Biomechanical investigation of a new core suture configuration and a new peripheral repair method for Zone II flexor tendon injuries

An experimental *ex vivo* study

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Declaration:

‘This thesis is the result of the author’s original research. It has been composed by the author and has not been previously submitted for examination which has led to the award of a degree.’

Signed:

Date: 10.10.13

“Unless you try to do something beyond what you have already mastered, you will never grow.”

Ralph Waldo Emerson

Abstract

Flexor tendon repair continues to provide surgeons with a challenge. Injuries occurring in zone II are a particular challenge due to the complex arrangement of the anatomy in this area. Poor results postoperatively are costly and lead to even more complex second operations, resulting in long time periods where patients are unable to use their hands fully. The repair techniques are numerous, as are the materials available for such repairs, and the training for surgeons in this field is largely based on practice on simulation models and the apprenticeship method. The literature review in this dissertation highlights the numerous repair variables, the type of repair, suture size and material, number of strands of suture crossing the repair site, as well as testing models. The use of a non-absorbable suture like braided polyester (Ticron) and a monofilament of polypropylene (Prolene) are commonly used in clinical practice, and these are both investigated for effect on tensile properties of the repair, as is the learning curve of one of the investigators.

One aim of this research was to develop a realistic, low cost bench-top model for tensile testing of tendon repairs, to enable the primary of the study; an investigation of a new core suture and a new peripheral repair method. A porcine model is investigated for anatomical similarity with human flexor tendons, and a mechanical testing protocol established, that allows investigation of tendon repairs for tensile tests as well as internal work of friction. A comparison of the well-known and tested Pennington Modification of the Modified Kessler (MK) technique is performed with a new repair type that incorporates a ventral locking loop to the standard Modified

Kessler (LMK). In addition the work introduces a PolyCaprolactone (PCL) sheet to the repair, as a new peripheral repair method.
Early findings showed higher values for the Ticron repairs and a clear learning curve with regards to mechanical properties of the repairs. The importance of conditioning of the tendons was also investigated in the preliminary studies and a clear effect on the work of flexion was found for each cycle tested.

No difference was seen in the tensile properties between the repairs, but for the two strand core suture repairs the LMK had lower work of flexion values. This was investigated further by measuring the cross sectional diameters of the two repair types to see if the change in repair technique had influenced the size of the repair. No difference was found. The four strand core suture repairs showed a higher force to gap formation of the LMK repair compared to the MK. The addition of the PCL sheet resulted in lower force to gap formation values. With regards to the work of flexion the PCL resulted in an almost doubling of the values seen with MK or LMK repairs, however no statistical difference was seen with regards to work of flexion between the MK and LMK.

This work has shown that the porcine model is similar to human and provides a good model for tendon repair. A mechanical testing procedure for tensile testing as well as work of flexion has been established that allows for testing of tendon repairs. The model would also allow for trainees to practice and get qualitative feedback on their repairs. No clear biomechanical advantage was seen with the addition of the ventral locking loop when compared to the MK, however the MK configuration used had an increased locking loop diameter than those presented in the literature which means that the LMK may provide a biomechanical advantage. Further studies are warranted to investigate how the LMK compares to MK with smaller locking loop areas, and other commonly used repair techniques. The addition of the PCL sheet to the repair, caused a large increase in the work of flexion of the repairs, and the PCL sheets showed large deformation after low loads during tensile testing which means that they would not have any clinical application for zone 2 tendon repairs. The possibility of using the PCL sheet as a tissue engineering scaffold/matrix for tendon repairs out with the hand could be investigated in further studies, however it is likely that modifications need to be done to the PCL before it proves useful in clinical practice.

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List of Publications

1. Zetlitz E, Hart AM, Nicol AC & Wearing SC. **Time-Dependent Conditioning Effects Are Important When Evaluating the Gliding Resistance of Flexor Tendon Repairs.** Proceedings International Federation for Medical and Biological Engineering. 2010; 31: 942-5.

2. Zetlitz E, Wearing SC & Hart AM. **Objective assessment of surgical training in flexor tendon repair: the utility of a low-cost porcine model as demonstrated by a single-subject research design**. Journal of Surgical Education. 2012 Jul;69(4):504-10.

3. Zetlitz E, Wearing SC & Hart AM. **Do continuous epitendinous sutures improve gliding properties and failure strength of Zone II flexor digitorum profundus repairs? revisiting the work of Wade et al (1986).** Proceedings of the International Society of Biomechanics – Published abstract.

List of Presentations

Zetlitz E, Hart AM, Black R & Wearing SC.
Gliding properties of a new Polymer wrap repair for flexor tendon repairs.
Accepted for presentation at: ISB XXIV Congress of the International Society of Biomechanics, Natal, Rio Grande do Norte, Brazil, August 04-9th 2013.

Zetlitz E, Wearing SC & Hart AM.
Do Continuous Epitendinous Sutures Improve Gliding Properties And Failure Strength Of Zone II Flexor Digitorum Profundus Repairs? Revisiting the Work of Wade et al (1986)
XXIII Congress of the International Society of Biomechanics. Brussels, Belgium, June 3-7th, 2011.

Zetlitz E, Wearing SC, Nicol AC, Wearing SC**.**
Time-dependent Conditioning Effects are Important when Evaluating the Gliding Resistance of Flexor Tendon Repairs
6th World Congress of Biomechanics, Singapore, 1-6th Aug 2010

Zetlitz E, Hart AM, Wearing SC, Watson SB & Nicol AC. **Internal work of flexion for a novel suture repair used in flexor tendon surgery.**
XXII Congress of the International Society of Biomechanics, Cape Town, South Africa, 5-9th July 2009.

Zetlitz E, Hart AM, Wearing SC, Watson SB & Nicol AC. **A learning curve for flexor tendon repairs: the value of formal training**.
13th Congress of the Federation of European Societies for Surgery of the Hand, Lausanne, Switzerland, 19th–21st June 2008. (Poster presentation)

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Abbreviations

FDP	Flexor digitorum profundus
FDS	Flexor digitorum superficialis
FPL	Flexor pollicis longus
MK	Pennington modification of the Modified Kessler
LMK	Ventral locking loop modification of the MK
PCL	PolyCaprolactone
MCP	Metacarpalphalangeal
MCPJ	Metacarpalphalangeal joint
PIP	Proximal interphalangeal
PIPJ	Proximal interphalangeal joint
DIP	Distal interphalangeal
DIPJ	Distal interphalangeal joint
A1	First annular pulley
A2	Second annular pulley
A3	Third annular pulley
A4	Fourth annular pulley
UTS	Ultimate tensile strength
WOF	Work of flexion
WOFint	Internal work of flexion

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Chapter 1 Introduction

Flexor tendon repair techniques have evolved as detailed information regarding the anatomical considerations and the internal structure of the tendon has become known. This is especially the case for zone 2 repairs that have received a lot of attention since Bunnell introduced the term “no man`s land” in 1934 (1). Throughout the 19th century primary repair of flexor tendons in zone 2 was advised against due to poor results mainly as a result of adhesions.

It still remains a technically demanding procedure, requiring the surgeon to have in depth knowledge of the anatomy of the hand and the implications the trauma and surgical repair will have on the biomechanics of the hand post-operatively.

A ventral locking loop (LMK) modification (section 5.3) of the Pennington modified Kessler (MK) repair has been used in clinical practice by two local Plastic and Hand Surgeons resulting in lower repair rupture rates then the MK repair (2) .

The primary aim of this work is to evaluate the tensile strength and work of flexion of a new core suture configuration, the ventral locking loop (LMK) modification (section 5.3) of the Pennington modified Kessler (MK) repair. Both 2 and 4 strand core sutures are evaluated in light of the on-going debate regarding the number of core suture strands required to withstand various active rehabilitation regimes (3). A new peripheral repair method, incorporating a sheet of Polycaprolactone with nano-grooves for tenocyte migration, is also evaluated for biomechanical properties.

The body of the thesis is broadly divided into a set of 3 phases, evaluating the primary and secondary aims of the project. The primary aims refer to experimental work comparing the two core suture repair techniques and the new peripheral repair method. The secondary aims of the work refers to the methodological issues raised during the project, like the setup of the testing protocol, the suitability of the porcine model, the effect of learning effect on the biomechanical properties of the repairs and the effect of conditioning of the tendon prior to WOF testing.

In the first phase of the PhD program, a preliminary study was undertaken to determine the suitability of porcine tendons as an anatomical model for human tendon repairs. A comparison of two strand MK and LMK, (Primary aim), was also investigated early on to evaluate the biomechanical testing model, as well as a potential learning curve in a trainee and the findings of this is reported in chapter 6. As a result of this preliminary work, the second phase of this thesis gives a detailed evaluation of the technical limitations of the biomechanical testing model as an analysis tool (Secondary aims). In particular the physical constraints imposed by the equipment were noted to influence the gliding resistance of the repair, which in turn may have limited the ability to generalise the experimental findings. As a result, the study outlined in Chapter 5.4.1.1 assessed the influence of temporal constraints on the acquisition of data, while the experiment conducted in chapter 5.4.1.2 addressed the effects of spatially constraining the repair in the flexor sheath. The third phase of the study describes the application of these techniques in evaluating both the two strand repairs (Chapter 5.5.1) and the novel four strand core suture repairs as well as the new peripheral repair method (Chapter 5.5.2).

Chapter 2 of this thesis presents the available current literature of tendon repairs, as well as the anatomy, healing process and testing protocols reported in the literature. The third chapter includes other repair methods investigated early on in the project, the use of Barbed sutures and Tisseel fibrin glue.

The findings of these studies are collectively discussed in Chapter 7, together with the limitations of the work, with the conclusions and future work that may be needed highlighted in chapter 8. For completeness, additional publications by the author have been cited within the body of the PhD, but provide only ancillary information regarding methodological aspects of the main studies.

Chapter 2 Literature Review

2.1 Tendon Injuries

Flexor tendon repair is one of the most common procedures in hand surgery, but carries a high complication rate, up to 17% (4-6), and remains a difficult procedure for surgical trainees to learn. This is exacerbated by varied clinical exposure to the problem, and current directives to shorten training periods and reduce working hours. There is also an emerging requirement for objective, structured assessment to support competitive-entry into, or competency-based progression in training.

Hand injuries account for 10 % of patients presenting to emergency rooms in the US. Of these, 61% are lacerations with 34.6% involving the flexor tendons (7). The incidence is reported to be one in 7000 people in industrialised countries (8). The epidemiology of flexor injuries show that most are located in zone II, and for open injuries the average age is 25-30 years (9-14). Men are four times more likely than women to sustain this injury, and the little finger seems to be the digit most commonly injured, with 60% of injuries occurring in the dominant hand (9-14).

In 1990, a total of 573 flexor tendon repairs were carried out in Norwegian hospitals (8). Although current figures on the financial burden imposed by flexor tendon injuries are not available, each repair was estimated to cost the Swedish health care system as much as SEK 153,458 (~£12,950) in 1998 (15). Thus, in Norway alone, flexor tendon injuries likely cost in excess of £7.4 million per annum. Rupture rates of repaired flexor tendons vary according to the suture and repair type used as well as the postoperative mobilisation regime used, and it has been as high as up to 17% in some reports (4, 11, 12, 16, 17). Dy et al. carried out a retrospective review of patients treated for flexor tendon injuries in the New York area in an eight year period and found that there was re-operation rate of 6% (18). This will increase the costs of the injury further due to subsequent surgical revision, extended rehabilitation times, greater hospital costs and an increased number of hours lost from work. A repair that is strong and smooth, performed in an atraumatic manner is thought to prevent complications, but the “perfect” repair has yet to be discovered. This means that surgical technique as well as repair type needs to be consolidated as a way of ensuring that each patient receives best possible treatment.

To date, in Norway and the UK there is no structured training for trainee surgeons to practice tendon repairs. Training in flexor tendon repair has traditionally involved observation of seniors, perhaps with practice on models such as dental rolls (19), silicon (20) or manufactured simulators (21), before performing repairs under supervision. However, the curtailment of training periods and reduction in working hours dictated by the European Working Time Directive has minimised the opportunity today’s trainees have to learn by the traditional apprenticeship model (22). For good governance, reduced training opportunities also necessitate increased in-training assessment to validate competency-based progression.

2.2 Gross and Histological Anatomy of the flexor tendons and pulley system

2.2.1 Macroscopic anatomy of the flexor tendons and pulley system

The hand is usually described as the part of the upper limb that is distal to the forearm. It is the most flexible part of the skeleton. It has a total of 29 bones, 8 forming the carpus, 5 forming the metacarpus, 14 forming the phalangeal section, and 2 sesamoid bones (23). The skeleton is held together by an extensive network of ligaments, and the flexion of the phalanges is mainly attributed to the flexor digitorum profundus (FDP) and flexor digitorum superficialis (FDS) muscles, although some of the movement is aided by the intrinsic muscles.

Based on the early work by Verdan (24) the hand can be divided into five anatomical zones, starting distally and moving proximally. These zones were initially used to describe the general prognosis of flexor tendon repair and healing in the areas. The zones, which are illustrated in Figure 2.1, involve only the lesser digits and do not include the thumb which has its own prognostic divisions.

Figure 2.1. Surface markings showing Verdan's 5 zones of the hand.

•Zone I extends from the insertion of the FDP on the distal phalanx to the insertion of the FDS proximally. It contains only one tendon, the FDP.

•Zone II extends from distal palmar crease superficially, or proximal border of the A1 pulley, to the insertion of FDS on proximal part of the middle phalanx. It contains the division of FDS into 2 slips and the FDP tendon. At the point of FDS division, the FDP tendon moves through Campers Chiasm (25) (Figure 2.2), and goes from a deep to a superficial tendon.

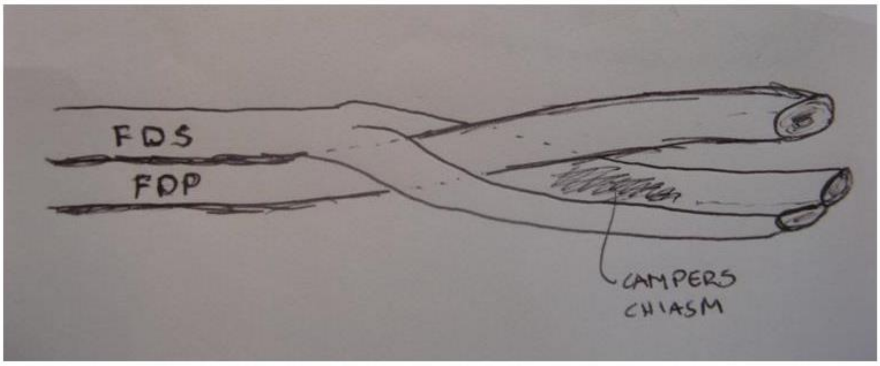

Figure 2.2. Camper’s Chiasm.

This zone was previously named “No Man`s Land” by Bunnell (24), due to the complex anatomical arrangement and interaction between the FDS, FDP and the tendon sheath with its pulleys. It is the zone that provides the biggest challenge for successful results after flexor tendon repair.

A further division of zone II into subzones 2A-2D (Figure 2.3) has been suggested by Tang (26).

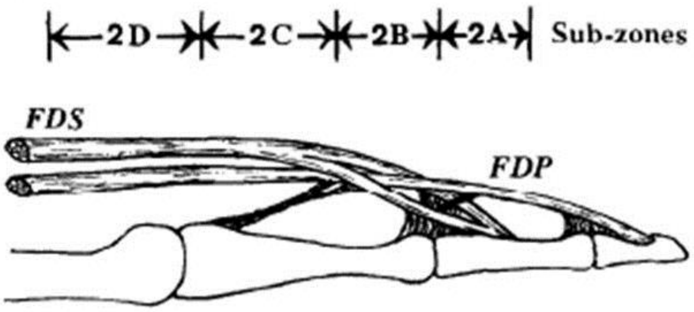

Figure 2.3. Subdivision of Zone II as described by Tang (26), (Copyright Journal of Hand Surgery (Am). Reprint permission obtained).

•Zone III — commences at the carpal ligament proximally and ends at the proximal aspect of the A1 ligament, which governs the start of zone II. Here both FDP and FDS tendons are present, FDP lying deep to the FDS.

•Zone IV — denotes the area of the carpal tunnel. It contains the median nerve, and 9 finger tendons, the FDS and FDP to digits 2-5 as well as flexor pollicis longus tendon (FPL).

•Zone V — starts at the musculotendinous junction of the flexor tendons in the forearm and continues to the start of the carpal tunnel.

2.2.1.1 Flexor digitorum superficialis and flexor digitorum profundus

The function of the digital flexor tendons is to transmit tensile forces between the muscle and the bone, in order to move the finger. The FDS, FDP and FPL muscles originate in the proximal part of the forearm, and their musculotendinous junctions are located in the distal forearm. The FDP originates from the upper three-quarters of the anterior and medial surfaces of the ulna, the interosseous membrane and deep fascia of the forearm (23). The FDS which is classically described as having two muscle heads; the humero-ulnar and radial, originate from the medial epicondyle of the humerus (the common flexor tendon origin), the ulnar collateral ligament of elbow and the coronoid process of ulna. The radial head, in contrast, originates at the radial tuberosisty, extending from the oblique line to the point of insertion of the pronator teres. The origins are interconnected by a layer of fibrous aponeurosis that covers the median nerve and the ulnar artery (23).

The FDP and FDS span the forearm and each divide into four separate tendons, one tendon to each of the fingers of the hand (excluding the thumb). In the carpal tunnel, the FDP tendons lies deep to the FDS tendons but become superficial to the FDS tendons at the level of the proximal phalanx due to the division of FDS tendon. The FDS tendon reunites again on the dorsal aspect and forms the Camper's Chiasm (Figure 2.2), named after the anatomist that described it, Petrus Camper (1722-1789). The FDS then splits again and insert to the base of the middle phalanx (25). The FDP tendon continues distally to insert on the proximal volar third of the distal phalanx. At the level of the MCP joint the shape of the tendon is more or less oval, and it becomes more triangular shaped at the mid-proximal phalanx, with the apex of the triangle located volar. The FDP tendon then continues to span out and separate into 2 bundles at the PIP joint and develops a volar groove. The radial and ulnar bundles are separate but connected by endotenon (27).

2.2.1.2 Neural innervations

The FDP is innervated by the median nerve, nerve roots C8, T1 motor innervations, with the anterior interosseous branch (AIN) supplying the radial half of the muscle. Bhadra et al. (28) reported that the AIN innervates the index and middle finger in 75% of the cadavers studied, and that the ulnar nerve innervates the middle, ring and little finger. Consequently, there are instances in which the FDP to the middle finger has dual innervations (28).

The FDS is innervated solely by the median nerve, nerve roots C7, C8 and T1.

2.2.1.3 Flexor sheath

Once the FDP and FDS enter the carpal tunnel, they are surrounded by a system of synovial and retinacular structures consisting of the transverse carpal ligament, the palmar aponeurosis pulley and the digital flexor sheath. The digital flexor sheath begins at the level of the metacarpal neck or distal palmar crease (29-31).

The transverse carpal ligament was referred to by Klein and Moore in 1992, as an important part of the digital flexor pulley system (32), which acts as the main flexor pulley at the wrist (31). The transverse carpal ligament covers the carpal tunnel and it extends from the carpal bones, where it stretches from the hook of the hamate and pisiform bones to the scaphoid and trapezium. It is a broad and strong fibrous ligament.

The palmar aponeurosis (PA) pulley was first described by Manske and Lesker (33) and consists of an arch of transversely orientated fibres that extend over the flexor tendons. The fibres then attach to the deep transverse metacarpal ligament by vertical fibres. The PA is believed to be important in preserving finger range of motion, after disruption of the first or second annular pulleys (31). These findings were confirmed by Doyle who describes the PA pulley as a form of substitution pulley if the A1 or A2 pulley (Figure 2.4) is lost.

The digital pulley system has been widely investigated. While the system is often described to consist of an orderly and repeatable set of five annular and three cruciate pulleys (Figure 2.4) starting in the distal palm and ending at the distal interphalangeal joint (31), identification of all pulleys is not always possible. The A3 and A5 pulleys are often reported as the most difficult to identify (33, 34). The main function of these pulleys is to maintain the tendon close to the bone/joint and thus improve the axis of motion and preventing bowstringing of the tendons (29, 31, 32, 34, 35). Doyle investigated the functional adaptations of the pulley system and found that the arrangement of annular and cruciate pulleys ensures significant flexion of the fingers without buckling of the retinacular system or impingement of the flexor tendons (31). The annular pulleys comprise of thicker areas of arching fibres, while the cruciate pulleys consists of thin, flexible areas of crisscrossing fibres (31).

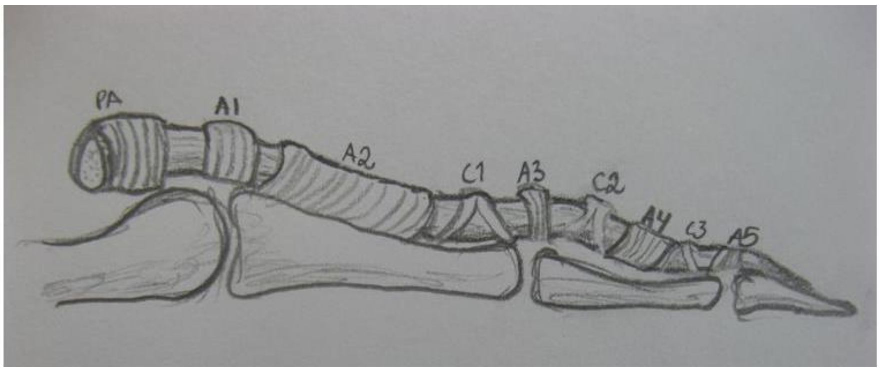

Figure 2.4. An illustration of the pulley system of the hand.

The annular pulleys:

•A1: Originates from the palmar plate and the base of the proximal phalanx. It spreads over the surfaces of the membranous sheath at the metacarpalphalangeal (MCP) joint.

•A2: This pulley originates from the proximal half of the proximal phalanx. The pulley fibres are arranged with oblique fibres superimposed on the annular fibres at the mid point of the pulley. This is the point where a blood vessel (vinculum) enters the digital pulley system in 90% of specimens investigated by Lin et al (29).

•A3: The third annular pulley originates from and attaches to the volar plate at the proximal-interphalangeal PIP joint.

•A4: Situated at the mid-point of the middle phalanx. It originates from the mid portion of the middle phalanx. Its composition is similar to that of the A2 pulley with oblique fibres superimposed over ring-shaped or annular fibres.

•A5: The most distal of the pulleys, the A5 pulley is relatively thin and overlies the distal interphalangeal joint (DIPJ). It originates from the volar plate of the DIPJ and its end marks the termination of the digital pulley system.

There are 3 cruciform pulleys that were recognized by Doyle (31), however, it should be mentioned that other authors describe a C0 pulley when there is a pliable, thin interspace between the first and second annular pulley (34, 36).

•C1: The first cruciate pulley is located at the distal end of the A2 pulley.

•C2: The second cruciate pulley is found between the A3 and A4 annular pulleys of the middle phalanx.

•C3: The third cruciate pulley sits at the distal end of the A4 annular pulley.

The pulleys various composition and size means that they have separate functions within the flexor tendon sheath. The thicker annular pulleys are located over the shafts of the bones, while the thinner and fine cruciform pulleys are located over the joints to allow for flexion (31). Doyle also describes an “anatomical accommodation” between the A1 and A2 pulleys. This is either in the form of a true separation between the pulleys, a thinning of the distal margin of A1 and proximal margin of A2, or triangular shaped windows on the lateral aspects of the pulleys to allow for flexion (31).

2.2.2 Structure and biochemical composition

2.2.2.1 Tendon

The hierarchical structure of the flexor tendons are similar to that of other human tendons, consisting of mainly of collagen I. The collagen is packed firmly in a parallel arrangement, along the length of the tendon. The tendon has a cellular and a non-cellular component. The cellular component consists mainly of fibroblasts and tenocytes that produce collagen, as well as reorganize the extra-cellular matrix. The non-cellular material is mainly composed of collagen (type I), water, proteoglycans and elastin (37-41). Each of the components can vary in their proportion, and this is recognised in reports where the constituents rarely add up to 100% (42). About 70% of the tendon is water, while the collagen content ranges from 60-85% of the dry weight of the tendon. The elastin content is about two percent, as is the cellular content, while proteoglycans only make up around one percent (37-42).

At the smallest level, procollagen is formed by cross linking of three chains of amino acids into a triple helix pattern, by covalent and hydrogen bonds. The amino acid glycine occupies the middle section of the helix and is vital for the precise folding of the triple helix, and this helical arrangement is a major part in the molecules ability to resist tension (37, 39). The process of procollagen formation occurs intracellularly, in the endoplasmic reticulum (39). The cell then releases the procollagen into the extra-cellular matrix. These molecules are soluble and contain an amine group ($NH_{2)}$ and carboxylic acid group (COOH) at each end of the molecule. Enzyme activity then breaks the bonds between the procollagen and the NH_2 and the COOH ends, thus producing tropocollagen (38, 39).

The molecule size is now about 300nm in length and 1.5nm wide. The tropocollagen molecules, via end to end fusion, then assemble into fibrils (39, 40). The fibrils gradually increase in both length and diameter, but remain the smallest structural unit. The fibrils are not in a straight arrangement, but follow a zig-zag pattern or “quarter stagger array”, and their diameter varies from 10nm to 500nm (43). The fibrils are referred to as crimped fibrils and follow a circular or elliptical shape (43). This crimped pattern allows a 1-3% elongation of the tendon (44). The fibrils pack in parallel bundles with proteoglycans and water to form a fibre (Figure 2.5). The fibres are covered by endotenon, and they then group to form fascicles. The fascicles are also covered by endotenon, a loose connective tissue supporting nerves and vessels. The septa of the endotenon join together to form a fine fibrous outer layer, called the epitenon or visceral layer (37, 40, 41).

