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THE BIOMECHANICS OF THE MEDIAL PATELLOFEMORAL LIGAMENT

by

Joanna Stephen

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Department of Mechanical Engineering,

Imperial College London

London SW7 2AZ

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Abstract

Injury of the Medial Patellofemoral Ligament (MPFL) occurs during patellofemoral joint (PFJ) dislocation. Reconstruction of this ligament is a common treatment for this patient population. This thesis is composed of a series of cadaveric experiments examining the MPFL.

The MPFL was found to originate from the midpoint between the medial epicondyle and adductor tubercle. Its length change pattern was close to isometric through knee flexion range. Transection of the MPFL resulted in significant increases in lateral patellar translation, tilt and increased lateral PFJ contact pressures in early knee flexion (all: $P<0.05$). Anatomical reconstruction of the ligament restored patellar motion and joint contact pressures ($P>0.05$). Femoral tunnels positioned too proximal resulted in increased medial contact pressures and patellar motion in deeper knee flexion, whilst distal tunnels caused increased medial contact pressures near extension ($P<0.05$). Anterior-posterior femoral positioning or varying position of the patellar attachment was not found to have a significant effect on ligament length change patterns ($P>0.05$). The radiographic femoral MPFL attachment point, assuming the anterior-posterior medial femoral condyle diameter to be 100%, was; 40% from the posterior, 50% from the distal and 60% from the anterior femoral border.

Reduction in medial quadriceps muscle tension was similar to the effect of MPFL transection causing a significant increase of lateral patellar tracking and lateral PFJ contact pressures (all: $P<0.05$). Progressive tibial tuberosity (TT) lateralisation caused increased lateral patellar tracking, lateral PFJ contact pressures and reduced patellar stability ($P<0.05$). TT medialisation did not cause corresponding excessive increases in mean medial contact pressures ($P>0.05$). With tibial tuberosity-trochlear groove (TT-TG) distances of up to 15mm PFJ mechanics were satisfactorily restored with anatomical MPFL reconstruction ($P>0.05$). However in cases with TT-TG distances greater than 15mm, MPFL reconstruction alone was not sufficient to restore PFJ contact mechanics and patellar kinematics ($P<0.05$).

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Chapter 3

Work from this chapter was published as 'Sectioning the medial patellofemoral ligament alters patellofemoral joint kinematics and contact mechanics' in Journal of Orthopaedic Research: Volume 31, Number 9, pages: 1423-1429.

Chapter 4

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CHAPTER 1

INTRODUCTION

Patellofemoral joint surgery: 'the black hole of orthopaedics'

(Dye et al., 1999)

1.1 Background

Patellofemoral joint instability is a condition which can often result in a disabling sequel of recurrent instability and pain in patients. The past decade has seen significant progress made in the understanding and imaging of patellar dislocation, with more detailed attention paid to the specific contributions that injured structures make to local patellar stability. However this remains a controversial area, highlighted by the excess of over one hundred procedures which have been described and performed for the treatment of patellofemoral disorders (Stefancin and Parker, 2007). Unfortunately surgical management in this population of patients has not been uniformly successful with widespread reports of mediocre results and failure evident in the literature. In recent years the role of the Medial Patellofemoral Ligament (MPFL) in patellofemoral joint stability during early knee flexion has become more recognised, and MPFL reconstruction is becoming a popular surgical technique undertaken worldwide. However inconsistencies in current literature highlight a lack of understanding of MPFL characteristics, meaning surgical techniques undertaken are often not fully informed.

There are a number of questions which are paramount to ensure success during treatment of an injured MPFL:

1. Is there a role for conservative treatment in patient management?
2. What is the native anatomy of the structure to be reconstructed?

3. What effect does the presence of abnormal anatomy in the region have on the surgical decision making for the reconstruction of the structure?

Then, if reconstruction is found to be needed:

4. What is the optimal surgical technique for reconstruction of the ligament?

Although accounts of the MPFL exist in the literature these are scant compared to more established procedures. Furthermore these are often inconsistent, and contradictory in nature. This thesis aims to clarify answers to the above questions in relation to the MPFL to help drive future refinements to the treatment of this complex group of patients and thus optimise clinical outcome.

1.2 Aims and Objectives

The main aim of this thesis was to obtain objective evidence relating to the anatomy, biomechanics and function of the MPFL which could be used to inform surgical practice. Specific objectives were as follows:

1. To investigate the native anatomy of the MPFL.

2. To obtain data on the length change patterns of the MPFL, using a variety of femoral and patellar attachment points.

3. To study the effect of MPFL sectioning on the patellofemoral joint mechanics.

4. To determine the effect of tunnel position, graft tension and graft fixation on outcome during MPFL reconstructive surgery, with the aim to propose the optimal technique to reconstruct the MPFL.

5. To establish the role of the medial quadriceps in patellofemoral motion and stability.

6. To examine the effect of progressive medialisation and lateralisation of the tibial tuberosity on patellofemoral joint tracking, mechanics and stability.

7. To investigate the role of the MPFL in the presence of abnormal tibial tuberosity-trochlear groove (TT-TG) alignment.

1.3 Organisation

This thesis presents a series of results from cadaveric testing conducted to try to provide answers for the aims and objectives outlined in section 1.2. Following this introduction, the second chapter provides an overview of the patellofemoral joint, exploring anatomical variations which could contribute to patellofemoral joint instability and highlighting the role of the MPFL. This review forms a fixed basis from which to explore the MPFL in greater depth. The following chapters are then each laid out in a similar format. They begin with a section on background literature and aims, highlighting the chapter objectives and summarising relevant prior work conducted in the area. The experimental method, analysis, results and discussion are then outlined. Each chapter concludes with a summary of the key findings from the work. Following the conclusion of each chapter there is an Addendum which explains in greater detail key decisions made regarding the methodology design and where indicated presents some further experimental results. This lay out permits a clear and concise presentation of the series of investigations undertaken whilst also providing the reader the opportunity, where interested, to gain a greater and more in-depth understanding of the decision making process related to the methodology.

The first set of the experimental reports (Chapters 3-5) relate to cadaveric work undertaken to explore the anatomy, biomechanics and surgical reconstruction of the MPFL. This section of the thesis presents literature relevant to patient populations who have suffered patellar dislocation in the presence of otherwise normal anatomy in the knee. In order to conserve resources, in particular cadaveric specimens, one set of knees was prepared and then used to answer the research questions asked by each individual chapter. The second part of the thesis begins with chapter six which explores the role of the medial quadriceps muscles on patellofemoral mechanics and patellar stability. Again this chapter contains data most relevant to those patients who have suffered patellar dislocation in the absence of any abnormal anatomy such as elevated TT-TG distances or trochlear dysplasia.The seventh chapter then investigates the effect of progressive medialisation and lateralisation of the tibial tuberosity on patellofemoral joint tracking, stability and contact mechanics. The role of the MPFL and its reconstruction in the presence of an abnormal TT-TG is investigated in chapter eight. These chapters both inform the reader regarding a different population of patients compared to the prior four. They concentrate on those patients who suffer more chronic patellar instability caused as a result of abnormal anatomy in the area. Again to be sensitive to the appropriate use of cadaveric specimens, chapters 6-8 were conducted on the same set of specimens. Finally chapter nine concludes the thesis by summarising the main findings from the work, discussing limitations of the investigations and outlining suggestions and directions for future work.

