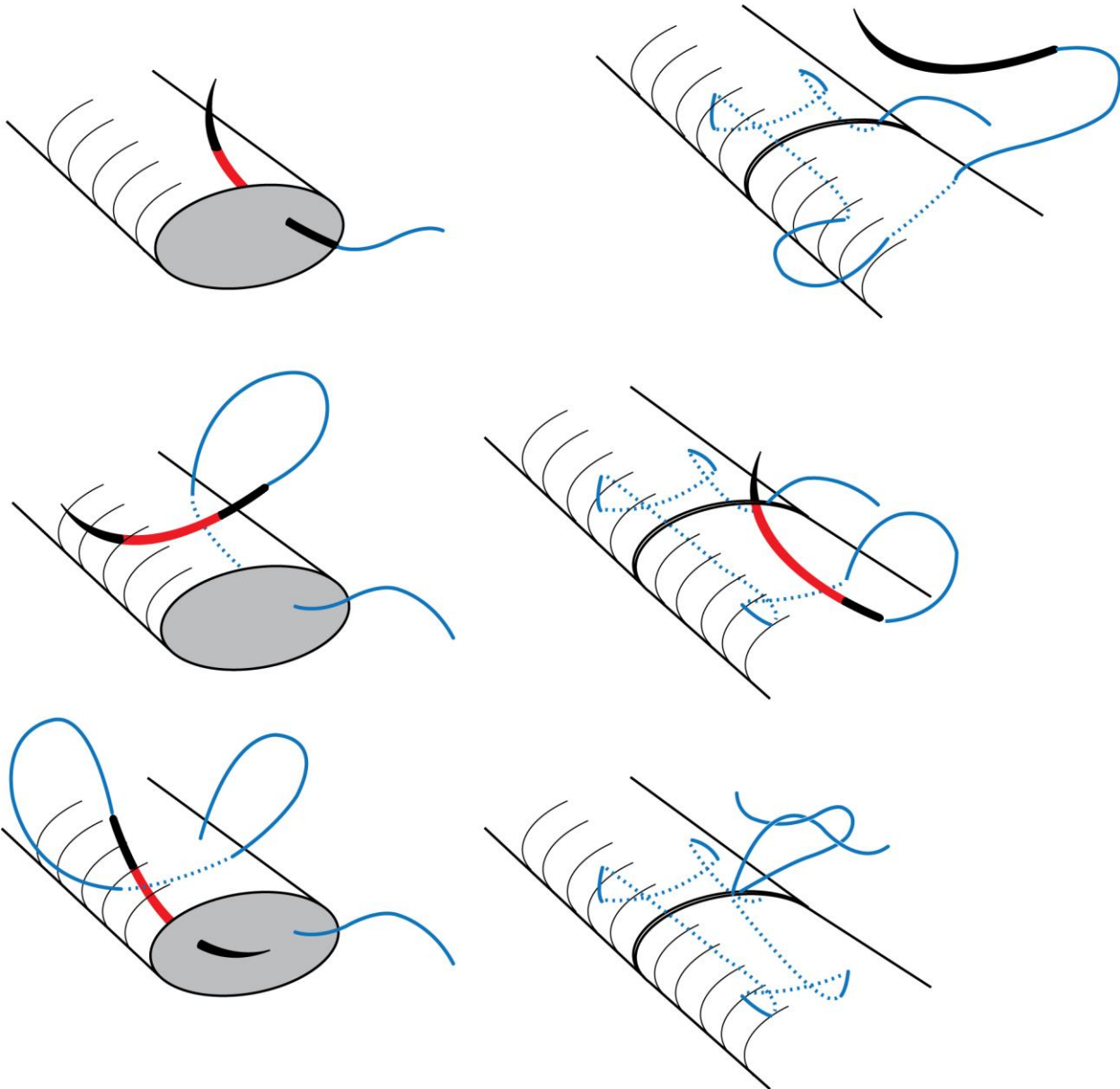
Figure 17: Extended heel wedge shoe. Image provided by R. Hanson.



Figure 18: Fetlock support brace as described by Whitfield-Cargile et al. in the image on the left (Whitfield-Cargile 2011). Figure used under Fair Use guidelines. A variation of the fetlock support brace without a dorsal splint to treat a tendon laceration in a hindlimb can be seen in the image on the right. Image on the right provided by R. Hanson.



Figure 19: Locking-loop suture pattern. Original illustration by J. McMaster.



J McMaster

Figure 20: Three Loop Pulley (3LP) repair pattern. Original illustration by J. McMaster.

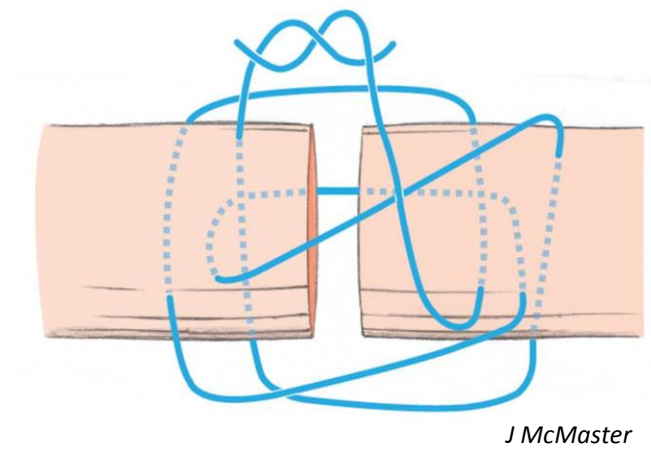


Figure 21: Soft tissue swelling of the distal limb at eight weeks post-repair (carbon fiber repair in the image on the left and the nylon repair in the image on the right) (Nixon 1984). Figure used under Fair Use guidelines.

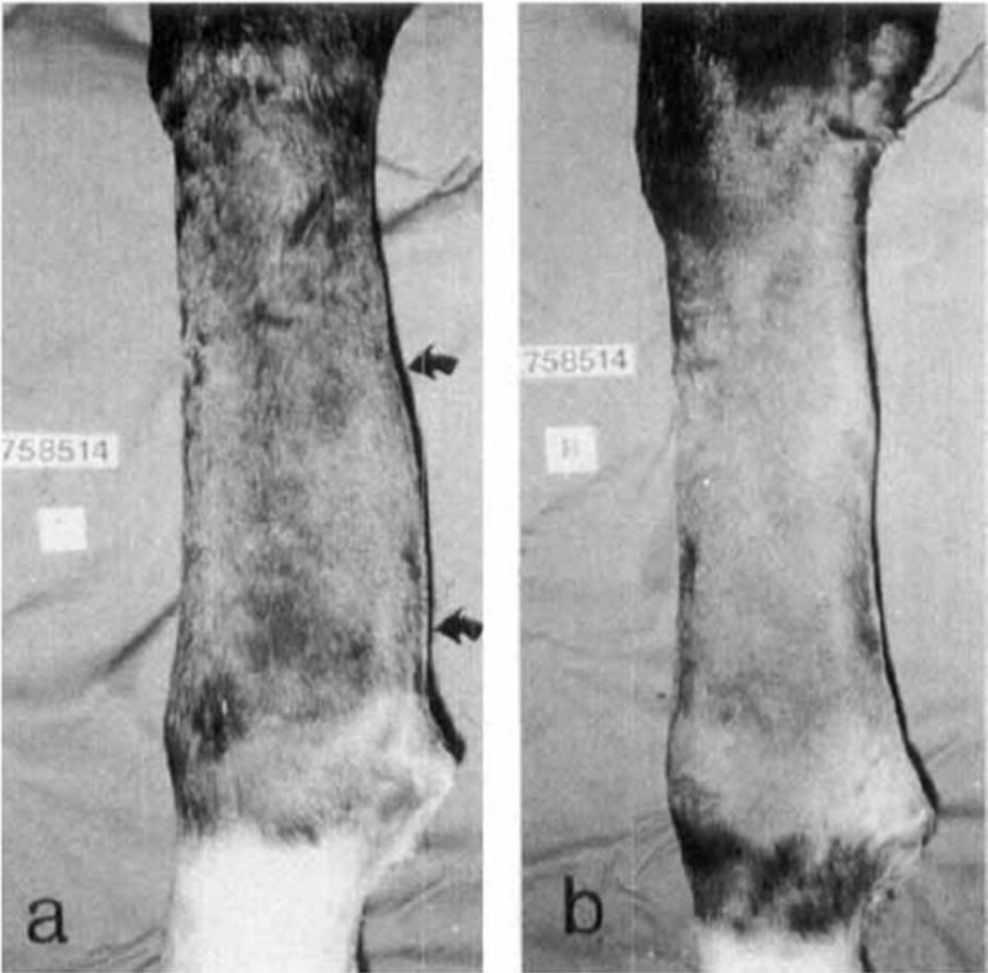
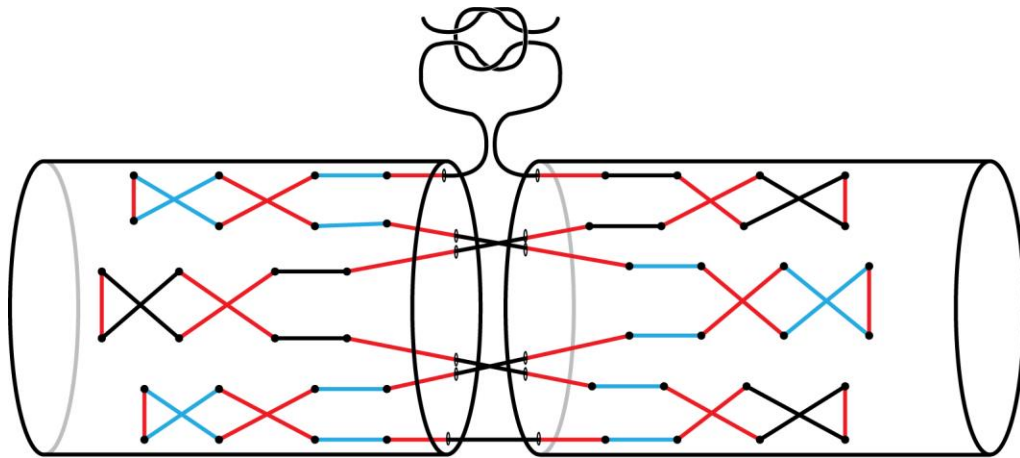


Figure 22: Six-Strand Savage repair. Original illustration by J. McMaster.



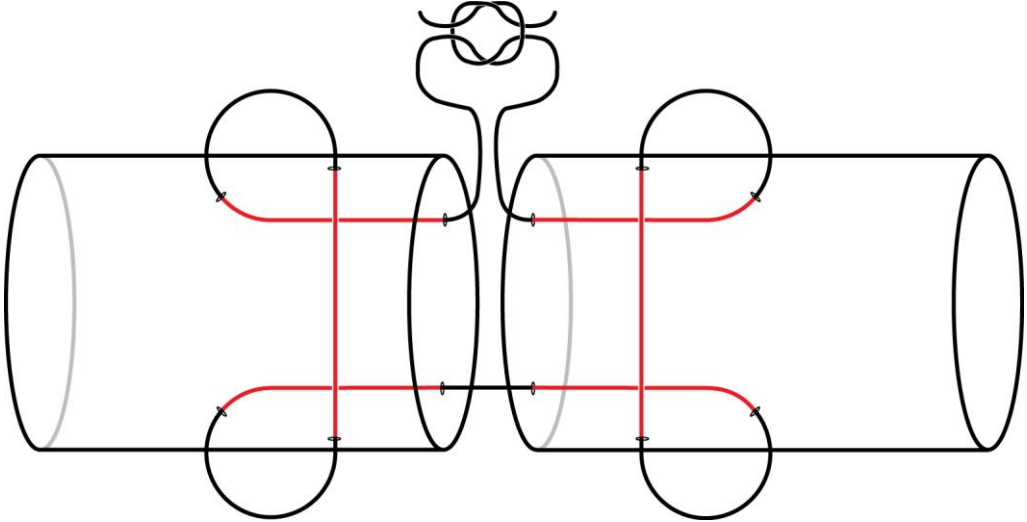
— Outside tendon, front

— Within central tendon

— Outside tendon, back

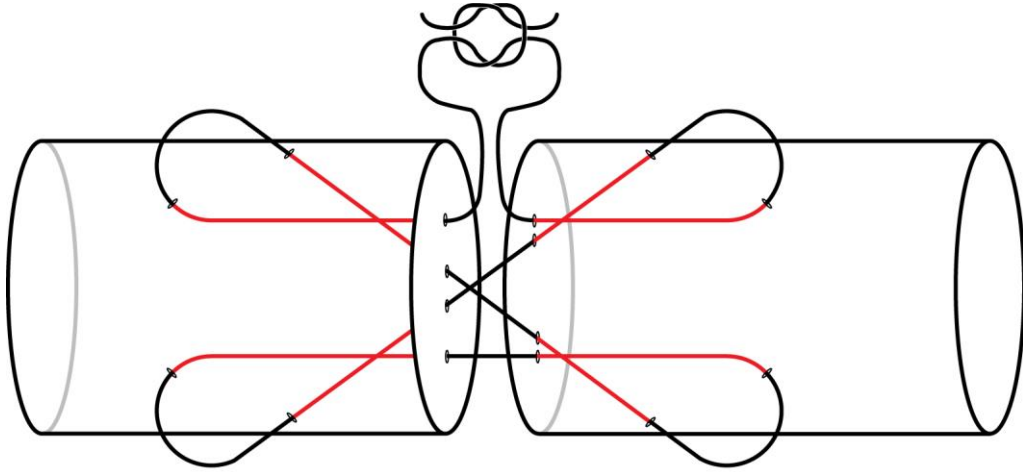
J McMaster

Figure 23: Modified Kessler suture repair. Original illustration by J. McMaster.



J McMaster

Figure 24: Cruciate suture repair. Original illustration by J. McMaster.



J McMaster

Figure 25: Tsuge suture technique (Angeles 2002). Figure used under Fair Use guidelines.

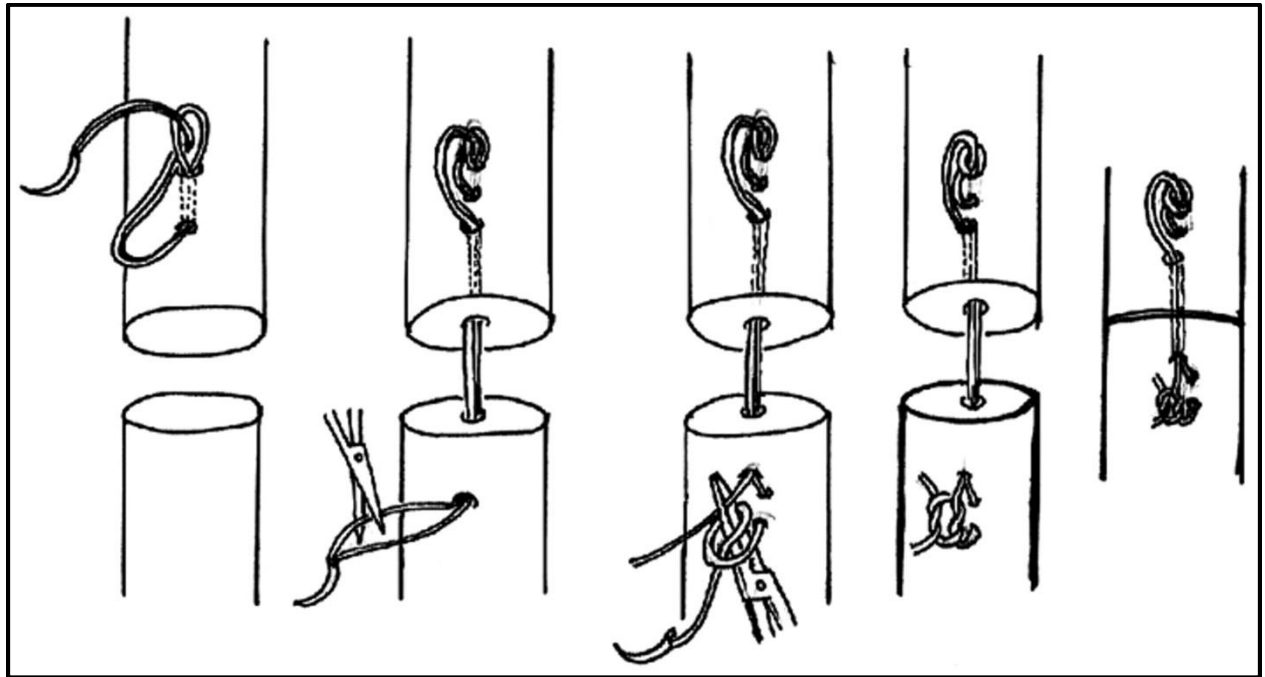


Figure 26: Intratendinous anchor developed by Gordon et al. (Gordon 1998). Figure used under Fair Use guidelines.

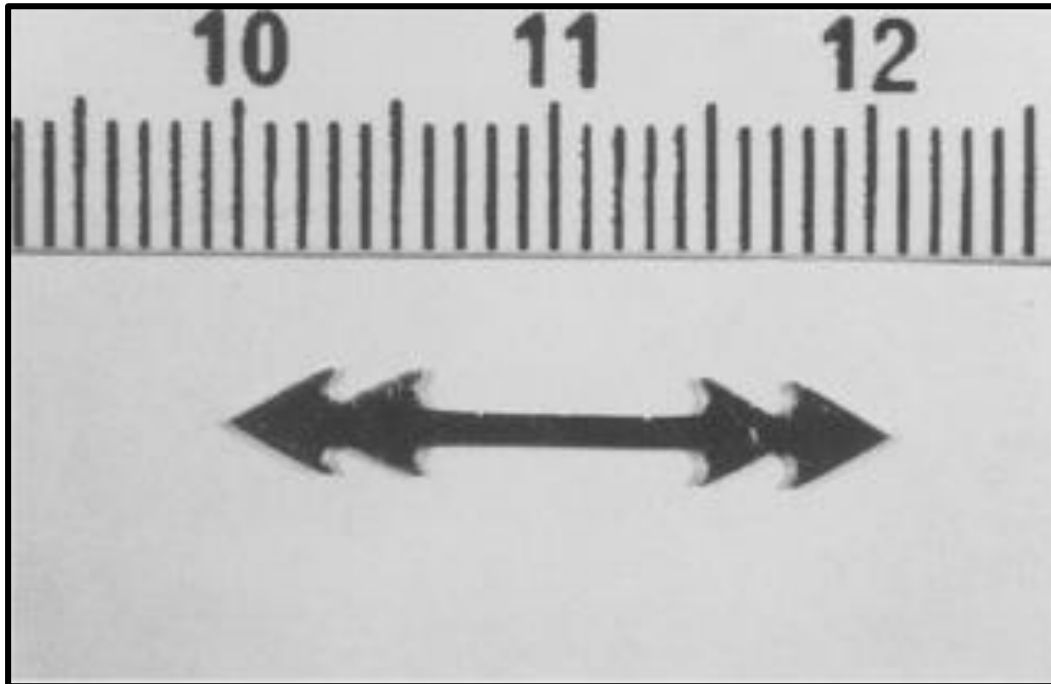


Figure 27: Stainless steel anchors designed by Erol et al. Model 1 can be seen in the image on the left and Model 2 can be seen in the image on the right (Erol 2007). Figure used under Fair Use guidelines.