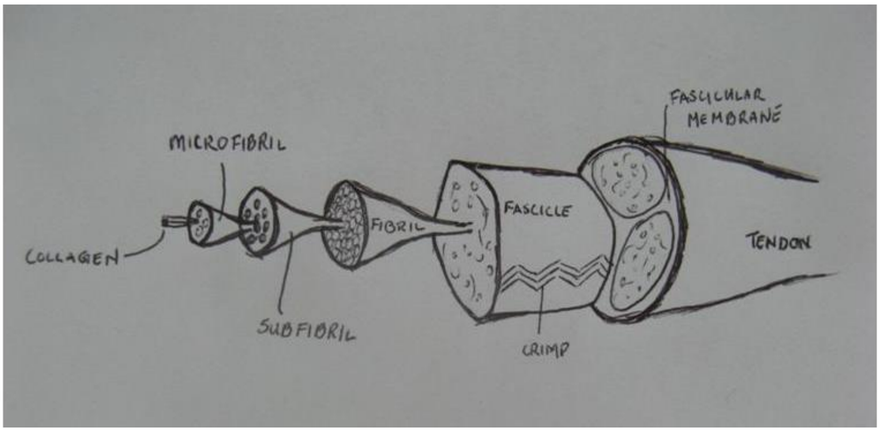

Figure 2.5. Schematic of the hierarchical structure of tendons

While all tendons share the same overall hierarchical structure, the relative proportion of their constituents differ according to their functional role and they have been shown to present with subtle histological differences along their length (45, 46). Three tenocyte subgroups have been identifies based on their histological nuclear morphology. The first are long thin type 1 cell second have slightly thicker cigar shaped nuclei whilst the third group are more condrocyte like. Tenocytes close to the frictional surface of pulleys are rhomboidally shaped, have a more chondrocyte like appearance, than the elongated appearance of tenocytes from tendon mid substance (47, 48). Fibroblasts are the main cells of the tendon, and within the tendons they are usually referred to as tenocytes. The main function of the tenocytes is maintenance of the extra cellular matrix in response to mechanical stimuli, through the synthesis of collagen and proteoglycans (39-41). The tenocytes align in the direction of the collagen to which they attach, and connect for the most part to one single fibre of collagen. It is thought that the tenocytes themselves do not contribute to the mechanical properties of the tendon, but there is no data to date confirming the amount of strain the tenocytes are exposed to (49).

Areas with different mechanical load have also been found to have variations in the extra cellular matrix (50-52). The dorsal aspect of the tendon where it is exposed to more tension and distortion has a higher rate of collagen synthesis and is therefore more adapt to tensile forces and more elastic (52). The collagen fibrils are thicker in this area and have reduced proteoglycan content (52). The opposite is seen on the volar aspect, where the tendon is in close relationship to the pulley system and is exposed to compressive forces. Here the tendon has greater proteoglycan content and water, and lower collagen content compared to the dorsal aspect of the tendon.

The most common, large proteoglycan in tendon is aggrecan. It can, via a glycoprotein, bind to hyaluronan (HA) (38). This molecular complex is important for the maintenance of osmotic pressure and hydration of the tendon. The highly charged side chains of the complex are hydrophilic, which results in swelling that enables the collagen matrix to resist compression (39, 53). Aggrecan also reduces the viscosity of the tendon in regions of tension. It acts to lubricate the fibrils, allowing them to slide over each other, and allows stretch response to sudden loading (53, 54).

Lubricin is a GAG protein that has recently been linked to tendon gliding. It is believed to have a key role in the natural surface boundary lubrication of tendons, and when it is missing, it results in increased intra-synovial adhesions syndromes. In a combination with HA, lubricin was found to improve gliding characteristics in an vitro in a study using human extra synovial (Palmaris longus) tendons (55). While animal models have also shown that addition of HA and Lubricin to the tendon during repairs results in fewer adhesions, in vivo work has suggested the combination had a negative effect on tendon healing and lead to reduced tensile strength of the repairs (56).

2.2.2.2 Flexor sheath

The digital flexor sheath has been described as a synovial lined fibro-osseus tunnel (35, 57). Amis and Jones have shown that the digital flexor sheath has a thin synovial lining in which pulleys are formed, and that the thickness of these areas/pulleys, cause them to stand out of their environment. The pulleys are collagen reinforcements of the thin synovial sheath (30). Surprisingly, this does not attach to the edges of the pulleys, but rather to their external/superficial surface, causing an overlap or a pocket, and a leading free edge of the pulley (Figure 2.6) (30).

Figure 2.6. Picture of the free edge of a pulley, from Amis and Jones 1988 (30). (Copyright The Journal of Bone and Joint Surgery Br. Reprint permission obtained)

The composition and number of layers within a pulley is controversial. Some authors have found what they believe to be three layer composition of the annular pulleys, layers that are not only different in structure but also in function (58). The inner layer is composed of a layer of fibroblasts that are thought to be responsible for the secretion of hyaluronic acid, and is similar to the surface layer of the FDP (57, 58). The distinction between fibroblasts and synovial cells, however, is hard to make in the absence of biochemical testing (58). The function of this sheath is to provide a low friction surface and allow optimal gliding conditions for the tendon, to provide the tendon with a fluid medium promoting nutrition, and the pulley portions ensure that the tendons are in constant relationship with the joint axis to provide efficient flexion of the digit (36). Chondrocyte like cells are seen within the pulley at high pressure areas, where tendon meets the pulley at time of digital flexion (59).

2.3 Tendon Vasculature and Nutrition

2.3.1 Blood supply

The blood supply to the flexor tendons and the digital flexor sheath was first described by Mayer in 1916 (60). This description and that of other authors describe three types of vascular supply originating from:

1) Musculotendinous junction,

2) The bony insertion or enthesis and

3) The mesenteric vincular vessels (60-62).

Zhang et al. described the blood supply with regards to non-synovial regions and synovial regions of digital flexor tendons after an injection study of human cadaver arms (63). Their study investigated intratendinous and extra-tendinous vascular sources. The intratendinous system stems from the musculotendinous junction and the enthesis, and run longitudinally in the interfascicular connective tissue of the tendon. They have short transverse anastomotic branches around the tendinous fascicles, which connect to the other longitudinal vessels, forming a vascular network. The vessel diameters are in the order of 10-25 μm for the longitudinal vessels and 25-40 μm for the transverse anastomosis (63). The extratendinous system refers to the surrounding tissue relating to the different parts of the tendon.

In the extra-synovial tendon the extratendinous system comes from the vessels of the microvacuolar system, previously called paratenon or loose connective tissue layer between the tendon and the skin (64). These vessels enter through the interfascicular grooves on the surface of the tendon and connect to the intratendinous longitudinal blood vessels. Zhang et al. found that there was generally even distribution of vessels on transverse section in this area (63).

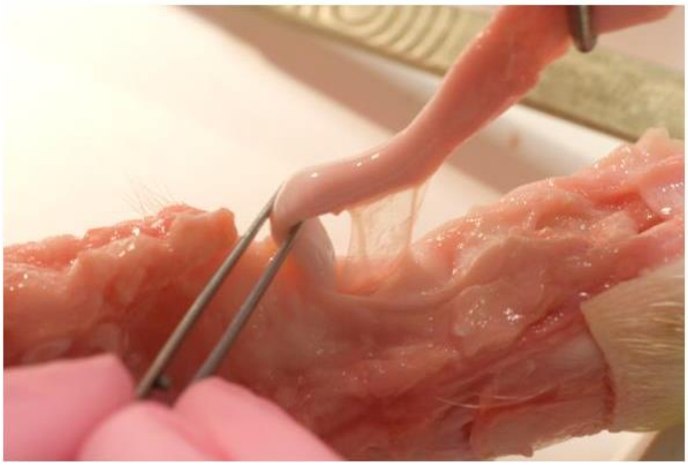

Figure 2.7. The vincula longa entering the dorsal aspect of the FDP tendon in the porcine model.

In the synovial part of the hand, including zone II, the extratendinous blood supply is derived from the mesotendon and its vincular vessels, vincula longus and brevis (one artery and two accompanying veins) (63). The vincula vessels attaches to the tendon on the dorsal side (figure 2.7), and the vessels then spread in a proximal and distal direction. These vessels, run along the interfascicular grooves on the dorsal side or the lateral sides of the tendons (Figure 2.8) resulting in a fine network of branches proximally and distally. These branches connect with the intratendinous system forming looped vascular nets. Based on the work of Zhang et al., it is clear that the numbers of intratendinous blood vessels are small in the synovial part of the hand. They feed from the vessels from the musculotendinous junction as well as from the bony insertion or enthesis. Moreover, the blood vessels in this area are primarily situated on the dorsal side of the tendon (Figure 2.8), with about one third to one-half of the volar aspect of the tendon devoid of vessels (63). Zhang et al. identified the segmental blood supply as originally proposed by Mayer (60) and Lundborg (65), and reported that in the non-synovial areas of the hand there was a greater number of intratendinous vessel. In the synovial regions, partial and segmented patterns were identified. The area of avascularity in the FDP involved the PIP joint, while in the FDS the MCP joint was relatively avascular (60, 61, 63, 65). This distribution may reflect the different loading conditions experienced at different sites of the tendon. The accumulation of the vessels on the dorsal side of the tendon prevents their compression during digital flexion; as compression is localized on the volar aspect of the tendon and inner aspect of the pulleys. At the interphalangeal joints, the vessels of the vinculi branch transversely and are thought to ensure blood flow to the tendon is maintained during full digital flexion (66, 67).

The digital flexor sheath is a highly vascular structure, with a well developed vascular plexus (59, 68). The frictional surface of the pulleys has no vessels, so gliding of the tendon in the flexor sheath occurs between 2 avascular structures (59).

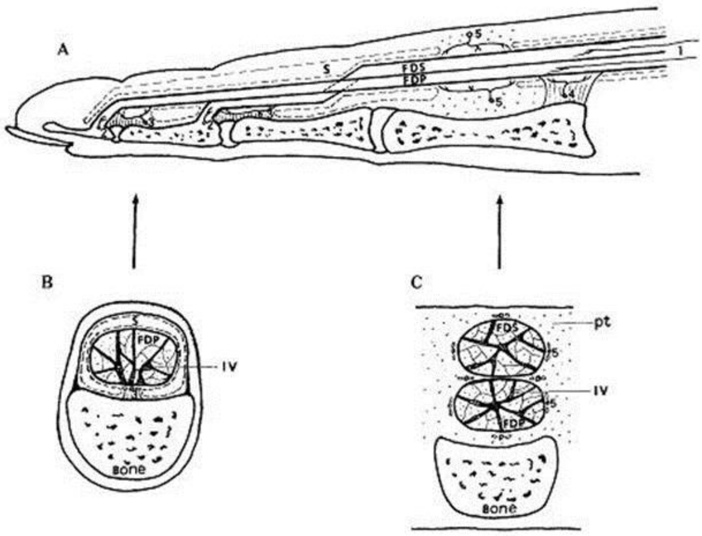

Figure 2.8. A diagram of the intratendinous blood supply of the flexor tendons of the hand from Zhang et al. (63) (Copyright Surgical and radiologic anatomy. Permission for reprint obtained.)

2.3.2 Synovial fluid

Although tendon nutrition is primarily supplied by the vascular system, Manske et al. (69), using a hydrogen washout technique, demonstrated that the tendon healing occurred despite complete disruption of vascular sources. Consequently, synovial diffusion has been hypothesized by many authors to play an important role in the delivery of the necessary nutrients to the tendon (63, 68, 69).

The effect of flexion on the tendon and flexor sheath system is to increase interstitial pressures which results in the extrusion of free or unbound fluid from the tendon. As the digit extends, the interstitial pressures are reduced and imbibition of fluid is free to occur (70). This process seems to complement the vascular perfusion of the tendon and thus delivering enough nutrients to the tendons. It resembles a synovial joint with proteins diffusing into tendon (59). Hence cyclic tensile loading of tendon results in a pumping mechanism, which has been suggested to enhance convective transport of nutritional and growth related factors. Although there is evidence that the load-induced fluid movement does not necessary influence the diffusion of small solutes, such as glucose, in the tendon (70), studies evaluating convective transport in cartilage and the vertebral discs suggest that fluid shifts induced by mechanical loading enhance the penetration of large solutes (in the order of 40kDa) (29, 34).

The synovial fluid is an ultrafiltrate of blood plasma, free from large proteins, and enriched with locally produced molecules like HA, chondrontin, glucosaminoglycans (GAGs) (71, 72). The synovial fluid aids not only in providing important nutrition but also in lubricating the tendon pulley interfaces (37, 68). A fixed lubricating glycoprotein, lubricin, has been identified on the surface of flexor tendons as well as in cartilaginous surfaces (73). Aggrecan, a high molecular weight glycoprotein, has also been identified which acts to separate the collagen fibrils and make the tendon surface more resistant to compression (73, 74). While lubricin has been shown to lower friction between the tendon and pulley when surgically applied to the tendon surface and reduce adhesion formation, it has also been shown to adversely impact the healing of tendons (56). (See section 2.2.2.1). The synovial fluid in the flexor sheath is similar to synovial fluid found in cartilaginous joints (71, 75).

2.4 Biomechanics of the Flexor Tendons

2.4.1 Biomechanical Testing Methods

Testing of the biomechanical properties of flexor tendons and their repairs involves many testing protocols using various models (37) (Table 2.1). Biomechanical testing establishing the ultimate tensile strength of repaired tendons is primarily done in a linear testing machine, where the tendon and repairs are exposed to a single linear load to failure pull (37). This methodology also allows the investigators to determine the force required to produce a gap between the tendon ends and subsequent failure of the repair. Static linear testing, *ex vivo*, is the most frequently used biomechanical method (37).

Table 2.1. Various testing methods and models reported in the literature.

Study	Model	Tendon	Machine	Test type
Miller 2007 (76)	Human	FDP	MTS	Static
Cao 2006 (77)	Porcine	FDP	Instron	Static
Cao 2005 (78)	Porcine	FDP	Instron	Static
Lawrence 2005 (79)	Porcine	FDP	Instron	Static
Tang 2005 (80)	Porcine	FDP	Instron	Static
Xie et al 2005 (81)	Porcine	FDP	Instron	Static
Xie & Tang 2005 (82)	Porcine	FDP	Instron	Static
Su et al 2005 (83)	Human	FDP	MTS	Static
Dona et al 2004 (84)	Sheep	FDP	MTS	Static
Tan & Tang 2004 (85)	Porcine	FDP	Instron	Static
Tanaka et al 2004 (86)	Human	FDP	MTS	Static
Tang et al 2003 (87)	Porcine	FDP	Instron	Static
Merrel 2003 (88)	Human	FDP	Custom	Static
Wang et al 2003 (89)	Porcine	FDP	Instron	Static
Angeles 2002 (90)	Human	FDP	Custom	Dynamic
Cao et al 2002 (91)	Porcine	FDP	Instron	Static
Xie et al 2002 (92)	Human	FDP	Instron	Static
Labana et al 2001 (93)	Human	FDP	Instron	Static
Slade et al 2001 (94)	Human	FDP	Custom	Static
Smith & Evans 2001 (95)	Porcine	FDP	Instron	Static
Tang et al 2001 (96)	Human	FDP	Instron	Static (curvilinear)
Taras et al 2001 (97)	Human	FDP	MTS	Static
Barrie et al 2000 (98)	Human	FDP	Custom	Static
Dinopoulos et al 2000 (99)	Canine	FDP	Instron	Static
Hatanaka & Manske 2000 (100)	Human	FDP	Instron	Static
Wada et al 2000 (101)	Canine	FDP	Maruto MZ-500s	Static
Gordon et al 1999 (102)	Human	FDP	Instron	Static
Hatanaka & Manske 1999 (103)	Human	FDP	Instron	Static
McLarney et al 1999 (104)	Human	FDP	Custom	Static
Gordon et al 1998 (105)	Human	FDP	Instron	Static
Lotz et al 1998 (106)	Human	FDP	MTS	Static
Stein et al 1998 (107)	Human	FDP	not stated	Static
Hotokezaka & Manske 1997 (108)	Human	FDP	Scott Tensile Testing machine	Static
Shaieb & Singer 1997 (109)	Rabbit	FDL	Instron	Static
Pruitt et al 1996 (110)	Canine	FDP	Scott TensileTesting mach	Static
Aoki, Pruitt et al, 1995 (111)	Canine	FDP	Scott TensileTesting mach	Static
Soejima et al 1995 (112)	Human	FDP	MMED testing machine	Static
Aoki et al 1994 (113)	Human	FDP	Scott TensileTesting mach	Dynamic
Noguchi et al 1993 (114)	Human & Canine	FDP	Instron	Static

However, physiological loading of a tendon repair is not reflected in static linear testing, as the result of repetitive physiological loading of the repair cannot be evaluated. It is a useful tool however for the testing of a large number of tendons to observe the repair site directly to record gap formation. It offers a method of testing that allows for easy comparison of the strength of the repairs and their mode of failure, and minimises other variables affecting repair strength (37).

Dynamic linear testing allows testing of repetitive loads on the strength of the tendon repair, thus providing a more physiologic evaluation of the postoperative mobilisation (115-119). With this type of testing the tendon repair is loaded cyclically, and has resulted in the formation of gaps at the repair site at significantly smaller loads than with static testing (116, 120). Dynamic testing protocols investigate the remaining strength of the tendon repair after a known number of cycles simulating the postoperative mobilisation regime where the repair is exposed to successively higher loads during the rehabilitation. The number of cycles varies from study to study with some investigators using 100 cycles per tendon (121), and others as many as 3000 cycles (122). It is a time consuming method and if large number of repairs are evaluated the practicality of the set up can be problematic, depending on the number of cycles each tendon repair is tested for.

During the postoperative mobilisation the tendon repair also undergoes angular pull as it passes under pulleys in the tendon sheath. A curvilinear tensile testing model, where the tendon repair passes under a single pulley has been used by some authors to investigate how the direction of tension influences tendon strength (87, 91, 96, 123).

Curvilinear tensile testing methods have found that angular pull testing produces gap formation at lower loads than the linear testing protocols (87, 96). This method does not include all the factors involved in finger motion and separates the pulley from the rest of the digit and sheath before testing. It cannot be used to evaluate influence of any adhesion formation.

2.4.2 Tensile Strength

2.4.2.1 Intact Human Digital Flexor Tendon

The biomechanical properties of the flexor tendons mirror those seen when testing collagen only (37). The mean breaking strength of normal healthy human FDP was investigated *in vitro* by Pring et al, who found that the FDP tendons fail by avulsion of the tendon from the bony insertion on the distal phalanx (124). They report a avulsion strength of 558 ± 69 N, and a strength of the tendon itself of 1175 ± 245 N, and a strain measurements of the tendon of 10-13% (124). The stiffness have been reported to 35.9 N/mm in human FDP tendons (106).

2.4.2.2 Repaired Digital Flexor Tendon

Several biomechanical and anatomical testing models have been used to investigate tendon properties (Table 2.1), as well as a number of different repair techniques. This has resulted in difficulties when it comes to directly comparing the results (37).

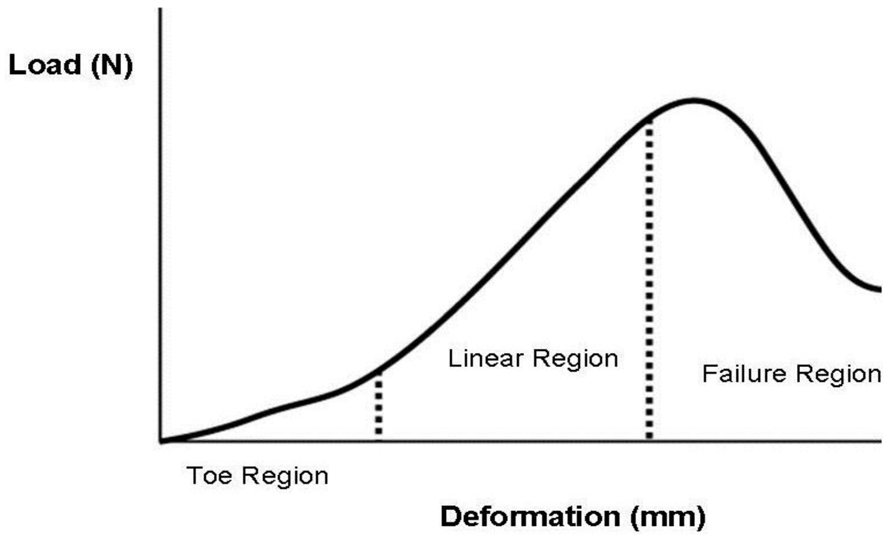

Figure 2.9. An example of a load deformation curve.

The load-deformation curve obtained when testing repaired tendons has 3 areas (Figure 2.9), 1) a toe region, 2) a linear region and 3) failure region (37). The toe region is thought to reflect the straightening of the wavy-crimp-like pattern of the elastin fibres (38, 40, 41, 43). Macroscopically it is the point where the tendon starts tightening, before stress starts (37). The linear region is believed to reflect elongation of collagen for a given load and the slope of the curve represents the stiffness (N/mm) of the tendon or repair, the tendons ability to resist deformation during loading, figure 2.10 (37). The third region is the failure region where permanent and irreversible changes occur (37). It includes the yield point, were the tendon continues to elongate at the same rate despite a decrease in load when under displacement control, and the ultimate failure point, where the tendon’s integrity is no longer intact.

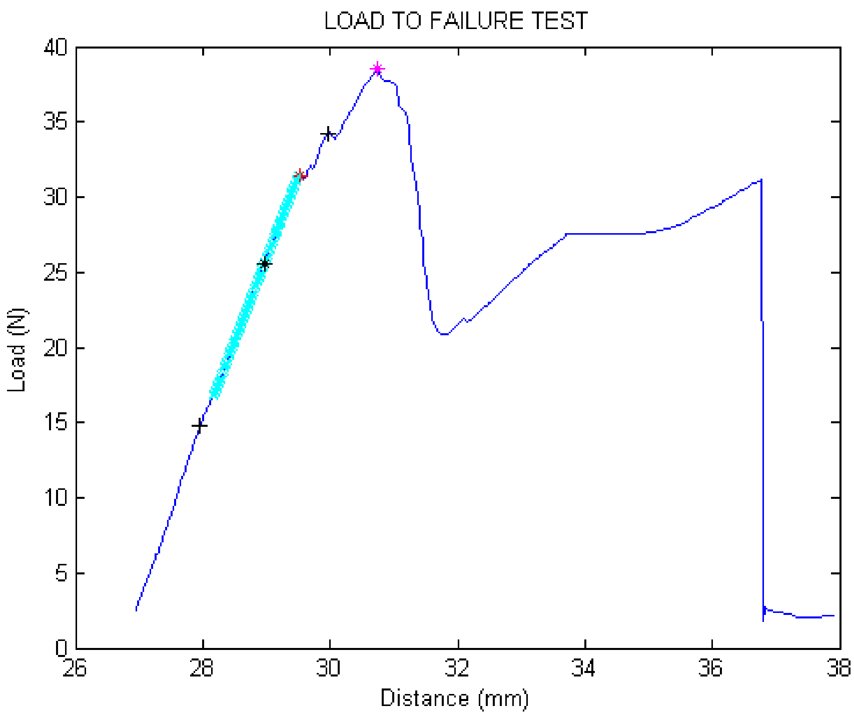

Figure 2.10. Example of a load deformation curve (16LMK3RP). 1mm, 2mm and 3mm gaping is indicated by +, *, + respectively. * indicates yield point. * indicates ultimate tensile strength. Thickened light blue line represents the tangent stiffness.

2.4.3 Tendon excursion

The concept of tendon excursion illustrates the distance the tendon moves along its path, and also the distance the tendon has from the axis of rotation of the joint. The pulleys, especially the annular pulleys, ensure that the tendon is close to the flexor sheath and thus prevent bowstringing of the tendons during flexion of the digits. Brand et al. described 3 different divisions for tendon excursion: potential excursion, required excursion and available excursion (125). The potential excursion of a tendon refers to the resting fibre length of the muscle and is independent of any connective tissue limitations. The required excursion is the maximum excursion for the muscle *in situ* and the available excursion refers to the maximum excursion of a muscle when its freed from its insertion (125).

Any factors that will increase the distance between the flexor tendon and the joint, like loss of a pulley, will increase the excursion required by the tendon to achieve the necessary joint rotation (37).

Several groups have looked at tendon excursion during active and passive movement, in different zones of the hand and in various anatomical models (126-129). The reports give measurements of tendon excursion that range between 10 and 33 mm during an active rehabilitation regimen, while tendon excursions for a passive regimen range between 1 and 21mm (126, 127, 130). Tendon excursion measurements performed *in vitro* in zone V by Brown and Peimer give an excursion of intact FDS of 30.21 ± 5.05 mm, and 26.12 ± 2.96 mm for FDP (131) . Tanaka et al. looked specifically at the excursion of the FDP tendon of the index finger in Zone II and reports it to be 20.9 mm (127). Korstanje et al. investigated the excursion of the flexor tendons *in vivo* using ultrasound (zone V), and report findings of 24 mm for active movement and 14 for passive movement (130). McGrouther and Ahmed measured excursion in human FDP with regards to various joints. They exposed the flexor tendon sheath and flexor tendons and stimulated active movement of the fingers by pulling on the tendons distally, and measuring flexion at the various joints (126) found that excursion of FDP in relation to DIPJ was on average 1.0 mm for every 10 degrees, and 1.3 mm in relation to PIPJ (126). The wide variety of measuring techniques as well as specimens may account for the differences reported, which makes a "gold standard" approach difficult to identify. For the porcine model excursions of 20mm ±3, have been reported (132). A measurement of the excursion is necessary for the calculation of the tendons biomechanical properties and to ensure that adequate tendon rehabilitation is achieved postoperatively.