This work is primarily directed towards surgeons working with populations of patients who suffer patellar dislocation, with both acute and chronic problems. It is hoped that the results will add to the existing evidence base to help to optimise management of these patients in the future.

CHAPTER 2

THE PATELLOFEMORAL JOINT

2.1 Bones / Joint

The patellofemoral joint is a highly intricate synovial articulation, comprising of the femur, trochlear groove and the patella (Fulkerson, 2002) (Figure 2.1). Its surface morphology is formed early during *in utero* life before standing and walking (Doskocil, 1985; Gray and Gardner, 1950). High loads, often many multiples of body weight, are accepted and re-directed by the joint (Brechter and Powers, 2002). Its asymmetrical geometry highlights its complex function and reflects the demand for it to withstand high compressive and tensile loads through a large range of motion relative to its size (Tecklenburg et al., 2006).

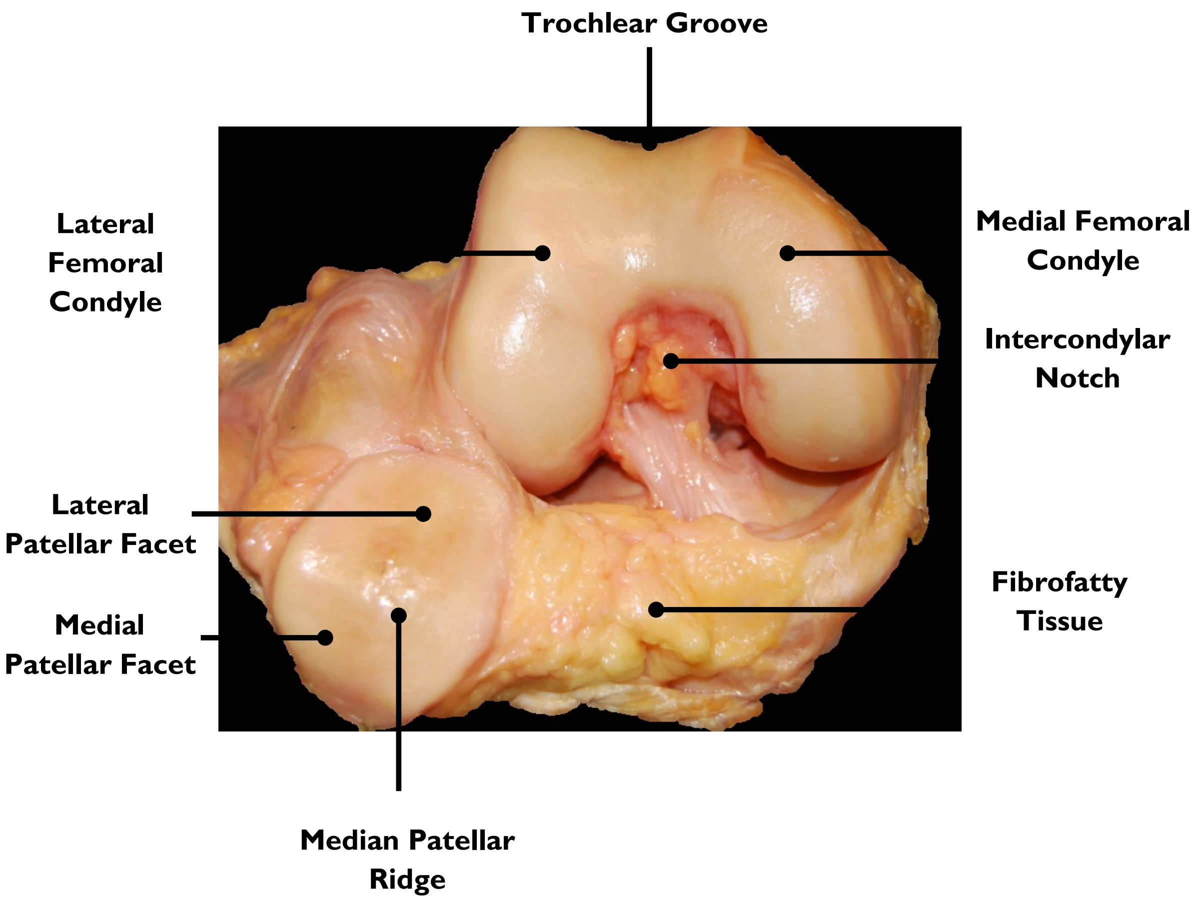

Figure 2.1 Photograph of the anterior view of the right patellofemoral joint in 90° flexion. Key anatomical features of the patella and trochlea are highlighted, with the patella cut away from the distal end of the femur and deflected to the lateral side.

Optimal function of the patellofemoral joint is enabled by the complex interaction of a number of local and global variables including the interaction of static (ligamentous and connective tissues), dynamic (musculature) and bony structures (patellar and trochlear geometry, femoral torsion) (Senavongse and Amis, 2005). Abnormalities in any one of these areas can result in patellofemoral dysfunction, manifesting clinically as pain or instability (Staubli, 1999).

2.1.1 Patella

The patella is the largest sesamoid bone in the human body and is located anterior and proximal to the trochlea in full extension where it rests on a fibrofatty pad (Staubli, 1999). It is enveloped by the quadriceps tendon anteriorly, with the patellar tendon emerging distally to insert into the tibia (Hungerford and Barry, 1979). The patella possesses the thickest articular cartilage of all the joints in the human body, approximately 5-9mm at its thickest, which reflects the large forces it has to accommodate over its relatively small surface area (Fulkerson, 1997; Grelsamer and Weinstein, 2001; Wiberg, 1941). This cartilage lines the articular surface of the proximal two thirds of the underlying surface of the patella and is intra-articular; the distal patellar pole attaches to the patellar tendon and is extra-articular (Tria and Alicea, 1995). The cartilage acts as a bearing surface, but is not sensate in nature (Dye et al., 1998).

The median cartilaginous ridge of the patella separates its medial and lateral facets (Figure 2.1). Its location has wide inter subject variability (Tecklenburg et al., 2006). It is reported to overlie the bony ridge in only 15% of cases, is positioned lateral to the ridge in 60% of subjects and medially in 25%; in some subjects it is barely visible whereas in others it is very pronounced (Staubli, 1999). There are three medial and three lateral facets on the patella, which are congruent with those of the trochlear surfaces lying beneath during progressive degrees of flexion (Tria and Alicea, 1995). A seventh odd facet is positioned on the medial border forming an articulation against the medial femoral condyle in deep knee flexion when the patella bridges across the intercondylar notch (Goodfellow et al., 1976).