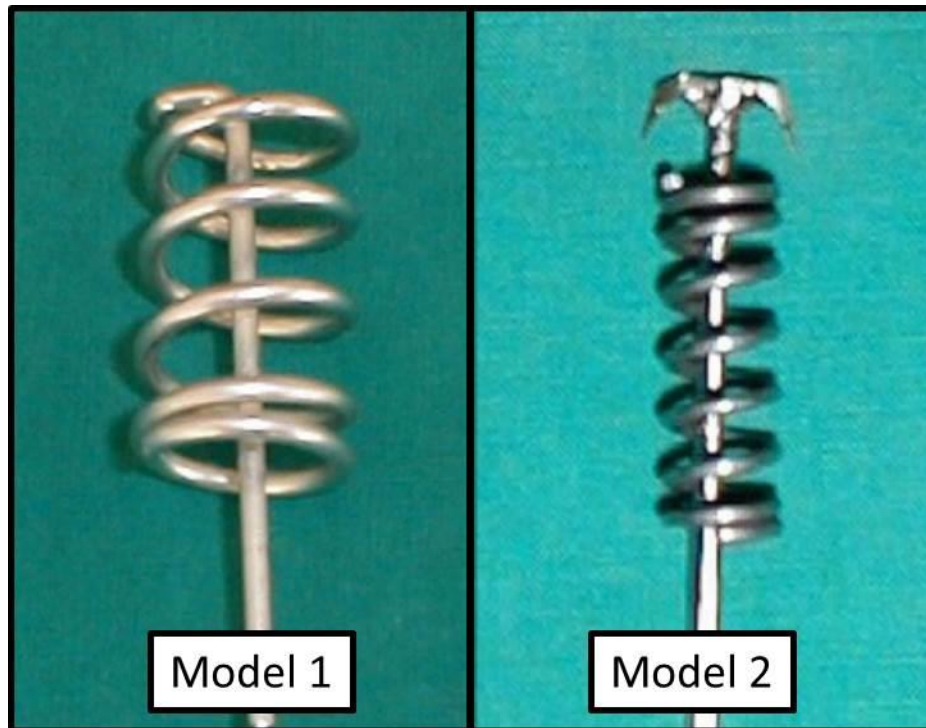


Figure 28: Nitinol barbed implant developed by Hirpara et al. (Hirpara 2010) Figure used under Fair Use guidelines.

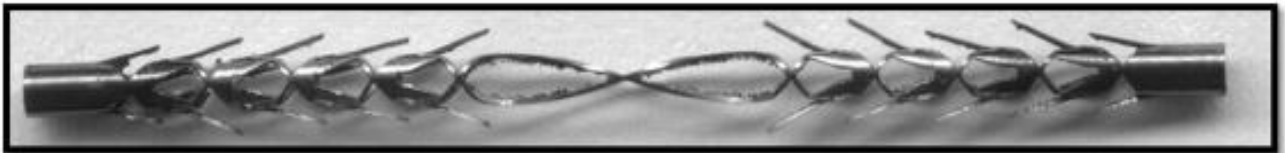


Figure 29: Assembled Teno Fix® implant (Su 2005). Figure used under Fair Use guidelines.

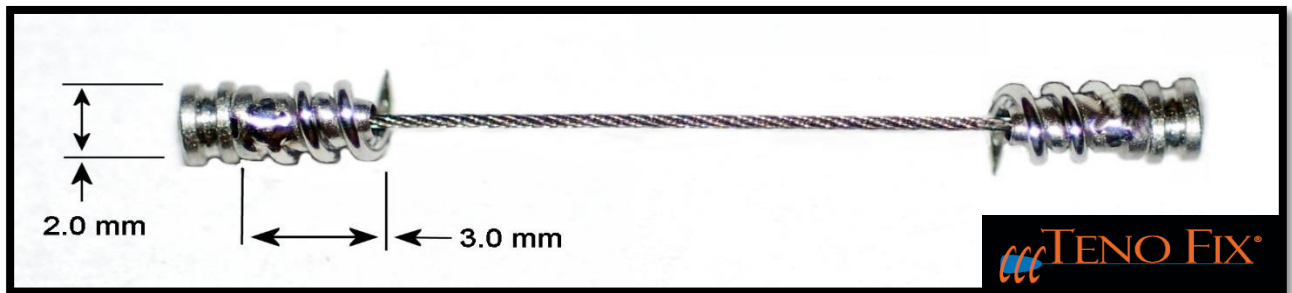


Figure 30: Placement of a stainless steel Teno Fix® anchor in a tendon using a Teno Fix® obturator (Su 2006). Figure used under Fair Use guidelines.

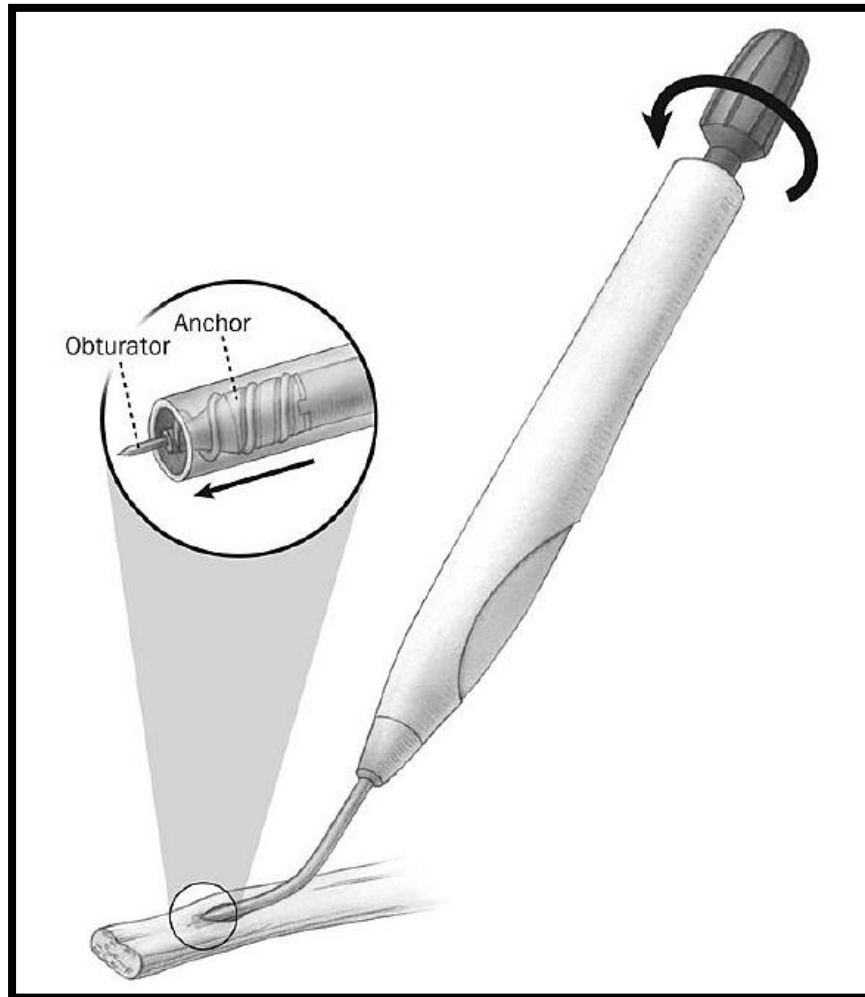


Figure 31: Threading stainless steel suture through anchors placed in tendon ends (Su 2006).
Figure used under Fair Use guidelines.

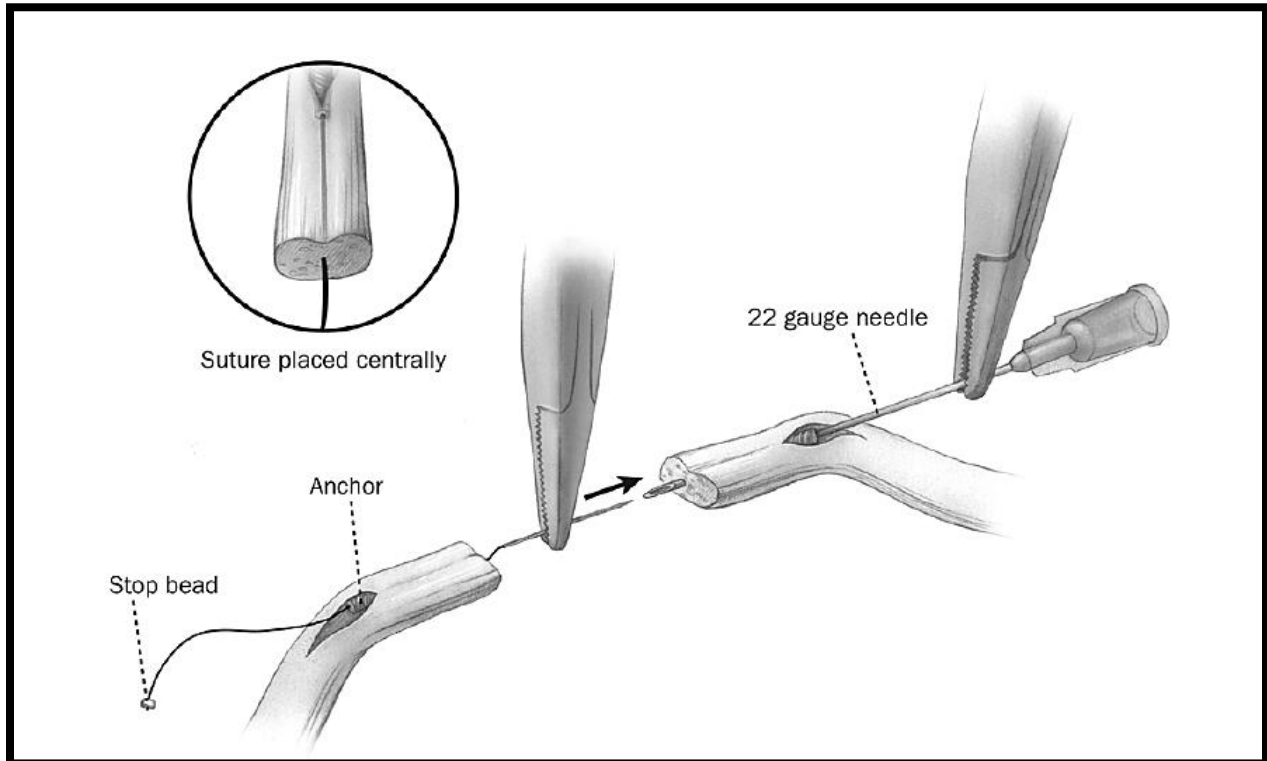


Figure 32: Placement and subsequent crimping of a stop bead to secure the Teno Fix® device in a tendon (Su 2006). Figure used under Fair Use guidelines.

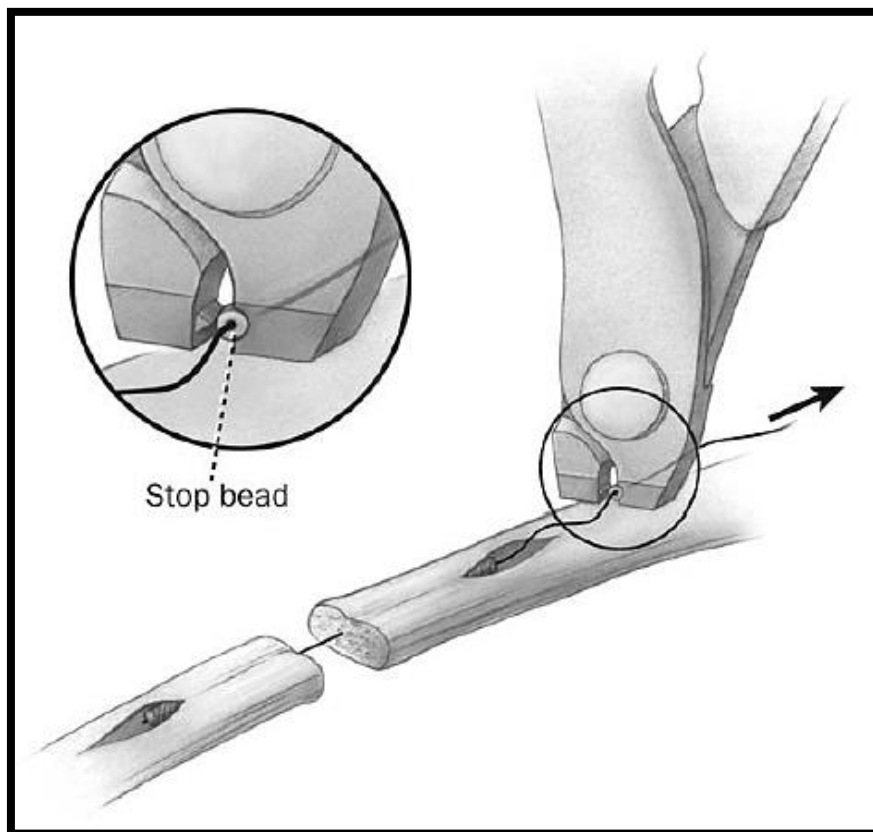
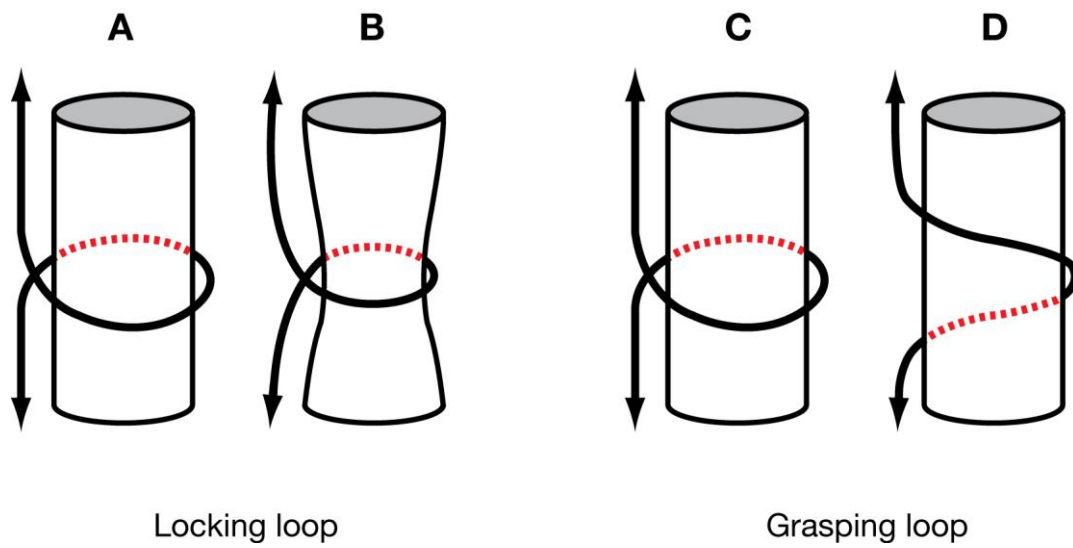


Figure 33: Schematic of a locking loop (A and B) versus a grasping loop (C and D). Original illustration by J. McMaster.



J McMaster

Figure 34: Schematic of suture configurations for grasping and locking loops (Hotokezaka 1997). Figure used under Fair Use guidelines.

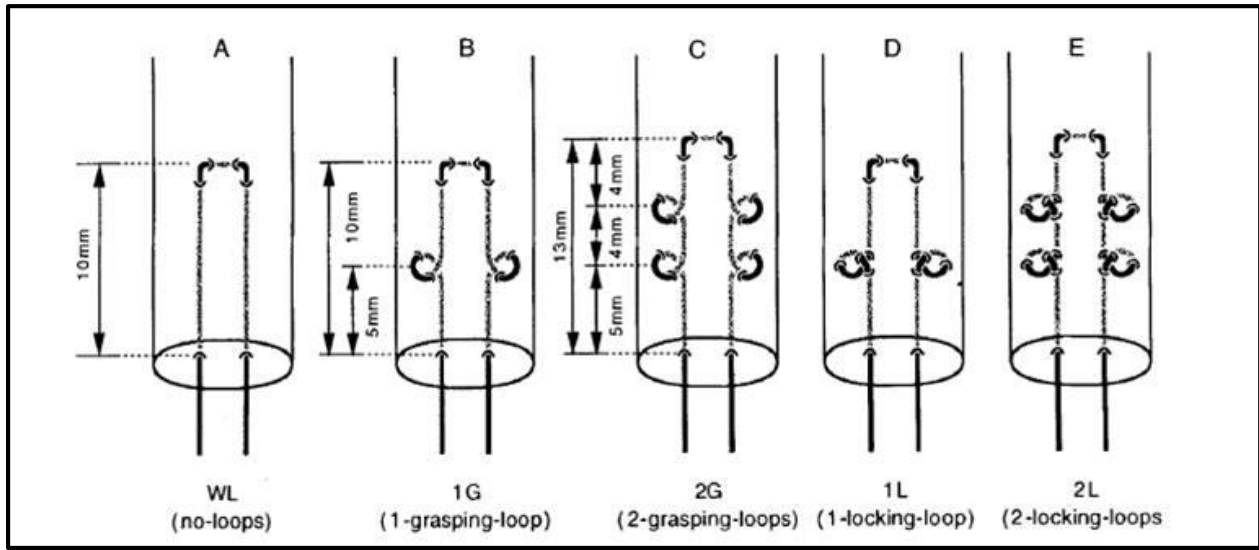


Figure 35: Schematic showing the passage of suture through a cut end of a tendon to form a locking loop. This schematic represents a suture bite incorporating 10% of the CSA of the cut portion of tendon (Dona 2004). Figure used under Fair Use guidelines.

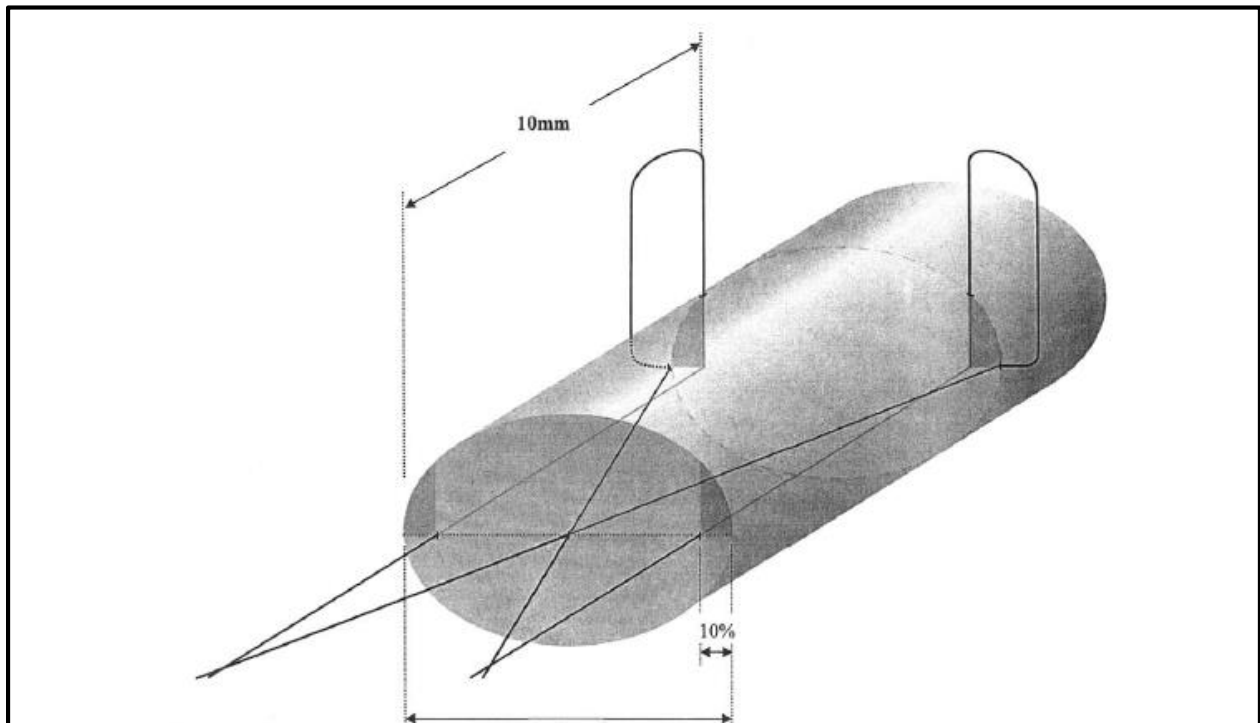


Figure 36: Schematic illustrating placement of a locking loop perpendicular to the longitudinal axis of a tendon (A) versus placement of a locking loop parallel to the longitudinal axis of a tendon (B) (Tan 2004). Figure used under Fair Use guidelines.