2.4.4 Forces during finger flexion

The forces required to flex the fingers have been explored by many authors (123, 133, 134). The range of force measurements found with passive flexion was 2-4 N. With active flexion of the fingers against some resistance the tensile force in the FDP increased to 10 N, while against moderate resistance tensile forces of 70 N were reported. Undertaking the tip pinch grip increased the tensile forces up to 120 N. *In vivo* flexor tendon forces have been investigated at the wrist level of healthy human tendons during carpal tunnel release, (134-137). Urbaniak et al. found that passive flexion-extension of the FDP resulted in a mean force of 2 N, while active flexion with some manual, though unspecified, resistance resulted in 9 N (135) which correlates with the results by Schuind et al. (134). Kursa et al. looked at FDP and FDS force measurements *in vivo* during carpal tunnel release surgery using buckle transducers placed on the tendons to the index finger. At the same time fingertip forces were measured at the rates of 1.5, 5 and 15 N/s. They report a higher force generated by the FDP tendon than by FDS when then same fingertip load is applied (136).

In vivo measurements of the force in uninjured digital flexor tendons shows that the force in the flexor tendons during passive and active finger motion with no resistance ranged from 0.2 to 27 N, and finger motion against resistance up to 5 N generated force, ranging from 2-48 N (137).

Tendon gliding is a measure of the friction that occurs between the flexor tendon and the surrounding flexor sheath in which it moves. No surfaces are without friction, and it is established that when two objects move in relation to each other, energy will be lost to the system in the form of heat.

While FDP tendons are exposed to considerable tensile forces during both passive and active digital movement, digital movement also results in friction or energy loss at the tendon-sheath interface. While the concept of the “work of flexion” (WOF) (Nm), which represents the sum of the forces (N) needed to move the tendon along its excursion (mm)(138), was first described by Lane et al. in 1976, it was not until the mid 1990’s that work of flexion measurements were used to characterise the tendon and tendon repairs gliding resistance (123, 139). Various methods are described for assessing the gliding resistance of tendon repairs in the literature. One method, described by Silva et al. (140) and Winters et al. (141), measures the tendon excursion and the flexion of the fingers during loading, while Aoki et als., (142) method measures the work necessary to move the tendon a certain distance. Uchiyama et al. investigated a testing method that measured the frictional force at the tendon-pulley interface by determining the force differential between the proximal and distal ends of the tendons (123). Although insightful, these initial tests measured the frictional force in relation to only one pulley interface and did not allow for the measurement of overall finger function or the effects of adhesion formation after a tendon repair (143, 144). Zhao et al., developed these methods further by introducing a new devise for measuring work of flexion and gliding resistance in a human cadaver digit (144). The set up described by Zhao et al., included a small load transducer attached at the distal phalanx and the measurement of the force of the proximal tendon with a ring transducer (144).

The total work of flexion (TWOF) characterizes the sum of all the forces that resist flexion of the digit during gliding and, as such, has an internal and an external component. The TWOF depends on the length of excursion of the tendon and the rotation achieved of the digit against a set load (90, 138, 142, 145-147).

The external components include joint stiffness, mass of the digit and resistance of the surrounding soft tissue and the resistance of the antagonist muscles. Post operative swelling of surrounding tissues can also increase the external work of flexion (148).
The internal WOF (WOFInt) is a combination of the surface friction between the tendon or tendon repair and the flexor sheath, and the bulk effect of fitting the tendon/ repair in the tendon sheath (size of the tendon/repair), and any adhesion formation that results after the repair. A friction coefficient for gliding of tendons in zone 2 has been reported to be 0.03 (123). The internal friction between a healthy intact tendon and A2 pulley has been reported to be 0.09 N (123, 149). Zhao et als., method measures WOF at the same time as the resistance between the tendon and the flexor sheath providing researchers with a tool to identify the involvement of the various resisting forces (144).

Investigators also report gliding resistance as a mean change for the tendon over the given excursion in N (86), or as a % change in WOF by using the formula (90, 142):

$$\%\text{ change in WOF} = \left(\frac{WOF_{after} - WOF_{before}}{WOF_{before}} \right) \times 100\%$$

2.5 Tendon Repair

Early tendon repair was viewed in a similar manner to nerve repair, and was performed using fine silk sutures to approximate the tendon edges. Immobilisation either by splinting or percutaneous pins through the tendon was mainstay (24). As evidence has emerged of the benefit of early mobilization/rehabilitation of the repair site, the focus was moved to a stronger repair technique. The Kessler technique is the core suture used most commonly by surgeons and trainees (150-152) (Table 2.2). The Pennington modification of the Kessler repair was introduced by Hatanaka and Manske (Figure 2.11), and has shown to be sufficiently strong for the forces involved in passive rehabilitation, but not those of early active motion, as increased rupture rates have shown (100, 103, 153). Other commonly used repairs listed in Table 2.2.

Table 2.2. Commonly used tendon repairs and features of the repairs. (See Figure 2.12 and Figure 2.14 for schematic).

Method	Strands	Knots	Knot location	Other information
Kessler	2	2	External	
Modified Kessler (MK)	2	1	Internal	
Tajima	2	2	Internal	
Bunnel	2	1	External	
Tsuge	2	1	Internal	Looped suture
Cruciate	4	1	External	
Strickland	4	2	Internal	
Double MK	4	4	Internal	
Savage	6	1	Internal	
Tripple MK	6	3	Internal	
Lee	8	2	Internal	Looped suture

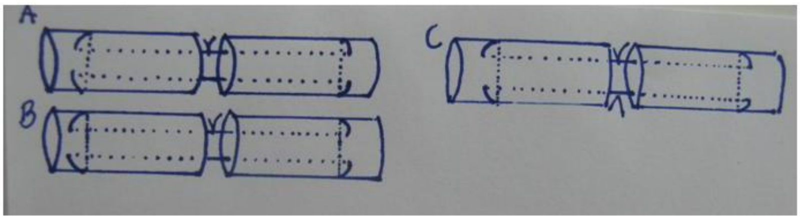

Figure 2.11. Modified Kessler repairs, A. Grasping Modified Kessler (154), B. Locking Modified Kessler or Pennington modification C. Locking MK with two knots (85).

2.5.1 Properties of Tendon Repairs

The ideal tendon repair should, be strong enough to withstand early active mobilization, be easy to perform, should not interfere with tendon vascularity, and ensure a smooth juncture at the tendon ends with secure knots and minimal gap formation (155). The magnitude of force experienced by a tendon repair will depend on the type of rehabilitation regime chosen. Different hand surgery units favour different rehabilitation regimes, ranging from Kleinert's "rubber band" technique to passive and active regimes. It has been estimated that a tendon repair needs to withstand forces that are 50% higher than those seen with healthy intact tendons (156), secondary to post-operative aspects, such as swelling, joint stiffness, damage to the gliding surface of the flexor sheath, addition of suture material to the tendon surface etc, that are likely to increase the gliding resistance (156). Savage recommended an initial tendon repair strength of 5 times that of the tension in profundus tendon (14.7-19 N), which means that the repair strength should be in the order of 73-95 N (157). Measurements in vivo of intact tendon forces by Schuind et al., indicate that tendon repairs should be able to withstand forces from 29-53 N, depending on the individual variations seen with unrestricted active mobilisation

(134). It is also noted that due to the tenomalacia seen at tendon ends after repair, the initial repair strength decreases, but less so with mobilisation (158). Work by Urbaniak et al. and Hitchcock et al. has found that immobilisation of the repair will reduce the tensile strength of the repair further (135, 159). The introduction of postoperative mobilisation has been shown to counter this decrease, this is seen both with early passive mobilisation of the repair (160-162) but even more so with early active mobilisation (146, 159, 163).

A flexor tendon repair contains a core suture and a peripheral suture, both of which contribute to the tensile strength of the repair (88, 106, 164). Studies have found that the peripheral suture fails prior to the core suture (88, 106, 117, 122, 164-166). Lotz et al identified an analytical model that was tested using MK core suture and deep versus superficial peripheral sutures and that the strength of the repair configuration did not change with the different peripheral suture techniques, using a 4/0 core suture and 6/0 peripheral continuous suture (106). They found that once the peripheral suture fails the entire load is transferred to the core suture. Should the strength of the core suture be greater than the force to which the tendon is exposed, the tensile strength measurement may still increase with correlating gap formation (106, 118). The failure of the core suture takes place by either suture pullout , or suture breakage, depending on the suture configuration used, if the holding capacity of the suture on the tendon exceeds that of the suture material (76, 97, 118). Miller et al report that 99% of their repair method, the Massachusetts General Hospital (MGH) repair fail by suture breakage, while 74 % of the Strickland repairs failed by suture pullout (76). With regards to the MK repairs the difference in locking or grasping suture configuration results in different failure mechanisms, with the locking repair failing by suture breakage, and the grasping configuration suture pullout (101) (97, 167, 168).

At the time of surgery, the tensile strength of the repair is mainly related to the properties of the suture material used, the suture knot and the grip of the sutures on the tendon ends. The biomechanical properties of the repair therefore depends on those of the material used, and can be strengthened by increasing the number of the suture strands used (157), as well as the size or calibre of the suture (97, 118). The grip of the suture relies on the suture configuration (80, 101, 118, 157), the grip size (81, 100), and the number of grips (169). These are explored in the following sections.

2.5.2 Core Suture

2.5.2.1 Number of strands

Increasing the number of strands across the tendon ends, to a multi-strand repair was performed by Savage (Figure 2.12, A-H). The biomechanical properties of this repair were investigated using a linear static biomechanical method. His six strand, central core suture repair (Figure 2.12 F) had enhanced gap resistance and ultimate tensile strength when compared to Bunnel, Kessler, Kleinert, modified Kessler and Becker repairs. Savage's results found that his new repair had sufficient tensile strength to withstand the estimated forces for early active mobilization (157). Multiple strands of the central core suture have been investigated by several groups, and four and six strands modified Kessler repairs (109, 170), Savage (170) and the Tsuge (171) sutures all demonstrated higher UTS and force to gap formation. In the Pennington modified Kessler and the Savage repair studies, an improvement was also found in the yield point and stiffness of the repairs, which has been explained by the higher material strength and hold that these repairs have on the tendon (172). Barrie et al. increased the number of strands to 8 and compared various 4 and 8 strand repairs using a linear cyclic testing protocol (117). Increasing the number of strands did improve the tensile strength of the repairs, but did not improve the gap resistance (117). As the number of strands of the repair increase so does the complexity of the technique, the number of needle passages and tissue handling of the tendon, and with this the risk of compromising tendon nutrition and damaging the tendon. Another important point with regards to the number of strands of the repair was highlighted by Trail et al., the tension across the various suture strands need to be equal in order to ensure balanced loading. Failure to obtain this will lead to a weakened repair and premature rupture (173).

Double stranded sutures have also been looked at as a way of increasing the number of strands of the repairs without an increase in needle passages. The most known is that of the Tsuge technique, or modified Tsuge. A four strand modified Tsuge was compared to 5 other four strand techniques by Angeles et al., and they found improved gliding, ease of placement and sufficient UTS to withstand early active rehabilitation with the modified Tsuge technique (90).

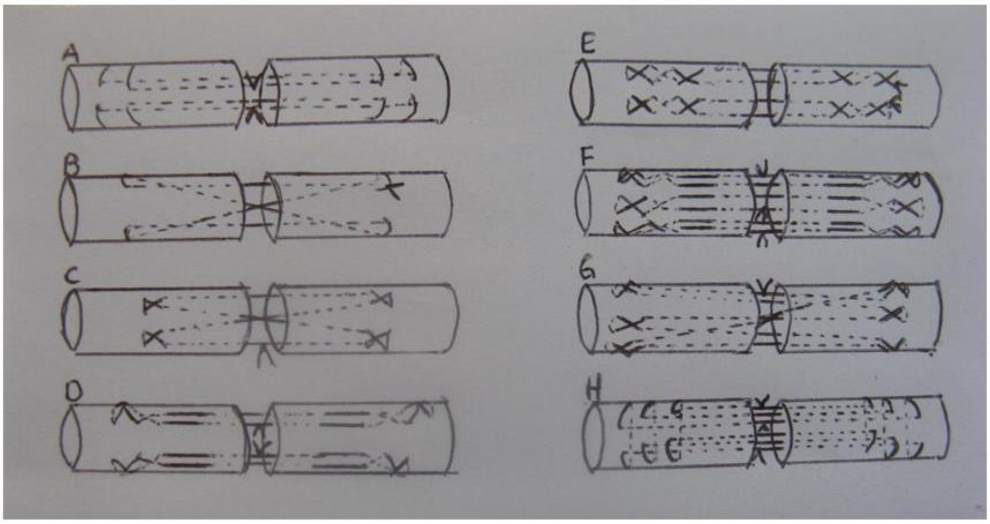

Figure 2.12. Core suture types using a multistrand suture method. A. Double modified locking Kessler (109). B. Cruciate, non-locking repair (104). C. Cruciate locked cross-stitch (98), D. 4-strand Savage (114), E. Modified Becker repair (MGH repair) (174), F. Modified Savage/Adelaide repair (157), G. 6-strand Savage (175) , H. Tripple modified Kessler (109).

2.5.2.2 Suture configuration: Locking versus Grasping

In addition to the strength of the core suture material and the number of strands used in the core repair, the configuration of the repair is also important in optimizing the tendon repair (109, 176). It has been modified over the years with respect to the bite purchase (103), and also with regards to locking versus grasping configurations (103, 143, 154). The first accurate description of the location of the longitudinal strands compared to the transverse strand in a modified Kessler technique (figure 2.13) was by Pennington (154). In his paper a locking suture repair is described as the placement of the transverse part of the suture on the volar aspect, or superficial to the longitudinal suture so that the suture grabs a hold of a bundle of the tendon fibres

(154), in contrast to a grasping configuration where the suture does not tighten around the tendon fibres and thus pulls out (108, 154).

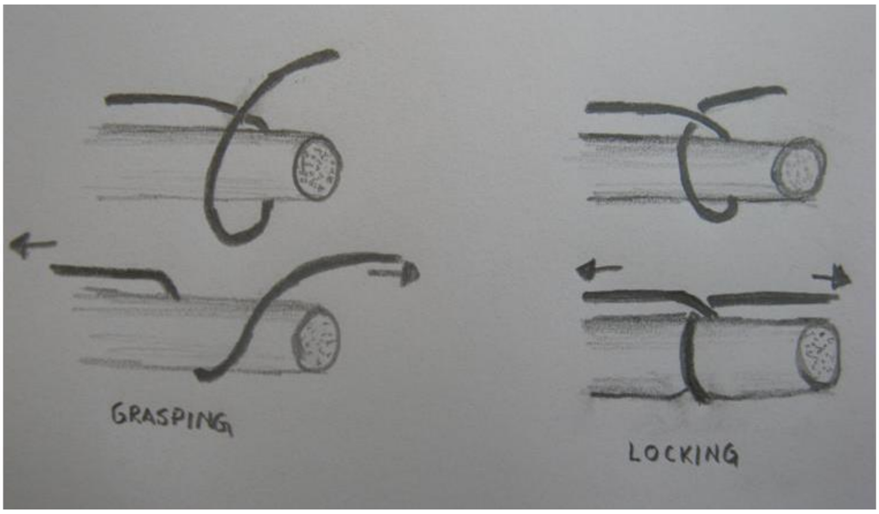

Figure 2.13: Illustration of the differences between a locking and a grasping suture configuration.

The biomechanical differences between a locking and grasping suture configuration, have been widely investigated (100, 101, 108, 117). A study looking at 1-locking, 2-locking and 2-grasping loop repair types found that the 2-locking loop configuration had a higher ultimate tensile strength and force to gap formation that the 2-grasping loops (108). This has however only been demonstrated using 3/0 or larger calibre suture material (97, 108). Using a canine model, Wada et al. examined the effect of locking versus grasping configuration on the 4-strand cruciate and 4-strand MK technique at time zero, and they found that the for both techniques the locking configuration resulted in the strongest repair. Of the two repair types it was the locking MK that was found to be the strongest, with a breaking strength of 45.9 ± 5.0 N, and had the greatest force to gap formation 24.6 ± 2.5 N, no difference was seen in the stiffness between the repairs (101). All the locking repairs failed by suture material breakage, while the grasping repairs pulled out through the tendon (101).

Hatanaka et al. used a similar model and investigated locking and grasping MK repairs *in vivo*, and found higher tensile strength and force to gap formation at days 0, 3 and 21 from the operation when using the locking configuration (161).

Two types of locking configurations have been described, the circle locking and the cross-locking (82). The locking modified Kessler (154), the Tsuge repair (177), and circle-loop models (82), are examples of the circle locking configuration. The cross locking repairs include the Becker and augmented Becker repair (174, 178), Savage (157) and modified Savage (92) and the locked cruciate cross-stitch repair (98, 117, 118). A further classification may be done considering the location of the cross-stitch on the tendon, exposed or embedded, both of which are seen with the Savage technique (figure 2.12). Xie and Tang evaluated the embedded and exposed cross-stitch with the circle locking repairs using a porcine model and found no significant difference in strength or force to initial and 2mm gap formation (82).

A further alteration in the repair configuration includes change to the cross-sectional area of the tendon that is incorporated into the repair. A study by Hatanaka and Manske looked at how an increase in the cross-sectional area of the tendon incorporated in the repair related to the strength of the repair (Figure 2.14) (103). In their study however, they only used this suture configuration in one tendon end (103). Hatanaka and Manske increased the area of the tendon incorporated into the modified Pennington repair from 5-15%, and found an increase in the ultimate tensile strength. A further increase to 25% of the tendon, without increasing the cross-sectional area (overlapping loops), had no significant impact on the tensile strength, but was found to increase the tendency for gap formation to occur (103). An investigation of increasing the cross-sectional area included in a 4-strand cruciate repair from 10%, 25%, 33% and 50%, demonstrated that the 25% area locking loops provided the optimal ultimate tensile strength, force to 2mm gap formation and stiffness (84). With regards to the size of the locking circles, Xie et al. found that 2 and 3 mm circles had a significantly higher ultimate tensile strength and force to gap formation compared to 1mm circles, in 2- and 4-strand repairs (81).

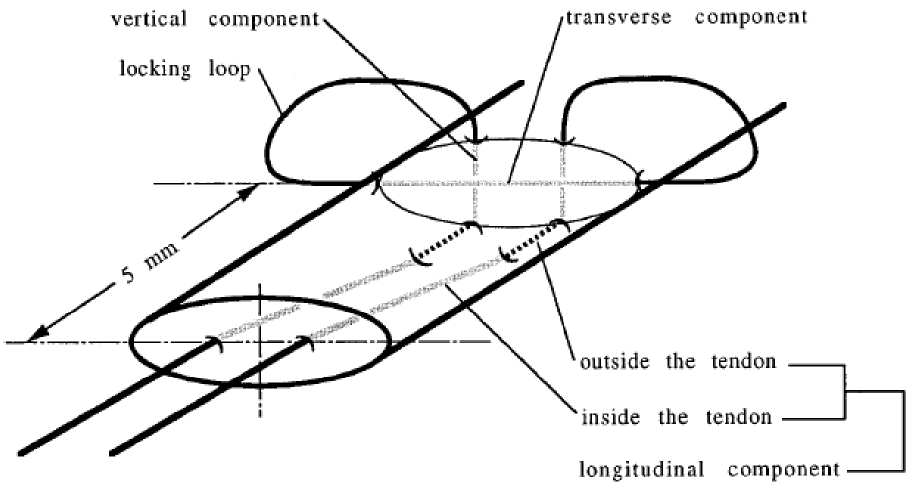

Figure 2.14. Schematic from Hatanaka and Manske's article showing the location and configuration of the loops used for their Tajima repair (Copyright Journal of Hand Surgery (Am) permission for reproduction obtained).

The orientation of the locking loops on the tendon also influences the strength of the repair as seen in Tan and Tangs study (85). By placing the locking loops perpendicular to the long axis of the tendon a greater ultimate tensile strength and force to 2 mm gap formation was found compared to placing the loops parallel to the tendons long axis (85). Presumably the effectiveness of the locking loop is dependent of the compression between the collagen fibrils and the suture. The parallel orientation of collagen within the tendon would mean that the strength of the parallel loop placement would be dependent primarily on cross-linking between the collagen bundles rather then bundles themselves.

Lotz et al. hypothesised that the imbalance of stiffness between the core and peripheral suture leads to overload and breakage of the peripheral suture, and therefore the UTS of the core suture therefore is not so important (106). They suggest modifying the strength and stiffness of the core and peripheral suture to the point where they reach ultimate tensile strength at the same time, by for instance increasing the stiffness of the core suture. Another suggestion would be to alter strain at the yield point of the peripheral suture, so that it is similar to that of the core suture. This was investigated by Merrel et al., where the purchase of the peripheral suture was increased, resulting in higher ultimate tensile strength and force to gap formation (88). Mishra et al. also looked at the value of maximising load sharing between the core and peripheral suture. They evaluated the effect of the more elastic nylon versus coated braided polyester using a modified Kessler repair and a continuous running peripheral suture of polypropolene. The nylon repair resulted in an increase in gap formation and earlier failure point (122).

2.5.2.3 Suture Materials

A strong mechanical connection that does not compromise the blood supply of the tendons is sought after when performing a tendon repair. The material used needs to be minimally or non-reactive to the tissues, inextensible to prevent gap formation and easy to use with good knot holding properties. The material also needs to sustain its tensile properties until the repaired tendon has gained enough strength (173).

Numerous studies using various materials have been reported. Stainless steel was considered a good material for repair providing a high tensile strength and low tissue reactivity, but difficulties with its handling reduces its value for tendon repairs (135). Other materials like Nitinol (NiTi) a shape memory alloy, has been evaluated as an option for flexor tendon repair in vitro. It was found to have a high strength (47 N) and stiffness (9.1 ±1.0 N/mm), and compared to braided polyester (Ethibond), and is reported to have easy handling (179, 180). Other newer suture materials like multifilament stainless steel suture (MFSS) and Fiberwire© was investigated by McDonald et al. (181). They found that a knotted suture of the MFSS (3/0) had a high UTS of 121 ± 3.2 N, and a high stiffness of 47.1 ±6.9 N/mm, compared to a UTS of Fiberwire© (3/0) 53 ± 15.8 N and a stiffness of 8.7 ± 0.8N. Although these materials may offer biomechanical benefits, they have not accepted general use in tendon repair surgery due to issues with handling characteristics (173), knot properties (182) and lack of in vivo testing (181).

The use of bio-absorbable suture material has not been recommended clinically, as the half-life of these sutures often do not have enough tensile strength to sustain postoperative mobilisation, and it is suggested that there may be higher tissue reaction and adhesion formation using this material (183, 184). Absorbable sutures with longer half life like Polydioxanone (PDS©) and polyglycolide-trimethylene carbonate (Maxon©) have been investigated *in vitro* in a canine model by Wada et al., and compared to a non-absorbable braided polyester sutures (Ethibond) (185, 186). This study found that there was a reduction in strength in the absorbable suture in first 14 days, with UTS decreasing from 54.4 N ± 7.4 to 35.7 N ± 6.7 (Maxon) and from 43.0 N ± 6.2 to 38.7 N ± 4.2 (PDS). The same loss was shown with force to gap formation reducing from 44.7N ± 5.7 to 31.8 N ± 6.5 (Maxon) and from 37.4N ± 5.3 to 32.3 N ± 4.8 for PDS. When comparing the PDS to non-absorbable sutures (Ethibond) in the same model over a 42 day period, a loss in UTS was seen with PDS in first 14 days (48.1 N ± 4.3 to 40.1 N ± 3.1), but not with Ethibond, no reduction in the force to gap formation was observed with either suture (185). These values correspond with the force measurements found by Schuind et al. for the active mobilisation of a digit (134), but the exact duration and intensity of post operative mobilisation in a canine model is not quantified and cannot be directly correlated to a human model. There is also concern regarding the tissue reaction of absorbable suture materials, as they exhibit an inflammatory response that is higher that that seen with non-absorbable sutures (184, 185).

Commonly used suture materials are non-absorbable synthetic materials, like coated and uncoated braided polyester, nylon and polypropylene (Prolene) (Table 1) (79, 173, 187). Load to failure measurements done of unknotted suture materials (gauge 4/0) by Trail et al., found that Ticron was slightly weaker than Maxon and monofilament stainless steel, 19.16 N (SD 1.09), versus 23.95 N (SD 3.78) and 21.12 N (SD 2.15) (173). Further investigation by the same authors found that once the materials were knotted the load to failure value of Ticron fell to 10.32 N (SD 1.80), likely due to the silicon coating of the suture(173). Lawrence et al. looked at non-absorbable sutures (gauge 4/0) and tested the sutures biomechanical properties when employed in a 4 strand repair in a porcine model. In their study, braided polyester (Ethibond) was compared to Prolene, nylon, stainless steel and fiberwire. They found the braided polyester, using a 4 strand cruciate cross stitch repair had a UTS of 65.6 N (SD4.7), force to gap formation 52.3 N (SD4.8) and stiffness of 10.2 N/mm (SD1.4), higher than both the nylon and polypropylene but lower than the stainless steel and fiberwire (79).

Momose et al. investigated the friction coefficient of monofilament nylon versus braided polyester with silicon coating (Ticron) and uncoated braided polyester (Mersilene), in a tensile testing machine. They found that nylon had a statistically lower friction coefficient (0.143± 0.008) than Ticron (0.162±0.009) and Mersilene (0.170± 0.010)(188).

Vizesi et al. report that coated braided polyester suture (Ticron) had the highest stiffness when tested at room (7.70 N/mm (SD 0.97)) and body temperature (7.59 N/mm (SD 0.75)), followed by Prolene 3.70 N/mm (SD 0.54) and 2.76 N/mm (SD 0.21) (187). They also investigated the viscoelastic properties of the suture materials and found that Prolene exhibited the highest values of stress relaxation ratio, both at room and at body temperature, 1.89 (SD0.03) and 2.31 (SD0.14), significantly higher that those seen with Ticron, 1.31 (SD 0.03) and 1.29 (SD 0.02) (187). With regards to creep ratio Ticron again had lower values (1.18 (SD0.01) and 1.21 (SD 0.02)), than those of Prolene (1.73 (SD 0.03) and 1.95 (SD 0.09)). Ticron did not show any temperature effect with regards to creep ratio or stress relaxation ratio (187).