A classification system has been proposed to sub categorise patellae based on the size and shape of their medial and lateral facets (Wiberg, 1941) (Figure 2.2). The least common configuration (approx. 10%) labelled Type I, occurs with concave medial and lateral facets almost equal in size. Type II describes a flattened or slightly convex medial facet smaller in size than the lateral facet and is the most common type (65%). Lastly Type III patellae (25%) have smaller medial facets and larger lateral facets, both of which are convex (Tria and Alicea, 1995; Wiberg, 1941) (Figure 2.2).

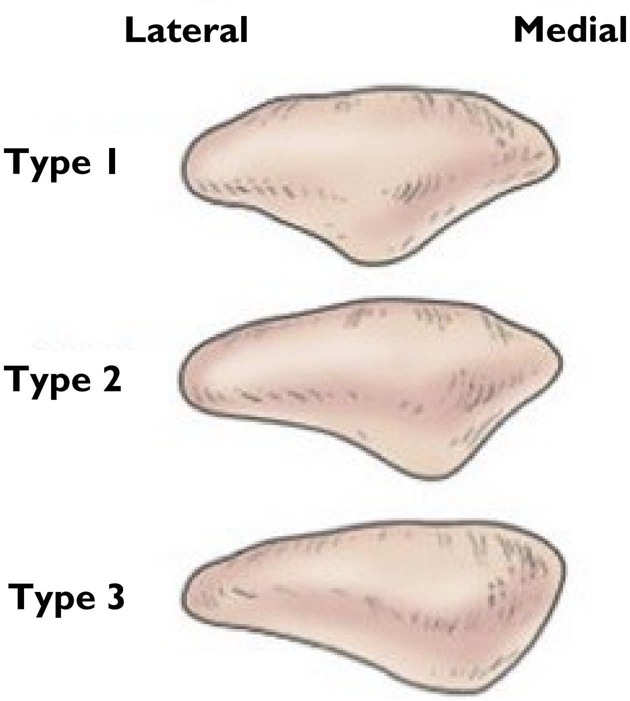

Figure 2.2 Wiberg Patella Classification (as adapted from Tecklenburg et al. (2006)).

The patella has a rich blood supply arising from six arteries, which form a vascular anastomotic ring positioned anterior to the patella (Kirschner et al., 1997). It is supplied by the descending genicular artery, the medial inferior genicular artery, the medial superior genicular artery, the lateral inferior genicular artery (together with the anterior tibial recurrent artery), and the lateral superior genicular artery (Kirschner et al., 1997). The inferior patella is known to have a rich vascular supply (Nemschak and Pretterklieber, 2012). However, the margins of the patella have a poor blood supply (Scapinelli, 1967).

Innervation to the patella is supplied by branches of the infrapatellar branch of the saphenous nerve and medial, intermediate and lateral cutaneous nerves of the thigh. These nerves all contribute to the patellar plexus of cutaneous nerves at the anterior thigh and patella (Kennedy et al., 1982). Intraosseous nerves have also been identified in the pereosteum and loose connective tissue around the patella, deriving from a medially based neurovascular bundle (Barton et al., 2007). A full understanding of the function of these has still to be established.

Investigators have studied the link between patellar shape and the clinical presentation of patellofemoral joint instability. One study found patients who suffered patella dislocation to commonly have underdeveloped medial facets and Wiberg type III patellae (Servien et al., 2003).

Panni at al (2011) identified a significant correlation between an increased patellar tilt and dysplastic patella. Meanwhile Pfirrmann et al (2000) found an increased Wiberg index and reduced medial patellar facet size (Type II and III) in trochlear dysplastic knees.

2.1.2 The Trochlea

The distal femur splits into two asymmetrical femoral condyles which are connected by a sulcus forming a groove referred to as the femoral trochlea (Figure 2.1). This happens *in utero*, with the trochlear shape evident on examination of foetuses (Gray and Gardner, 1950). The femoral diaphysis is vertical in the newborn baby, and then as weight bearing begins in early childhood years, a femoral obliquity angle develops inducing a valgus force on the extensor mechanism of the knee (Fulkerson, 1990).

The trochlea provides a surface for the patella to glide over during knee flexion and extension (Fulkerson, 1990). Continuing posteriorly and distally as the knee flexes, the trochlea ends in the intercondylar notch, a non-articulating groove (Figure 2.1). Similar to the patella, and reflecting its need to provide load distribution, the articular cartilage has been reported to be thicker centrally over the sulcus compared with the outer trochlea (van Huyssteen et al., 2006). Sulcus angle is a common measure of trochlear geometry and can be examined with MRI or axial radiographic images taken in 30˚ knee flexion. It is measured as the angle between the tangential line to the medial and lateral trochlear slopes (normal range: 135˚-145˚) (Carson et al., 1984; Hughston, 1968) (Figure 2.3A). Trochlear depth is measured from MRI images with the maximum Anterior-Posterior (A-P) distance of the medial and lateral femoral condyles and the minimum A-P distance of the deepest point of the trochlear groove from the line parallel to the posterior outline of the femoral condyles (Pfirrmann et al., 2000) (Figure 2.3B). Trochlear depth is calculated with the formula (([a+b]/2)-c). Mean trochlear depth distances from 5mm-6mm have been reported in populations of individuals with normal trochlear anatomy (Malghem and Maldague, 1989; Pfirrmann et al., 2000).

The geometry and size of the femoral condyles and the trochlea demonstrate wide inter subject variability. On axial view, generally the lateral condyle is larger; with the lateral part of the trochlear groove typically wider and more anterior compared to its medial counterpart (Shih et al., 2004). When the knee is flexed, the medial facet becomes more prominent as the skyline moves distally and posteriorly (Shih et al., 2004). In extension, the patella lies proximal to the trochlear groove, before it engages with the groove at approximately 30° flexion (Christian et al., 2006). The shape of the lateral facet in extension therefore helps to guide the patella into the trochlear groove as the knee flexes. Relatively uniform patellar contact pressures are reported to be present with large lateral facets as a result of the laterally acting vectors on the patella (Hehne, 1990). A measurement of the lateral femoral condyle is the Lateral Trochlear Inclination (LTI) taken from MRI scans. It is calculated as the angle between the line tangential to the cartilage surface of the lateral trochlear facet and the line tangential to the posterior condyles of the femur (Carrillon et al., 2000) (Figure 2.3C). The mean LTI value in subjects with no history of patellofemoral joint pathology has been reported as 17° (Carrillon et al., 2000).