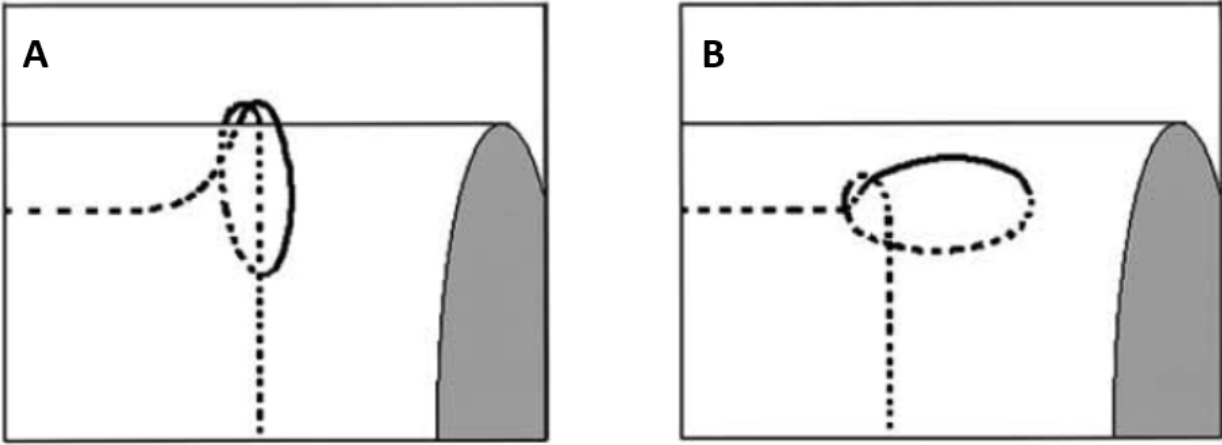
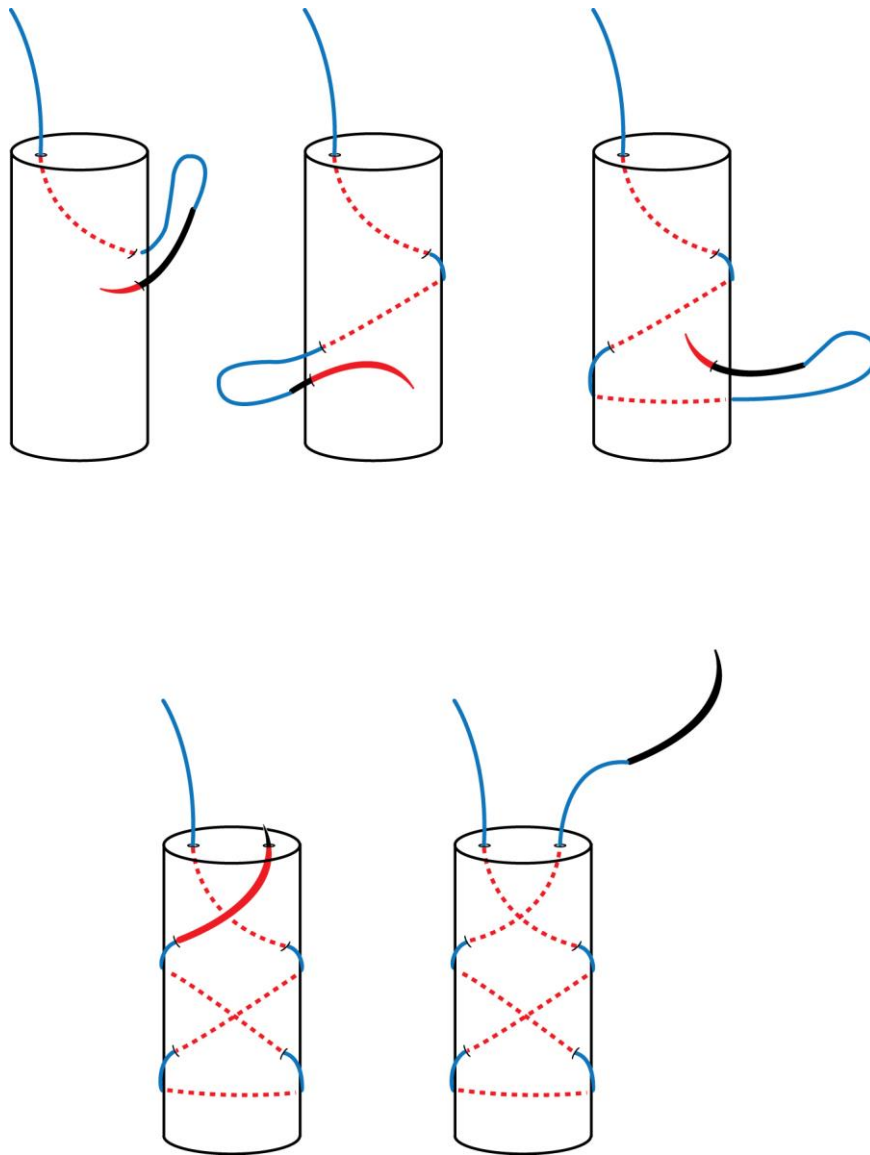


Figure 37: Placement of a Bunnell suture. Original illustration by J. McMaster.

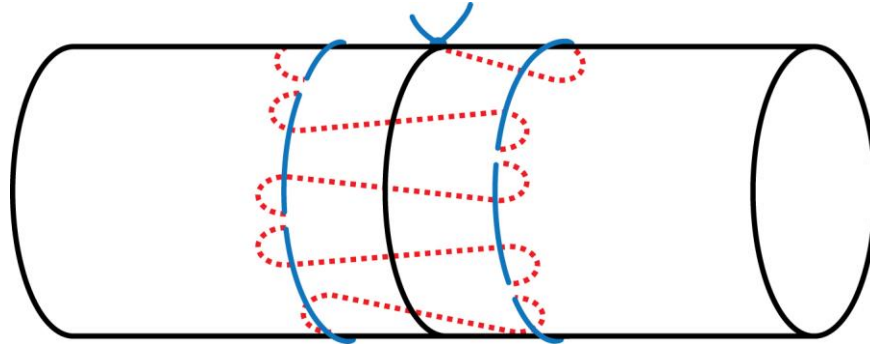


J McMaster

Figure 38: Bioabsorbable Poly-L-Lactic acid tendon plate (Jenson 2005). Figure used under Fair Use guidelines.



Figure 39: Lin-locking peripheral suture. Original illustration by J. McMaster.



J McMaster

Figure 40: Teno Fix[®] device pattern used by Barrett et al (Barrett 2014). Figure used under Fair Use guidelines.

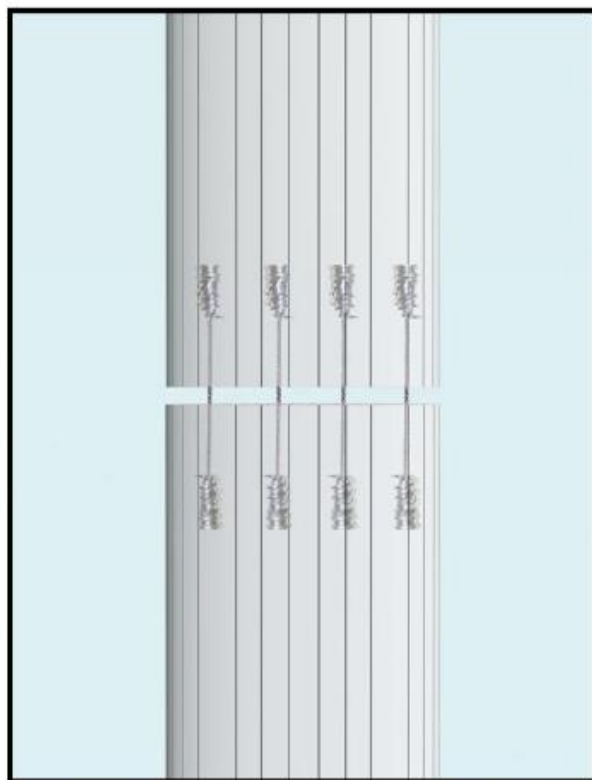
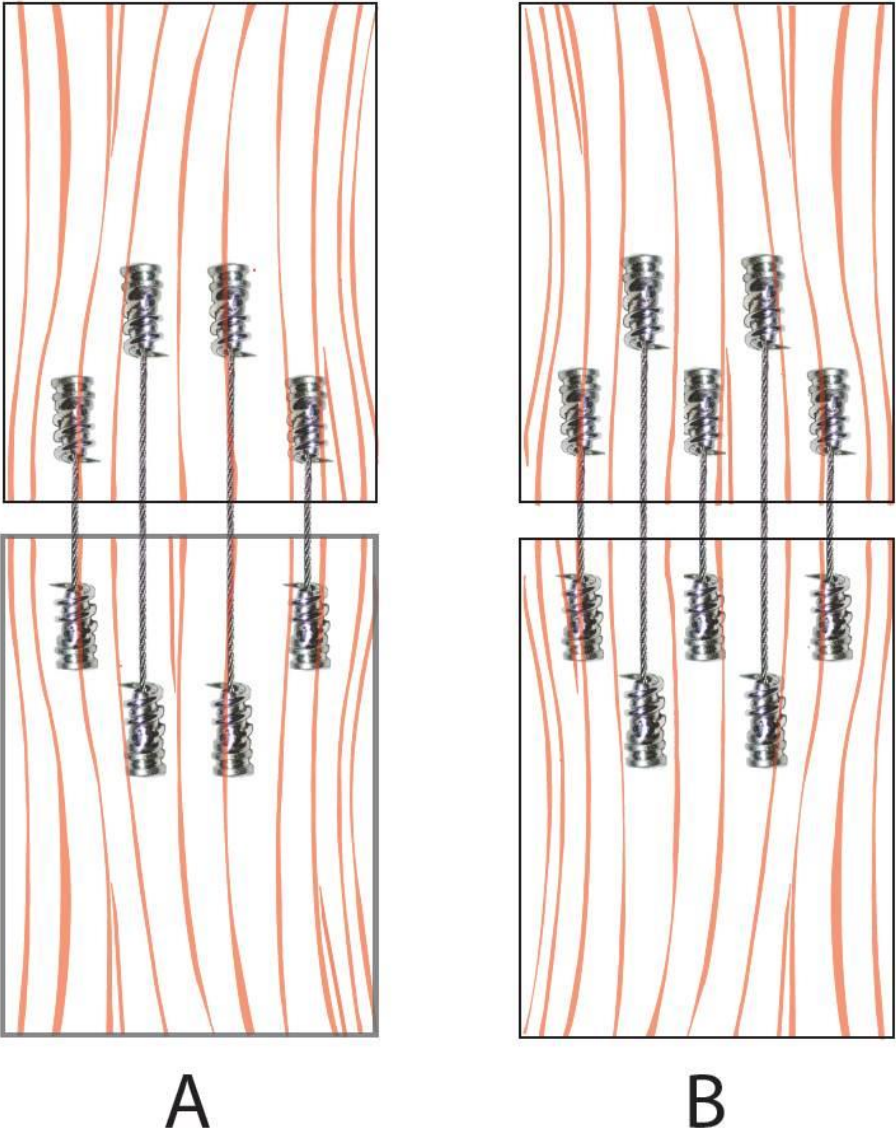


Figure 41: TF repair patterns used in this study; 4TF can be seen on the left and 5TF can be seen on the right. Original illustration by J. McMaster.



J McMaster

Figure 42: Custom-designed stainless steel clamps for static tensile testing of repairs. Image provided by R. Hanson.

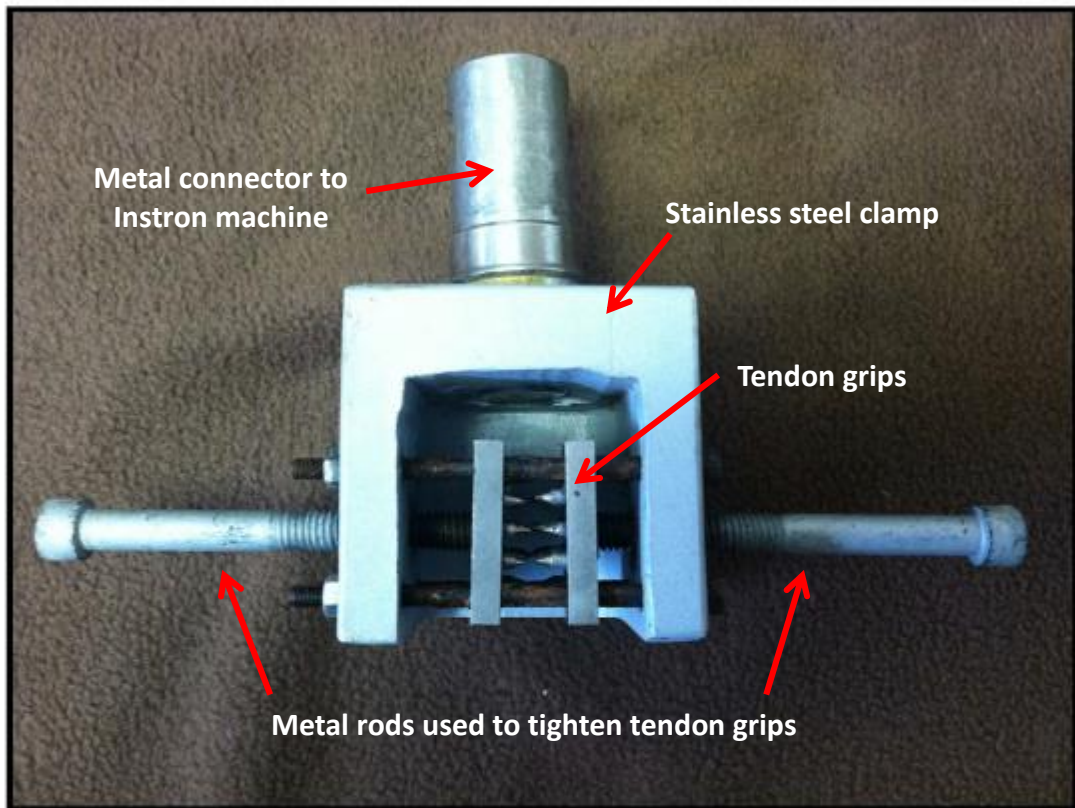
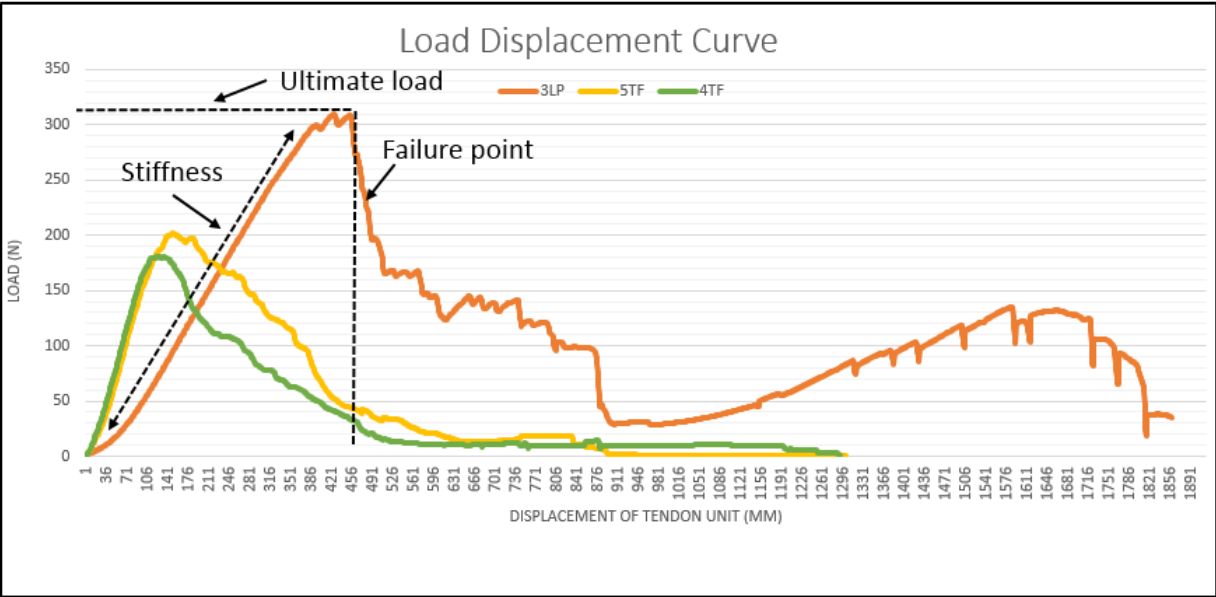


Figure 43: Load displacement curve of the 3LP (Orange), 4TF (Green) and 5TF (Yellow) of this study.



Appendix 2: Tables

Table 1: Prognosis of horses that sustained flexor tendon lacerations as reported by Foland et al., Taylor et al. and Jordana et al.^{2, 4, 167}

Study	Number of Horses	Percent alive one year following injury	Percent return to previous level of activity	Percent return to reduced level of activity/pasture or breeding sound
Foland et al. 1991	35	82%	18%	82%
Taylor et al. 1995	90	78%	54%	46%
Jordana et al. 2011	106	82%	55%	45%

Table 2: Failure of Tendon Repairs. Mean (95% confidence interval) ultimate load to failure (N), load to a 2mm gap, gap at failure and stiffness for tendons repaired with a 3LP, 4TF or 5TF repair pattern. The frequency of the mode of failure is reported. Within each column, means with the same superscript are not significantly different ($P>.05$).

Repair	Ultimate Load (N)	Load at 2mm Gap (N)	Gap at Failure (mm)	Stiffness (N/mm)	Mode of Failure
3 LP	355.8 ^a (318.7-392.9)	184.8 ^a (165.9-203.7)	38.1 ^a (37.8-38.4)	9.6 ^a (8.8-10.4)	Suture pull-out (18) Suture breakage (2)
4 TF	193.6 ^b (177.5-209.6)	56.9 ^b (25.3-88.5)	20.8 ^b (18.6-23.0)	21.1 ^b (18.9-23.2)	Anchor pull-out (9) Anchor failure (1)
5 TF	222.8 ^b (207.8-237.7)	85.7 ^b (58.6-112.8)	22.3 ^b (19.2-25.4)	21.7 ^b (19.0-24.4)	Anchor pull-out (7) Anchor failure (3)

Table 3: All data collected for 3LP tendons in study

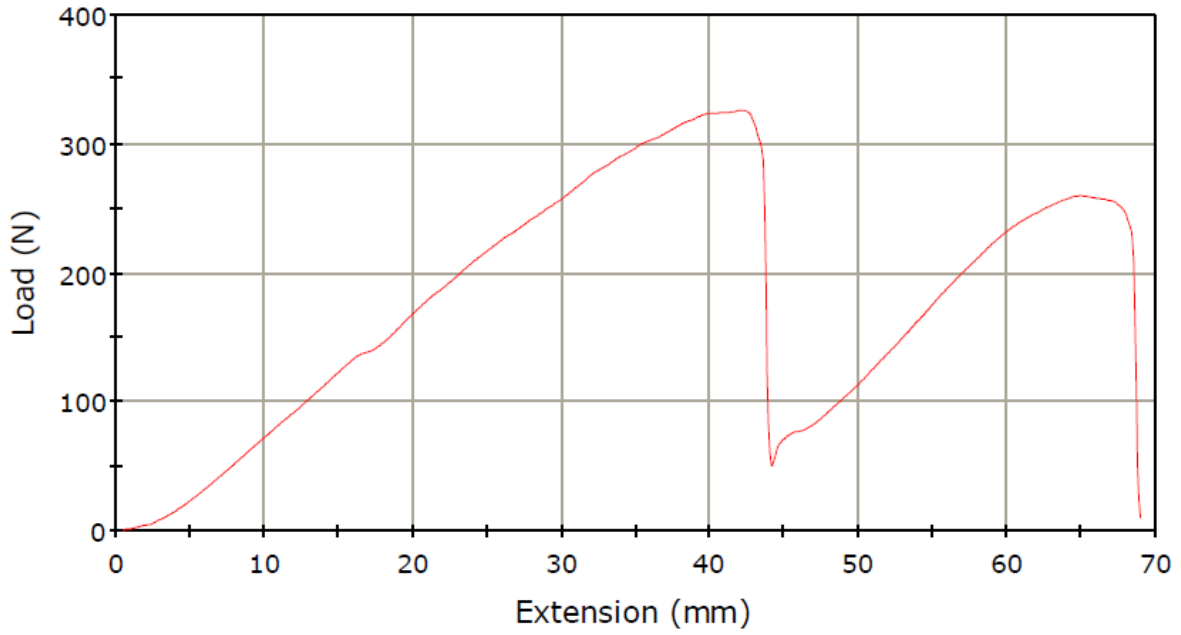
Horse	Repair	Breed	Age	Sex	Weight	C.S. Area (mm ²)	Load at Failure (N)	Mode of Failure	Gap at failure (mm)	Load to 2mm gap (N)	Stiffness (N/mm-1)
1	3LP	AQH	12	Gelding	544.32	123.2	325.78	Pull-through	37	208.28	8.7
2	3LP	Appaloosa	11	Gelding	589.68	129.2	323.51	Pull-through	44	175.55	8.3
3	3LP	Appaloosa	17	Gelding	408.24	130.5	406.07	Pull-through	41	223.95	9.6
4	3LP	AQH	28	Gelding	544.3	122.9	384.99	Pull-through	32	203.36	11.5
5	3LP	Cob	10	Gelding	420.51	125.7	372.75	Pull-through	40	206.35	8.9
6	3LP	AQH	10	Mare	457.61	124.8	264.5	Pull-through	35	161.25	7.1
7	3LP	Hanoverian	4	Gelding	550.32	130.6	369.18	Pull-through	31	220.8	10.8
8	3LP	TWH	8	Gelding	430.77	134.5	420.44	Pull-through	35	221.22	11.9
9	3LP	Arabian	20	Mare	453.6	131.7	367.5	Pull-through	44	148.18	10.9
10	3LP	Belgian	9	Gelding	592.6	137.2	254.1	Pull-through	27	166.90	8.9
11	3LP	AQH	23	Gelding	476.28	128.6	279.49	Pull-through	36	163.56	7.9
12	3LP	Hanoverian	7	Mare	635.04	129.1	453.21	Pull-through	47	213.94	10.2
13	3LP	AQH	10	Gelding	454.23	127.3	351.04	Pull-through	45	131.23	8.5
14	3LP	TB	20	Gelding	453.6	128.2	385.47	Pull-through	41	196.10	9.3
15	3LP	AQH	10	Gelding	460.7	125	342.75	Pull-through	42	198.85	9.7
16	3LP	Cob	18	Mare	430.56	127	301	Pull-through	36	135.12	9
17	3LP	TWH	15	Gelding	440.71	130.5	435.9	Pull-through	42	218.78	10.3
18	3LP	TB	10	Gelding	450.12	129.7	441.8	Breakage	40	169.23	11.6
19	3LP	AQH	20	Gelding	445.26	128	259.54	Pull-through	31	137.53	8.4
20	3LP	AQH	12	Gelding	460.78	125.8	376.2	Breakage	36	196.30	10.4