The biocompatibility of the non-absorbable synthetic sutures often used in flexor tendon repair has been good both in experimental and human tissue models (189-191). The reaction seen in the tissues with uncoated braided polyester likens that of inert stainless steel and nylon. The addition of Teflon© to the braided polyester resulted in an increased response of lymphocytes and histiocytes. The tissue reaction seen with prolene is close to that of nylon sutures, however there has been reports of fragmentation of the suture with and without bone/cartilage depositing around the suture material (191).

Coated braided polyester provides a strong, relatively inextensible and easy handling suture (79, 173, 187), with acceptable tissue reactivity (79, 122, 190, 191) and using 3-5 throws is recommended to prevent slippage of the knot (173).

2.5.2.4 Core suture purchase

The suture purchase refers to the amount/bite of tendon that is included in each the repair (135), and this determines the part of the tendon that is incorporated in the repair (192). The ideal core suture purchase for the different repair types has been investigated by several investigators (77, 80, 85, 193). The studies have looked at the biomechanical implications of 0.3 cm, 0.4 cm, 0.7 cm, 1.0 cm and 1.2 cm core suture purchase length on the force to gap formation and ultimate tensile strength, using 2 and 4 strand repairs, locking and grasping repairs in a porcine and human model (77, 80, 85, 193, 194). These studies have found that increasing the core purchase from 0.4 cm -1.0 cm result in a stronger repair, both when looking at UTS and force to 2 mm gap formation. Tan and Tang found that an increase in the core suture purchase beyond 1.0 cm, to 1.2 cm reduced the force to gap formation but their findings were not statistically significant (85). Tang et al. report that a purchase bite of the core suture smaller than 0.4 cm weakened the repairs (80), and that the UTS and force to gap formation of the repair remained constant from 0.7-1.0 cm. The 1.0 cm purchase length was suggested to increase the strength further but the findings were not statistically significant (80). Their recommendation is to use a core suture purchase in the range of 0.7-1.0 cm (80).

Kim et al. investigated the core suture purchase length in a porcine model, with a four-strand Kessler technique and 3/0 Ethilon© suture. They found a further increase in ultimate tensile strength employing a 1.33 cm purchase length (81 N) compared to a purchase length of 1.0 cm (66.7 N) (195), yet due to the impracticality of such a suture purchase length, the authors support Tang et al.'s suggestion of a purchase length between 0.7-1.0 cm (80, 195).

Lee et al. studied the effect of core tendon purchase on work of flexion in addition to force to gap formation and strength (194), in a human cadaver study. Their findings on force to gap formation and ultimate tensile strength correlate with previously published results, a core suture purchase of 1.0 cm resulted in the highest force to gap formation and tensile strength. With regards to WOF they found that a suture purchase of 1.0 cm had the smallest average percent increase (5.2% ± 9.9%), and the 3 mm purchase the largest (22.1% ± 9.3%) (194).

2.5.2.5 Suture calibre

Despite a wealth of literature investigating the biomechanics of various repair techniques, concise studies looking at the importance of suture calibre on the mechanics of flexor tendon repairs are sparse. Although a greater suture calibre has as expected been shown to increase the ultimate tensile strength of tendon repairs during static testing (97, 100, 196, 197), a corresponding increase in the yield or force to gap formation has not been reported. Taras et al. investigated 2 strand repair techniques including a grasping modified Kessler using different calibres of braided polyester (Ethibond), in a linear testing protocol in a human model (97). Their work showed an increase in the strength of all repair types with the increase in the suture calibre, least for the modified Kessler technique that showed an improvement in strength of 167% increasing the core suture from 5/0 to 2/0.

In a study investigating a 2/0 suture calibre and 2-strand MK (Pennington) repair, a greater force to gap formation was reported (100). The calibre 3/0 suture has been observed to fail by means of suture pullout and suture rupture (76, 97, 100, 103, 118, 198), and the 4/0 calibre mainly by suture rupture (117, 118, 121, 198). In addition the strength of the 4/0 has been reported as having a holding capacity inferior to both grasping and locking configuration repairs of 2-,4- and 6-strand repairs (81, 97, 117, 118, 121).

Alavanja et al. looked at various suture calibres of braided polyester, 2/0, 3/0, 4/0, in a four-strand cruciate cross-stitch with a locking configuration. They found no increase in the force to gap formation with the larger suture calibre, using cyclic and static tensile testing, but found a significant increase in the change of WOF of a 2/0 gauge

(0.51± 0.27) versus a 4/0 gauge (0.31 ± 0.11) calibre suture, but no difference in the change of WOF between a 3/0 and 4/0 calibre (197).

A suture calibre of 3/0 or 4/0 is often used clinically, depending on the size of the tendon to be repaired. The above studies validate the use of either calibre.

2.5.2.6 Placement of core suture, volar versus dorsal

The dorsally based blood supply inside the tendon has limited the placement of sutures in this area. However, investigations into placing the sutures dorsally has also been recommended (199). Comparison of biomechanical properties with dorsal versus volar suture placement has not been conclusive. Komanduri et al. first investigated dorsal versus volar placement of core sutures, using both a linear and a curvilinear testing protocol (200). They reported a statistically significant increase in the tensile strength of the repairs with sutures placed dorsally. Soejima et al. found that a MK repair with the sutures placed dorsally (35.48 N ± 1.965) gave higher ultimate tensile strength than with volar placement (28.04 N ± 1.89), a 26.5% increase in UTS (112). In a further study they looked at the biomechanical, biochemical and histological properties of the dorsal versus volar aspect of the tendon, and found distinct differences between the two. They report a lower amount of collagen cross-linking, and a greater cross-sectional area of each single bundle on the dorsal side compared to the volar side, in addition to greater strength (52). They conclude their work with the recommendation that core sutures should be placed dorsally (52, 112). Stein et al. however, found no difference in any of their repair techniques when they compared volar and dorsal placement of sutures. No significant difference was seen with Kessler repair were the UTS with volar placement of sutures was 27.85 ± 1.11 N versus 28.85

± 1.56 N with dorsal placement, there was no significant difference found in the force to gap formation either (107). They suggest that the differences seen by Soejima et al. may be due to the intrinsic differences in the dorsal aspect of the tendon, where the longitudinal collagen fibers are enclosed by a thin epitenon, and they propose that "this gives a fascial or Chinese finger trap effect to the longitudinal fibers" (107).

Aoki et al. investigated the influence of the placement of sutures on the WOF of three repair techniques, using 4/0 and 5/0 braided polyester (Ethibond), and found that placing the sutures dorsally resulted in a lower gliding resistance than volar sutures (142).

In an attempt to avoid or minimise the trauma to the tendon, core sutures are usually placed on the volar aspect in the clinical setting (24, 201-204), preventing damage to the vascular supply. This is despite work that has shown that tendon healing can occur with synovial diffusion rather than vascular supply (59, 63, 69). Although studies indicate that there may be biomechanical benefits seen with placement of core sutures on the dorsal aspect of the tendon (52, 112, 142, 200), there is so far a lack of *in vivo* work to support a change in clinical practice, and with several of the repair types currently used dorsal placement will increase the technical difficulty of the repairs.

2.5.2.7 Placement of suture knots

Studies looking at the biomechanical properties of suture material have shown that the suture will rupture at the knot (157). How the suture knots of the various repair techniques are placed (Table 2.2) has been found to affect the strength of the repair in *ex vivo* studies using static tensile testing (111, 157). Placement of the knots on the outside of the repair/tendon and reducing the number of knots increased the strength of the tendon repair, when compared to repairs where the knots are located between the tendon ends (111) (Table 2.2). In an *in vivo* canine model the placement of knots between the tendon ends and on the tendon surface was evaluated (110). Pruitt et al. found that at week one, both techniques showed a decrease in the tensile strength by 85% of the initial strength measurement. At 3 weeks an increase from the initial values were seen with both techniques to 120%, while at 6 weeks the group with the knots placed between the tendon ends had increased their values further to 167% of the initial value, whilst that of the knot outside group remained at 120% of the initial value (110).

Placement of the knots on the tendon surface will also increase the gliding resistance of the repair significantly (166, 188, 205). Momose et al. investigated *ex vivo*, the effect of knot placement on gliding resistance in tendons and found that here was a maximum change in the gliding resistance of 0.78 ± 0.35 N when one knot was placed on the volar aspect of the tendon versus 0.60± 0.23 N with on laterally place knot and 1.06 ±0.44 N with 2 lateral placed knots (188).

Barbed suture materials have been introduced to surgical practice to overcome problems associated with knot reaction not just in regard to tendon repair, but more commonly in cosmetic procedures, where for instance barbed suspension sutures are used for face lifts (206). Early studies using barbed sutures for flexor tendon repair were reported in 1967 by McKenzie using a unidirectional barbed steel wire (207). More recently McClellan et al., investigated the use of a four strand bidirectional barbed suture repair (Figure 2.15) (no knots) in flexor tendon repair (208) using a porcine model. They investigated a suture configuration which had elements of the modified Kessler as well as the Savage repair, against the standard MK and Savage repairs. In their investigation all barbed sutures failed by pulling out of the tendon, the UTS was reported as 72.39± 15.16 N with a force to 2mm gap formation of 62.84±17.30 N. Their investigation also compared the cross sectional area of the repairs using calliper measurements, and found that to be significantly smaller than the MK and Savage repairs that were tested (208). The introduction of the barbs to the suture leads to a decrease in the tensile properties of the suture, and it is recommended that one uses a calibre size greater than for other sutures, so a 2/0 barbed suture instead of a 3/0 Ticron (209).

Figure 2.15. Schematic of a bidirectional barbed suture.

2.5.3 Tendon repair devices

While tendon repair using sutures remains the mainstay of surgical treatment, other methods have been investigated as a means to providing a stronger and easier repair of flexor tendon injuries and the active mobilisation following repairs. Silfverskiold and Andersson looked at using a 10-15 mm long, woven polyester material, Mersilene mesh sleeve, that was pulled over tendon and fastened with a 6/0 peripheral cross-stitch sutures at each end. They report an ultimate tensile strength of 103 N for this repair (202), with failure of the repair involving mesh tearing rather than peripheral suture disruption (202). Aoki et al. used a Dacron splints with a thickness of 0.22 mm, of various widths (2-6 mm), either internally or dorsally across the tendon ends (113). They reported good results for the strength of the tendon splint repairs with UTS measurement of 82.9 N for the internal splint, and 79.4 N for the dorsal splint, and higher force to gap formation measurements, 20.1 N for the internal splint and 30.9 N for the dorsal splint (113). No measure of the WOF for these devices has been reported.

Other attempts have been tried using metal splints and anchors. A stainless steel internal anchor was tested by Gordon et al. for strength, and compared to the Kessler, Becker and Savage techniques. The anchor measured 20 mm in length, 3 mm width and 1 mm in thickness. This device has two anchors on each side of the repair, and was fastened with 2 sutures at each tendon end at a 90 degree angle to the anchor's long axis. The ultimate tensile strength of this repair was significantly higher (75.2 N) than the other repair techniques (21.3-50.4 N), and they found that the stiffness of the steel anchor and the Savage technique was significantly higher than the other repairs (105). A modification of this concept, with two intra-tendinous anchors connected by a single multifilament 2/0 stainless steel suture (TenoFix©) has been investigated using an *ex vivo* canine model in static and cyclic tensile testing (83, 119). *In vivo* repairs using the same device in a canine model were looked at by Su et al. (2006), and they also evaluated the repair in a multi-centre, randomized and blinded clinical study (210). Biomechanical testing of the repair found that it had a higher force to 2 mm gap formation and stiffness when compared to a 4/0 suture 4-strand cruciate locking repair. It also exhibited higher peak force differential during measurement of gliding resistance (83). The inflammatory response of the TenoFix© in a canine model was minimal, and no interference was seen with tendon healing (211). In their clinical study, Su et al. found that the TenoFix© repair method was a safe method for tendons that had sufficient size and were the exposure was adequate. They report good or excellent results in 67% of the TenoFix© and 70% of the controls, and a rupture rate of 0% with TenoFix© and 18% with the MK control group (210). The device is however not recommended for placement in tendons close to the joints where there could be limitations on the flexion and perhaps pain on movement (105). They are also not recommended for small tendons, for instance in children, due to the diameter of the device, and should not be used in possibly dirty wounds or complex injuries (210), as the incorporation of metal in such a wound can increase the risk of infection.

2.5.4 Peripheral suture technique

The peripheral suture started out as a “tidying up” suture to aid the gliding of the repaired tendon within the sheath and under the pulley (212). It was then found that the peripheral suture also aid in the strength of the repair and in force to gap formation. Work by Wade et al. (1989) compared a simple, continuous epitendinous repair with simple interrupted loops and vertical mattress sutures (Halsted repair) and found that the Halsted repair increased the load at which gap formation occurred by 93% (164). The term epitendinous suture was changed to peripheral suture as newer repair techniques grasp the tendon substance as well as the epitenon.

Further improvements in the force to gap formation, stiffness and ultimate tensile strength, have been seen as new repair techniques have been introduced (88, 122, 164, 169, 202, 213-219). The repairs reported to be the strongest are the cross-stitch (202), embedded cross-stitch (219), interlocking cross-stitch and interlocking horizontal mattress (218), Horizontal mattress (164), Lin continuous locking (213), and the horizontal intra-fibre (220), however they are all also quite technically demanding which limits their use in the clinical setting.

The repair technique most used and evaluated is the simple, running (continuous) peripheral suture (Figure 2.16E) (212), this is a relatively simple repair method. The strength of this repair can be increased by deeper purchase of the tendon (216). A biomechanical study showed that by increasing the depth of the suture into half of the radius of the tendon, compared to that of the epitenon only, the ultimate tensile strength of the repair increased by 80%, and the stiffness by 90% (216). Lotz et al. found that the yield force of the repair also improved (106). An increase in the penetration of the peripheral suture from 1mm to 3mm (Figure 2.16F) has also had a positive effect on the ultimate tensile strength and force to gap formation in a repair using modified Kessler and a continuous peripheral repair (88). Kubota et al. also found that when the number of passes of the suture becomes higher, this results in increased force to gap formation and strength, in simple, continuous repairs and other repair types (169).

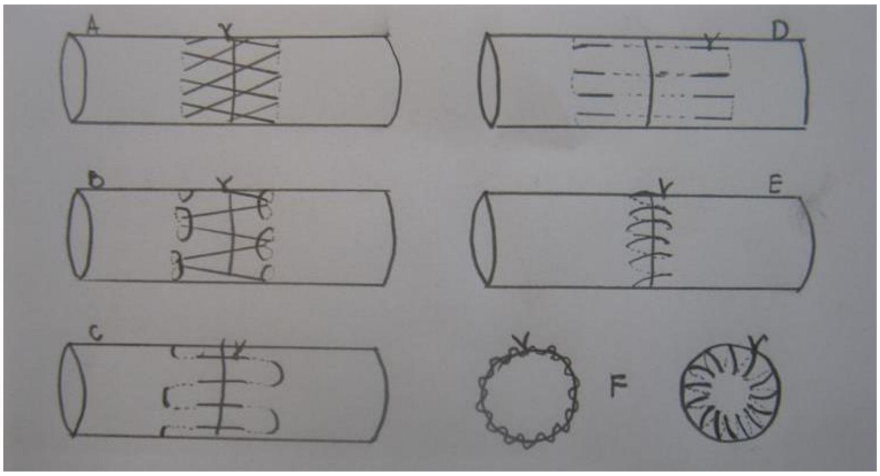

Figure 2.16. Schematics of various peripheral suture techniques. A. Silfverskiold or cross-stitch repair (202), B. Lin repair(213), C. Halsted repair (164), D. Horizontal intrafiber repair (220), E. Simple, continuous running (9), F. Superficial and deep simple, continuous running (216).

Moriya et al. looked at the biomechanical properties of five different peripheral tendon repairs and found no difference between the groups with regards to force to 2 mm gap formation, maximum strength or stiffness (205). The repair using a simple running repair with the knot on the outside of the tendon had significantly higher gliding resistance than the repair with the knot between the tendon ends. This correlates with the results reported by Momose et al. A correlation between the number of throws of suture material and gliding resistance was not found by Moriya et al. (205).

The application of a microfabricated scaffold to the surface of the tendon to improve the healing properties was reported in a rat model, leading to successful tendon repair (221), this was used instead of the peripheral suture. The authors used a biodegradable polydiaxone (PDS) sheath with microgrooves to guide cell migration after tendon laceration or crush injury. The repaired tendons and control groups were harvested and evaluated at 3, 7, 14, 28, 42 and 73 days. The collagen was found to reform and realign along the grooves running longitudinally. The authors suggested that the PDS sheet ensures that there is no adhesion formation between the tendon and the flexor sheath (221). In a recent study, Kaphoor et al., investigated tenocyte migration on micropatterned surfaces with grooves ranging from 50μm to 250 μm, and found that the 50μm grooved surfaces had better and denser orientation of tenocytes (222). More recent studies have described tubular constructs of type I collagen (223), and the production of 2D and 3D electrospun scaffolds for tendon repairs which may be used in a clinical model. Biomechanical information on tensile strength and work of flexion on the application of such conduits in an animal or human model is not reported.

Nanogrooved sheets of Polycaprolactone is another material that has been tested for neurite orientation where constructs with nano-grooves have shown increase in neurite growth and orientation (224). This material is detailed in the following section.

2.5.4.1 Polycaprolactone

Chemical properties

The polymer, ε-polycaprolactone (PCL) is an aliphatic type of polyester consisting of hexanoate repeat units, Figure 2.17 (147). The thermal, physical and mechanical properties of PCL are determined by the compounds molecular weight and degree of crystallinity (225).

In its semi crystalline form it has a glass transition temperature of around −60°C and a low melting point of around 60°C (147). PCL is prepared by ring opening polymerization of ε-caprolactone using a catalyst such as stannous octanoate.

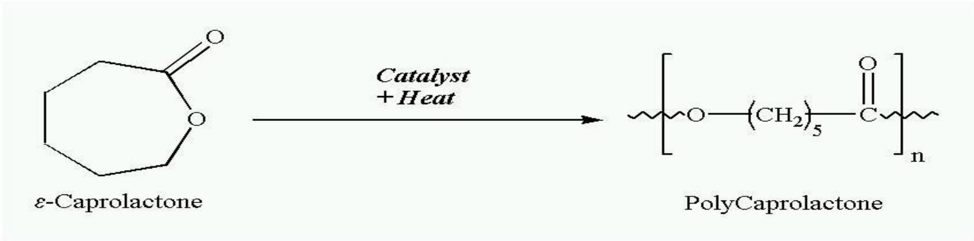

Figure 2.17: The chemical structure of PolyCaprolactone

PCL is soluble in a number of compounds and has the property of being miscible with numerous other polymers like polycarbonates, poly-vinyl chloride and nitrocellulose to name a few (225).

Biodegradation

PCL is a slowly biodegradable polyester, with a biodegradation that spans from several months to several years (226). The degradation is dependent on the molecular weight, the conditions of biodegradation and the level of crystallinity. Two phases of biodegradation are recognised. Phase one, or the amorphous phase, is recognised by a reduction of molecular weight and increase in solubility, without compromise in the structure or loss of mass (226). Once the molecular mass falls below 5000, due to the cleavage of the ester bonds, this represents the start of phase two. Pitt et al. found that at this stage the compound started to break apart and a loss of mass occurred. They further predicted that this is the stage where the compound will slowly absorb and excrete the polymer (226). Studies of implantable devices of PCL, micropellets and rods has not found a difference in the rate of degradation between the different shapes of PCL (227), however these studies did not evaluate the tissue engineering scaffolds. No clear evidence has been found of an enzymatic component of the phase one degradation *in vivo*, however in the environment degradation of PCL has an enzymatic component (226).

Biocompatibility

Polycaprolactone is a readily used polymer and has been tested both *in vitro* and *in vivo*, and has been used clinically for about 30 years (226). As far back as 1970's PCL was tested as a wound dressing (228). It has been FDA approved for implantable and drug delivery devices, due to its complete biodegradation and biocompatibility.

2.5.5 Gliding properties of Tendon Repair

Several authors have investigated the friction force or gliding properties of repaired tendons and also the excursion, in a range of models involving human (114, 144, 229, 230) and animal (149) (226) tendons (See section 2.4.4). At smaller angles of flexion the friction is less than at higher arc of contact (73). This means that the friction of a flexed digit will be higher depending on the degree of flexion.

When a tendon is injured and subsequently repaired, any modification or repair to the tendon will increase the gliding resistance, or WOF. The gliding resistance of a repair is influenced by the number of sutures used in the repair, the amount of suture and knots exposed on the surface of the repair, including the configuration of the suture technique, the size of the sutures (197) and the suture material used (226). Zhao et al report a internal work of flexion of 0.017Nm for an intact human tendon, and 0.065 Nm for their Modified Kessler repair (144).

Research into the effect of the number of strands of the core suture crossing the repair on gliding resistance is equivocal. While some studies have suggested that increasing the number of core sutures (from 2-4) results in an increase in gliding resistance (229), others have reported no effects of strand number on gliding resistance (226), despite a measured increase in the cross-sectional area of the repair with increased number of suture strands (230). Winters et al. investigated a 8 strand MK repair using double sutures in an *in vivo* canine model and although gliding resistance was not measured, they report reduced motion for the 8 strand repair compared to 2 and 4 strand repairs, suggesting that the increased amount of suture may affect movement of the repair postoperatively (231). While comparisons between the studies is difficult due to differences in testing protocols, repair techniques and specimens, it has been suggested that it is the amount of exposed suture material on the surface of the tendon that has the greatest influence on gliding resistance or WOF (226). This will be determined by the type of repair used, knot placement and peripheral suture technique.

Repair techniques like the modified Becker and Savage repair have high tensile strength but also high gliding resistance (232), and they have more suture material on the tendon surface (Figure10). In their study Momose et al., found that the Becker repair had an increase in mean gliding resistance of 0.58 N ± 0.13, compared to mean gliding resistance of 0.27 N± 0.05 of MK (2strand). Increasing the core sutures of the modified Becker repair (Figure 2.12) to four or six strand and having two or more exposed knots and more exposed suture material on the tendon surface increased the gliding resistance most of the techniques tested (149, 166, 233). This resulted in a gliding resistance of 0.58 N±1.52 .

The influence of the placement of sutures on the volar versus dorsal aspect of the tendon has been investigated for work of flexion by Aoki et al. They found that a tendon splint from woven Dacron placed dorsal increased the gliding resistance less (20.5% ± 18.3%) than a 4 strand Savage repair with the sutures placed volar (37% ± 21.7%) (142).

The placement of the suture knots has been investigated in an intact tendon where one volar knot was compared to one or two laterally placed knots (188). Momose et al. found that the placement of two knots on the lateral surface of the tendon resulted in a higher mean change in gliding resistance (0.63 N ± 0.25), while one knot placed laterally had the smallest change in mean gliding resistance (0.35 N ± 0.14) (188).

The different repair types result in different amount of suture material exposed on the tendon surface. Noguchi et al. looked at the angular rotation of the digits after tendon repair and found that placement of the suture knots externally reduced the angular rotation of the digits compared to when the suture knot was placed between the tendon ends (114).

Peripheral suture techniques influences the gliding resistance of the repair in two ways; by placing suture material on the tendon surface, and by preventing formation of gap formation which can cause the moving tendon to catch on the pulley and increase the peak gliding resistance. A balance must therefore be found. Moriya et al. found that all peripheral suture repairs increased the mean gliding resistance of the repair and that the simple, running suture (knot inside) had a change in mean gliding resistance of 0.65 N after 1000 cycles, and with the knot outside a change in mean gliding resistance of 0.84 N after 1000 cycles (205). In a canine model Zhao et al. found that the simple, running peripheral suture resulted in the smallest increase in mean gliding resistance (0.33 N) when compared to measurements of an intact tendon (0.08 N) (149). Gap formation at the repair site has been found to increase the peak gliding resistance. Measurements of one, two and three millimetre gap formation have found that the repair is likely to get caught on the free edge of the pulley and cause triggering of the repair (234). It is not know how big a gap will allow for triggering of the tendon and thus interrupt tendon healing, but it in the clinical setting it is not believed that a gap of 10 mm or more will be tolerated (86), in fact 3 mm gap formation or greater is believed to represent high risk of rupture for tendon repair (235). Some authors have reported that a gap formation of up to 10 mm will not affect the tendon healing (226).

A repair combination of the two-strand modified Kessler technique with one knot between the tendon ends, and a simple running peripheral suture seems to have the lowest gliding resistance (226). Of the other repair techniques that have more suture material exposed on the surface of the tendons the mean gliding resistance is increased, modified Kessler with two exposed knots (0.8-1.07 N), Tsuge repair with one exposed knot (1.12 N) (226).

No considerable difference in gliding resistance has been found when comparing a locking versus a grasping loop repair. Tanaka et al. investigated the four strand modified Kessler (grasping) with a mean gliding resistance of 0.79 N, and the Pennington modification of the modified Kessler (locking) 0.89 N using human cadaver tendons (86).

The gliding of the repaired tendon will not only be affected by the repair technique but may also be affected by the tissue swelling that occurs to the tendon, sheath and/or surrounding tissue as a result of injury (226). Lane found that if a tendon is immobilised straight away there is a ten fold increase in the force needed to obtain full flexion of the digit, and a 90-fold increase in the total work of flexion an hour after injury. The peak resistance appears to be between day four to seven after injury, and then it gradually decreases (226).

2.6 Tendon healing

2.6.1 General Healing Process

The process of tendon healing was for a long time considered to be solely through in-growth of external granulation tissue or adhesions, without input from the tendons themselves (236). However, there is considerable evidence, albeit in animal models, that tendons can activate and partake in the healing process without external adhesion formation (226). Moreover, Mass and Tuel showed in 1989 that human flexor tendons can themselves stimulate an intrinsic healing process (237). Clinically both intrinsic and extrinsic healing mechanisms are recognized to occur at the tendon repair site. Their input seems to vary in contribution, and this is related to the injury itself, repairs and post operative mobilization regime (226).

Flexor tendons heal in accordance with other healing tissue and the process is usually divided into three main stages (226).