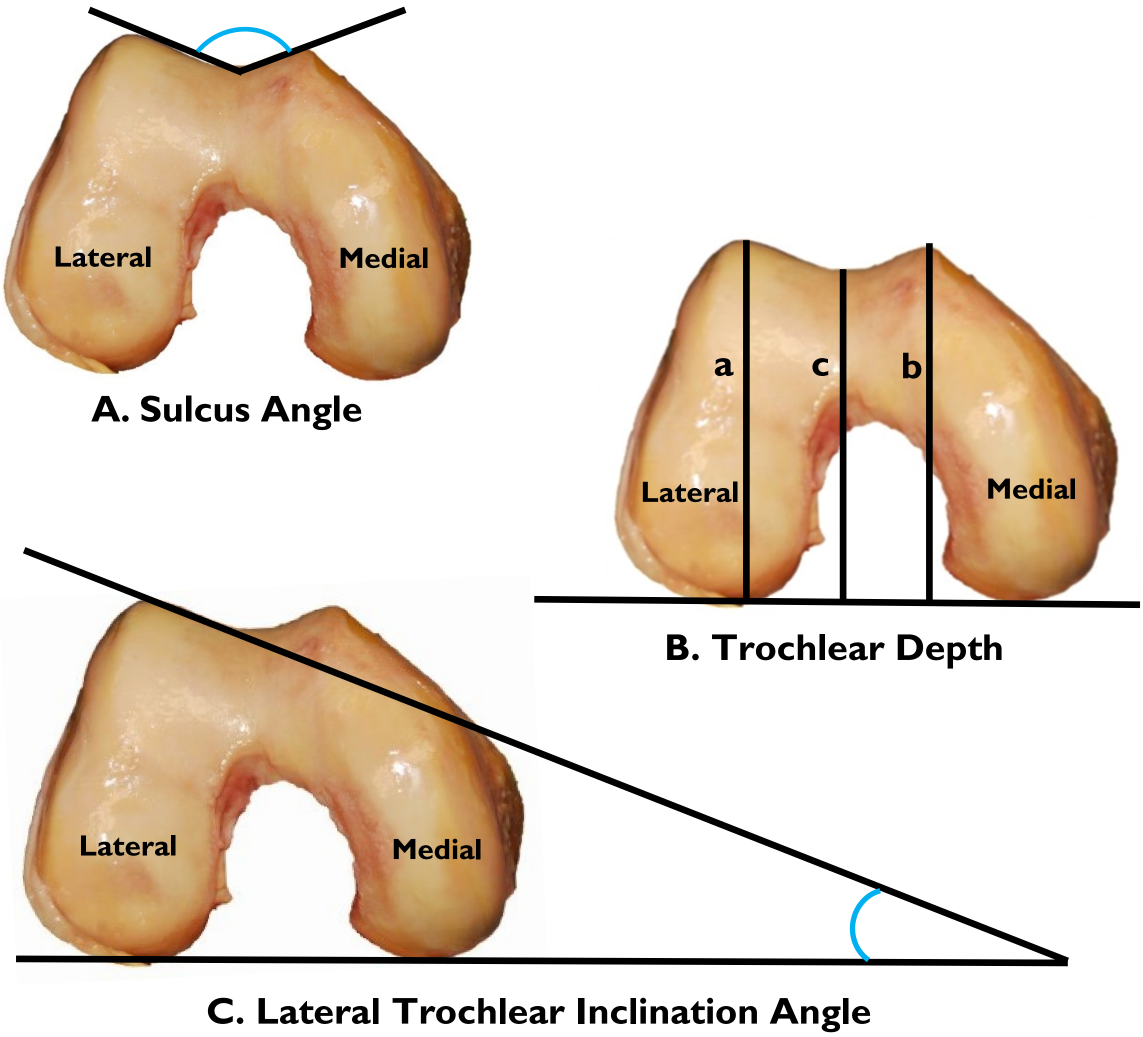

Figure 2.3 Anterior view of the trochlear groove flexed to 90°:

A: Sulcus Angle, **B:** Trochlear Depth, **C:** Lateral Inclination Angle.

The most frequent pathology identified in populations of recurrent patellar dislocators is trochlear dysplasia, where the femoral trochlea loses its concave shape and patella congruence and is instead flattened or convex (Figure 2.4). Tecklenburg et al (2006) describe a sulcus angle of greater than 150˚ and a LTI of above 11° in this population. Trochlear depths ranging from -0.6-2.7mm have been reported in populations with trochlear dysplasia (Pfirrmann et al., 2000). Dysplasia is reported to be present in up to 96% of patients suffering patellar dislocation (Dejour et al., 1994).

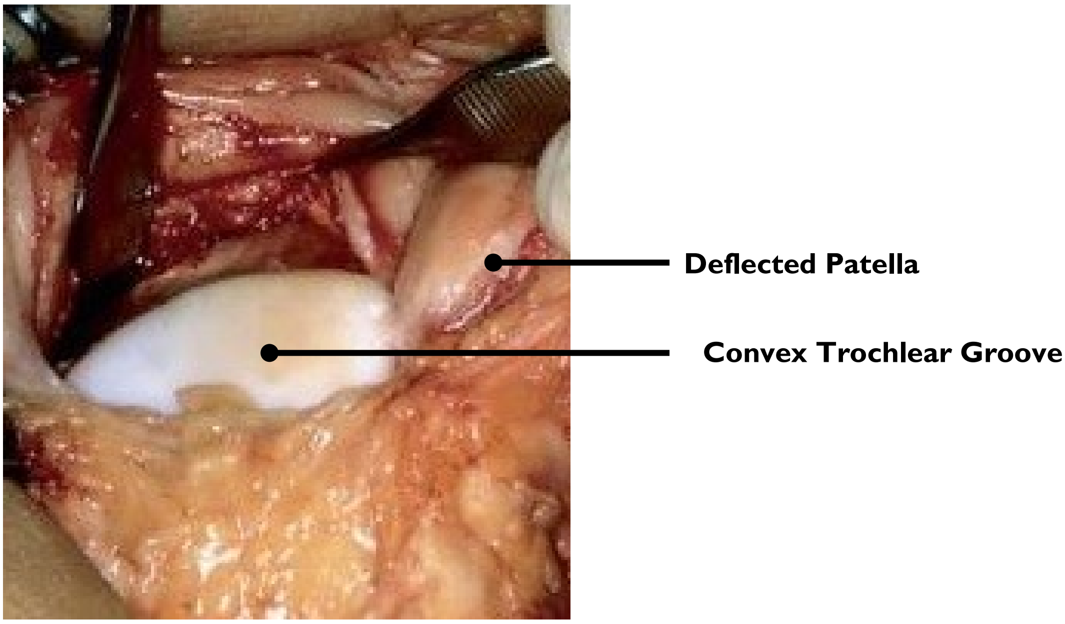

Figure 2.4 Intra-operative photograph of trochlear dysplasia. The convex groove is evident (as adapted from Sanchis-Alfonso et al. (2006)).

2.1.2.1 Axes of the femur

Two axes pass through the long proximal-distal length of the femur (Bull et al., 2002) (Figure 2.5). The anatomical axis refers to the line passing along the centre of the shaft of the femur to the centre of the knee joint. The second axis, referred to as the mechanical axis, relates to the lower limb in a weight bearing position. It runs from the centre of the femoral head to the centre of the knee joint and top of the intercondylar notch (Bull et al., 2002). The angle formed by these axes has been reported to range from 5˚-7˚ (Yoshioka et al., 1987).