Table 4: All data collected for 4TF and 5TF tendons in study

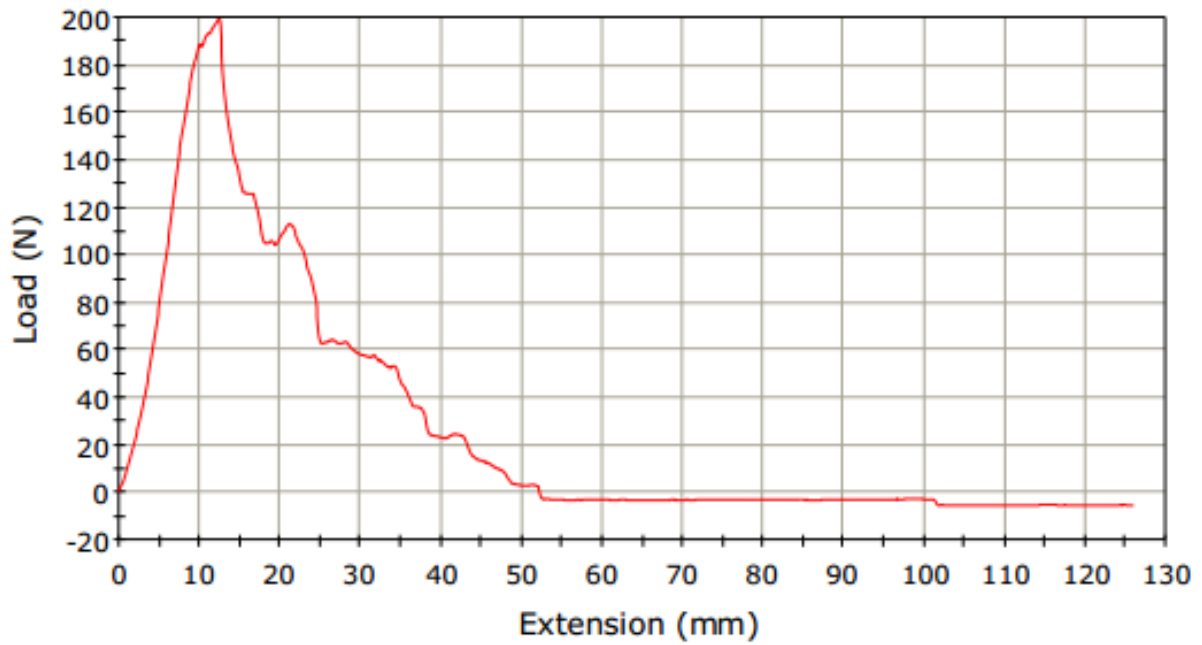
Horse	Repair	Breed	Age	Sex	Weight (mm ²)	C.S. Area (N)	Load at Failure (N)	Mode of Failure	Gap at failure (mm)	Load to create 2mm gap (N)	Stiffness (N/mm-1)
1	4TF	AQH	12	Gelding	544.32	123.2	199.11	Pull-through	23	39.09	18.8
2	4TF	Appaloosa	11	Gelding	589.68	129.2	210.67	Pull-through	22	37.34	20.3
3	4TF	Appaloosa	17	Gelding	408.24	130.5	187.37	Pull-through	21	58.02	18.9
4	4TF	AQH	28	Gelding	544.3	122.9	165.04	Pull-through	24	49.13	26.1
5	4TF	Cob	10	Gelding	420.51	125.7	170.99	Pull-through	27	1.06	19.9
6	4TF	AQH	10	Mare	457.61	124.8	169.1	Pull-through	17	62.23	14.7
7	4TF	Hanoverian	4	Gelding	550.32	130.6	254.25	Pull-through	14	150.84	22.1
8	4TF	TWH	8	Gelding	430.77	134.5	215.3	Pull-through	22	3.88	21.7
9	4TF	Arabian	20	Mare	453.6	131.7	179	Anchor failure	18	150.87	21.3
10	4TF	Belgian	9	Gelding	592.6	137.2	184.99	Pull-through	20	16.29	27.8
11	5TF	AQH	23	Gelding	476.28	128.6	261.08	Pull-through	30	97.61	27.8
12	5TF	Hanoverian	7	Mare	635.04	129.1	251.45	Anchor failure	18	40.72	23
13	5TF	AQH	10	Gelding	454.23	127.3	203.11	Pull-through	23	45.48	18
14	5TF	TB	20	Gelding	453.6	128.2	246.54	Pull-through	29	145.81	19.5
15	5TF	AQH	10	Gelding	460.7	125	193.28	Pull-through	28	1.18	21.7
16	5TF	Cob	18	Mare	430.56	127	202.78	Anchor failure	20	96.62	17.6
17	5TF	TWH	15	Gelding	440.71	130.5	244.97	Pull-through	21	66.25	20.8
18	5TF	TB	10	Gelding	450.12	129.7	213.36	Pull-through	23	133.19	30
19	5TF	AQH	20	Gelding	445.26	128	196.98	Anchor failure	16	109.86	15
20	5TF	AQH	12	Gelding	460.78	125.8	214.45	Pull-through	15	120.26	23.8