<u>1. The inflammatory stage</u>

This is activated by tissue trauma, and has a duration of two to three days. The injury to vessels in the flexor sheath and tendon activates the coagulation cascade. A clot is then formed around the injured tendon or tendon ends. There is an increase in surface cells migration into the area of trauma within 24 hours of injury. A thickening is seen of the epitenon cell layer lying adjacent to the injury (238), with an increase in fibronectin levels, which continues into the proliferative stage. Banes et al. found that fibronectin was closely related to surface cells of the tendons- or epitenon cells, and proposed that it has a role in binding of hyaluronic acid in healthy tendons to improve lubrication (239). Fibronectin is also thought to initiate cellular chemotaxis and migration (226). Damaged cells and an influx of platelets to the area results in the release of growth factors, including platelet derived growth factors (PDGF), transforming growth factor-β (TGF- β) and epidermal growth factors (EGF) as well as chemical mediators of inflammation. PDGF is believed to mediate the phenotypic transformation of fibroblasts (240) and EGF has been shown to influence the migration of fibroblasts and increase the production of collagen in a healing wound (241). These mediators also signals migration of neutrophils and macrophages to the area. The main role of these are to remove injured and necrotic tissue and to further release growth factors that initiate the second stage; the proliferative stage (226).

2. The proliferative stage

This is a longer stage that can last for an estimated four weeks. In this stage there is a further influx of fibroblasts and fibroblast proliferation, with the making of collagen and other extracellular proteins and further macrophage invasion. Neovascularisation also occurs during this stage, usually by day 21 (226). In the intrinsic part of the healing process a large cell proliferation is seen in the epitenon fibroblasts. This usually occurs in the three to seven day period from tissue injury (226), and results in a continuous layer of epitenon covering the tendon injury site (226). Apoptosis at the injury site results in a preliminary reduction in the cellularity of the endotenon and of the tendon core cell (226). Type III procollagen levels are increased in the epitenon cells by third day after trauma, but the type I collagen is decreased at this time and gradually increases to normal levels by week four (242). This relative increase in the ratio of type III to I collagen in the early stages of tendon healing is considered a part of normal tissue healing. When pathological solid adhesions are formed a reduction in the amount of type III collagen is seen (226). In this stage the injury becomes scar like.

The process of revascularisation of the tendon begins at day 3, when in-growth of vessels from the epitenon to the area of trauma is seen. It continues and reaches it peak at about day 21 post injury (238, 243, 244).

3. The remodelling stage

This represents a period from week four to anywhere between weeks 12-20. In this stage a reduction is seen in the cellular activity, while maturation and alignment of collagen takes precedence (226). The relative amount of type III and I collagen stabilize within the 12-24 weeks after repair in a standard tendon healing process. Once the remodelling stage is completed, only small differences between normal tendon and repaired tendon have been found histologically (242).

2.6.2 Adhesion formation

Adhesions that form between the repaired tendon and the tendon sheath cause difficulties in the healing and mobilisation of flexor tendons. It increases the work of flexion of a repaired tendon and reduces the function of the digit involved (155) . Factors influencing adhesion formation have been widely investigated and a comprehensive review is beyond the scope of the current thesis. Nonetheless, there is some evidence to suggest that tendon injury, in and of itself, is not sufficient to cause adhesion formation. Previous research has shown that the free ends of transected tendons become smooth and even and do not form adhesions inside the flexor sheath, unless the tendon ends are sutured or immobilised (115, 245). In a rabbit model with partial FDP lacerations, Matthews and Richards (115) demonstrated that the suture repair created the most side-effects, in form of decreased cell viability and a larger zone of regeneration, when compared to sheath excision and post-operative immobilisation. When comparing two of the factors the highest rate of adhesion formation was seen with a sutured and immobilised repair. If all three factors where present the result was a very solid and unrelenting tendon adhesion (115).

Increasing the number of core sutures have been postulated to affect the formation of adhesions by increased handling of the tendon, increasing bulk at the site of the repair, external suture material and possibly ischemia at the repair site (155, 246). A number of various flexor tendon repair techniques have been tested to evaluate the trauma load to the tendon from the repair itself and the resulting effects on tendon healing and formation of adhesions. *In vivo* tests in a canine model compared the modified Kessler and Tsuge techniques, with immobilisation post-operatively (247). No differences were found in the histological examination or intravascular dye injection results (247).

Strick et al. looked at two-and four strand repairs (MK) with and without epitendinous sutures in a chicken flexor tendon repair model, where the repair was immobilised for four weeks (226) . They found a marked increase in adhesion formation with both the two- and four- strand repair, and no difference was found in the adhesion breaking strength or the histological investigation. They also found no influence from the epitendinous suture on adhesion formation. A comparison was done between the two-strand MK and four-strand modified Becker repair in a canine model, with a passive mobilization regime, and the 4-strand repair had a much higher adhesion strength at both three and six weeks (248). The modified Becker repair (Figure 2.11E) includes a larger amount of suture material exposed on the tendon surface, was seen to increase the gliding resistance, decrease the tendon excursion throughout rehabilitation and lead to an increase in adhesion formation (248).

The influence of gap formation on the development of adhesions was reported as early as 1940`s by Mason and Allan. In 1960 it was confirmed that gap formation in combination with immobilisation of the tendon repair resulted in a not so favourable result (249). With regards to what size gap formation will result in failure there are conflicting views. It has been reported that gap formations as large as 8.5-10 mm and with adequate mobilisation may not infer a functional loss in form of adhesion formation (226) . It is however generally accepted that gap formation between the ends of the repaired tendon increase the gliding resistance, and may through that result in reduced gliding and build up in adhesions during the post operative mobilisation (234). A gap between the tendon ends as small as 2 mm has been shown to increase the peak gliding resistance by 100% compared to that of intact tendon (234). A further increase in gap formation till 3 mm or more leads to triggering of the tendon on the pulley edge, with a resulting massive increase in the peak resistance. In an in vivo canine model with passive mobilisation it was found that a gap formation of 3 mm or more obstructed the necessary strengthening of the repair, resulting in higher risk of further gap formation and failure of the repair (235). In the clinical setting with controlled mobilisation, a direct association was found between gap formation and the formation of adhesions, and tenolysis was performed in all cases where the gap formation was 4 mm (250).

2.7 Summary

The MK technique is a well known and recognised repair with good clinical results (152) , yet rupture rates are quoted as high as 17% (11). Despite good surgical technique and high tensile strength repairs, complications arise. Hand surgeons are therefore still searching for repair techniques that can eliminate or reduce complications further, following tendon repair. Discussions are still ongoing with regards to what a suitable tensile strength is for a flexor tendon repair and a number of Hand Surgery units recommend a 4 strand repair over a 2 strand repair. This is mainly as a result of change to an active mobilisation regime for post operative mobilisation in some hand surgery units. The American Society for Surgery of the Hand changed their recommendation in 2007, and advised that tendon core sutures should include at least 4 strands (3) to ensure adequate tensile strength. This increase in core suture strands is thought to increase the bulk of the repair (section 2.4.4), however the effect will depend on the configuration of the repair and whether it is locking or grasping (section 2.5.2) (188, 232). Suture calibre is also a factor that may affect the size or bulk of the tendon repair within the flexor sheath (section 2.5.2.5). Increased calibre size is reported to increase the ultimate tensile strength of repairs (97), but a balance must be struck as it is recognised that larger calibres of suture can increase the bulk of the repair (197). The use of a 3/0 or 4/0 calibre suture is mostly used clinically and is validated as a good size from the studies presented in the literature (section 2.5.2.5). Recommendation as to what suture material to use is outlined in section 2.5.2.3, which shows that braided polyester (Ethibond or Ticron) provides sufficient strength, suitable viscoelastic properties and handling properties for flexor tendon repair.

With regards to the peripheral repair numerous techniques are presented in the literature. The primary function of the peripheral suture is to ensure that the tendon ends are smooth and will not catch or snag as the tendon moves through the flexor tendon sheath (212) (reviewed in section 2.5.4). The peripheral suture consisting of multiple suture loops that fail by pulling through the tendon (section 2.5.4) prior to the core suture failing presumably as a result of a large amount of force being transferred to the peripheral suture. Gap formation in the tendon repair is likely to increase the risk of rupture and complications though it is clear from section 2.5.4 that there is no agreement as to what size of gap is clinically relevant. A tissue engineered sheet could potentially be introduced to the repair configurations as a potential matrix for tendon migration over a potential gap following tendon mobilisation. The biomechanical properties of such a repair need to be investigated to determine how it would compare to other peripheral repairs.

Zone II of the human hand consists of 2 tendons, a deep and superficial, in close relationship to each other and the fibro-osseous canal, or flexor sheath that surrounds them and the pulleys that ensure that the flexor tendons are kept in close proximity to the underlying bones and joints (Section 2.2). In addition to this the neurovascular bundles run in close proximity to the flexor sheath on either side, and the majority of the intratendinous blood supply located on the dorsal aspect of the tendon. Any trauma to the deep flexor tendons in this area is likely to impact on the other structures, making it technically demanding surgery. The placement of sutures is commonly on the volar aspect of the tendon due to the intratendinous blood supply and due to the complexity of some of the repair types making placement on the dorsal aspect very difficult (section 2.5.2.6). This means that the surgeon must be familiar with the complex anatomy of the area, atraumatic tissue handling and proper repair techniques to ensure best possible outcome. Patient compliance post operatively with the post operative rehabilitation will also influence the final result of the surgery.

Biomechanical testing of flexor tendon repairs are performed using a range of models including rat, canine, porcine and human, and a range of testing machines and set ups (table 2.1). The most commonly reported biomechanical properties reported for flexor tendon repairs in the literature are ultimate tensile strength, yield point, 1-3mm gap formation and in some more recent reports, stiffness (section 2.4).The majority of studies available in the literature use a static linear testing set up which allows for testing of a large number of samples over short time period (Table 2.1).

Total work of flexion is also reported as a measure of all the energy required to overcome movement of the finger in flexion, and is often used as a measure of the gliding properties of a repair. Various methods are described in the literature, with the first measurement described in 1976 by Lane (138). Zhao et al, have developed a method were WOF is measured at the same time as the resistance between the tendon and the flexor sheath providing researchers with a tool to identify the involvement of the various resisting forces (144).

Chapter 3 Early Experimental Work

A series of pilot tests were performed early on in the project, developed to home in on the research questions of the thesis. Although not directly addressing the research aim, these experiments are briefly reported here.

3.1 Barbed Sutures

As the discussed in the literature review, section 2.5.2.7, the suture knots affects the work of flexion of repairs as well as the strength of the repairs. The use of bidirectional barbed sutures where investigated early in the PhD project by using 2/0 Quill sutures (Angiotech, Vancouver, BC, Canada). The theoretical advantage of such repairs is that as tension occurs across the repair site the barbs will anchor onto the tendon fibers and prevent failure. 5 tendons were repaired with the 2/0 Quill sutures, using a MK repair. A 2/0 suture calibre was used as this corresponds to a 3/0 braided polyester suture, the introduction of the barbs causes a reduction in the tensile properties (209). The samples were tested for WOF and UTS in the Instron tensile testing machine using same testing protocol as for 2 strand repairs (Section 5.5.2 and 5.5.3).The sutures were also tested on their own in the Instron by clamping them 2cm proximal and distal to the centre of the suture (Figure 3.1), to see what the failure strength was.

Figure 3.1. Schematic of a bidirectional barbed suture.

The results of the study showed that the Barbed sutures pulled out of the tendon (all samples) and by doing so caused disruption to the tendon. When testing the suture alone in the Instron machine it failed in the center of the suture where the direction of the barbs changed (area in red on figure 3.1).

Values for UTS of the sutures alone were comparable to those published by the company (Angiotech), 26.3N (251), and for the tendon repair results revealed UTS of 25-30N (original data lost due to corrupt hard drive).

Table 3.1. WOF and Peak force differential for barbed suture testing

	WOF1	WOF2	WOF3	Force1	Force2	Force3
Mean Intact	0.005	0.008	0.007	0.754	0.584	0.585
sd Intact	0.008	0.005	0.005	0.897	0.469	0.440
Mean Barbed MK (n=5)	0.044	0.021	0.018	4.233	1.563	1.201
sd Barbed MK	0.020	0.005	0.005	2.278	0.664	0.506

The damage caused by the barbed sutures when they pulled out of the tendons means that any secondary repair of such a failed tendon would be very difficult, and it would likely result in further tendon repair requiring grafting in order to re-establish function. This would mean more complex surgery in the form of reconstruction and increasing time of work for any patient involved. In light of these findings the project was cancelled.

3.2 Tissue glue

Another option for the peripheral repair was investigated using tissue glue, Tisseel (Baxter AG, Vienna, Austria) over the tendon ends after the core suture was placed. Tisseel is a 2 component fibrin sealant which contains human fibrinogen and a synthetic fibrinollysis inhibitor. This is to prevent premature breakdown of the formed fibrin clot. Clinically it is used as an adjunct in surgery, to ensure hemostasis (252) , as tissue adhesive for wound closure (253) , scaffold for tissue engineering (254) and as an alternative and/or adjunct to sutures in nerve repair (255).

Three tendons were dissected and repaired as described in section 5.3, with a 2 strand LMK core suture repair, or 3/0 Ticron. Tisseel was applied to the tendon ends instead of a simple, continuous peripheral repair. To prevent spillage of the Tisseel into the flexor sheath the tendon repair site was surrounded by sterile latex free glove which ensured that the Tisseel only contacted the surface of the tendon (Figure 3.2).

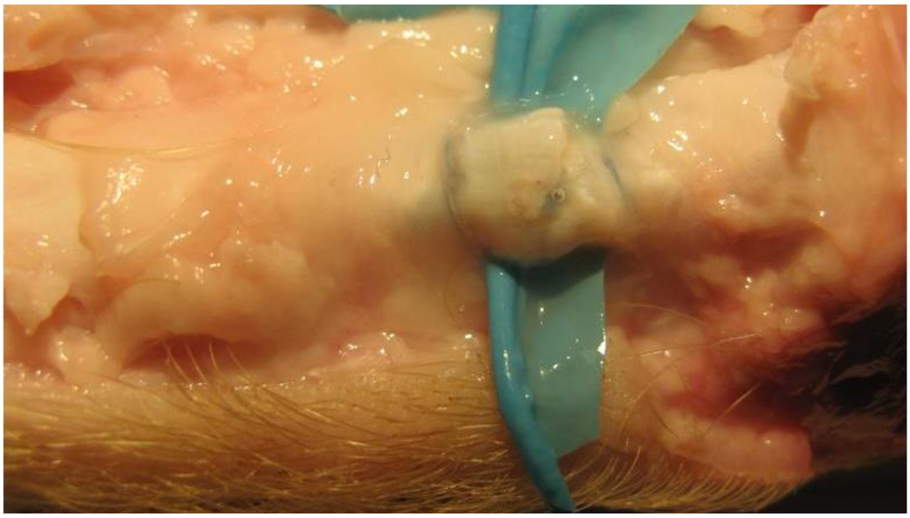

Figure 3.2. Tisseel applied to the tendon repair, ensuring a smooth surface

After 3 minutes the glove was removed from the tendon repair site, and these were tested for WOF in as described in section 4.5.3, with 3 cycles. This resulted in the Tisseel being disrupted from the tendon repair site as the repair passed under the A2 pulley, leaving a irregular surface (Figure 3.3). Further testing of the tensile properties of the tendon repairs was not pursued.

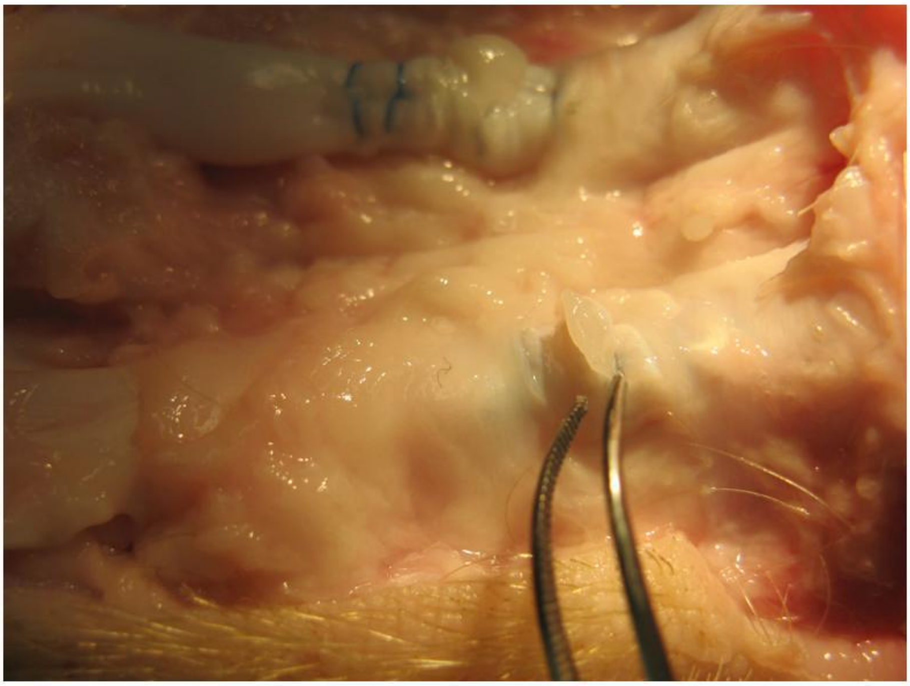

Figure 3.3. The Tisseel lifting from the tendon repair site after flexion – extension cycle through the flexor tendon sheath and A2 pulley.

The implications of increasing the irregularity of the repair as seen in figure 8.3 would be a higher WOF and Peak force resistance. The affect of the fibrin on adhesion formation inside the flexor sheath may also be affected, although in a recent study by

He et al, using a rabbit model no effect on range of motion testing was seen at 3 or 8 weeks after tendon repair with Tisseel (256), no histological examination was performed. The animals were left to move around freely post operatively. In their study, no work of flexion measurements were done, a "range-of-motion apparatus" was used to measure flexion at the proximal and distal joints and they report reduced adhesions. Others have reported an increased in adhesion formation using the same rabbit model (257). Restriction of movement at the initial time of surgery would be expected with the results seen in this small sample number as the Tisseel folded back on the tendon causing an increase in the bulk of the repair at this site. It is also possible that the entire applied Tisseel sample would loosen with more flexion-extension cycles and thus migrate within the flexor sheath. In the study by He et al., they report that no Tisseel was seen at the site of the repair, and the authors have concluded that this is due to the rapid degradation of the Tisseel in flexor tendons, examined at 3 and 6 weeks. It may however be that the Tisseel has broken apart from the repair site as our findings indicate and degraded elsewhere. More work using a live model and histological studies are indicated to determine this, which goes beyond the scope of this PhD thesis.

Chapter 4 Project Rationale

4.1 Background

The choice of which flexor tendon repair technique to use clinically varies from surgeon to surgeon, and is often related to the tradition of the various surgical units and surgeon preference. The ventral locking modified Kessler technique (described in section 5.3) was pilot tested biomechanically in small number of porcine extensor and flexor tendons. Tensile tests were performed by a plastic surgeon in collaboration with the Biomedical Engineering department at the University of Strathclyde. An Instron Tensile testing machine was used, and the suture calibre used was 3/0 Prolene for the core suture and a 5/0 Prolene for the peripheral repair. The MK repair used incorporated only about 15 % of the tendon surface, with a suture bite purchase of 0.7cm for the core and 3 mm for the peripheral suture. These initial tests showed a statistically significant difference in the ultimate tensile strength of the repairs compared to the standard Modified Kessler technique. This finding led to the LMK repair being introduced in the clinical setting by two local Plastic and Hand Surgeons. The feedback was positive on the technical aspect of the repair, with the repair being easy to perform and not perceived as adding time to the operation (2). The clinical results confirmed that seen in the laboratory testing, with less ruptures of the LMK repair than with the MK repair. A more in depth investigation is sought to determine the exact biomechanical properties of the repair not only with regards to tensile testing

but also with regards to work of flexion. A measure of the cross-sectional area of the repairs would be useful as an adjunct to the WOF measurements.
Nano-grooved PCL sheets will be investigated for biomechanical properties to see how this peripheral repair method (section 2.5.4) might compare to a standard peripheral suture repair.

The literature review has highlighted the various testing models and set ups described in the literature (section 2.4.1 and 2.4.4) and using this information a suitable anatomical model should be identified that provides similar anatomy to the human hand (sections 2.2). An investigation of porcine forelimb anatomy will be performed through dissections. The biomechanical testing will be performed using the Instron tensile testing machine as will the WOF testing. The WOF testing set up is based on the report by Zhao et al (144), using the Instron testing machine and a custom made ring transducer made within the department.

4.2 Aims

The Primary aim of the research is to compare the biomechanical properties of a conventional tendon repair technique to a new core suture configuration and a new peripheral repair method.

The primary aims of the research project are as follows:

1. Evaluate the mechanical strength and work of friction of two 2-strand repair techniques, MK versus LMK.

2. Evaluate the mechanical strength and work of friction of the ‘gold standard’ 4-strand repair technique (MK) relative to a novel 4strand repair method (LMK).

3. Evaluate the mechanical strength and work of flexion of a novel peripheral repair method incorporating a nano-grooved PCL sheet.

<u>The secondary aims:</u>

The secondary aims of the project are associated with the methodological aspects of the research, and are developed and investigated to evaluate the level of error associated with the measurements for the primary objectives of the project. They are identified as follows:

1. Identify a suitable animal model that effectively replicates human finger tendon anatomy.

2. Set up a mechanical testing procedure for evaluation of the ultimate tensile strength (UTS) and work of flexion (WOF) of flexor tendons.

3. Evaluate the effect of surgical training on the model.

4. Evaluate the effect of conditioning on the work of flexion measurements.

Chapter 5 Materials and Methods

5.1 Specimens

Fresh porcine forelimbs were obtained from commercially slaughtered animals on the day of slaughter (mean age 6 months) and were refrigerated at 4°C prior to dissection and repair. Dissections and repairs were performed over a period of 3 days during which specimens were refrigerated (4 °C). The accessory digits were disarticulated through the tarsal-metatarsal joints and the central two rays exposed. A sharp incision was made in the flexor sheath and the FDP tendon about 5 mm distal to metatarsophalangeal joint; a level corresponding structurally to flexor tendon zone II, or IIB by Tangs classification (26). The FDP tendons were repaired within the flexor sheath, and were frozen, for 8 ± 3 weeks at -12°C, until the time of testing according to study by Hirpara et al looking at the effect of freezing on tensile properties of porcine tendons (258).

5.2 Anatomical Observations

An initial 15 porcine trotters were dissected on the day of slaughter for observation of the anatomy of the tendons, pulleys and vascular supply. This allowed for dissection practice, familiarity with the anatomy and ensured that tissue handling was gentle and without undue trauma to the tendons and repairs in further samples. The 2 central rays of the porcine forelimb were studied, and the 2 accessory digits removed. The tendons of these were cut at the level of disarticulation. The volar skin over the 2 central rays was incised and retracted to expose the underlying soft tissue and the flexor sheath. The overlying soft tissue was dissected off the flexor tendon sheath starting just proximal of the MCP joint, to the distal phalanx. The number and location of the condensations in the flexor sheath were noted. The sheath was then incised to reveal the underlying tendons, the location and insertion of the vincula, as well as the relationship between the 2 tendons, and their insertions.

5.3 Core Suture Repair Techniques, MK vs LMK

Matched specimens from each trotter were randomly assigned to either a Pennington modified Kessler (MK), or a new ventral-locking, looped modified Kessler (LMK) repair technique (Figure 5.1). All repairs were performed with a locking configuration as per Penningtons definition (154). The longitudinal components of the core suture lay deep to the transverse component (locking technique) which was placed 0.7-1.0 cm from the cut tendon end (154). Each locking loop comprised approximately 25% of the cross-sectional area of the tendon (103, 108). Knots were placed between the cut tendon ends, and consisted of one double throw and three single throws (259). All repairs included a 5/0 polypropylene continuous simple, running peripheral suture in which each suture grasp visually penetrated about 3 mm into the tendon (216). Peripheral sutures contained on average 12 loops. All the repairs were performed by the same investigator.

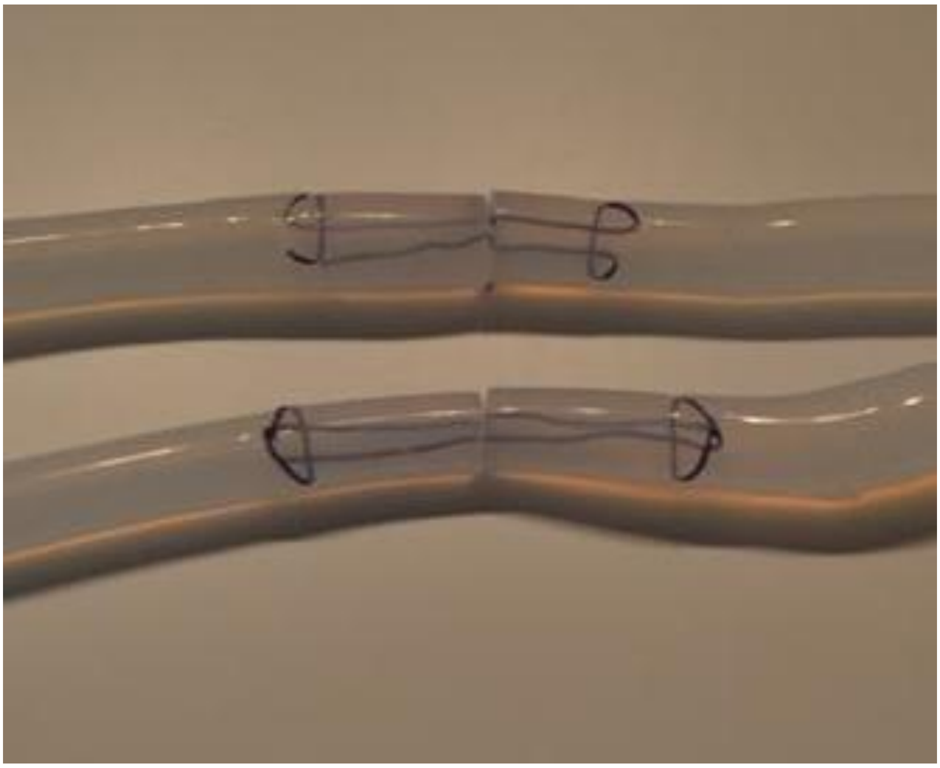

Figure 5.1: Silicone model of the two suture techniques, MK (top) versus LMK (bottom).