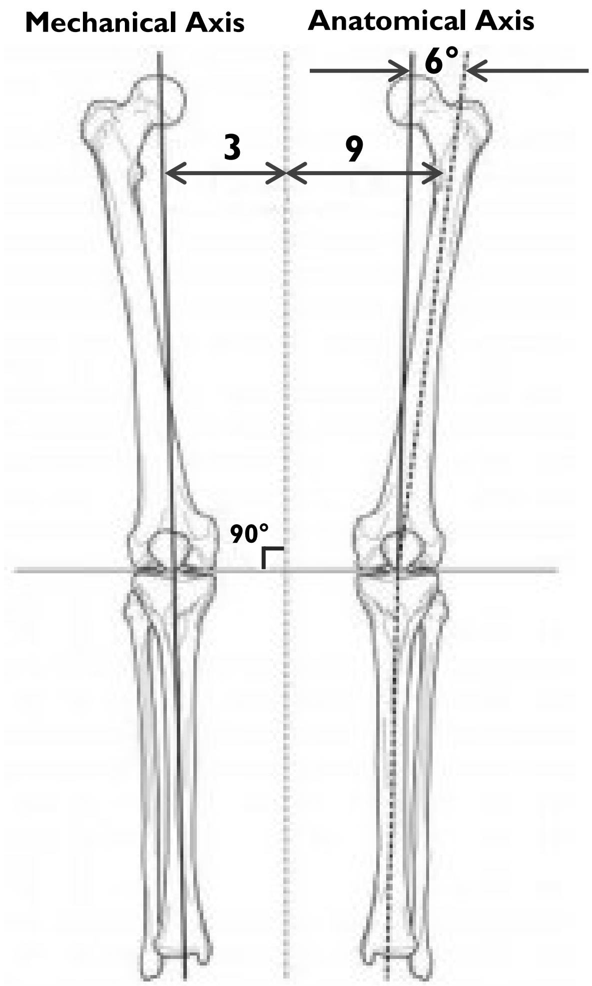

Figure 2.5 Mechanical (solid line) and anatomical (broken line) axes of the femur, showing the 6˚ angle they form.

2.1.2.2 Q-Angle Alignment

The patella articulates with the distal femur where it acts as a fulcrum, enabling the transmission of load from the quadriceps to the tibial tubercle throughout knee flexion range (Fulkerson, 1990). The position of the femoral axes mean a resulting natural six degree angle in the coronal plane, therefore any quadriceps contraction applies a continual lateral vector to the patella (Hehne, 1990). The lateral stabilisers are somewhat stronger and more fibrous than their medial counterparts, and this is hypothesised to further accentuate the laterally directed vectors acting on the patella (Powers, 2000).

The Quadriceps-angle (Q-angle) was originally proposed as a measure of patellar stability by Brattström (1970). It is formed by the line connecting the anterior superior iliac spine with the centre of the patella, with a line that connects the centre of the patella with the tibial tubercle (Figure 2.6). It is greatest in full extension due to the screw home mechanism of the knee (Hallen and Lindahl, 1966). This angle is proposed to quantify knee and hip alignment, and reflects frontal plane forces acting on the patella (Powers, 2003). In the presence of lower limb malalignment, an increased Q-angle has been considered to predispose the joint to a lateral patellar drift and instability (Fulkerson, 1990; Grelsamer, 2000; Lieb and Perry, 1968). An early cadaveric study identified significant changes in patellofemoral joint contact pressures as a result of varying the Q-angle (Huberti and Hayes, 1984). Contention exists regarding the value of 'normal' Q-angle measurements and consensus has not been universally agreed. Many research papers continue to cite the average Q-angle for males as 10^0, with 15^0 reported for females (Messier, 1988; Mizuno, 2001). The measure has been criticised for its static nature failing to reflect muscle activation during dynamic activities. Direct correlation between increased Q-Angle and patellofemoral measures indicative of instability has not been established and clinicians have been cautioned that angles above 'normal' values do not always equate to the development or perpetuation of patellar instability (Biedert and Warnke, 2001; Cooney et al., 2012).

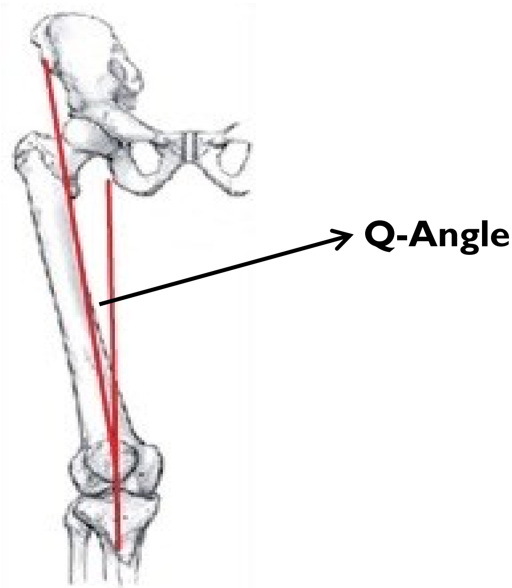

Figure 2.6 Quadriceps Angle (Q-Angle): measured as the angle between the line connecting the anterior superior iliac spine to the central patella and the line drawn from the centre of the patella to the tibial tubercle.

2.2 Passive Structures

2.2.1 Medial Retinaculum

The medial retinaculum of the knee is a band like structure made up of three layers which provide stability against lateral motion of the patella (Warren and Marshall, 1979) (Figure 2.7). The superficial layer of the retinaculum, often referred to as fascia, has a close relationship with the vastus medialis obliquus fascia but contributes very little to patellar stability (Baldwin, 2009). It is joined to layer 2 near the medial patellar border proximally and distally. The second layer is comprised of fibres from the superficial medial collateral ligament aligned vertically, the MPFL lying horizontally and the medial patellotibial ligament (MPTL) which is conjoined to layer 1. Beneath this, Layer 3 constitutes the deep medial collateral ligament, the medial patellomeniscal ligament (MPML) and the joint capsule. The retinaculum is tightly held to the medial side of the patella to ensure that there is a constant medial force on the patella to enhance patellar stability particularly during early knee flexion.

The patella is connected to the tibia distally via the MPTL and MPML, both ligaments are thinner than the MPFL (Thawait et al., 2012). The MPTL inserts into the inferior and medial margin of the patella and proximal patella tendon, attaching to the inferomedial tibial joint line. The MPML extends from the inferior and medial patella, attaching to the anterior rim of the medial meniscus (Dirim et al., 2008). The anatomy of the MPFL is explored in greater depth in Chapter 3.