Appendix 3: Load displacement curves for each tendon

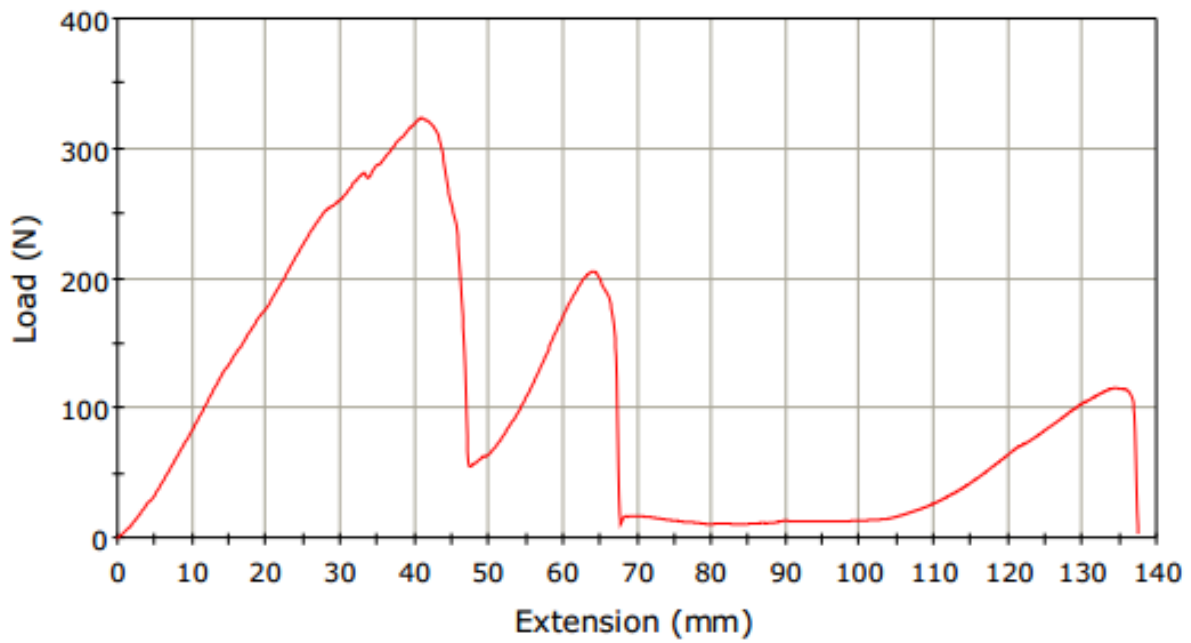
Horse 1- 3LP



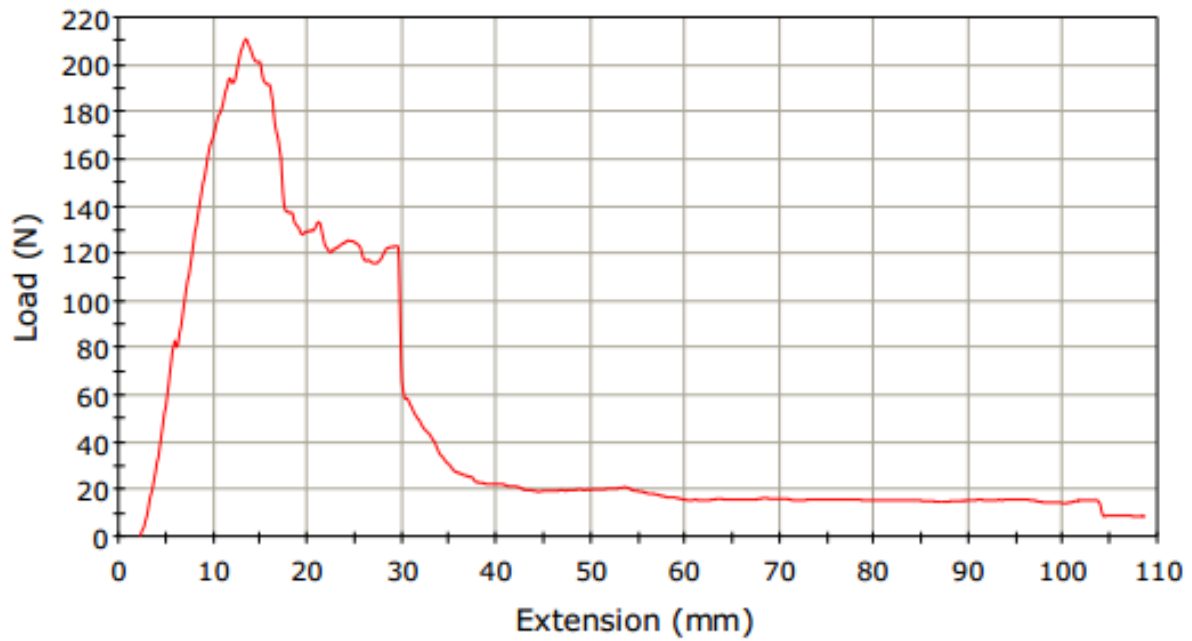
Horse 1- 4TF



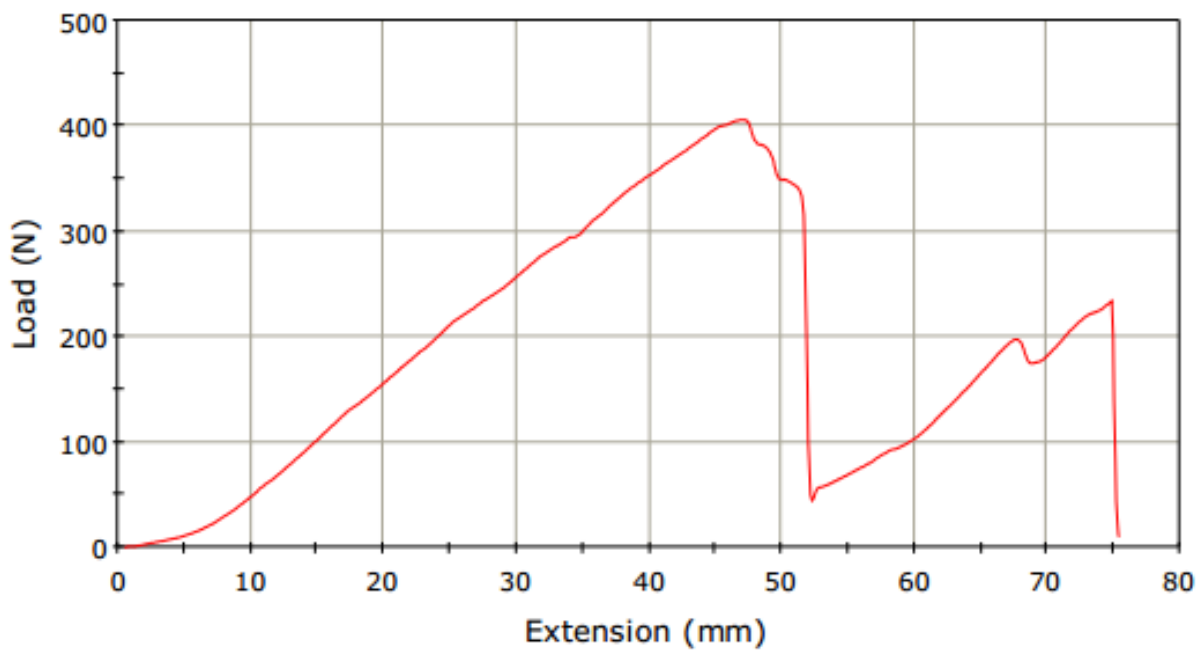
Horse 2- 3LP



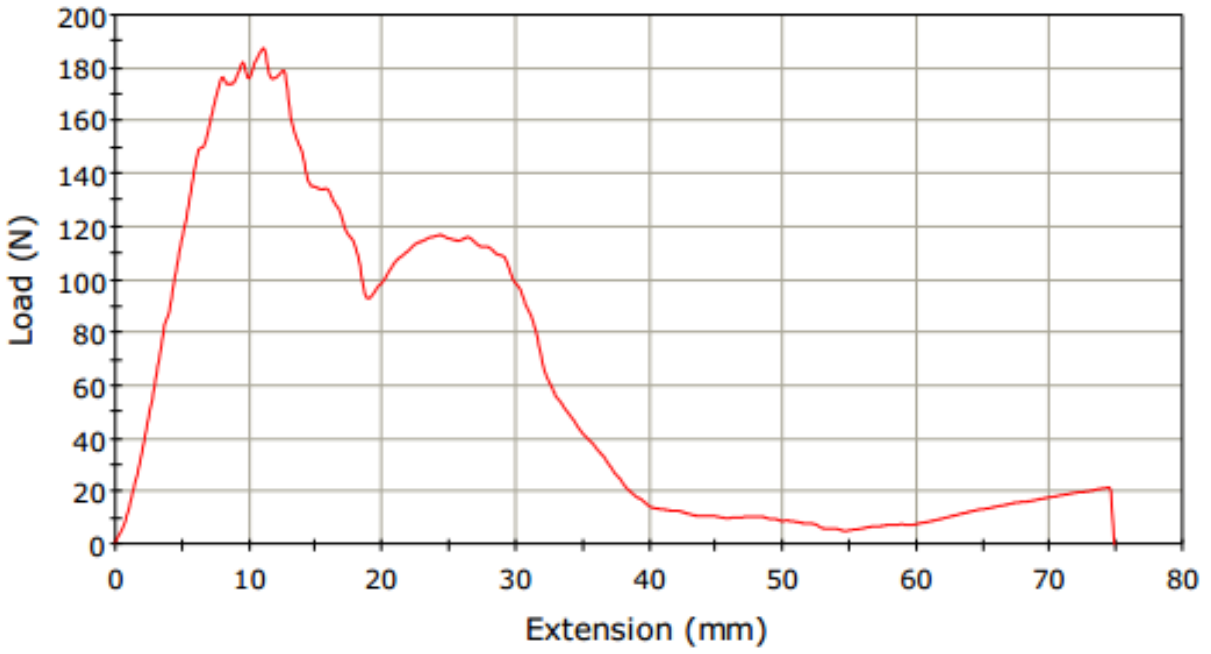
Horse 2- 4TF



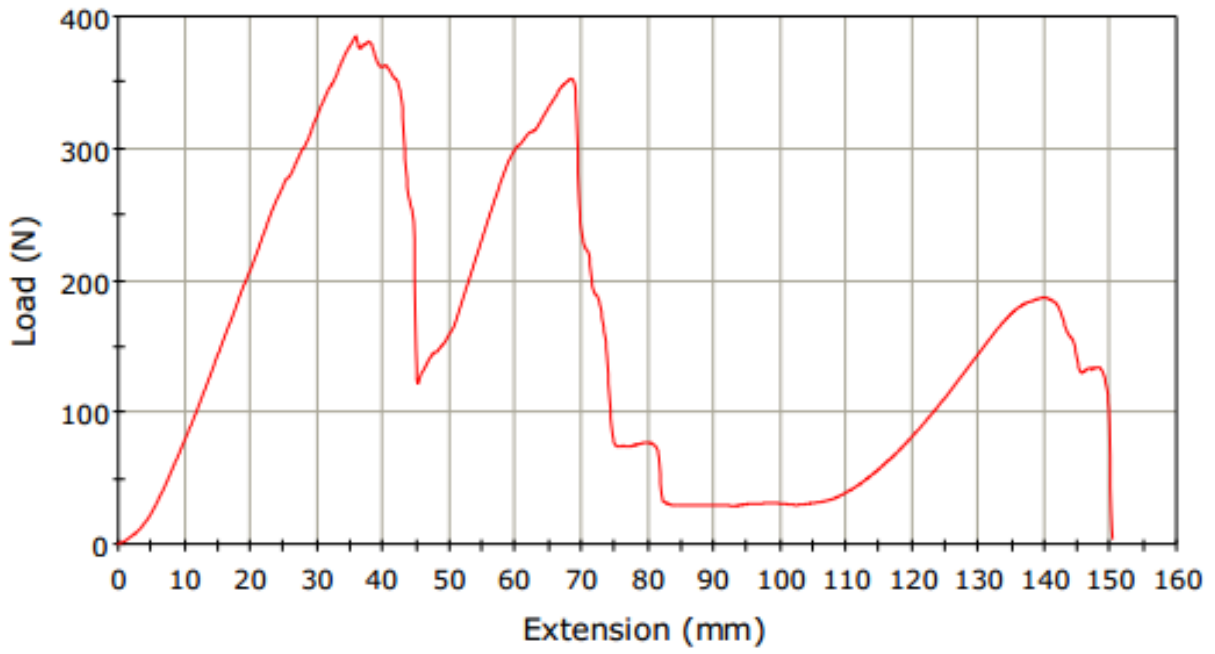
Horse 3- 3LP



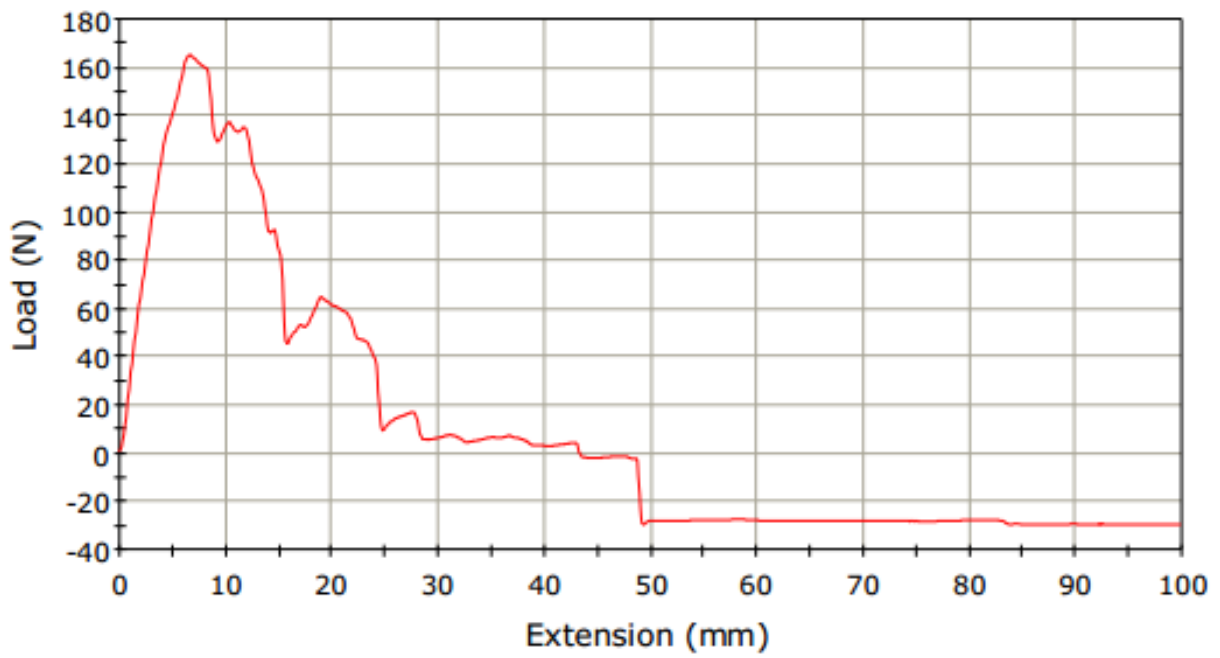
Horse 3- 4TF



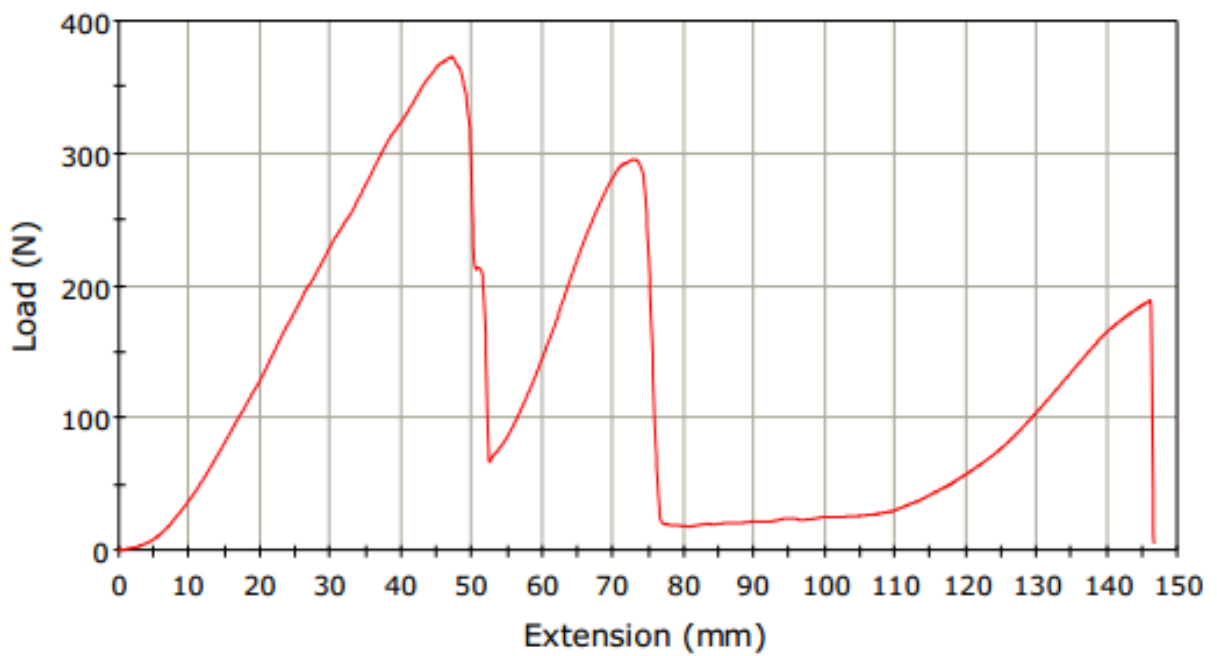
Horse 4- 3LP



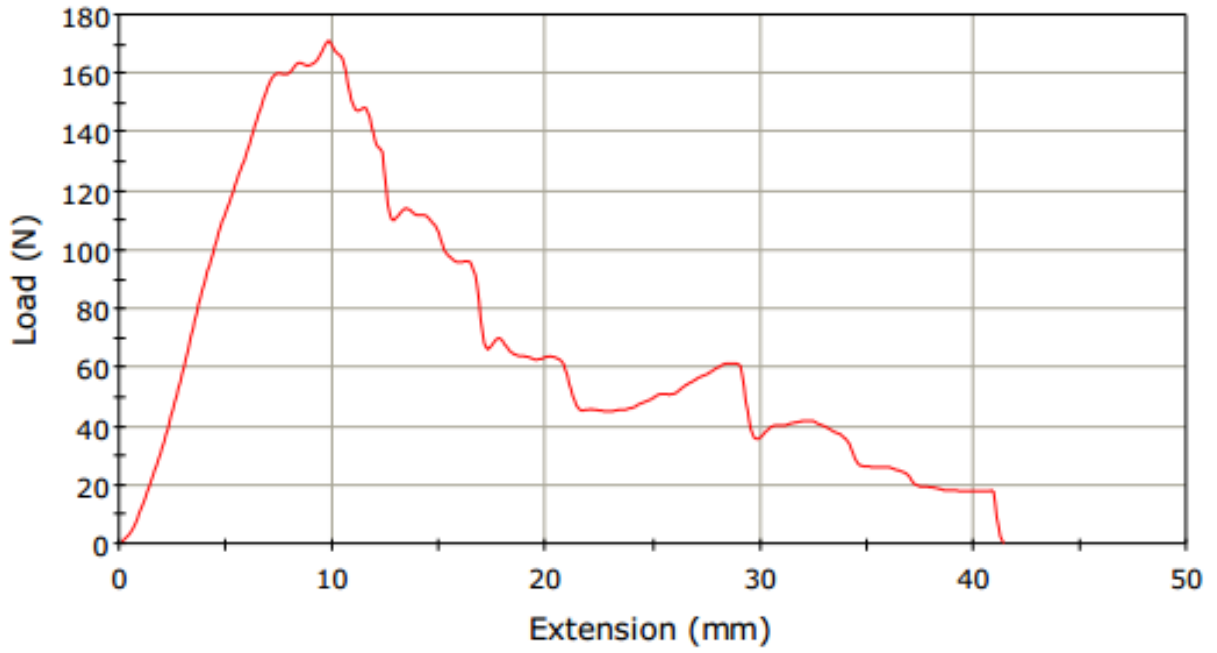
Horse 4- 4TF



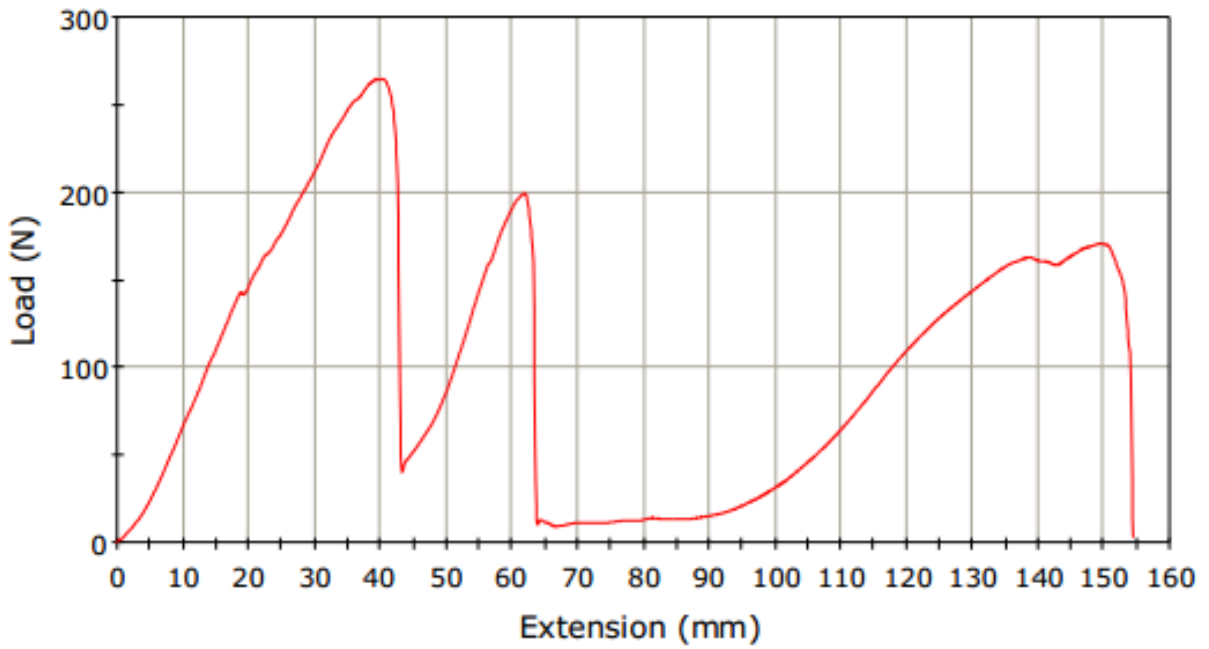