Further samples were dissected and prepared as for two strand repair, inside the digital flexor sheath. An additional suture strand was incorporated into each repair type, resulting in a four strand core suture repair, MK and LMK (Figure 5.2). The repairs contain two knots inside the tendon ends, and were performed using a 3/0 Ticron suture. The first suture was placed centrally on the volar aspect of the tendon, with approximately 9 mm purchase, and the second suture was carefully placed parallel to the first, with an approximate purchase of 7 mm, without disrupting the first suture. A 5/0 prolene continuous simple running peripheral repair was then placed as in the two strand samples.

Figure 5.2: 4 strand LMK repair without the peripheral suture placed.

5.4 PCL Peripheral Repair Method

A new peripheral repair type was introduced using a 4 strand LMK repair, were the simple continuous peripheral suture was replaced by a PCL sheet embedded with nano-grooves (figure 5.3). The PCL was fabricated at Glasgow University, from PCL pellets that were melted under pressure between 2 glass plates after they were washed with methanol, and dried (224). Using hot embossing technique (224) grooves and ridges were embedded on the PCL sheet, with a groove/ridge width of 12.5μm/ 5.0μm. The tendons and flexor sheath were prepared in the same manner as for the 4 strand repairs. The PCL sheets were placed so that the centre of the sheath covered the peripheral suture, with its grooved side placed onto the tendon and grooves/ridges in line with the tendon fibres. The sheath was placed around the tendon with the grooved surface in continuous contact with the tendon and sutured to the tendon with a continuous simple running suture, using 6/0 Prolene at the proximal and distal edges. This ensured a firm fit of the sheet to the tendon.

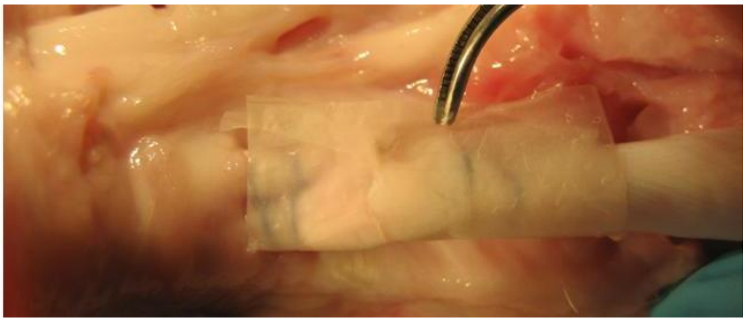

Figure 5.3. Picture showing the placement of the PCL sheet around the tendon repair.

5.5 Methods and Instrumentation

5.5.1 Methodological Considerations

5.5.1.1 Evaluation of Potential Learning Effects

Initially 80 tendons were repaired using a 2 strand repair technique of either MK or LMK configuration. The first 10 repairs were done with supervision and were subsequently discarded from further testing and analysis. 3/0 Prolene or 3/0 Ticron was used as the core suture material for each repair technique as studies have shown that using 3/0 suture increases the tendon repair fatigue strength compared to 4/0, independent of peripheral suture technique (118). The repairs were done over a 30 day period, and no formal feedback was given during this time. The samples were then tested in an Instron Tensile Testing machine for UTS, Yield, 2 mm gap formation and stiffness. They were then evaluated for a learning curve, by dividing the repairs into 3 groups (n= 70); 1. Repairs performed in the first ten days (early repairs n= 24) 2. Repairs performed on days 11 -20 (intermediate repairs n= 29). 3. Repairs performed after 20 days (late repairs n= 17).

5.5.1.2 Factors influencing the measurement of internal work of flexion.

Number of testing cycles

Each tendon and repair was tested in three cycles to evaluate the degree of conditioning over the testing period, which early on during testing seemed to have an effect on the WOF measurements.

Sheath Repair

The initial intact tendon was tested within the sheath as a baseline measurement of the WOF. The repair was subsequently cut, and the system retested to evaluate for any influence of sheath repair. The same procedure was performed after the tendon had been repaired; the system was tested with the flexor sheath open and repaired.

5.5.1.3 Evaluation of properties of PCL sheets

The PCL sheets were tested in an Instron tensile testing machine with a 10 N load cell. There were three specimen types tested. One group of non-grooved sheets, a second with the grooves oriented parallel to the testing axis (figure 5.4) and a third group with the sheets perpendicular to the testing axis, to evaluate the effect the grooves/ridges have on the mechanical properties of the PCL sheets. The PCL sheets were cut using dermatome and secured with pneumatic grips (80psi) in the Instron machine. The samples were preconditioned using 10 cycles at 1.5 mm (4.3%) at a rate of 72 mm/min). Specimens were then taken to 105 mm (300% strain) at the same rate, at this point there was significant deformation of the specimens and the testing stopped. A small number of PCL sheets where evaluated for UTS (n=5), at the same rate.

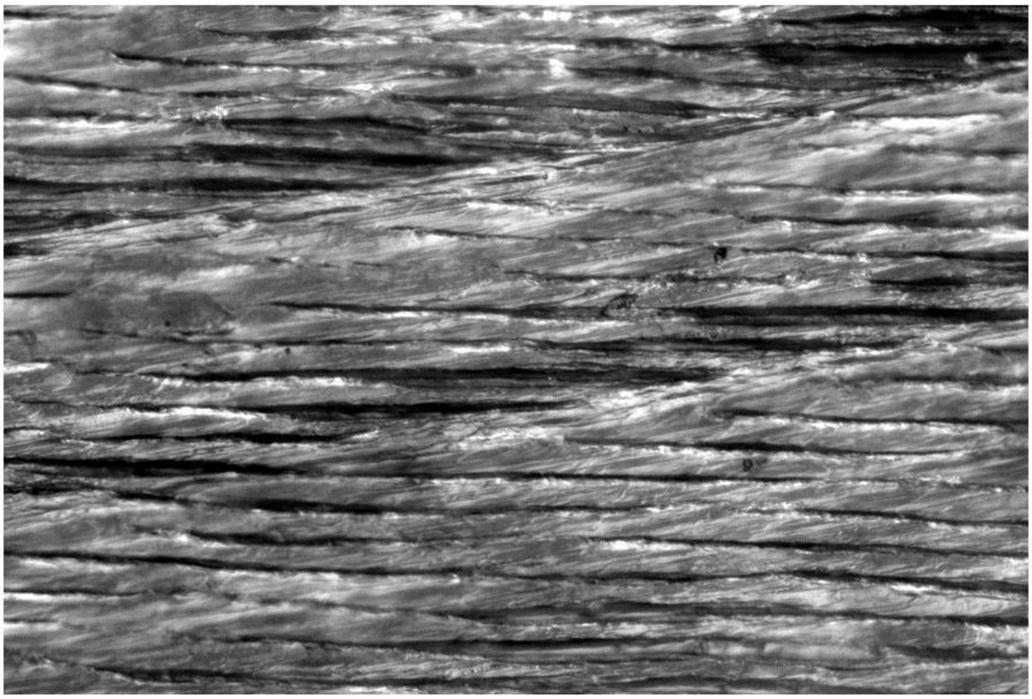

Figure 5.4: Electronmicroscope photo showing the grooves and ridges of the PCL sheets. The dark lines shows the grooves and the lighter lines the ridges.

5.5.2 Tensile Testing Protocol

The samples were thawed at room temperature on the day of testing. At this time the FDP tendon was dissected free from the FDS tendon and the flexor sheath. The FDP tendon was left attached to the distal phalanx, the attachment to the middle phalanx was severed. Proximally the tendon was divided centrally at the common tendon. After thawing, repaired tendons were mounted for mechanical testing within an Instron materials testing machine (Instron Materials Testing Machine 5800R, Instron, High Wycombe) fitted with an 100 N load cell (Force resolution 0.1 N). The distal end of the tendon-bone specimen was potted into steel tubing using a two-component styrene-based filler (Isopon P38, U-Pol, Wellingborough, Northants UK) and clamped within a mechanical, screw-action grip. Care was taken to ensure the tendon was not in contact with the potting mix. (Figure 5.5). The proximal end of the tendon was mounted in the uniaxial material testing machine with the longitudinal tendon fibres aligned with the axis of loading and secured within customised cardboard-lined wedge action grips. This gripping technique has been shown to be effective in limiting slipping at the tendon-grip interface (260). Once mounted within the materials testing machine, samples were preloaded to 2 N as a nominal, no-strain, no-load condition. Planar circular markers, 5.25 mm in diameter, were fixed to the tendon using 0.25 mm diameter tissue pins, which have shown to have negligible influence on the stress-strain curve of soft tissue (226). The markers were positioned approximately 10 mm each side of the repair and permitted optical estimation of tendon strain. Preloading at the time of fixation ensured minimal out-of-plane motion during testing. An additional marker set, consisting of two planar markers separated by 21.01 mm, were rigidly fixed to the lower grip and provided a calibration distance for later calculation of tendon gapping. Marker displacement during testing was subsequently recorded at 25 Hz by digital video (Canon XM1 3ccd Digital Video Camcorder, Canon Europe, Amstelveen, Netherlands), which was synchronized with the Instron load cell via a light emitting diode (LED) positioned within the field of view.

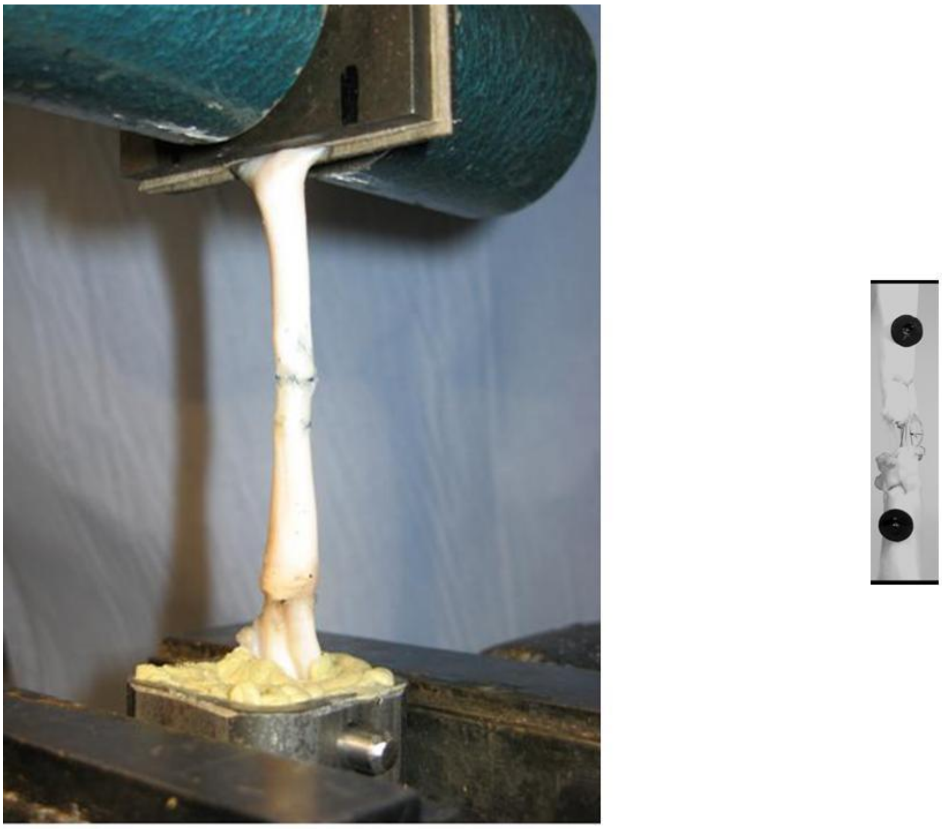

Figure 5.5. Picture of the test set up in the Instron machine of the tendon repair, at the start of tensile testing on the right, and image of the tendon repair and markers at the end of testing (left).

Prior to failure testing, specimens were preconditioned between 1-10 N for five cycles. Intact tendons (n= 4) were initially tested in the Instron to failure(pullout from the grips), using a 1000N load cell. Tendon specimens were then distracted until failure at a constant rate of 25 mm/min (78, 92).

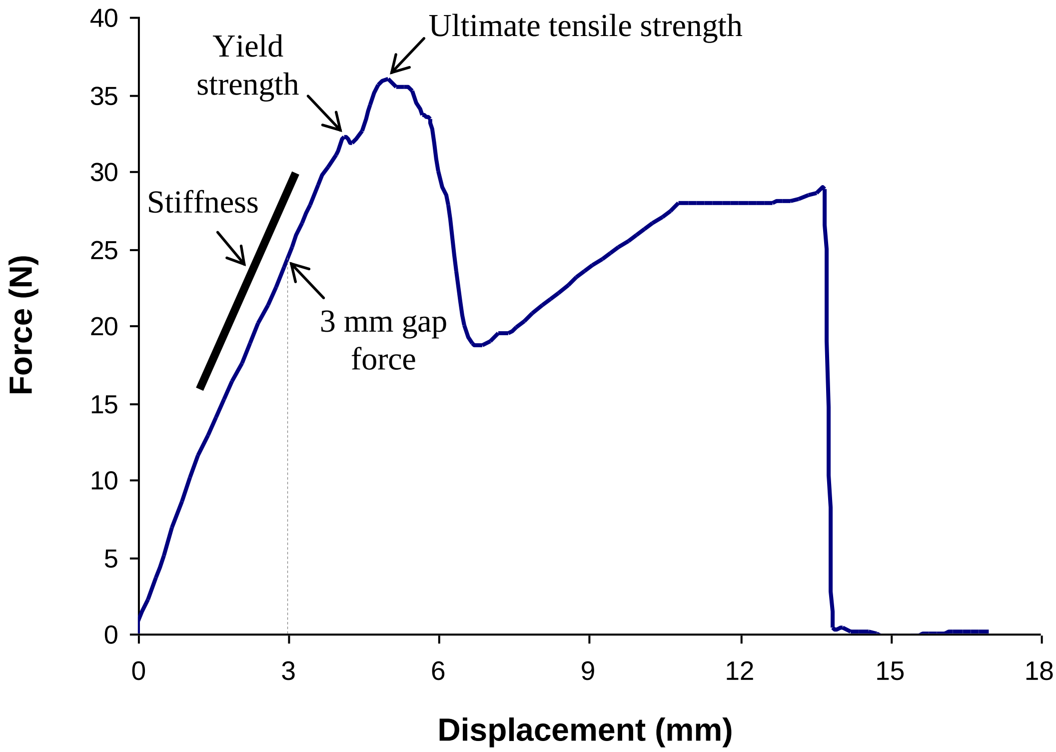

Figure 5.6. A typical force (N) versus displacement (mm) curve.

Video sequences (digital AVI files) were decompiled into tagged image format files using customized software. Grey scale images were segmented to binary images using the OTSU thresholding method. For each image, customized Matlab software was used to calculate the co-ordinates of the centroid of each marker and tendon displacement was subsequently calculated. Force –displacement curves were subsequently generated (Figure 5.6) and the following tensile parameters were calculated.

1. **Gap formation:** Tensile force at marker deformations of 1, 2 and 3 mm, representing ‘repair gapping’, were recorded.

2. **The ultimate strength** (UTS) of the repairs was defined as the peak force recorded during the load to failure test.
3. **Yield** was defined as the point where slope of the load-displacement curve first clearly decreased, (defined as the force at which the slope of the load displacement curve is less than or equal to zero (261).
4. **Stiffness** (force per unit of distraction) was defined as the slope of the linear region of the force-displacement curve and was identified by fitting a linear regression line to one third of the stress-strain curve while minimising the error using a least squares approach (170).

5.5.3 Work of Flexion Protocol

Paired bone-tendon samples were initially dissected severing the distal insertion of the FDP to distal phalanx and the volar plate attachment. Proximally the tendon was divided centrally at the common tendon, allowing the tendons to the two rays to move independently. The tendons were then repaired as described in section 5.3. The two central rays were then disarticulated taking care not to disrupt the attachments of the pulley and sheath. The distal attachment of the tendon to bone was dissected free. A WOF mechanical set up was constructed based on the set up by Zhao et al, using the Instron tensile testing machine and a small custom-made, lightweight, force transducer (aluminium ring (HE30) configuration with full Wheatstone bridge) made within the department (144). The ring transducer was calibrated (Figure 5.7), and then secured to the distal tendon using a loop of Ticron suture via a bone plate and a steel cable (Figure 5.7).

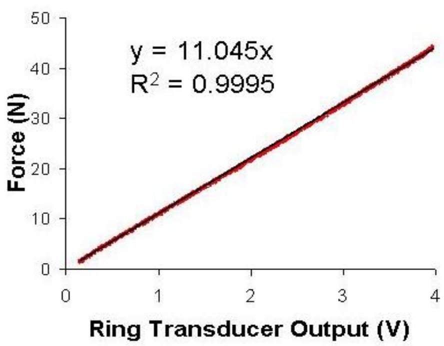

Figure 5.7. Calibration curve for the ring transducer, and close up picture of the transducer.

The proximal phalanx was subsequently mounted within a test frame and the proximal end of the tendon secured to a 100 N load cell of an Instron Uniaxial materials testing machine (Instron Materials Testing Machine 5800R, Instron, High Wycombe). Initial pretension was applied to the tendon by a 100 g weight connected to the distal transducer (Figure 5.8), which also ensured full extension of the digit. The tendon was then pulled proximally over a 30 mm excursion, causing digital flexion, at a rate of 2 mm/s and the force between the proximal and distal ends of the tendon recorded. The samples were tested for 3 cycles. Given that the applied loads were small (<10 N), it was assumed that deformation of the repair was minimal. Thus the peak force differential and the internal work of flexion (differential-force-tendon-excursion integral) were calculated as measures of gliding resistance (Figure 5.9). The peak force differential was related to the peak of the curve corresponding to the repair passing under the pulley. Differences between repair techniques were evaluated using a one–way analysis of variance (ANOVA). Statistical significance was set at 0.05.

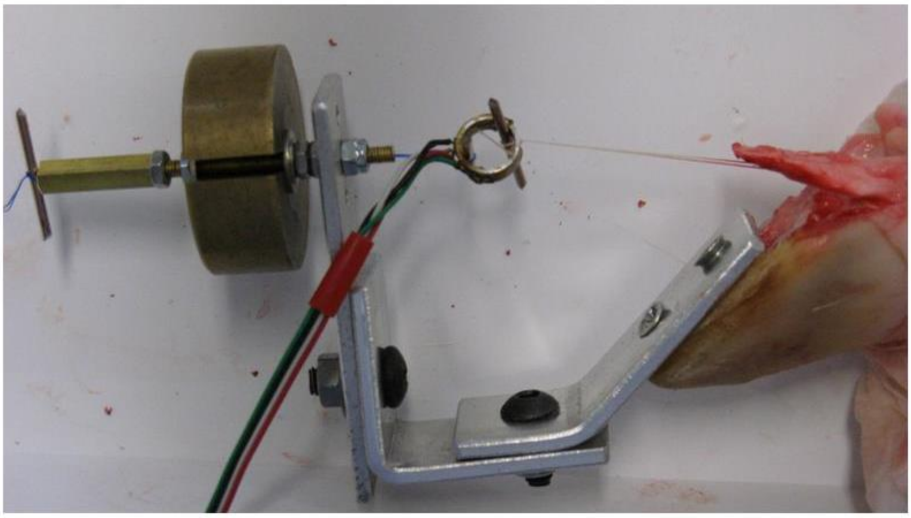

Figure 5.8. Set up of ring transducer and the weight connected to the distal phalanx

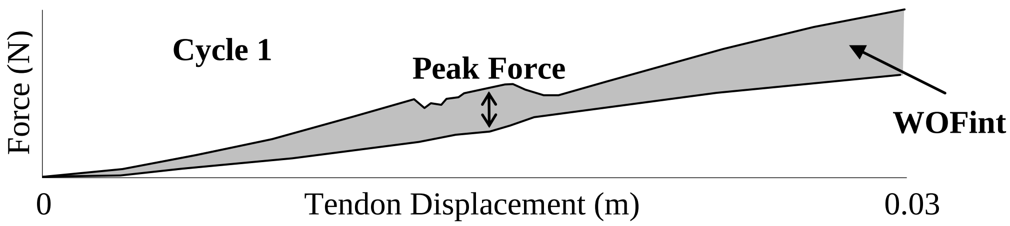

Figure 5.9. Illustration of a typical force profile measured at the proximal (upper line) and distal aspects (lower line) of the Flexor Digitorum Profundus (FDP) tendon during the first flexion cycle. The area between the two curves (shaded) represents the internal work of flexion (WOFint), while the maximum difference in force throughout the flexion cycle force represents the peak force.

5.5.4 Tendon & Repair Dimension Measurements

The tendons were dissected free from the tendon sheath and placed in a plasticine mould. The cross-sectional area of intact FDP porcine tendon at the level of the vincula (proximal to the repair) and at the repair site was determined using a previously published method (262), in which an impression of the tendon is made using an alginate compound (Trylon Ltd). The repaired specimens were placed in a rectangular mould lined with plasticine. An expansion was made in the plasticine at the area of the repair and at the level of the proximal vincula. This was designed so that there was an area around the repair site of wider size to allow the alginate mix to completely surround the repair. A mixture of one part alginate powder to one and a half parts of water was prepared immediately before application to the tendon. This was then used to cover the tendon and repair site circumferentially. After the alginate had set (approx 2 min) the plasticine-alginate/tendon cast was incised. A 1-2 mm slice using a dermatome blade was cut of the plasticine, alginate- tendon complex at the level of the peripheral repair site, i.e. the bulkiest point of the repair, and another slice (1-2 mm) was taken at the level of the proximal vincula (approximately 1 cm from the end of the repair). These were then photographed (Figure 5.10) and using the same custom built- software as used in section 5.4.2, the cross-sectional area for the bulkiest part of the repair calculated using Otsu thresholding method described in section 5.4.2, and the number of pixels for tendon cross section determined.

To assess the alginate compound (Trylon LTD) casts were initially made of a metal rod with known diameter (5.5 mm diameter).We repeatedly photographed these alginate casts over time at stable room temperature and humidity, to see how quickly the material would contract. Using the software previously described we found there to be insignificant contraction of the casts for the first 30 min, correlating to the work by Goodship and Birch that found no significant contraction for 10 min they tested (262).

All cast slices of the tendon as well as the calibration samples were photographed with a calibration disc of known internal diameter, within 15 min period.

The images were then converted to greyscale and using custom made software the area of the repairs was calculated.

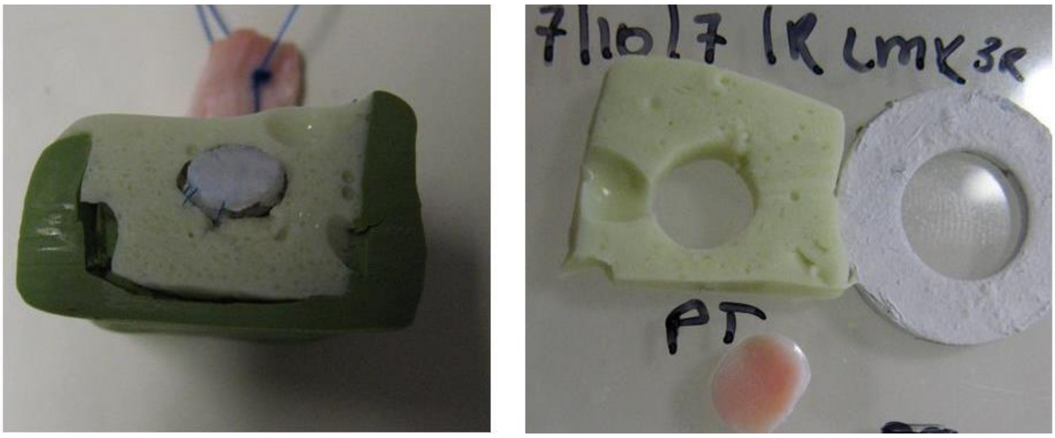

Figure 5.10. Picture on the left shows tendon repair in alginate/plasticine mould. On the right the calibration disc, alginate and tendon photographed, which was then used for calculation of the tendon repair dimension.

5.6 Research Design

5.6.1 Two strand repairs

After the initial testing of tendon repairs for a learning effect was completed, a further 74 samples where tested using the tensile testing protocol. 39 tendons were repaired using the Modified Kessler (MK) technique, of which 20 were repaired using a Prolene core suture, and 19 using Ticron. 35 tendons were repaired using a ventral locking looped modified Kessler repair (LMK), of which 18 were repaired with a Prolene core suture, and 17 with Ticron. Both 3/0 Ticron and 3/0 Prolene was used for the core suture in the initial testing set up, as these are both used in clinical practice.

The WOF protocol was set up and a total of 89 two strand repairs were performed for the WOF testing. The cross sectional areas of the repairs were evaluated as set up in the previous section, section 5.4.4.

The statistical calculations were all performed using SPSS, and a 2 (repair) x 2 (suture) way ANOVA was used to compare the two strand data.

5.6.2 Four strand repairs with Novel PCL technique

A stainless steel cable and hooks were introduced to the WOF testing protocol to replace the Ticron loop used for the two strand core suture testing to try to eliminate any error in the testing from suture extension. For the initial internal work of flexion test the tendon sheath was left intact, and the intact tendon and intact tendon sheath was tested to measure friction in the system (SIntact). A window was then cut in the flexor sheath, zone II, between A2 and A3 pulley and the sample retested with the sheath incised (Cut S). After this the tendon was divided in same manner as before, at the same area as described in section 4.3 and repaired with either a MK or a LMK repair using a 3/0 Ticron suture, and a simple running peripheral suture 5/0 Prolene. It was then tested for work of flexion. After this the sheath was repaired with three mattress sutures, using 6/0 Prolene and the specimen retested. A 6/0 suture was used instead of 5/0 suture due to the fragile nature of the PCL. For all the samples the peak force differential and the internal work of flexion was calculated and statistical significance calculated using a 1 way analysis of variance (ANOVA) as used for the 2 strand repairs.