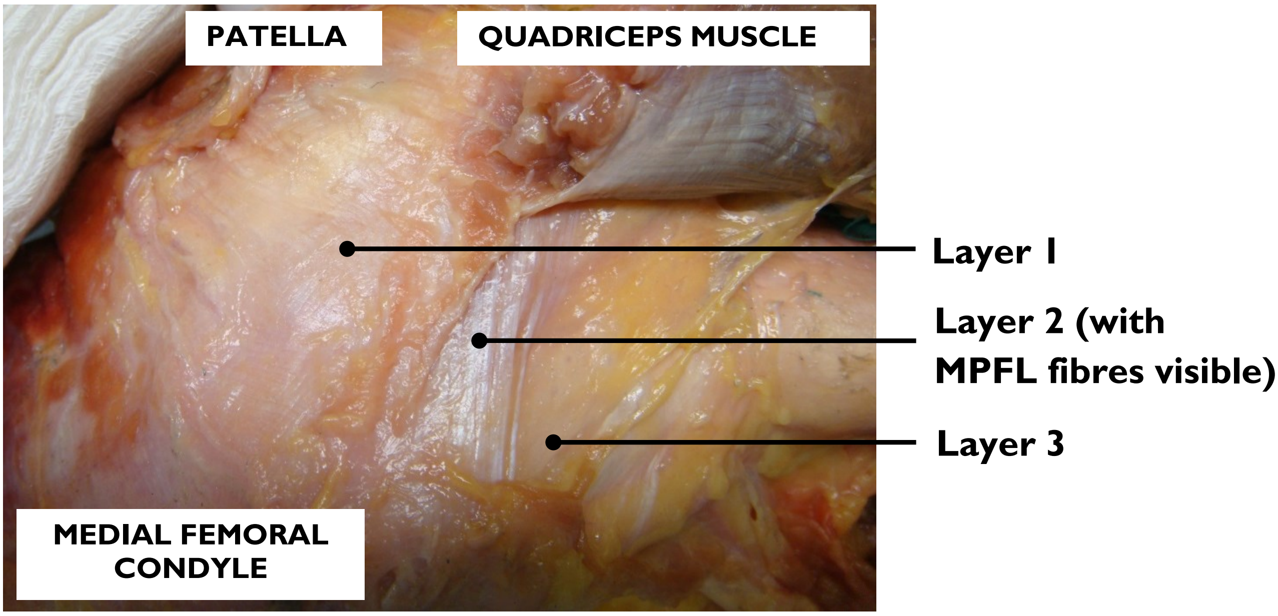

Figure 2.7 The three dissected layers of the medial retinaculum, with the vastus medialis muscle sectioned and elevated.

2.2.2 Lateral Retinaculum

The lateral side of the knee has a multi-layered retinaculum also consisting of thin ligaments which provide passive restraint to the patella. However the ligaments are less distinct from the overall retinacular geometry than on the medial side. The lateral retinaculum comprises three layers; superficial, intermediate and deep and as its name implies is positioned on the lateral side of the knee (Figure 2.8). The sequence and exact arrangement of the layers as well as the dimensions and attachments of the bands has been contentious and traditionally difficult to delineate given its converging, multi-layer presence (Last, 1950). However a recent review has brought some clarity to its description. The superficial layer is directly subcutaneous and constitutes the deep fascia which does not attach to the patella, but thickens laterally forming the iliotibial band attaching to the proximal part of the lateral condyle and blending with the quadriceps aponeurosis (Merican and Amis, 2008). The superficial layer separates easily from the patella given its lack of direct attachment; however it does provide a bracing resistance to lateral patellar motion. The most substantial layer, the intermediate layer constitutes derivatives of the iliotibial band and quadriceps aponeuroses. Iliotibial band fibres predominantly insert to Gerdy's tubercle distally, with some of the anterior fibres blending completely with the quadriceps aponeurosis, adhering to the patella and distal patellar tendon (Thawait et al., 2012). These fibres notably lack direct femoral attachment, however their orientation and position suggests they have a function in providing lateral patellofemoral restraint. Beneath this layer lies the deep layer, primarily constituting the joint capsule. The lateral patellofemoral ligament has been inconsistently reported to form a strand of the deep joint capsule, however in contrast to its medial counterpart it is not thought to be the primary restraint to medial patellar motion. Instead the iliotibial band patella fibres, present as a consistent short segment despite their lack of fixed femoral attachment. A further attachment from the inferolateral patella to the lateral meniscus has also been reported (Merican and Amis, 2008). The main role of the lateral retinaculum is as the primary restraint to medial patellofemoral displacement of the patella in relation to the femur. It contributes to patellofemoral tracking and helps with the distribution of medial and lateral compressive loads acting on the patella (Recondo et al., 2000).

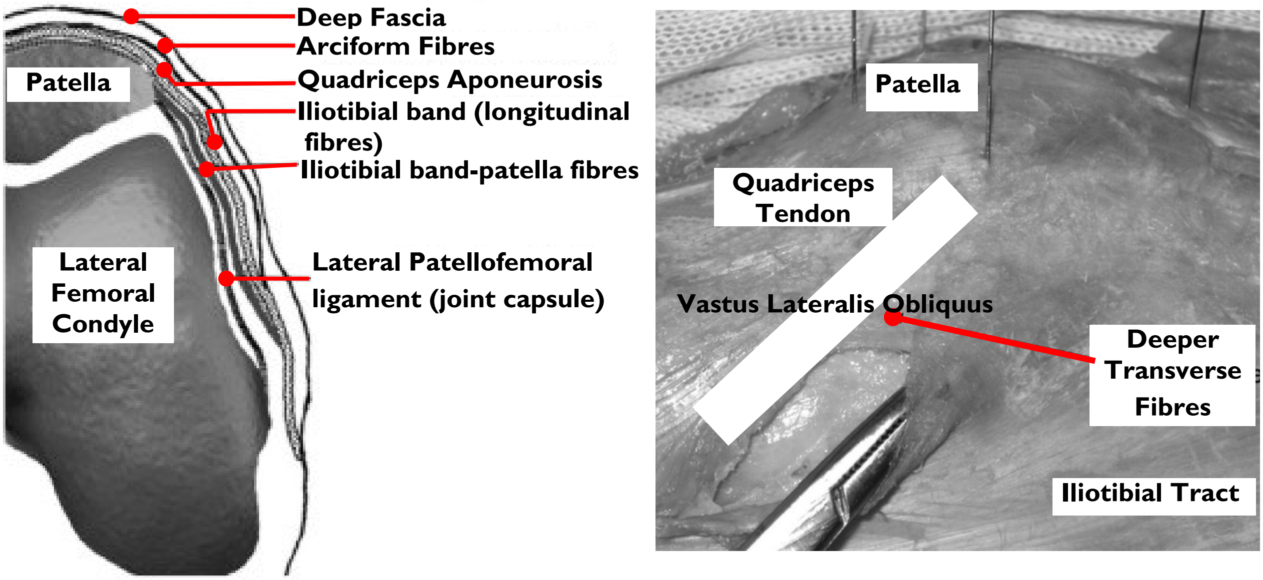

Figure 2.8 Cross-sectional diagram indicating components of the lateral retinaculum at the level of the lateral femoral epicondyle (left image) and a photograph of vastus lateralis obliquus tendon and the iliotibial tract (right image) (as adapted from Merican and Amis (2008)).