Horse 5- 3LP



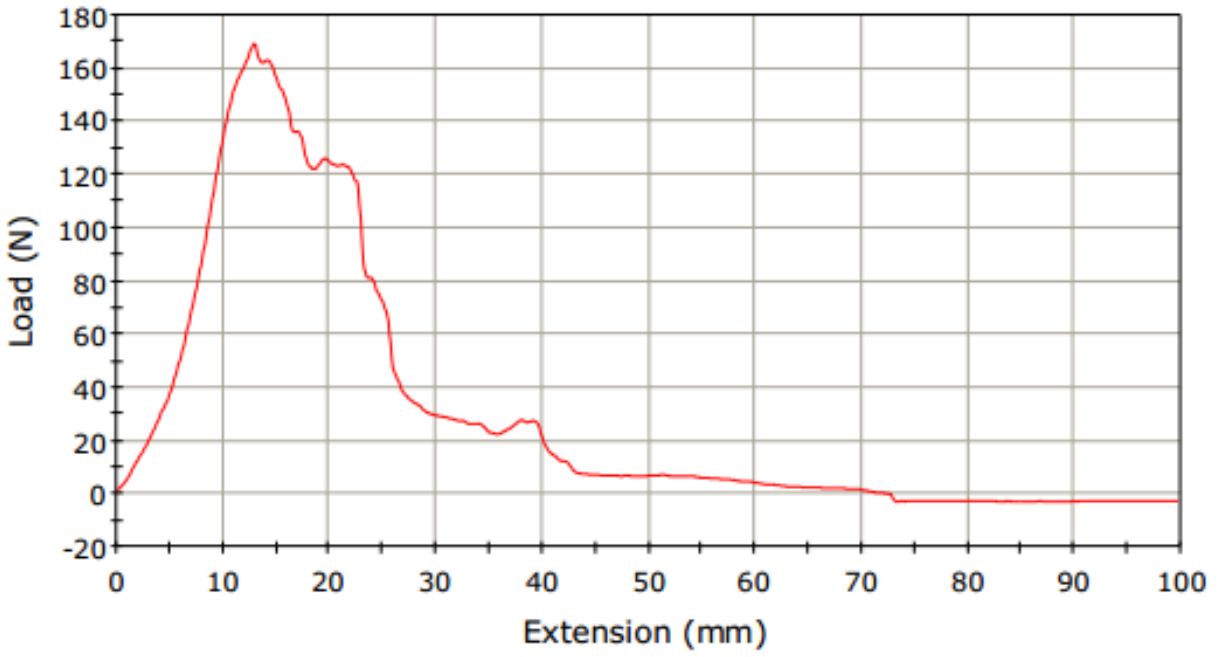
Horse 5- 4TF



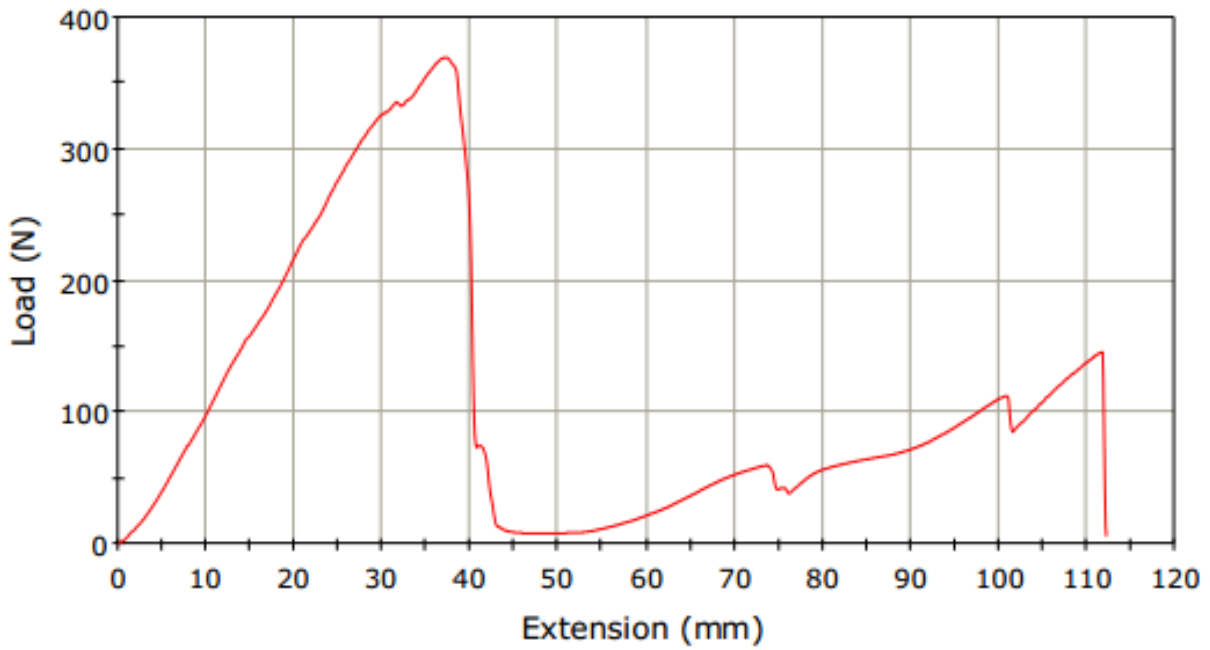
Horse 6- 3LP



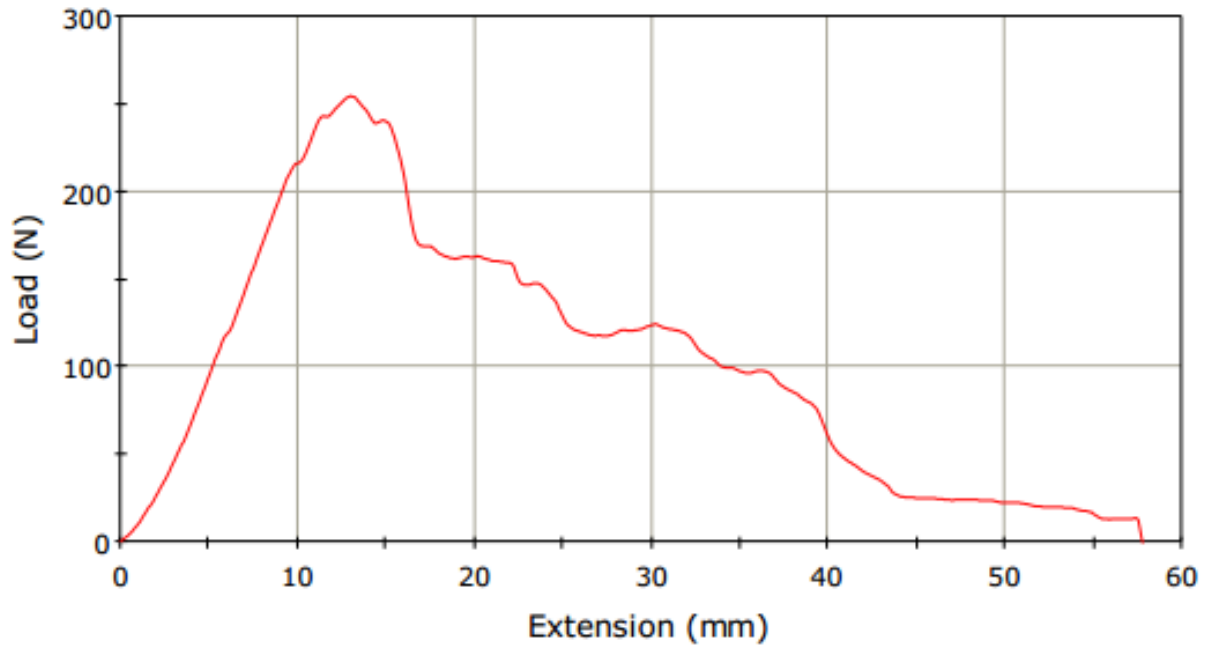
Horse 6- 4TF



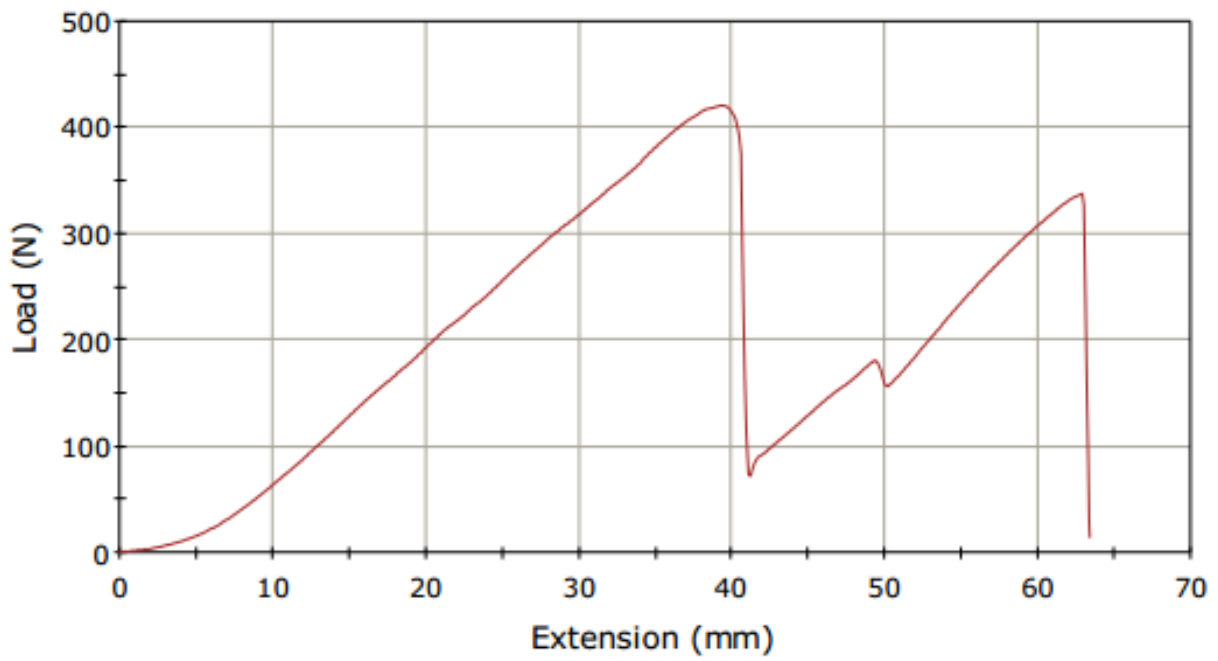
Horse 7- 3LP



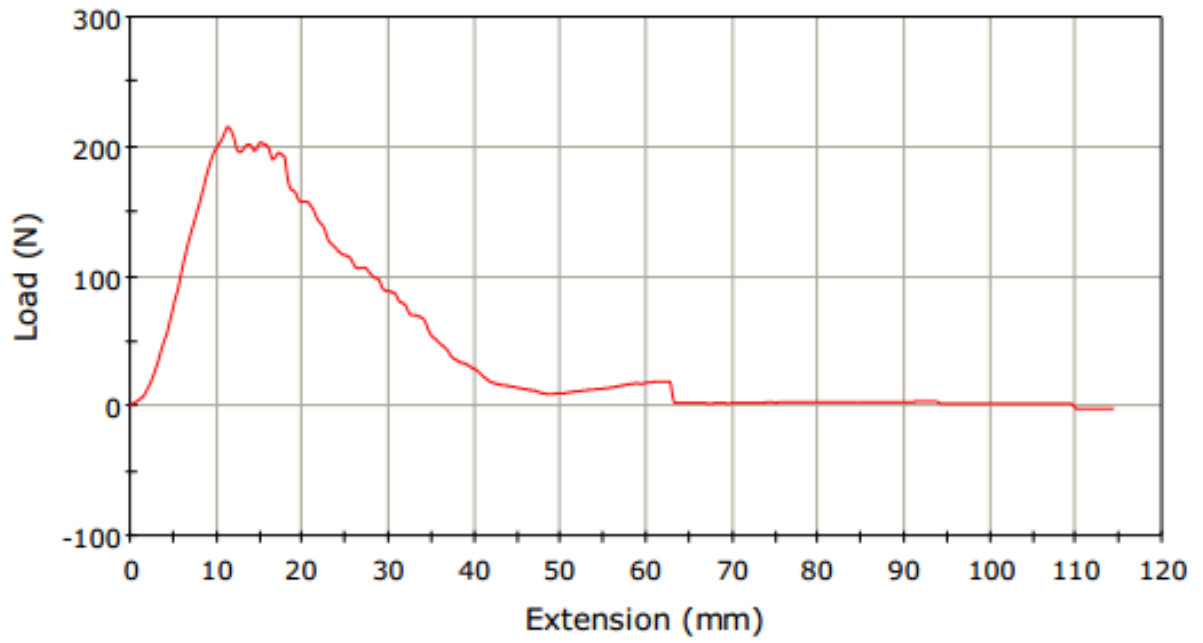
Horse 7- 4TF



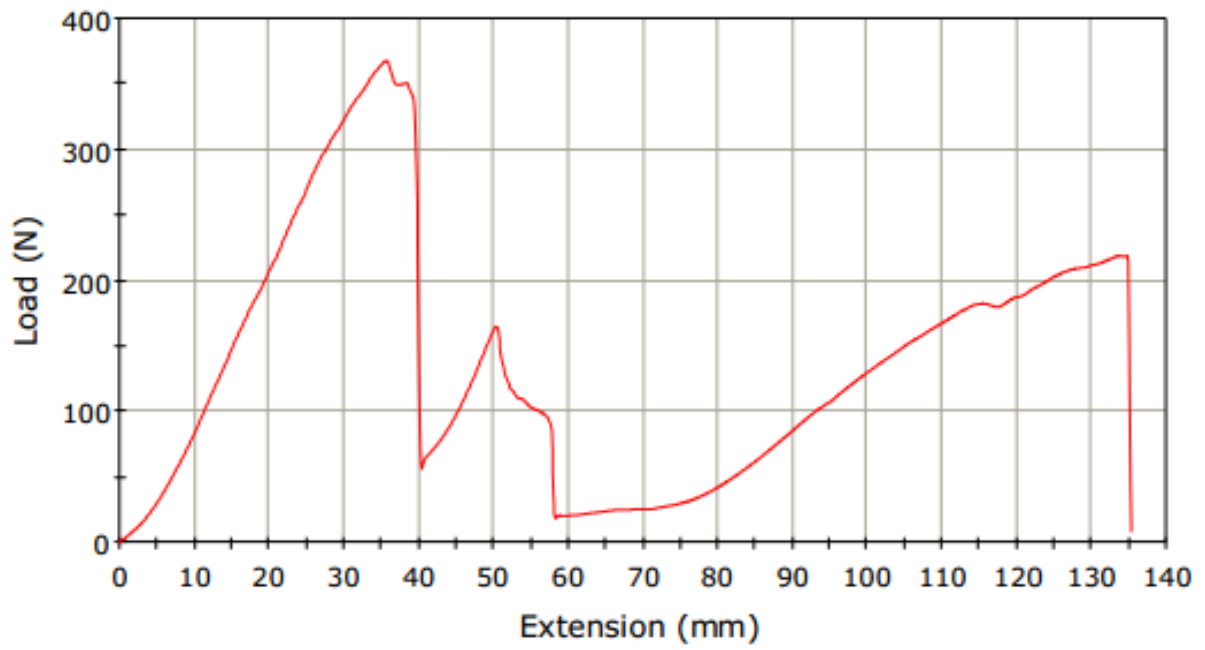
Horse 8- 3LP



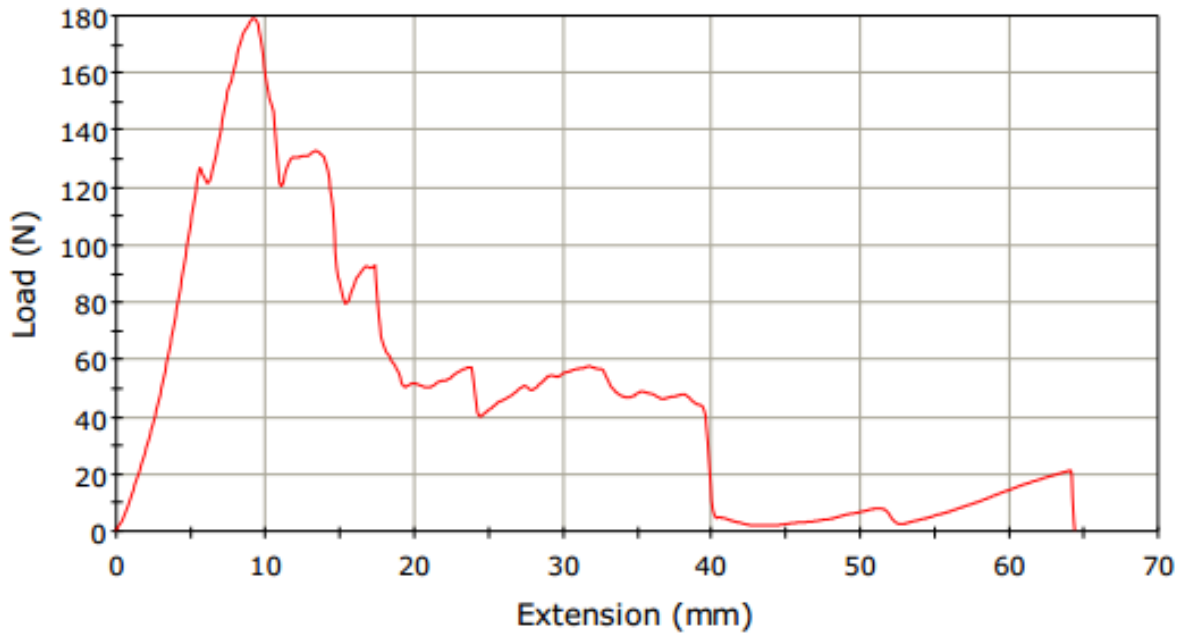
Horse 8- 4TF



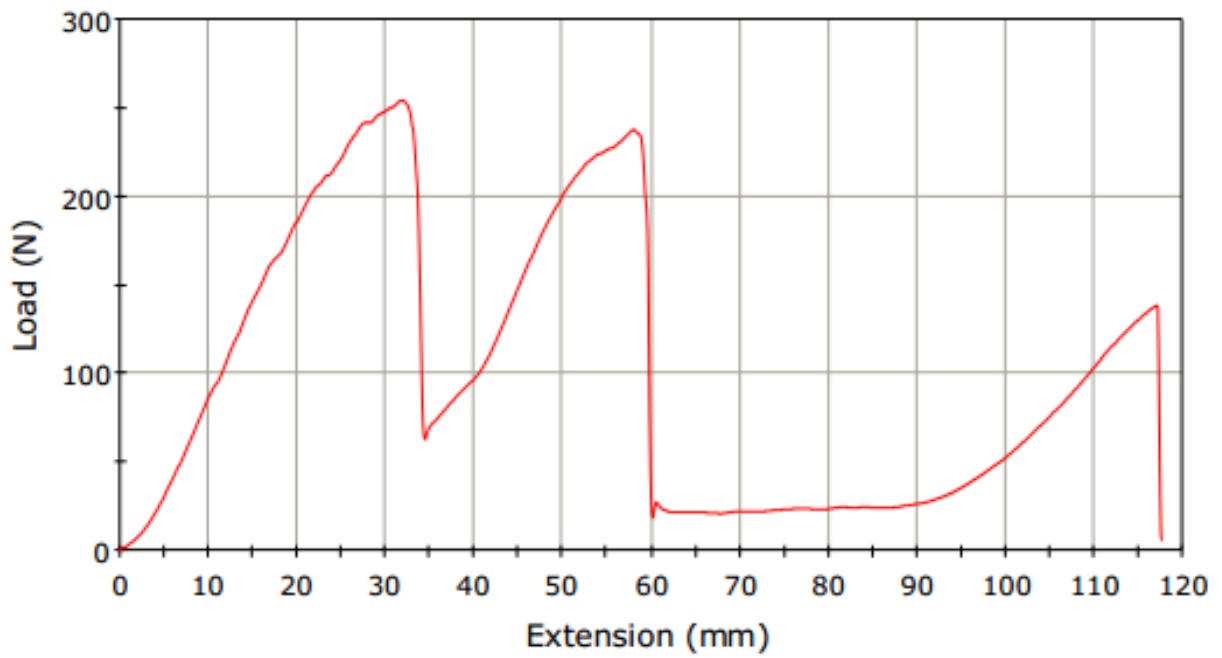
Horse 9- 3LP



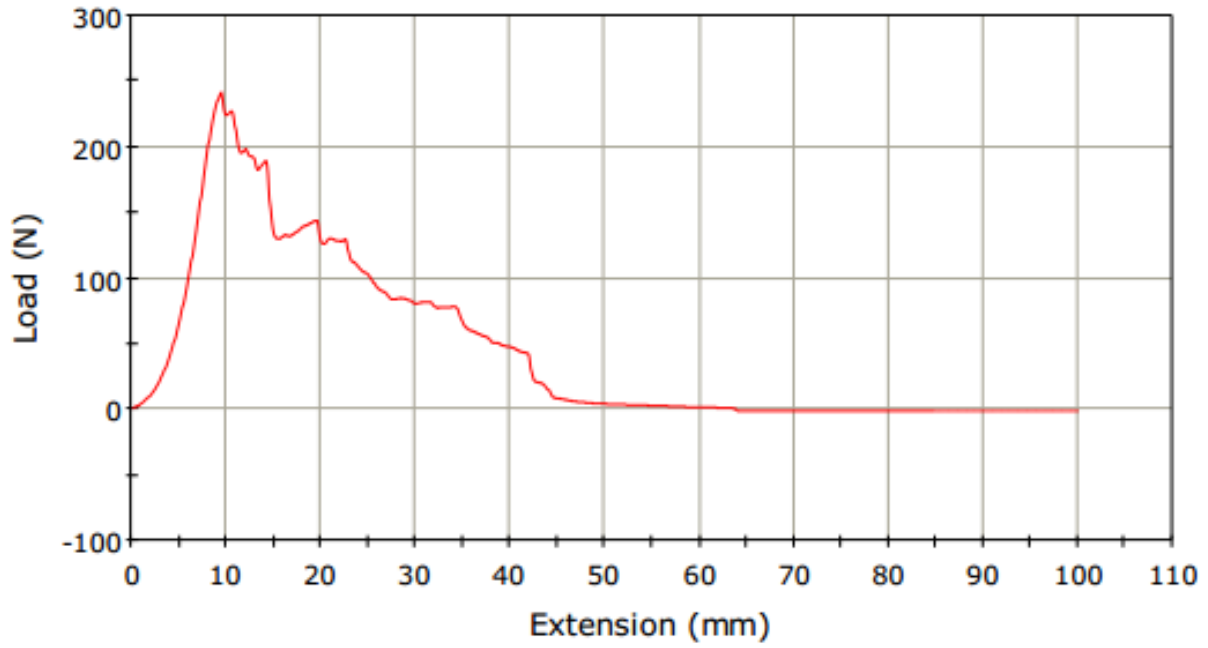
Horse 9- 4TF



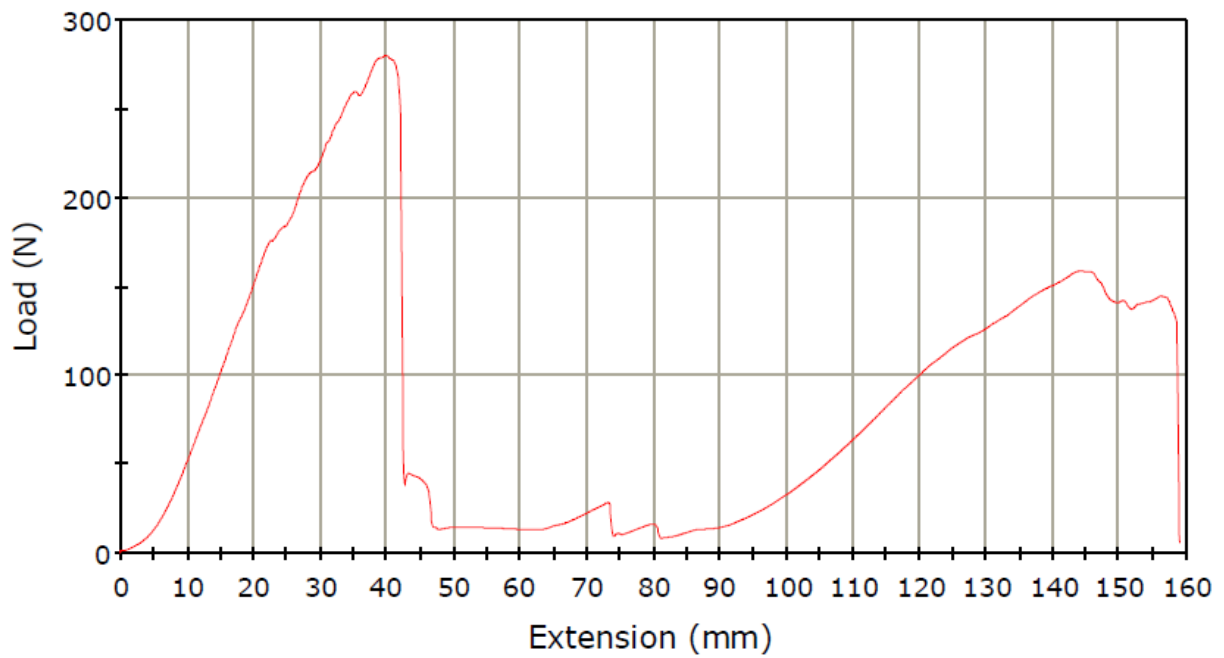