Finally the tendons were mounted in the Instron machine and the tensile properties were determined as for the 2 strand repairs in section 5.4.1.

The data obtained from the tensile testing and WOF was analysed in the same manner as for the 2 strand repair, using SPSS software (SPSS Inc. Chicago, Illinois) the same statistical calculations (ANOVA) were performed.

The 4 strand repairs where evaluated using a previously described method (section 5.4.3), with the following modifications. In addition to measuring the central repair and the area of the vincula, a third area in between these, was measured to see if there was any element of compression from the suture configurations in this area. The tendons were removed from the mould by gently incising the mould on the dorsal side of the tendon, care taken not to interrupt or cut any of the sutures on the surface of the tendon. The mould was then cut using same technique as in section 5.4.4 to obtain the 3 cross sectional areas of the tendon and repairs. Using the same software and camera set up as earlier, the diameter of the tendon and repairs were calculated.

The PCL sheets were subsequently introduced to the LMK repair and tested using a slightly modified testing sequence from that of the 4 strand repairs:

1. the intact system tested
2. the tendon sheath was cut and a 2.0 cm long sheet of PCL ,30μm thick, was wrapped around the intact tendon centred where the tendon would be lacerated (Step 3).
3. The tendon was then cut in zone 2 as for previous tests (sections 5.4.1) and repaired using a LMK repair of 3/0 Ticron and tested
4. A 2.0 cm long sheath of PCL was wrapped around the tendon at with the centre of the sheath over the repair site, with the grooved surface facing towards the tendon, covering all the sutures on the tendon surface. The repair was then tested with and without the tendon sheath being closed, as with previous testing.

Throughout the testing the tendon and PCL was kept moist by spraying the samples with Ringers solution, or covering the samples with gauze soaked in Ringers solution.

The samples where then evaluated for the same tensile properties as for the 2 and 4 strand repairs (sections 5.4.2). The same set up was used with the markers attached to the tendon in the area coved by the PCL, and the same parameters as in previous tests were measured and calculated. The failure mechanisms of the repairs were noted. No cross sectional diameter measurements were performed.

Chapter 6 Results

6.1 Anatomical Comparison

The dissection part of the study showed that the porcine model has a very similar anatomy to that of human digits. Deep and superficial flexor tendons were enclosed in a flexor sheath with specific and consistent condensations in annular and oblique configurations in all dissected specimens. The pulley configurations were found to be comparable to those described by Smith et al., in their study looking at porcine forelimbs (132). The first annular pulley (A1) is situated over the MCP joint, followed by A2, A3 and A4 (132). The pulleys attached to the phalanx itself or to a sesamoid bone (A1). The A1 pulley was found to be the largest of the pulleys identified, followed by the A2 pulley which was located over the proximal part of the proximal phalanx in the majority of the dissections, with some of the A2 pulleys attaching to the MCPJ. The smallest of the pulleys identified was the A4 pulley was located over the DIPJ. As previously described by Smith et al. (132) the number of cruciate pulleys were less defined with one cruciate pulley present in all the specimens between the A3 and A4 pulleys, while in some specimens a thin connection was seen between the A2 and A3 pulleys, but this was only clear (with 3.3x loupe magnification) in about 50% of the specimens dissected (Figure 6.1).

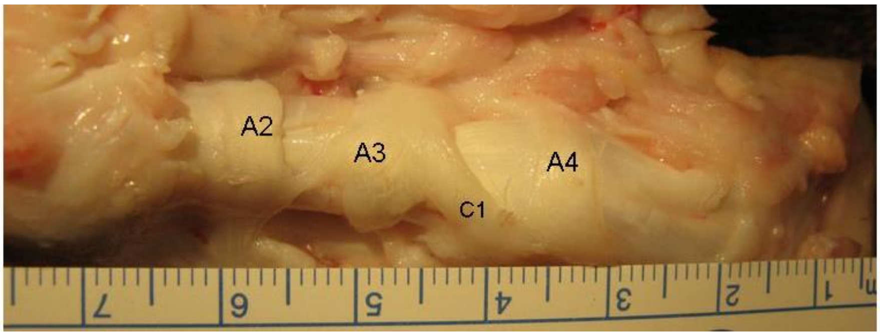

Figure 6.1. The porcine pulley system.

The superficial flexor tendon (FDS) attaches to the middle phalanx on the volar aspect after decussating. The insertion of the FDS was on the middle phalanx distally after the decussation. At this point the FDS tendon envelopes the FDP, as the FDP becomes more superficial. The FDS attaches deep to the FDP in all the specimens. The profundus tendon inserts on the distal phalanx, with a clear and significant attachment to the volar plate at DIP joint (Figure 6.2).

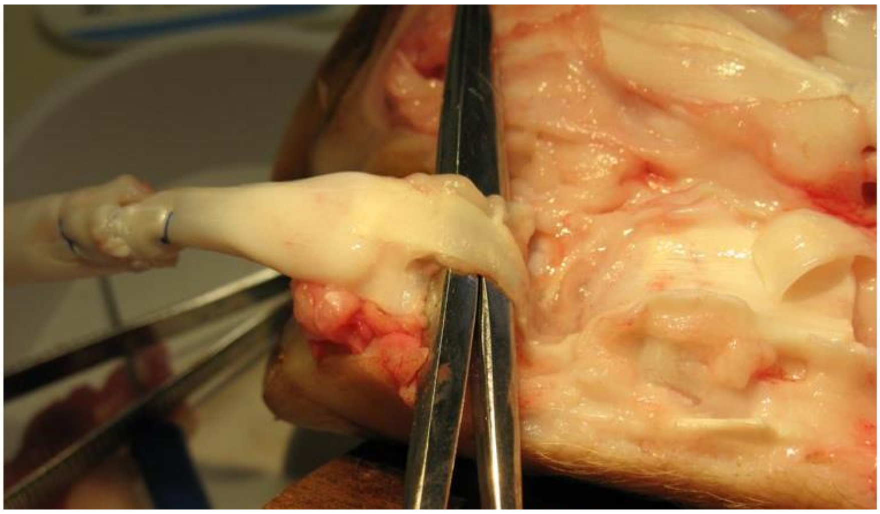

Figure 6.2. The attachment of the FDP to the distal phalanx, showing a slip attaching onto the volar plate.

More proximal to the insertion of the FDS, the FDS envelops the FDP, resulting in FDP not only moving through the flexor sheath/pulley system but also through the FDS tendon over the MCPJ area, zone 2. As the tendons enter into the vola the FDP tendons from the 2 digits join to form a common tendon (Figure 6.3) and they also join with the FDP going to the accessory digits.

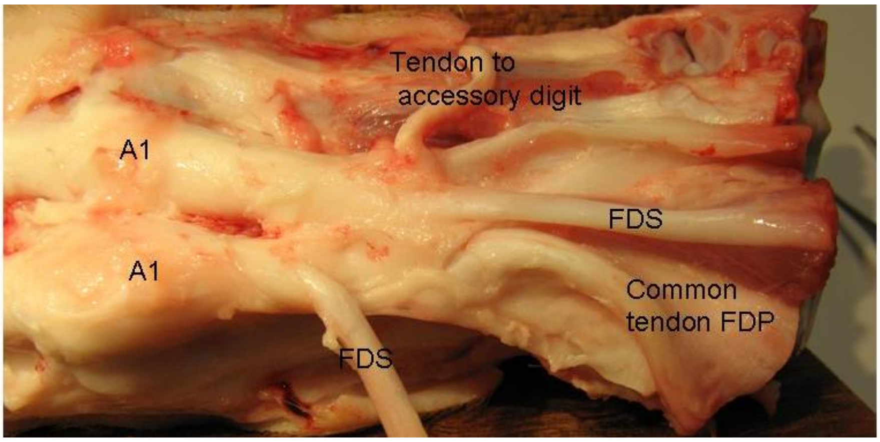

Figure 6.3: Anatomy of the FDS and FDP in the proximal part of the porcine forelimb

At the point where the FDS tendon splits and allows the FDP tendon to come through, the vincula was observed attaching to the dorsal aspect of the both the FDP (vincula longa) and FDS (vincula brevis) tendons (Figure 6.4).

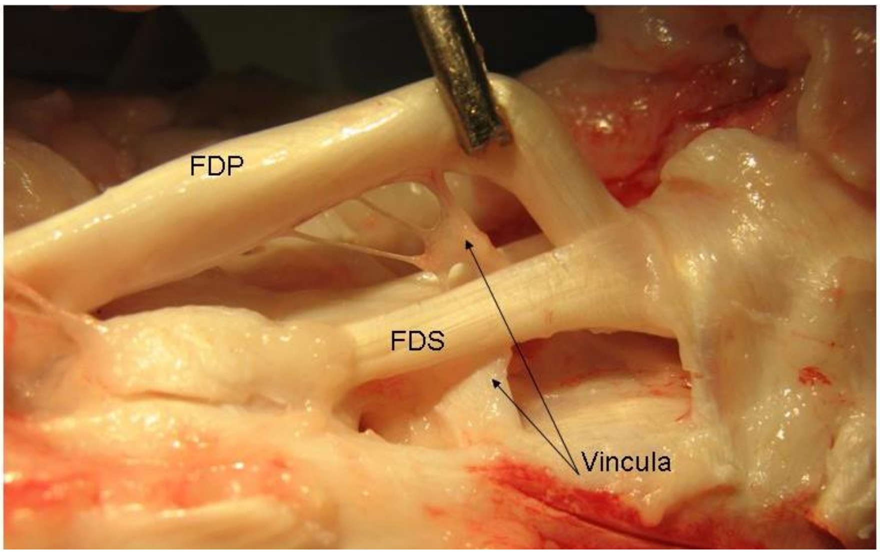

Figure 6.4. Decussation of the FDS with the FDP passing through and the vincula longa visible between the tendons attaching on the dorsal aspect.

6.2 Methodological Considerations

6.2.1 Evaluation of potential learning effects of tendon repair

The statistically significant differences of the mechanical properties over the training period are presented in Figure 6.5 a)-d). There was a statistically significant increase in the force to 3 mm gap formation, UTS and yield for the late repairs as well as a significantly lower stiffness with the late repairs (P< 0.05).

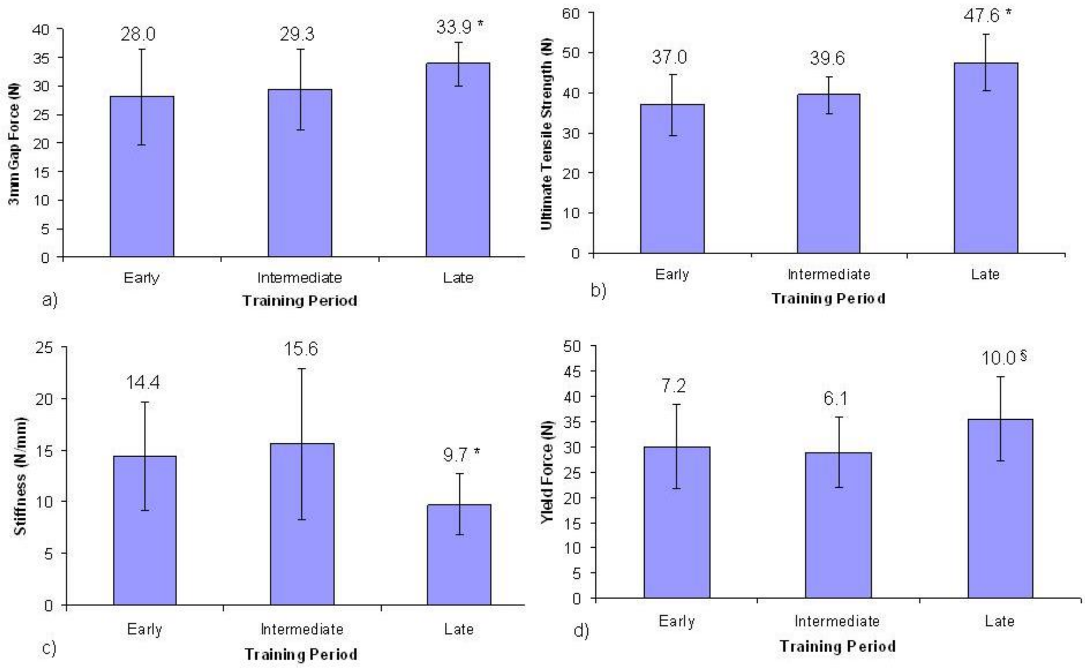

Figure 6.5. The effect of training period on the force to 3 mm gap formation (a), UTS (b), stiffness (c) and yield force (d).

6.2.2 Factors Influencing Measurement of the Internal Work of Flexion

6.2.2.1 Effect of Cycle Number

Representative forces at the proximal and distal tendon with three flexion-extension cycles are shown in Figure 6.6. Looking at the 3 cycles of the WOF a significant difference was seen between cycle 1, 2 and 3, with a reduction in the area between the proximal and distal force versus excursion graphs. For statistical analysis a repeated measures ANOVA was used, looking at the repair state and cycles with the statistical significance set at 0.05.

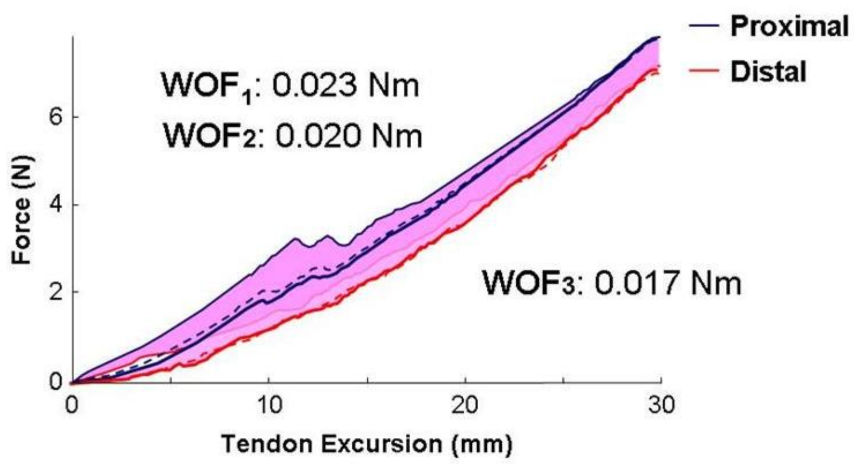

Figure 6.6. Graphs showing the effect of conditioning on WOF after 1, 2 and 3 cycles.

The effect of conditioning on the WOFint is presented in table 5.1. The conditioning effect reduced the WOFint of FDP tendon by 12% during cycle 2, and 4% during cycle 3, Figure 6.7. This was opposite to the effect of the surgical repair. The conditioning effect was also more pronounced with the surgical repair, and following the 1st cycle.

Table 6.1: **Mean (SD) Internal Work of Flexion for the three cycles of testing of the samples and pooled repair types, showing the effect of preconditioning on the system**.

	Cycle 1	Cycle 2	Cycle 3
Intact	0.017 (0.007)	0.016 (0.005)	0.015 (0.005)
Incised sheath	0.015 (0.006)	0.015 (0.005)	0.015 (0.005)
Repaired tendon(pooled techniques)	0.030 (0.012)	0.024 (0.008)	0.021 (0.009)
Repaired tendon and sheath	0.026 (0.008)	0.024 (0.006)	0.023 (0.006)

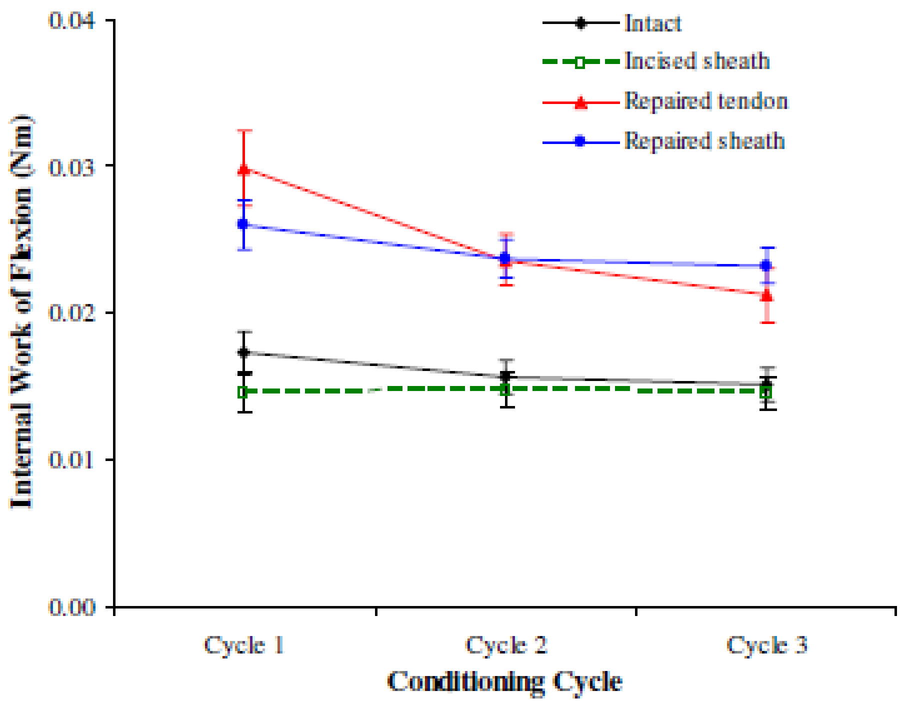

Figure 6.7: The effect of conditioning and the repair on the internal work of flexion, error bars representing the standard deviations.

6.2.2.2 Effects of sheath incision and repair

No difference was seen in the WOF between the intact system and the incised flexor sheath, with the value for both 0.015 Nm (SD 0.005).

No statistically significant difference was seen in the WOF or the peak force differential with repair of the flexor sheath (Table 6.2).

Table 6.2. Mean (SD) Internal Work of Flexion and Peak force differential for Tendon Repairs at cycle 3

	Ticron	
	MK	**LMK**
N	11	10
WOFint (Nm) core suture, w/o repaired sheath	0.0196 (0.005)	0.023 (0.009)
WOFint (Nm) core suture, with repaired sheath	0.022 (0.004)	0.023 (0.006)
Peak Force Differential (N)core suture w/o repaired sheath	0.96 (0.33)	1.10 (0.49)
Peak Force differential (N) core suture with repaired sheath	1.12 (0.35)	1.10 (0.54)

6.2.3 Evaluation of the Structural Properties of PCL sheets

The structural properties of PCL sheets with and without microgrooves are shown in table 5.4. The PCL specimens were found to fail at tensile force of 5.83± 1.32 N at a strain of 640.7± 69.4%. The orientation of the grooves resulted in statistically different stiffness and yield force ($P<0.05$). Figure 6.8 shows the PCL sheets at start, middle and end of tensile testing.

Table 6.3: Structural Properties of PCL sheeting extended to 105 mm (~300% strain)

	Non Grooved	Orthogonal	Parallel
N	11	10	11
PCL length (mm)	35	35	35
PCL Width (mm)	10	10	10
PCL thickness (micron)	30	30	30
Stiffness (N/mm)	1.40 (0.21)	1.00 (0.21)	1.59* (0.06)
Yield Force (N)	2.40 (0.36)	1.82† (0.24)	2.46 (0.27)
Displacement at Yield (mm)	2.8 (0.6)	3.1 (0.2)	2.7 (0.7)
Peak Tensile Force (N)	2.58 (0.27)	1.85† (0.31)	2.71 (0.14)
Displacement at Peak Tensile Force (mm)	58.5 (49.6)	89.5 (30.2)	65.3 (48.3)

UTS measures were performed and the PCL ruptured at 5.83 +\- 1.32 N at a strain of 640.7 +\- 69.4 %.

Figure 6.8. The PCL sheets at various points in the tensile testing, showing the deformation of the material from the start of the testing to failure.

6.3 Tendon and Repair Dimensions

A comparison was performed of the dimensions' of the 2 strand tendon repairs, to see if there was a difference between the MK and LMK repairs. The mean cross sectional area of the LMK repair was 33.9 mm^2 versus 32.7 mm^2 for the MK. Results are outlined in Table 6.4. No statistically significant difference was seen between the 2 repair types, but the central repair cross sectional area was significantly larger than the proximal cross sectional area (P = 0.001).

Table 6.4. Mean (SD) cross sectional area (mm^2) for the 2 strand repairs, at the central repair and proximal tendon

	MK	LMK
N	28	27
Proximal	30.5	29.9
	(4.8)	(3.3)
Central	32.7	33.9
	(6.1)	(4.7)

Average dimension of the 4 strand repairs are outlined in Table 6.5. No statistically significant difference was seen between the repairs. The mean cross sectional area of the MK 4 strand repairs was 32 mm^2 versus 34.5 mm^2 for the LMK. There was a significant difference seen between the proximal and central cross sectional tendon measurements ($P<0.05$), with the proximal site smaller than the central site.

Table 6.5 Mean (SD) cross-sectional measurements (mm^2) of the tendon and the 4 strand repairs

	MK	LMK
N	10	10
Proximal	27.0	27.8
	(4.5)	(3.3)
Central	32	34.5
	(3.2)	(6.5)

6.4 Two Strand Repairs

6.4.1 Tensile Properties

Tensile properties of the two strand repair techniques are shown in table 6.6. No statistically significant effect was found in ultimate tensile strength between the 2 repair types tested. There was however a significant difference in UTS between Ticron and Prolene suture material (Table 6.6). The two repair types did not differ in force needed to produce 3 mm gap formation in stiffness, rigidity, work or yield either, but there was a difference found between the suture materials. Ticron showed a statistically significant difference with regards to UTS, force to gap formation, stiffness and rigidity (see Table 6.6).

Table 6.6: Tensile testing results for 2 strand repairs using 2 suture types, n=74.

		MK Ticron	**MK Prolene**	**LMK Ticron**	**LMK Prolene**
N		19	20	17	18
L1mm	(N)	24.0 (4.6)	20.4 (3.3)	23.9 (3.8)	21.4 (3.3)
L2mm	(N)	30.7 (5.7)	28.1 (4.4)	31.6 (3.4)	26.5 (5.8)
L3mm	(N)	32.9 (7.3)	29.3 (7.9)	31.6 (4.8)	26.9 (7.2)
UTS	(N)	43.5 (6.6)	38.7 (6.6)	42.5 (6.6)	37.8 (8.6)
Displacement at UTS	(mm)	7.4 (3.2)	5.7 (2.8)	7.9 (2.4)	6.4 (2.9)
Stiffness	(N/mm)	14.8 (6.5)	12.1 (3.4)	16.3 (8.1)	11.5 (4.6)
Rigidity	(N.mm.mm-1)	392.3 (160.2)	327.3 (106.3)	431.7 (185.9)	313.0 (122.5)
Yield	(N)	32.7 (9.3)	31.1 (8.5)	30.6 (8.3)	28.9 (6.8)
Displacement at Yield	(mm)	1.9 (1.1)	2.3 (1.1)	1.8 (1.0)	2.0 (1.0)

6.4.2 Work of Flexion

A difference in internal work of flexion (WOFint) (Nm) was seen between the two repair types, with LMK configuration having statistically significant (P < 0.05) lower work of flexion (0.026 Nm) than the MK (0.029 Nm) (Table 6.7). When looking at the peak force differential for the two repairs, a significant difference was seen here too (Table 6.7), with LMK having lower mean value 1.35 N versus 1.85 N for the MK repair (P < 0.05).

Table 6.7: Mean (SD) Internal Work of Flexion for Tendon Repairs and Peak Force Differential.

	Prolene		Ticron	
	MK	**LMK**	**MK**	**LMK**
N	24	25	22	18
WOFint (Nm)	0.033 (0.015)	0.026 (0.010)	0.029 (0.011)	0.026 * (0.008)
Peak Force Differential (N)	2.25 (1.50)	1.54 (0.73)	1.85 (1.26)	1.35 * (0.51)

6.4.3 Repair Failures

The repairs were all observed and filmed during testing, and the peripheral suture was the first part of the repair to fail by pulling out of the tendon ends. The core sutures pulled out in 15 of the samples and in the remaining samples the core suture failed by suture rupture.

6.5 Four Strand Repair including PCL

6.5.1 Work of Flexion

The results for the intact tendon and tendon sheath system showed a WOFint of 0.015 Nm, and this did not change after the tendon sheath was incised.

The addition of a MK or LMK core suture plus a peripheral repair increased the WOF to 0.0196 Nm (30.7 % increase) for the MK and 0.023 Nm (53.3 % increase) for the LMK (Table 6.8). No statistically significant difference was found between the repair types, and repairing the tendon sheath did not influence the WOF significantly. No significant difference was seen in the peak force differential of the repair types (Table 6.9).

Table 6.8: Mean (SD) Internal Work of Flexion for Tendon Repairs at cycle 3.

	Ticron	
	MK	**LMK**
N	11	10
WOFint (Nm) core suture, w/o repaired sheath	0.0196	0.023
	(0.005)	(0.009)
WOFint (Nm) core suture, with repaired sheath	0.022	0.023
	(0.004)	(0.006)

Table 6.9: Mean (SD) Peak Force Differential for Tendon Repairs after cycle 3.

	Ticron	
	MK	**LMK**
N	11	10
Peak Force Differential (N)core suture w/o repaired sheath	0.096	1.10
	(0.33)	(0.49)
Peak Force differential (N) core suture with repaired sheath	1.12	1.10
	(0.35)	(0.54)

A significant differences was seen when testing the intact tendon versus intact tendon + PCL (P <0.05), an increase of 80 % was seen in the WOFint when the PCL sheath was added to the tendon (Table 6.10). The addition of the PCL was also found to increase the overall WOFint for the repair compared to the LMK core suture alone, and a total increase of 193% was seen when LMK core suture and the PCL sheet was used.

Table 6.10: WOFint for the 4strand LMK repair and PCL.