2.3 Active Structures

The primary active stabilisers of the patella are the quadriceps muscle group which insert onto the proximal patellar pole, continuing distally to cover the anterior patellar face, merging with the patellar tendon distally (Waligora et al., 2009). The quadriceps group consists of a number of muscles in differing orientations to help meet the variety of functional demands placed on the knee joint (Fulkerson, 2002). The rectus femoris lies on the vastus intermedius centrally, runs parallel with the femur, and inserts into the patella, with its superficial layer extraarticular (Staubli et al., 1999; Terry, 1989) (Figure 2.9). Taking its position as the deepest quadriceps tendon, the vastus intermedius beneath inserts into the proximal patella, with its posterior layer lined with the articular synovium distally (Staubli et al., 1999; Waligora et al., 2009). The Vastus Medialis, in accordance with its name, inserts into the superomedial third of the patella, with the angulation of its fibres meaning it is commonly separated into two portions (Terry, 1989). The Vastus Medialis Longus (VML) has a reported pennation angle of 15° and cross sectional area of 15%, and the Vastus Medialis Obliquus (VMO) has a pennation angle of 47° and cross sectional area of 10% (Farahmand et al., 1998a) (Figure 2.9). The Vastus Lateralis meanwhile inserts into the proximal patella at its lateral aspect and has also been defined as two separate groups, relating to its architecture. The Vastus Lateralis Longus (VLL) has a reported cross sectional area of 34% and pennation angle of 14° and the Vastus Lateralis Obliquus (VLO), a 35° pennation, and 10% cross sectional area (Farahmand et al., 1998a). The quadriceps tendon consists of three layers: a superficial layer with rectus femoris, middle layer with the vastus medialis and lateralis and the deep layer containing vastus intermedius (Waligora et al., 2009). Disuse atrophy in the quadriceps has been found to result in alterations in the pennation angle of the quadriceps components by over 5° (Bleakney and Maffulli, 2002).

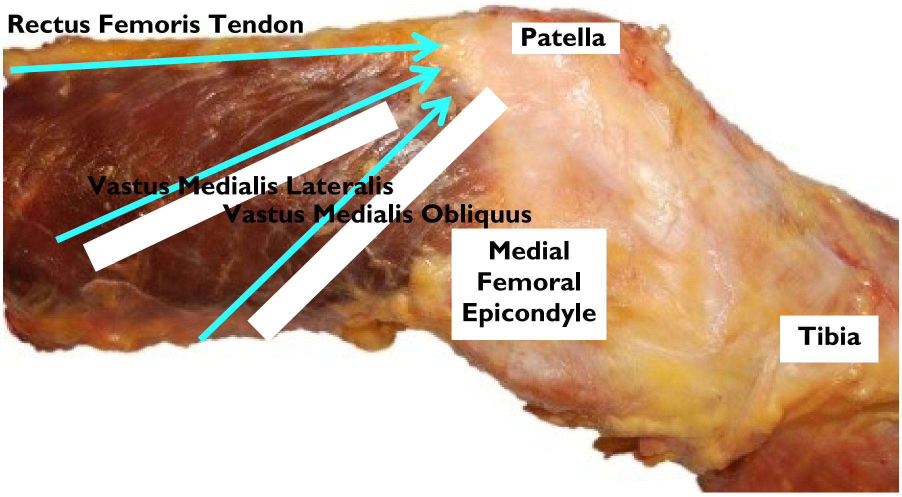

Figure 2.9 Photograph of the medial view of a left sided knee flexed to 45° showing the medial quadriceps muscles and their angulated orientation along the medial border of the patella. The vastus intermedius lies deep to rectus femoris.

2.4 Patellofemoral Biomechanics

The patella is a crucial component of the extensor mechanism, which enables transmission of quadriceps muscle tension to the patellar tendon. Its articular surface and cartilage allow the patellofemoral joint to bear very high compressive loads with low friction force, thus allowing it to distribute contact forces effectively. This is a critical role ensuring protection of the underlying highly innervated bone of the patella and trochlear groove, which could result in pain if damaged (Biedert, 2005).

A second role of the patella, alongside protection, is to improve the effective extension capacity of the quadriceps muscle by elevating the line of patellar tendon tension away from the femur (Tecklenburg et al., 2006). The moment arm is defined as the perpendicular distance from the centre of rotation of the joint to the line of action of the patellar tendon tension. The presence of the patella in Figure 2.10 can be seen to increase the moment arm 40% and, since the knee moment is the product of the patellar tendon tension and the moment arm, decrease the patellar tendon tension required to produce a knee extension moment. This is examined in greater detail in the following section.

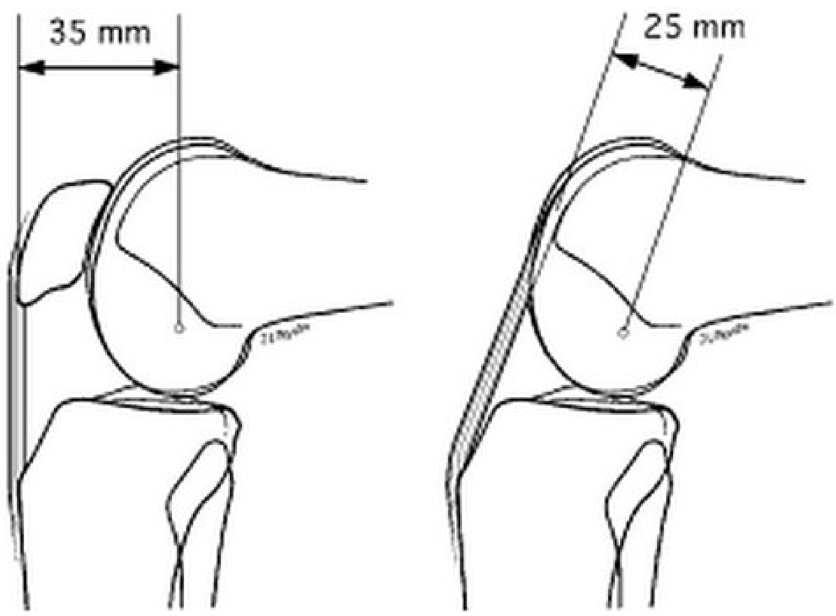

Figure 2.10 The knee moment arm is seen to decrease following patellectomy (right image), thus a larger patellar tendon tension is required to have the same knee extension strength as when the patella is present (left image) (as adapted from Amis and Farahmand (1996)).

2.4.1 Patellar Motion

The patella is permitted a large range of motion around the distal femur and has six degrees of freedom: three linear translations and three rotations (Amis et al., 2006). Distal and posterior translations of the patella as the knee flexes are the largest motions it undertakes, meaning most movement occurs in the saggital plane. Tibiofemoral flexion takes place approximately 20%-30% ahead of patellar flexion before the patella begins flexion around the femur (van Kampen and Huiskes, 1990) (Figure 2.11). This is due to the patella's distal translation on the trochlea and the change in patellar tendon orientation as the knee is flexed.