Horse 10- 3LP



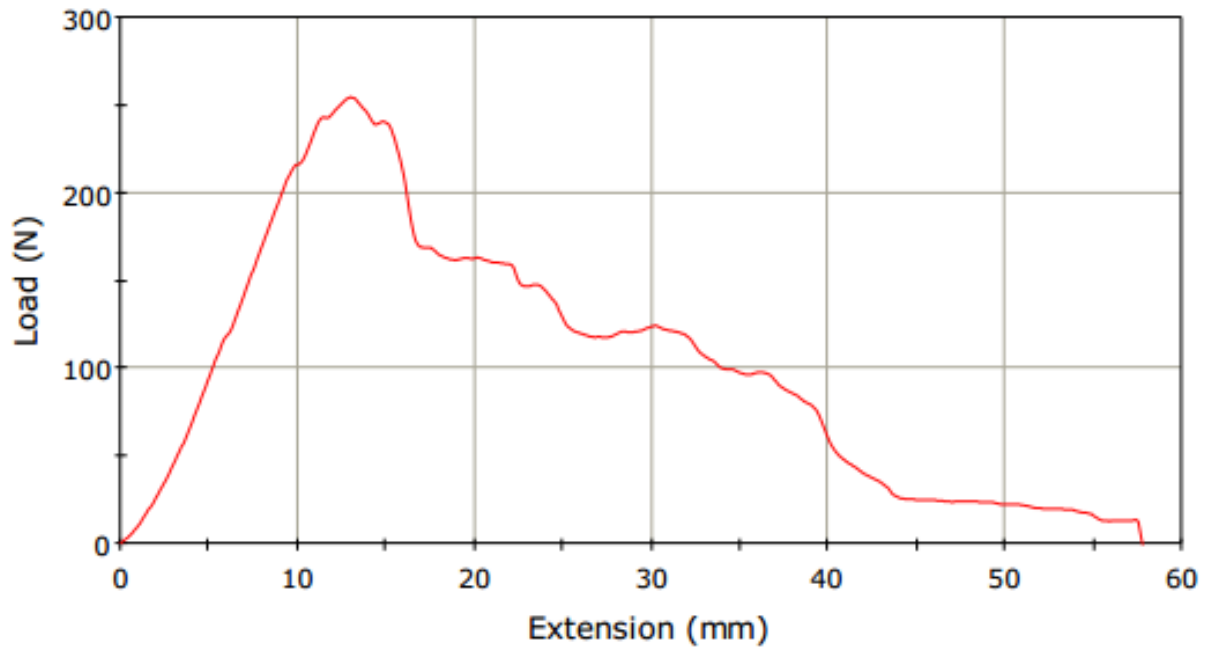
Horse 10- 4TF



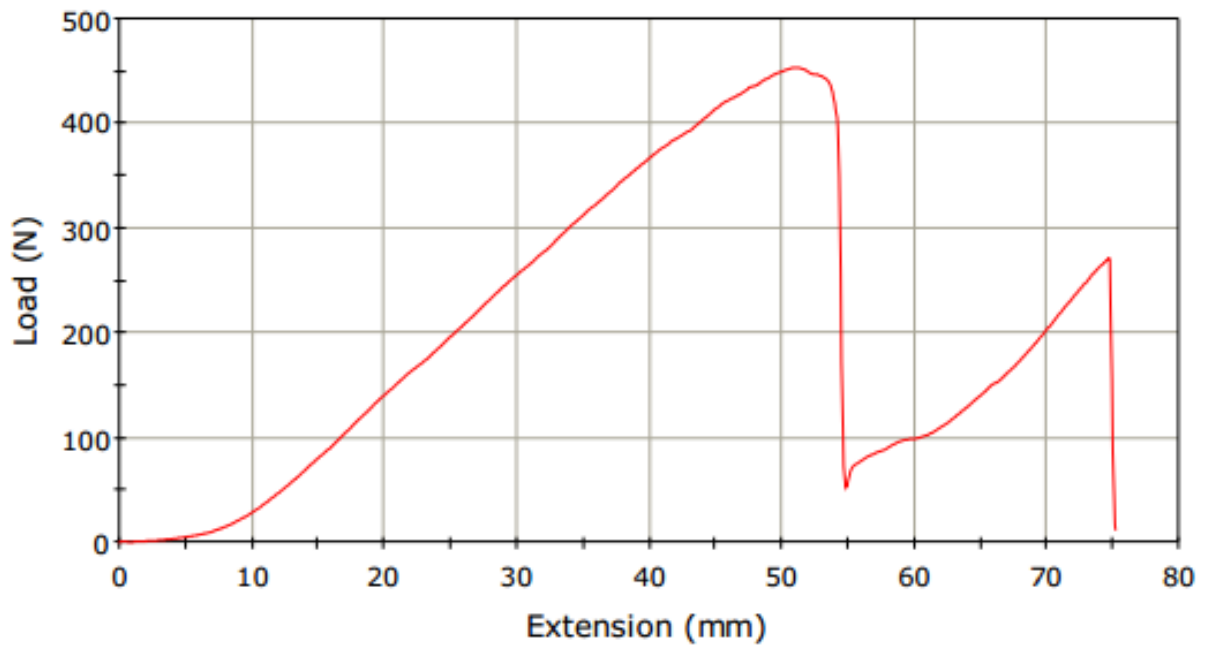
Horse 11- 3LP



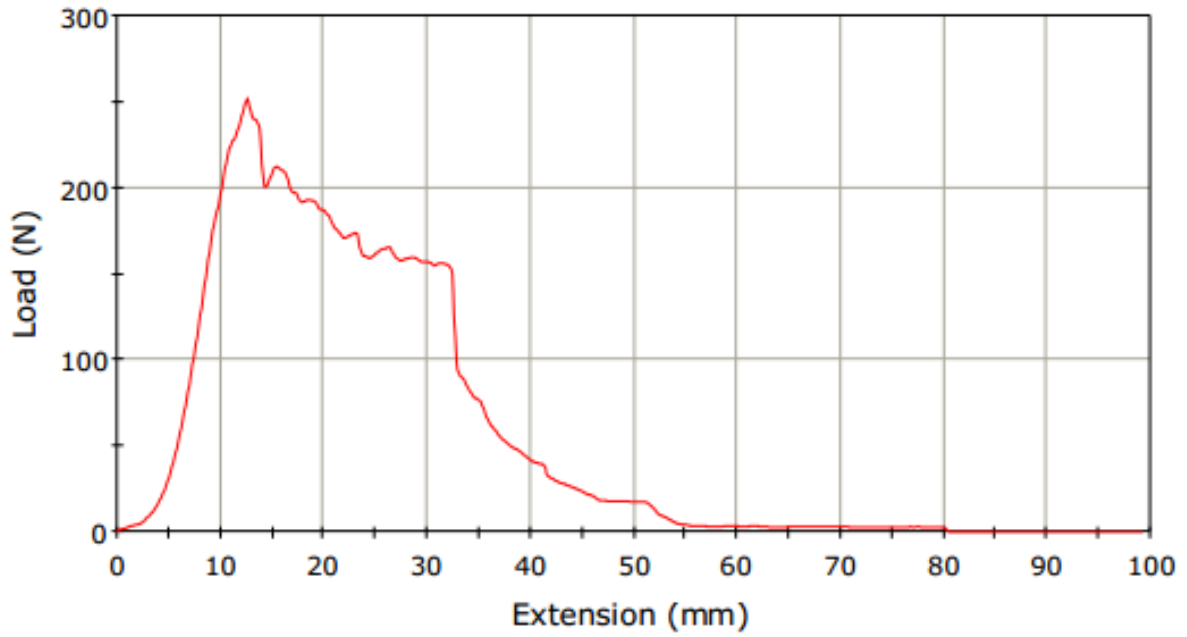
Horse 11- 5TF



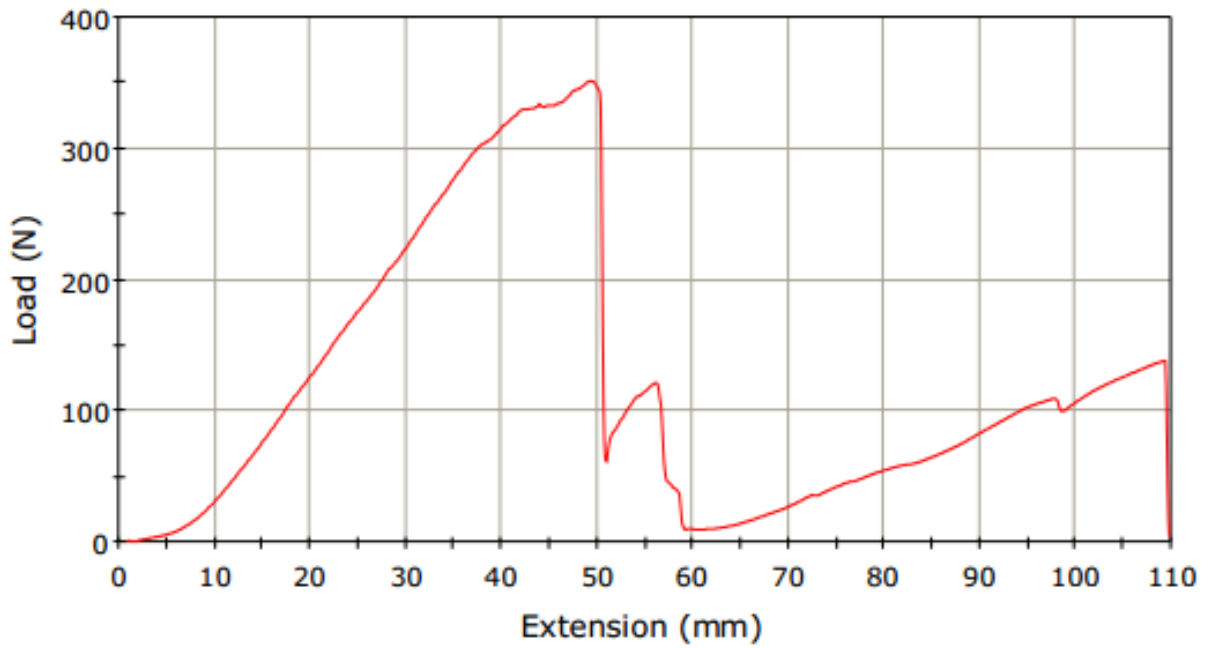
Horse 12- 3LP



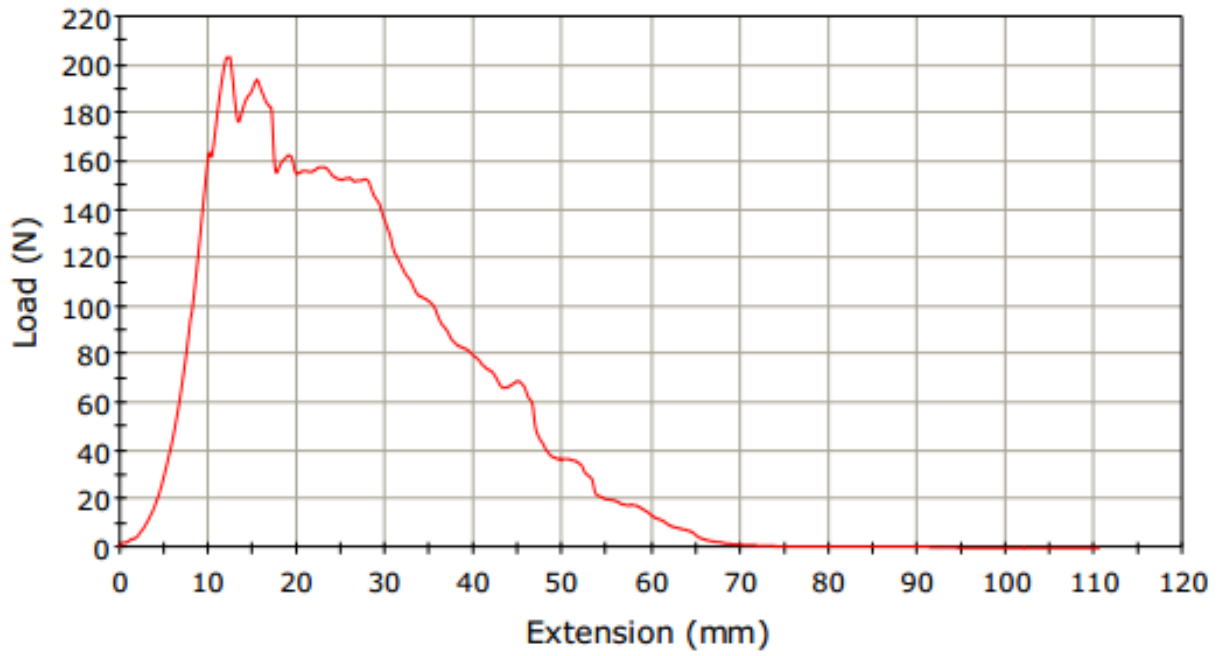
Horse 12- 5TF



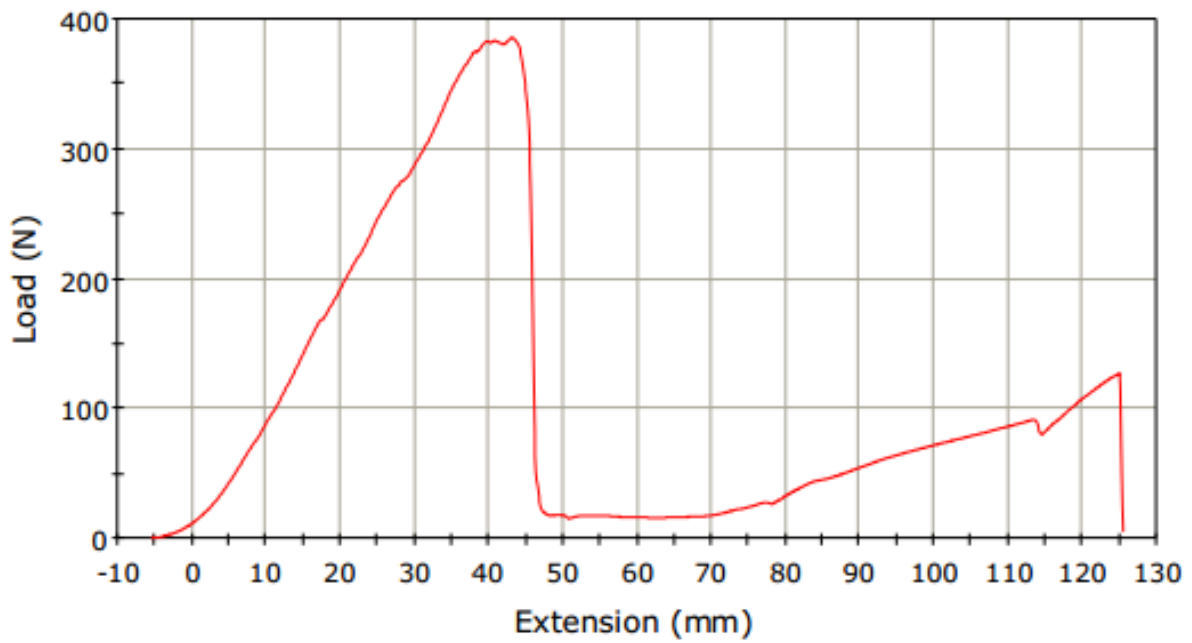
Horse 13- 3LP



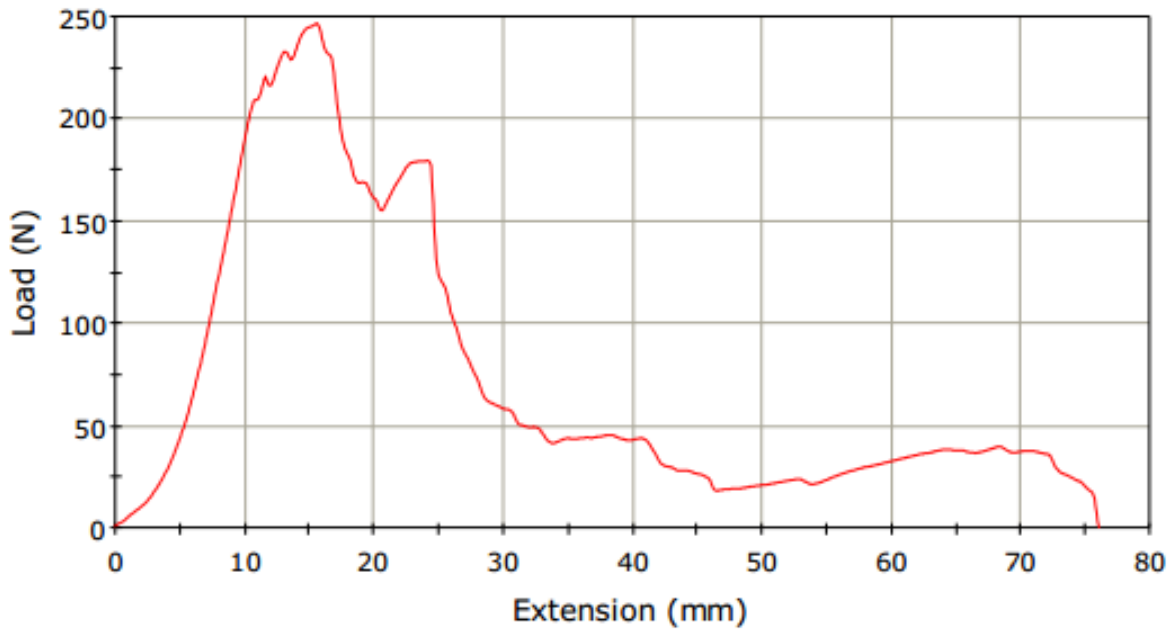
Horse 13- 5TF



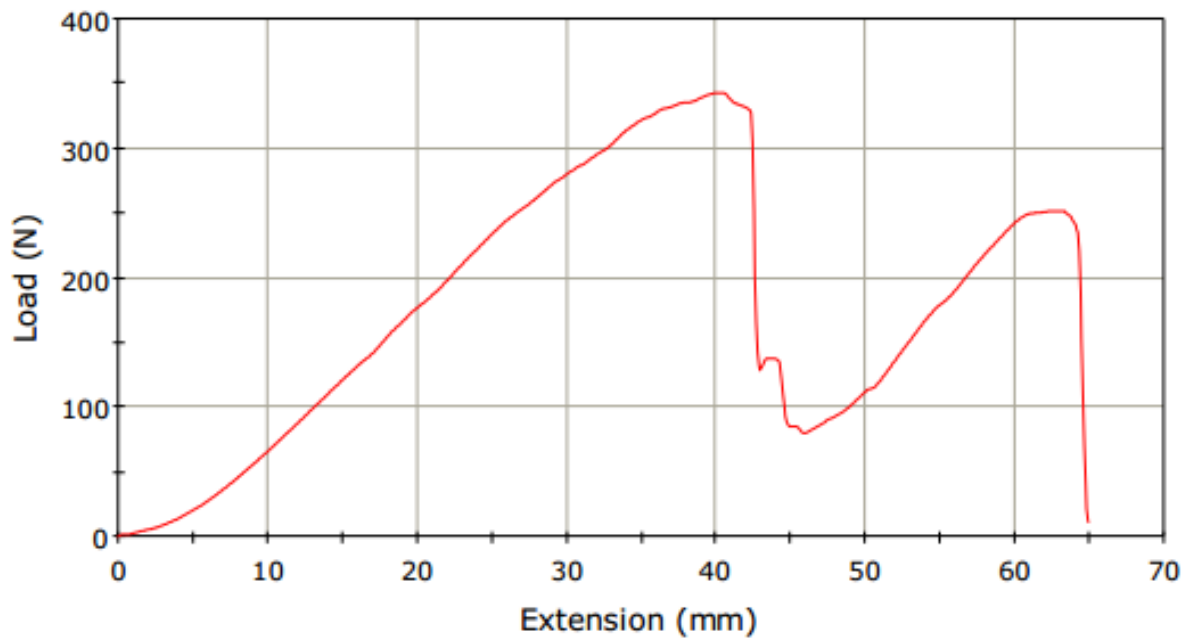
Horse 14- 3LP



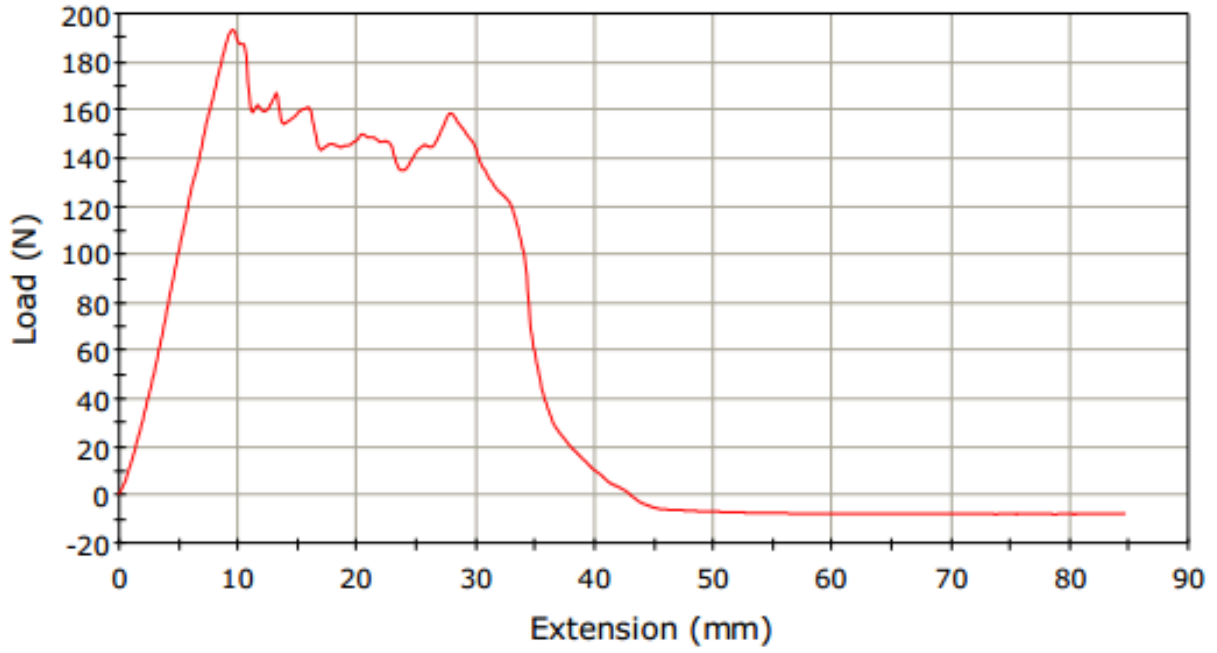
Horse 14- 5TF