	LMK
N	18
WOFint (Nm) intact tendon	0.015 (0.006)
WOFint (Nm) intact tendon, PCL	0.027 * (0.017)
WOFint (Nm) core suture, no peripheral suture	0.024 (0.009)
WOFint (Nm) core suture and PCL	0.044 * (0.013)

The Peak Force differential results correlate with that of the WOFint, with a marked increase seen in the mean peak force with the addition of the PCL (Table 6.11).

Table 6.11: Mean peak force differential for the 4 strand LMK repair and PCL.

Repair Type	LMK+PCL
N	18
Peak Force Differential (N) intact	1.05 (0.47)
Peak Force Differential (N) intact + PCL	1.46 (0.77)
Peak Force Differential (N) LMK repair	1.66 (1.15)
Peak Force differential (N) with repaired sheath	2.60 * (1.18)

6.5.2 Tensile Properties 4 strand and PCL

The results of the tensile testing of the 4 strand repairs are outlined in Table 6.12. There was a statistically significant difference between repairs in the force required to produce a 1 mm gap ($F_{2,35} = 27.0$, $P < 0.001$). Post hoc analysis (Tukeys test) revealed the significantly lower force to produce a 1 mm gap in the Polymer repair compared to the other techniques ($P < 0.05$), this was also seen with the force to 2 mm gap formation. The LMK repair showed a significantly higher force to 3 mm gap formation than the MK repair, and the combined LMK/PCL repair. The LMK also exhibits a higher displacement at UTS than the MK.

Table 6.12. Results of the tensile testing of the 4 strand MK, LMK and LMK+ PCL repairs (SD).

		MK	LMK	PCL	
N		11	10	18	
1 mm Gap formation	(N)	28.7 (3.5)	27.5 (3.3)	20.9 (2.5)	*
2 mm Gap Formation	(N)	39.5 (4.7)	41.3 (2.7)	30.8 (4.1)	*
3 mm Gap Formation	(N)	41.6 (4.4)	50.5 (5.1)	39.0 (5.3)	†
UTS	(N)	66.7 (9.1)	68.0 (5.7)	68.8 (9.5)	
Displacement at UTS	(mm)	7.8 (1.7)	6.1 (1.5)	7.0 (1.1)	‡
Stiffness	(N/mm)	19.5 (5.5)	16.3 (2.8)	14.9 (6.6)	
Rigidity	(N.mm.mm-1)	446.3 (126.1)	383 (65.7)	487 (224.1)	
Yield	(N)	32.7 (7.9)	40.9 (9.4)	35.8 (15.8)	
Displacement at Yield	(mm)	1.3 (0.5)	1.9 (0.7)	2.6 (1.7)	

The combined LMK/PCL repair had a significantly higher displacement at yield than the MK, but this was not seen when compared to the LMK. With regards to the UTS no difference was seen between the 3 repairs, but for the strain at UTS there was a difference seen between all the repairs, with MK having the highest value (34.1%), versus 26.1% LMK and 21.6 % for the LMK/PCL repair.

A comparison was also run of the simple, running peripheral repair versus no peripheral repair of the 4 strand sutures. The results are outlined in table 6.13.

Table 6.13. **Results of tensile testing comparing tendon repairs with and without a peripheral suture.**

		No Peripheral suture	Peripheral suture	
N		17	21	
3 mm Gap Formation	(N)	39 (5)	46 (7)	*
UTS	(N)	69 (9)	67 (8)	
Displacement at UTS	(mm)	7.0 (1.1)	7.0 (1.8)	
Yield	(N)	36 (16)	37 (9)	
Displacement at Yield	(mm)	2.6 (1.7)	1.6 (0.7)	*
Stiffness	(N/mm)	15 (7)	18 (5)	
WOFint	(Nm)	0.025 (0.008)	0.021 (0.009)	
Peak force differential	(N)	1.7 (1.2)	1.1 (0.4)	*

6.5.3 Repair Failures

All of the repairs failed by pullout of the peripheral suture followed by rupture of the core suture. The PCL repairs all failed initially by the 6/0 Prolene sutures pulling through the PCL sheath (Figure 6.9), followed by the core sutures breaking.

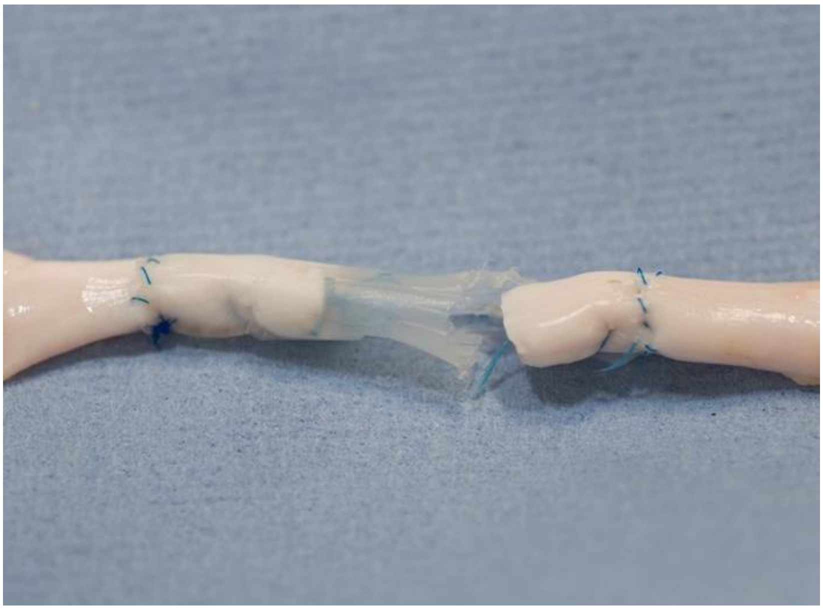

Figure 6.9. Failure of the PCL sheet.

Chapter 7 Discussion

7.1 Methodological Considerations

The research project was undertaken in order to compare the biomechanical properties of flexor tendon repairs in zone II, the MK repair, with a new core suture configuration incorporating a ventral loocking loop the LMK repair, and a new peripheral repair technique incorporating a PCL sheet. A porcine model was chosen due to the similarity of the anatomy of the porcine digits to that of human digits, its availability and low cost (donated fresh porcine trotters). Throughout the project over 200 tendons were dissected and tested. The findings of the dissections show that there are only minor differences in the cruciate pulley anatomy compared to human digits, validating this as a suitable anatomical model. Smith et al. presented a short review of the anatomy of the porcine forelimb tendon and flexor tendon system, and their results compare with the findings presented in this research, both with regards to the anatomical similarities and the suitability of the porcine model (132). A more recent anatomical and biomechanical study of flexor and extensor tendons of porcine hind legs confirmed that the biomechanical properties were also comparable to that of human tendons (263). The authors of this study conclude that the use of porcine deep flexor tendons from the hind legs, are similar to human, as are the biomechanical properties. Walbeehm et al. advocate the use of superficial flexor tendons in the porcine model as a better size match to human FDP tendons. The use of superficial flexor tendons will, however, not allow for accurate measurement of tendon repair gliding resistance (151).

The mean cross sectional areas of the intact porcine tendons measured in our study are comparable with the diameter reported by Mishra et al. in 2003, of 5.5 mm - 6 mm (122), which compares to the diameter of the middle finger of adult hand (77). With regards to the *in vivo* behaviour of porcine tendon repairs, the work presented in this thesis does not address this; further studies are warranted to determine how the healing of a porcine tendon compares to human tendon.

A static linear testing model was set up and evaluated for testing of porcine flexor tendon repairs. This is a commonly used biomechanical testing method for evaluating the strength, force to gap formation and failure of tested tendons (37). One study described cyclic loading of tendon repairs resulting gap formation at lower loads than static testing (120). This represents a more accurate picture of the physiological cyclic loading that occurs during post operative mobilisation, however the project was set up to look at the immediate post operative repair strength and not the effect of mobillisation. The static linear set up used proved reliable, easy to use and allowed for comparison of the biomechanical properties of a large number of the various repair types. The results are a representation what occurs immediately post operatively (at time zero), where the core suture carries the load.

The initial testing investigated the effect of training on the objective mechanical testing of tendons, and evaluated the effect of learning on one trainee (EZ). A clear learning curve was seen to influence the mechanical properties of the tendon repairs. A significant difference was seen between the early and late repairs tested with regards to UTS, gap formation, yield force and stiffness. The late repairs had higher UTS, force to gap formation and force to yield than the early repairs, and a lower stiffness. The repairs reached values for published results after 30 repairs had been completed. Looking at the force to 3 mm gap formation, it is clear that the early and intermediate repairs would not withstand an active post operative mobilisation regimen (29 N peak force). The tests were all done at time zero, representing the time of surgery, so loss in strength over the postoperative period was not evaluated. The higher stiffness values seen in the early and intermediate repairs were thought to represent over-tightening of the repair, possibly leading to "bunching" at the central repair, or an increase in the tendon diameter at this point. This might then influence the WOF of the tendons.

This work emphasizes the need for structured training of surgeons encompassing simulation modalities. The model presented here is easily reproducible and can accommodate a significant number of tendon repair testing in relatively short time. In comparison to the clinical setting where tendon injuries present irregularly, a much steeper learning curve and improved clinical results can be accomplished by combining simulation models with clinical exposure. The final result of the repair is not determined for weeks after the surgery, resulting in delayed or even absent feedback for the surgeon, making quality assurance difficult. The amount of time passed between each repair will also vary, and the effect of this on the surgeon's skills is not determined. A negative effect from this may also affect complication rates. This may be reduced by introducing this teaching model into the surgical training, allowing for regular exposure to tendon repairs. The tendon repairs performed by the trainee for this study were performed without any structured feedback during the procedure. It is our belief that continuous, structured feedback not only on the surgical technique but on biomechanical properties of the repair will be valuable for trainees and more experienced surgeons, and we hope to take this work further. The porcine tendons were frozen after repair and thawed out prior to testing. No significant effect of this has been seen on the biomechanical testing (258), which means that the repairs can be performed in one centre at one time, and tested somewhere else at a later time.

The initial tests also showed that there was a significant difference in the biomechanical properties of repairs performed with braided polyester (Ticron) versus monofilament polypropylene (Prolene). The two sutures were also very different in their handling. Prolene allows for easier tensioning during the repair (187). The use of a monofilament suture is often reported as easier to use for tendon repairs as it handles more easily when the repairs are tensioned (171, 188). By moistening the Ticron suture with normal saline solution and employing a meticulous and gentle technique there was no difficulty using the Ticron for the repairs. This is part of the learning effect when starting out with tendon repairs and trainee surgeons must be aware of this. The LMK also require that the repair is tightened as it is placed; it cannot be tightened once both tendon ends are repaired as the ventral loop locks the material in place, which will eliminate any sliding of the suture material and tightening of the loops. It is important to remember that the tendon repair initially reflects the properties of the suture material used, which means that the UTS of the tendon repair initially cannot exceed that of the suture and its knot.

Cao et al. investigated purchase length using Nylon sutures in a 4 strand core sutures repair. They report UTS value from 33.5 N to 45.2 N depending on the repair type and purchase length (0.4 cm and 1.0 cm) using 4/0 Ethilon, in a porcine model (77). The 2 strand repair results presented in this work also compare to their 4 strand results with regards to force to 2 mm gap formation, which highlights how the suture calibre and material affects the properties of the repairs. Braided polyester has a higher ultimate tensile strength than nylon sutures (79). Previous work by Vizesi et al. has also shown that Prolene has very different viscoelastic properties than Ticron, with higher creep and stress relaxation ratio (187). This means that the Prolene sutures would have a higher propensity for gap formation during mobilisation which could weaken the repairs.

One of the aspects investigated in the early testing was the effect of pre-conditioning of the tendons and the repair during internal work of flexion measurements. A time dependent conditioning effect is recognised in tensile testing of tendons (264), yet the literature regarding preconditioning of tendon repairs is scarse when it comes to gliding resistance measurements. The results from this thesis showed that there was a marked reduction in the WOF and the Peak force differential from cycle 1 to 3, with the largest reduction seen from cycle 1 to cycle 2. Additionally, a larger difference was found between the cycles for the repaired tendons than for the intact tendons, which may indicate that surgical repair or operative alteration may enhance micro-structural changes seen in conditioning.
The results presented here are contrary to those found by Momose et al, who report an increase in canine WOF with 300 cycles of flexion and extension by 45% (226), versus a significant decrease in WOF with each cycle presented here. This could be due to the *in vitro* testing, and the use of a porcine model compared to the canine model used by Momose et al. Canine models investigating gliding resistance have shown different characteristics from human models so a direct comparison to our work using a porcine model may be difficult (114). After 300 cycles it is also possible that the WOF measurements in this study would increase, due to for instance tendon dehydration or GAG disruption, which can be determined in further investigations. Tang et al, more recently recommend that six cycles of conditioning are used to stabilise measurements of tendon repair gliding resistance (265). This

recommendation was based on their work examining the loading of the proximal tendon, and did not take into consideration the distal tendon or the internal work of flexion. Other factors than the location of the repair, and type of tendon repair, are likely to influence the reaction to conditioning, such as temperature, hydration of the tendon and strain magnitude (226).

Other studies have employed a single cycle testing protocol for gliding resistance between tendon, tendon repairs and individual pulleys (226), while the work presented here consistently used multiple cycles for gliding resistance measurements. Clinically this means that great care must be taken by therapists on the first flexion-extension movement post operatively. Our results also emphasize the importance of preconditioning of the samples when investigating gliding resistance of tendon repairs.

7.2 Biomechanical Comparison of the Repairs

The primary aim of the project was to compare the commonly used and evaluated MK repair (104, 152, 157, 266, 267) to a modification of the MK which incorporated a ventral looking loop, LMK. The early testing found that Ticron had a greater force to gap formation and UTS when compared to Prolene in both of the repair configurations tested (Section 7.1). The comparison of the 4strand MK and LMK was therefore conducted using Ticron. The addition of the ventral locking loop to the MK was tested to see if this suture configuration had any biomechanical advantages over the standard MK, possibly by improving the load sharing over the tendon repair. The loading of the corner loops, seen with the locking MK repair, has been suggested to constrict the tendon fibres as a result of longitudinal force on the tendon in this area. This again may affect the blood supply to the area (151). This constriction of the corner loops may also result in lengthening of the longitudinal component of the MK and cause gaps to form between the tendon ends. Locking the transverse component of the repair may reduce this. Variations such as the Savage repair and modified Tsuge technique has employed this with favourable results (157, 170, 177). If more than one locking loop is used, the load sharing may be unequal, which may cause one loop to loosen as the other is tightened and thus cause loops and suture to move or slide, leading to gap formation and weakening of the repair. Hotokezaka and Manske found no increase in the UTS when increasing the number of locking loops (108), and that may be as a result of inequality in the load sharing, or due to the locking loops kinking under load which causes failure. By introducing a locking component on the ventral surface of the repair this may be reduced. We found no significant difference in the cross sectional area at the central repair area and 1 cm proximal to this, and therefore believe that the constriction of the tendon in this area is minimal, perhaps as a result of the ventral locking loop that is introduced. The force to gap formation (1, 2 and 3 mm) was not significantly different between the MK and the LMK, and no significant difference in the resistance to gap formation (stiffness) was seen between the repair types, for the 2 and 4 strand repairs. It is however possible that there is a constriction of the tendon despite the ventral locking loop, and a difference in the dimensions out with the areas that were measured. A histological examination might reveal more information. Further work is needed to determine this, perhaps using an *in vivo* model.

An investigation of the size and diameter of the locking loops were performed by Xie et al, using two and four strand repairs, where they compared 1, 2 and 3 mm locking loops and found that the UTS increased with increase in the diameter of the locking loops. Only the increase from 1 to 2 mm showed statistically significant results with regards to UTS. The force to 2 mm gap formation was also found to be highest for the 2 mm diameter locking loop. In their study they employed a simple circular lock suture situated perpendicular to the long axis of the tendon (81).

Hatananka and Manske investigated the effect of the cross sectional area of the locking loops used with the Pennington modification of the MK repair and found that the UTS improved with an increase in the cross sectional area of the locking loops, but that the larger cross sectional area loop repairs had a higher tendency for gap formation (103).

The 2 strand core suture LMK repairs were found to exhibit the same biomechanical properties as the standard 2 strand MK repairs in the initial study. No significant difference was seen in UTS, with both repair types reaching values of over 40 N (MK 43.5 N and LMK 42.5 N) using Ticron sutures. This compares to published studies that report UTS values ranging from 21-43N, depending on the repair technique, suture type and suture calibre used (85, 87, 91, 95). No study has been identified where the same combination of repair type, suture calibre and type, model and testing machine is used which makes direct comparison difficult. No difference was seen in the force to gap formation, yield, stiffness or rigidity of the repairs. A statistically significant difference was seen between the two suture types, Ticron versus Prolene, with the Ticron repairs showing higher values for the all the investigated parameters studied, as discussed in section 7.1.

The yield point of the repair is as, if not more important than UTS as yield represents the point at which most peripheral sutures fail, therefore interrupting the core suture/peripheral suture repair complex, which cause an imbalance in the load sharing of the repairs (88, 106, 170). The 2 strand repair yield point was reached at about 30 N which may not be sufficient to sustain forces seen with active post-operative mobilisation. The repairs were therefore extended to 4 strand core suture repairs to ensure that the properties at time zero were sufficient enough to endure the 29 N forces expected to be seen during an active postoperative rehabilitation regime. The 4 strand data did show significant difference in the force to 3 mm gap formation for the LMK and MK repairs, with the LMK being significantly higher. No significant differences were seen in the UTS, yield, stiffness or rigidity of the repair types between the MK and LMK as expected from the 2 strand results. The 4strand repairs had mean yield values of 32.7 N for MK and 40.9 N for the LMK, and UTS for the MK repair 66.7 N and 68.0 N for the LMK. In the literature the UTS values from 4strand repairs depend on the suture configuration (ie repair type) as well as the calibre and type of suture, and have been reported to range from 43.4N- 87N (78, 79, 81). The results obtained with the 4strand repair in this study fall in this range, and should withstand the forces seen with an active mobilisation regime as reported by Schuind et al. (section 2.5.1) (134).

The LMK repair did not differ in the force to gap formation compared to that of the MK repairs using 2 strand core sutures. The 4 strand LMK did have higher force to 3 mm gap formation (50.5 N) than both the MK (41.6 N) and the LMK+PCL (39.0 N) repair. The difference between MK and LMK may be explained by the addition of the ventral locking loop, while the reduction seen with the addition of PCL is likely due to the pull out of sutures from the PCL sheet.

In regards to measurement of the stiffness of the repairs this can be considered the “resistance to gapping” (108), or “ the force required to produce a unit of elongation of a given segment of material" (77), and thus provides valuable information of the repair. Stiffness for the 2 strand repairs were 14.8 N/mm (MK) and 16.3 N/mm (LMK), and 16.3 N/mm for 4 strand LMK and 19.5 N/mm for MK. It was also noted that the late repairs in the initial study had a higher stiffness than that of the earlier repairs.

The internal WOF for the 2 strand repairs found that the LMK had a significant lower internal WOF (0.026 Nm) than the MK (0.029 Nm). The cross sectional areas of the repairs were measured to see if a difference in WOF could be explained by a difference in the cross sectional area of the two repair types, but no significant difference was found in the sizes of the repairs. The reason for the lower WOFint for the LMK is therefore not clear, and may represent a chance finding. The 4 strand repairs did not show the same statistical difference between the LMK and MK with regards to the WOFint as seen with the 2 strand repairs, and cross sectional area measurements did not show significant difference between the 4 strand MK and LMK The incorporation of the ventral locking loop does not seem to have a clear biomechanical advantage over the standard MK repair, but may provide some benefit with regards to load sharing. This may influence how this repair type responds in the clinical setting to the various rehabilitation regimes and needs to be assessed in further studies.

The PCL sheets were tested without grooves and with the grooves oriented parallel to the axis of distraction, and orthogonally to the axis of distraction. The sheets were found to have a low yield force 2.40 N for the non-grooved and 2.7 N for the sheets with the grooves parallel to the axis of pull, and a large displacement, ranging from 58.5mm to 89.5mm, at peak tensile force. The grooves underwent deformation at low

loads (yield) and the material started to unwind, which means that the grooves were lost thus limiting the effect they may have on tendon healing.
With regards to the UTS of LMK + PCL versus LMK alone repairs no difference was seen, which reflects the values of the suture material and calibre used for the core sutures. The force to gap formation for 1 mm gap was significantly lower for the PCL repair, and it had consistently lower values than the other repairs although not statistically significant for 2 and 3mm gap formation.

The addition of a PCL sheet to the LMK resulted in lower forces to gap formation, reduction in the stiffness of the repair and exhibited a greater strain at low forces than the other repairs, which may indicate that the PCL is not as efficient as the simple running peripheral suture.
The WOF of the LMK + PCL repairs were as expected much higher than those the LMK and MK with a simple running peripheral suture. The PCL almost doubled the WOF and the peak force differential for the intact and repaired tendon.
The PCL repairs all failed by the pullout of the sutures holding the sheath in place. Inequality in the loading of the repair, resulting from failure of the peripheral suture or PCL first, will result in inequality of the load on the core suture as proposed by Lotz et al. (106), and may cause gapping. The use of the PCL or similar construct would allow for more gapping of the ends as these are covered by the PCL and this would perhaps allow for tenocyte migration and thus tendon healing despite gap formation >3 mm gaps. Theoretically the sheath could still be clinically useful if it provides matrix for the tenocytes to grow, this as well as work investigating a surface modification to reduce the WOF will need to be introduced to the PCL before it would be introduced in clinical practice. As a tool for flexor tendon repairs especially in zone

II the clinical usefulness of the PCL sheet in its current form is low. It may however prove to be a useful adjunct to tendon repairs outwith the hand, where the WOF or gliding resistance is less of a concern.

7.3 Limitations

There are limitations of this work that need to be considered. The research uses a bench model versus a live model, and only measures biomechanical properties immediately post-operatively (time zero). The effect of postoperative swelling and loss of tensile strength due to tendon softening is not evaluated as the bench model is not able to mimic the changes in the surrounding soft tissue that happen both at the time of injury and after as the healing process begins. A porcine model with similar anatomy to a human model has been used (122, 132) yet some differences exist. The size of the porcine tendons are known to be larger that those of human (268), however the tendons used for this research has similar size to those tested by Havulinna et al, and are comparable to human FDP tendons of the index , middle and ring finger (226). The size of the pulleys are also different from that seen in human hands were the largest pulley is the A2, versus the A1 in the porcine model.

The porcine tendons were frozen at -12°C after dissection and repair, and then thawed out at room temperature before testing. The effect of freeze-thawing on porcine tendons has not been found to have an effect on the mechanical properties of tendons when tested in a linear testing machine (258).

All repairs undertaken in this research used a simple, running peripheral suture. Previous research has shown that locking peripheral sutures may impart greater tensile properties than simple peripheral sutures (213). However, this study found that the impact on the UTS with simple peripheral suture is minimal, and that the peripheral suture influences the force to gap formation and displacement at yield of the repair. A running peripheral suture is a commonly used method clinically (10, 269, 270), and the same technique has been employed for all testing.

Disruption of distal tendon insertion from bone was done in order to attach the ring transducer and may alter the moment arm during extension and flexion. This was investigated by Zhao et al., where WOF was investigated after resecting and reconnecting the distal tendon, and they found a minimal effect in human model (144).The exact mechanical set up was used for all test protocols, and alignment of the tendon to its insertion distally was kept constant Thus it is thought that the influence of this is minimal.

The tensile testing did not include cyclical testing, which has been shown to influence repair failures (90, 117, 118). However, the use of a linear testing protocol enabled the comparison of a large number of tendon repairs at time zero as set out in the aim of this study. Further tests looking at the repair during cyclic testing may be of interest.

For the WOF measurements it is assumed that there is no deformation of the repair since the applied force is so low (<10N), less than that seen to produce gap formation in the early tests performed. Although the elongation of the repair during the testing was not formally assessed the repairs were visually inspected for gapping at the end of each repair.

Chapter 8 Conclusions and Future Work

<u>From the results of this dissertation the following is concluded:</u>

Primary Aims:

The primary work of the project showed that the use of a ventral locking modification of the Pennington Modified Kessler technique has biomechanical properties that compare to the standard MK technique at time zero, and is strong enough to sustain the forces expected in the postoperative rehabilitation period.

. The addition of nano-grooved PCL to the repair resulted in lower tensile testing values and higher WOF and peak force differential values than seen with LMK and MK. This limits the clinical use of the construct in its current form for zone 2 flexor tendon injuries of the hand.

Secondary aims:

This research has confirmed that the porcine model is a suitable and easily available model for testing and investigating flexor tendon repair types and biomechanical properties.

The work showed that there is a substantial learning effect associated with FDP repairs and the model used here could be useful for benchmarking.

The biomechanical testing set up compares to similar set ups reported in the literature.

Ticron suture was found to be superior to Prolene with regards to tensile properties and should be used for FDP repairs.

Pre-conditioning of FDP repairs was found to have a significant effect on WOF testing, and should be incorporated in testing protocols.

No effect was seen on the WOF of the tested FDP repairs when the flexor sheath was repaired.

The addition of a simple, running peripheral suture did not show any effect on the UTS of the repair, but did increase the force to gap formation and yield.

<u>Future work might include:</u>

1. Study to evaluate the differences in the LMK and MK using cyclic testing to evaluate the effect of more physiological loading on the repair.

2. Any histological differences between the MK and LMK may be determined using an *in vivo* model. This could help determine how the repair reacts to post operative mobilisation, how the tissue reacts and the strength and gliding over time.

3. The testing set up and procedure could easily be used to evaluate trainees throughout their training period and to ensure that surgeons are performing repairs that are up to the standards set.

4. Further work looking at ways to improve gliding properties of the PCL will need to done before it can be tested clinically.

5. Further evaluation of the PCL for tendon repairs in areas out with the hand may be of interest.

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“If we knew what it was we were doing, it would not be called research, would it?”
Albert Einstein