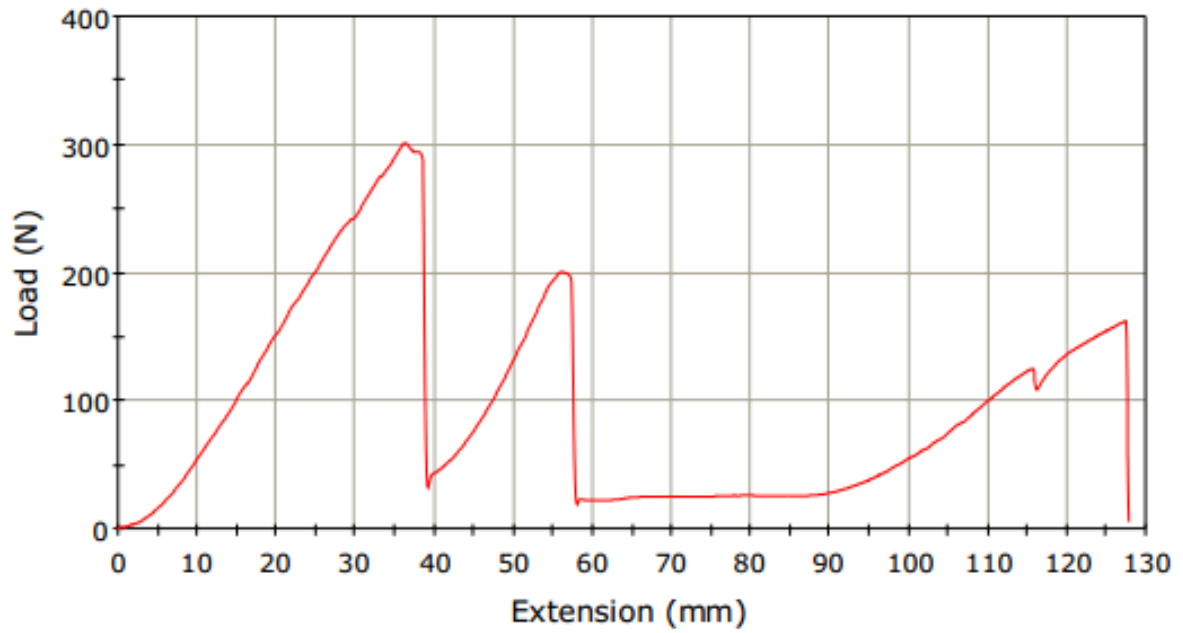
Horse 15- 3LP



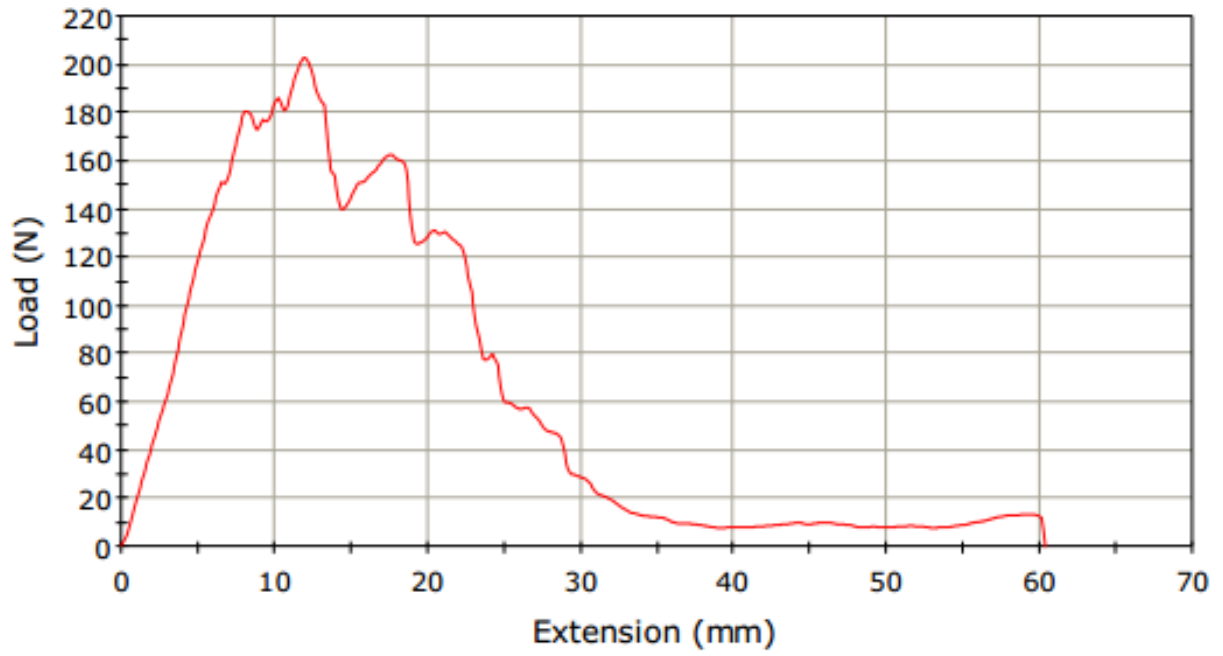
Horse 15- 5TF



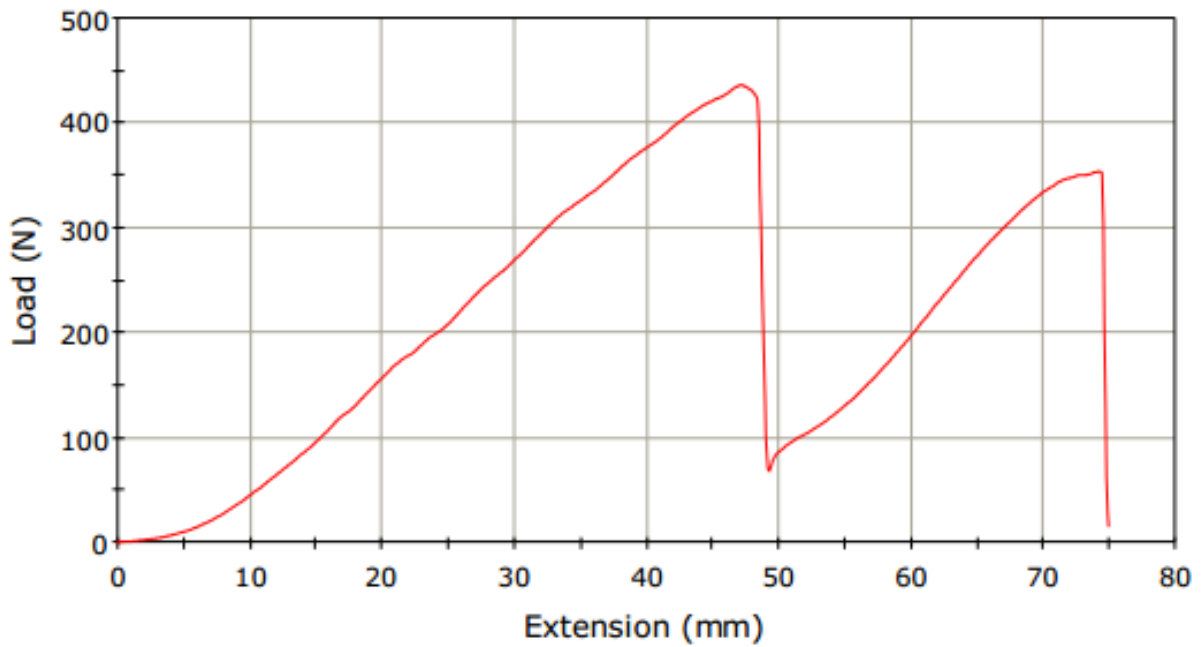
Horse 16- 3LP



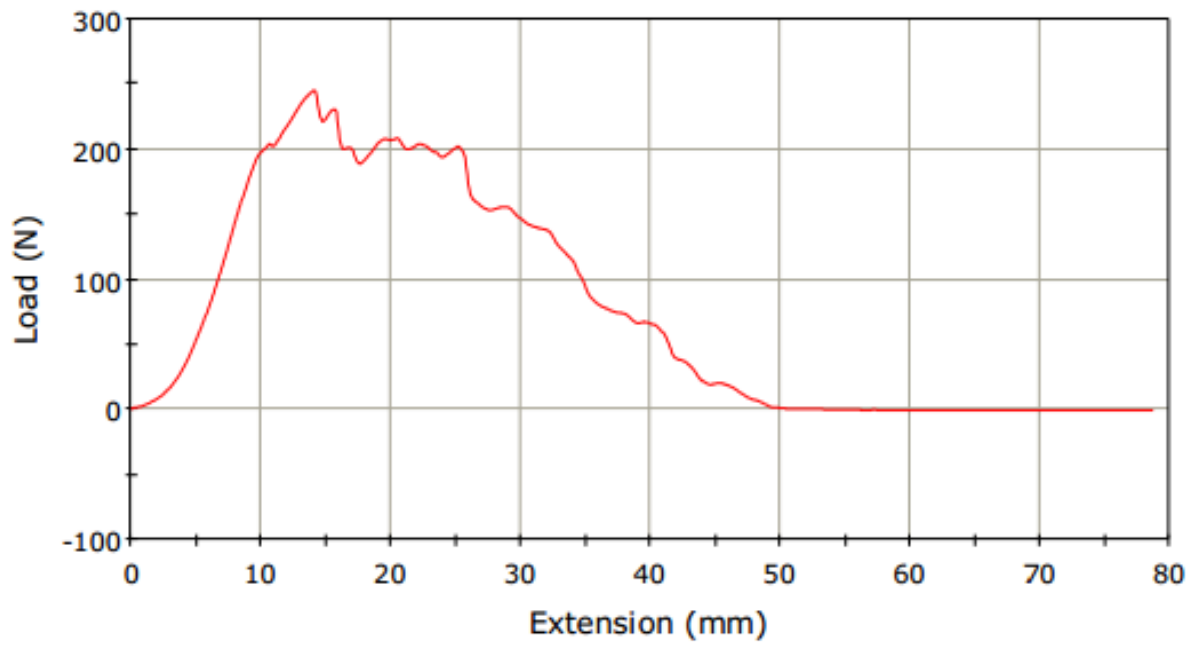
Horse 16- 5TF



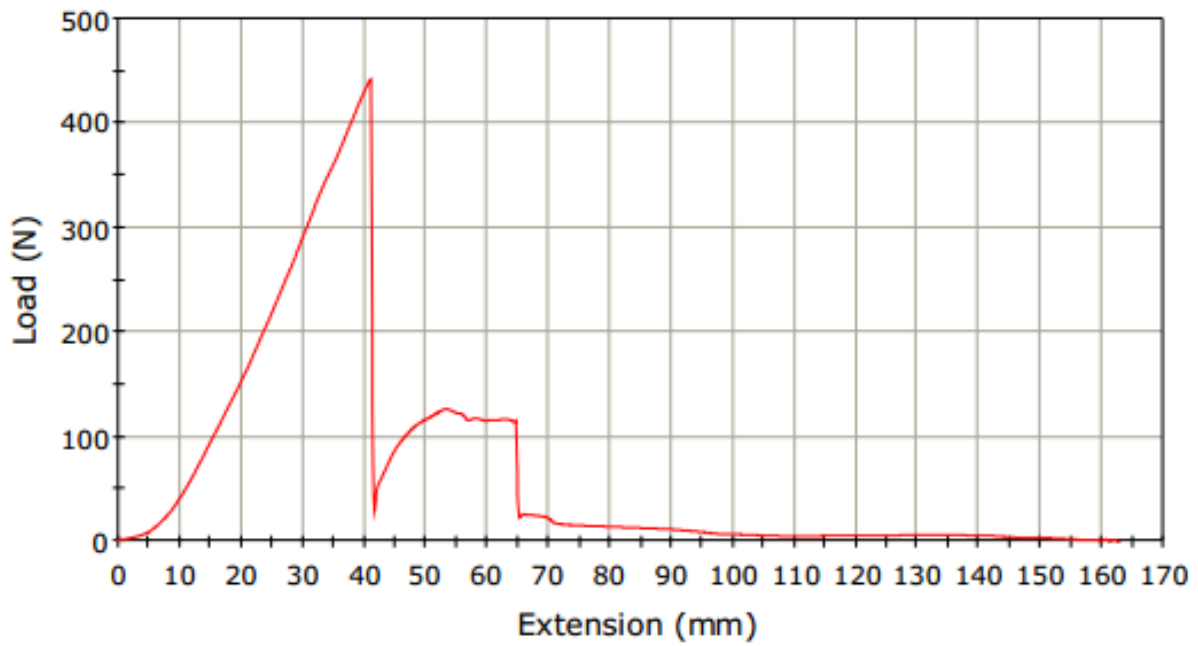
Horse 17- 3LP



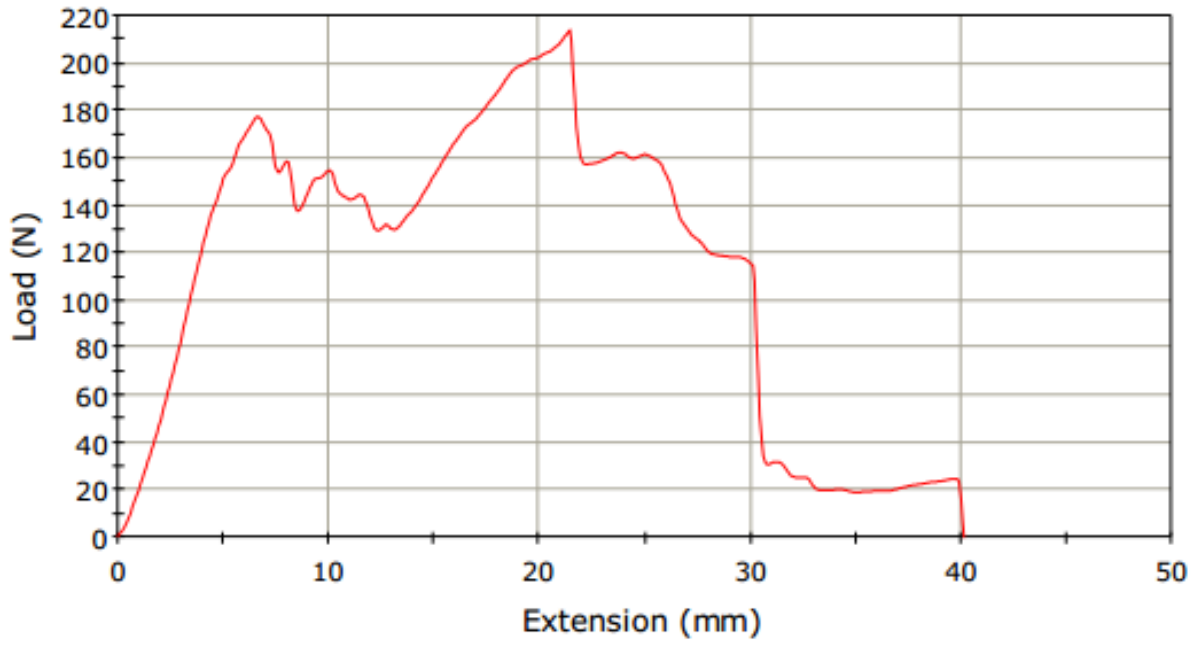
Horse 17- 5TF



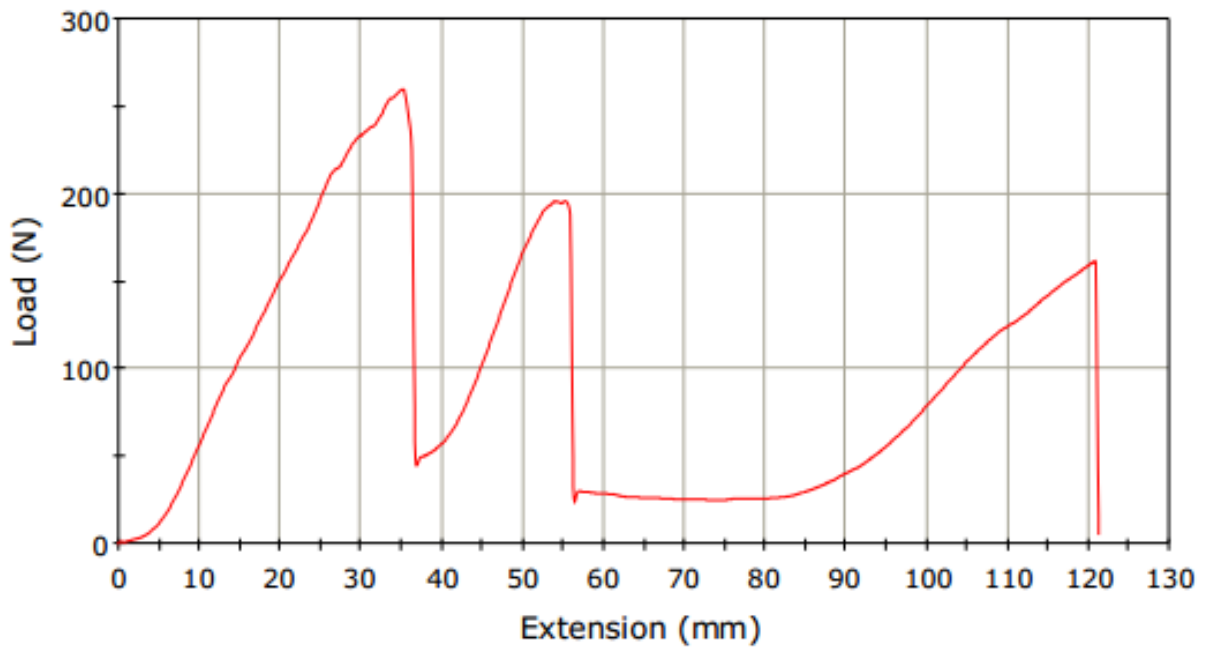
Horse 18- 3LP



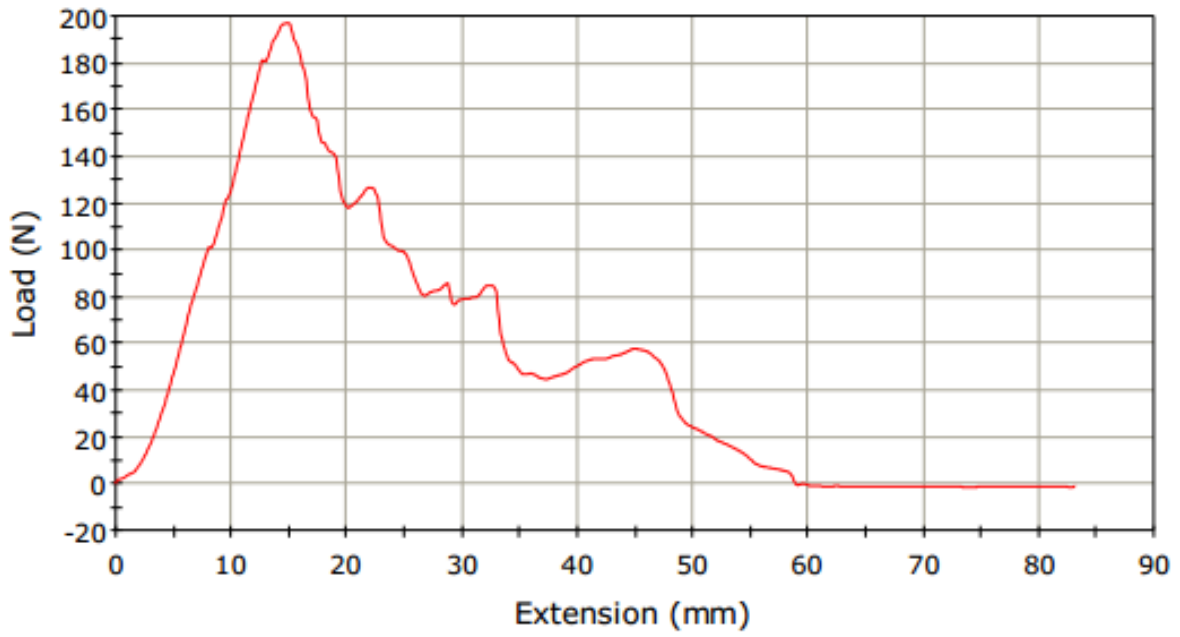
Horse 18- 5TF



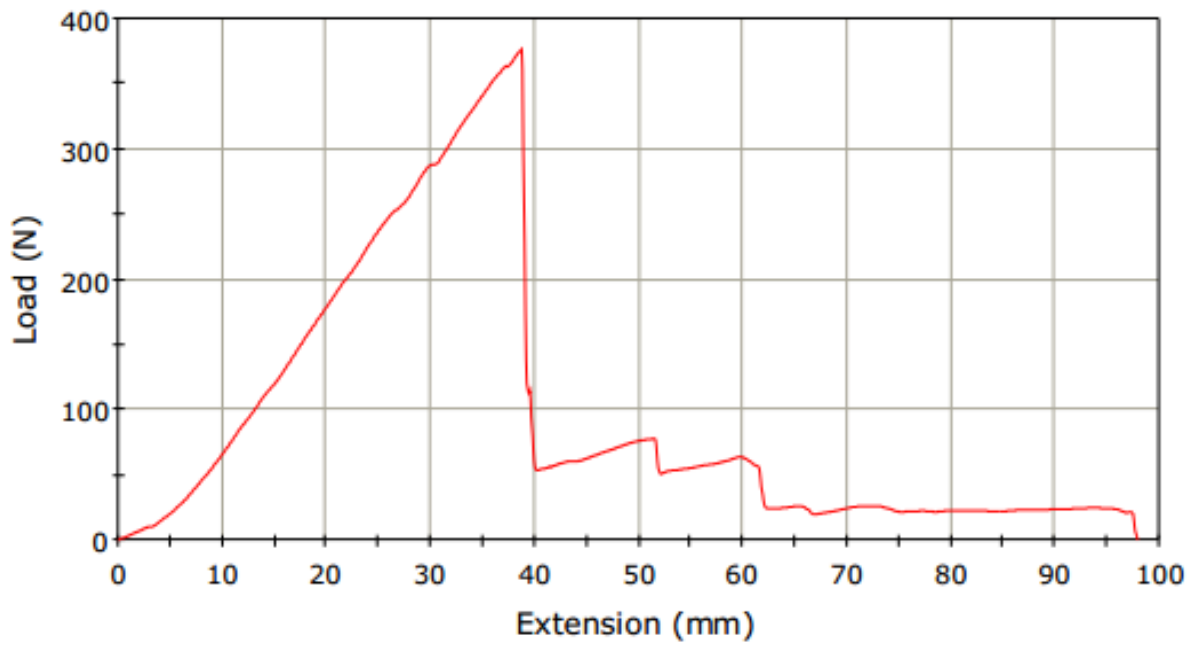
Horse 19- 3LP



Horse 19- 5TF



Horse 20- 3LP



Horse 20- 5TF