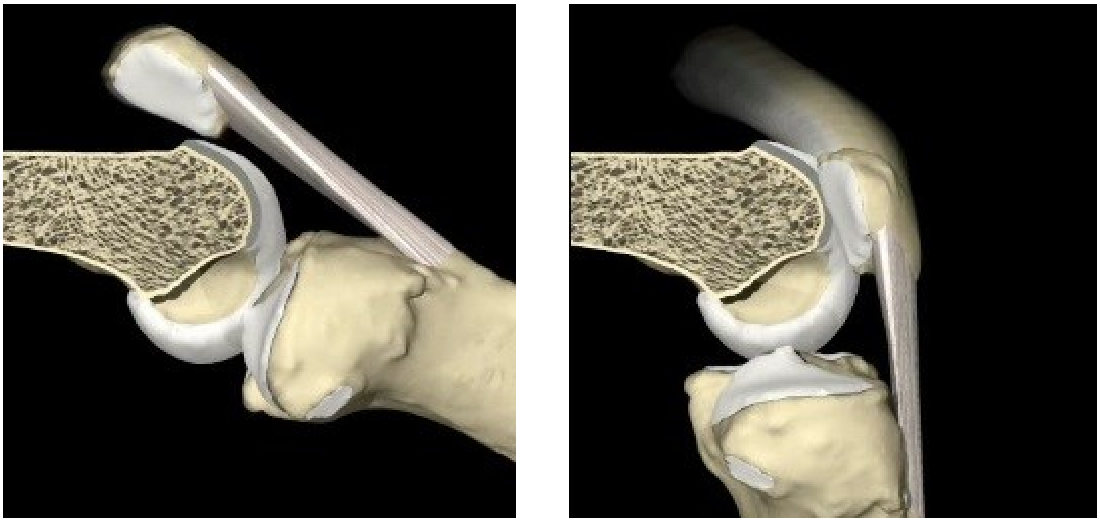

Figure 2.11 Patellofemoral joint flexion: patella flexion lags behind the tibiofemoral joint in early flexion (left side image), before moving in a distal and posterior direction by 80° flexion (right image) (as modelled by SIMM Musculoskeletal Graphics, Inc. Chicago, IL USA).

Arguably the most important motions of the patella are medial-lateral translation and patellar tilt, since these permit excursion of the patella from the trochlear groove (Laurin et al., 1978) (Figure 2.12). Consequently these measures are recognised as clinical measures of malalignment. Initially during the first 10°-20° of knee flexion the patella is typically reported to track slightly medially as the patella engages with the trochlea; this is followed by progressive lateral translation up to 80°-90° knee flexion (Amis et al., 2006; Katchburian et al., 2003). Patellar tilt is commonly described over a relatively small arc of motion, with movement in some knees thought to be close to zero (Amis et al., 2006; Katchburian et al., 2003). In pathological knees however, this value is often significantly increased (Amis et al., 2008). Finally patellar rotation (Figure 2.12), around an axis perpendicular to the patellar plane, appears to show wide inter subject variation and is thought to have little clinical significance (Amis et al., 2006). Variability in the terminology and axis of motion when defining patellar motion is evident on review of the literature and this is likely to contribute to conflicting *in vivo* and *in vitro* reports. This is discussed in greater detail in Chapter 4.

Typically most patellofemoral disorders occur in early knee flexion, before 30º (Fulkerson, 1990). This is because the patella typically enters with the trochlear groove at some stage in this range, and is more vulnerable before it engages with the trochlea since it is reliant on only soft tissue and muscular restraint (Senavongse and Amis, 2005).

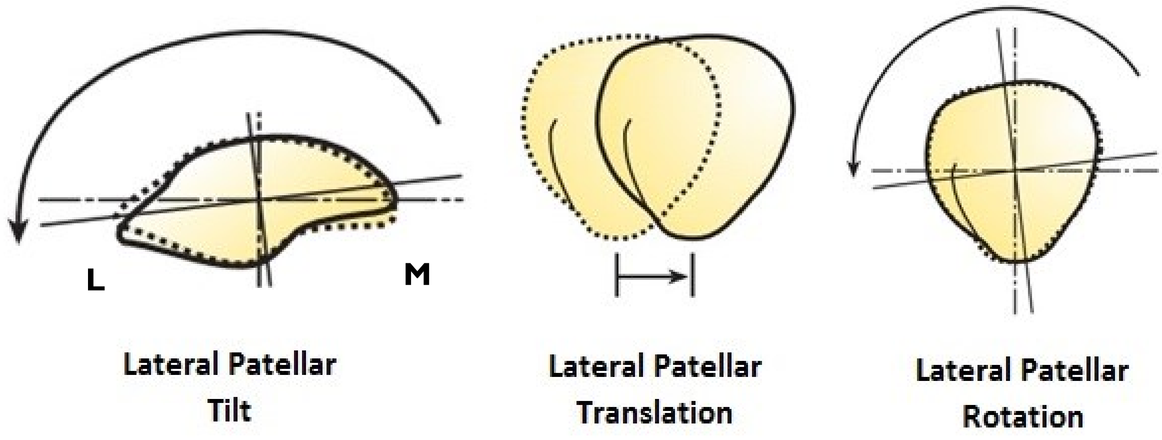

Figure 2.12 Patellar motion of a left patella (as modelled by SIMM, Musculoskeletal Graphics, Inc. Chicago, IL USA). Left image shows an axial patellar view of lateral patellar tilt, the centre and right images show the anterior patellar view of lateral translation and rotation.

2.4.2 Patellofemoral Contact Area

Contact areas of the patellofemoral joint tend to be small as a result of patellar and trochlear incongruence, particularly during early flexion (Fulkerson, 1997). The contact area on the patella moves from distal to proximal (Figure 2.13), whereas on the trochlea it moves from proximal to distal with increasing flexion angle (Feller et al., 1993) (Figure 2.14).

In terminal knee extension the patella is typically not in contact with the trochlear groove (Smidt, 1973). During initial knee flexion the distal end of the patella makes a small contact area with the proximal sulcus which rapidly spreads across the width of the distal patella as it engages with the trochlear groove at about 20° (Besier et al., 2005; Powers et al., 1998). The contact area initiates on the medial margin of the medial facet, before extending in a broad band across the patella to reach the lateral aspect of the lateral facet (Leszko et al., 2010). This contact area moves proximally with deeper flexion as patellar and trochlear congruence increases (Powers et al., 1998). From 30°-60° contact centres across the middle of the patella, with maximal patellofemoral joint contact area reported to occur at 90° when contact moves towards the proximal patellar pole (Huberti and Hayes, 1984). Beyond 90° knee flexion, the patella typically moves laterally and the medial facet becomes largely loaded on the odd facet, which rests against the lateral facing aspect of the medial condyle. Often beyond 90° knee flexion the patella will form a double contour contact, bridging across the medial and lateral condyles (Feller et al., 1993). This double contact pattern has been observed during early knee flexion in symptomatic knees (Seedholm et al., 1979). It has been suggested to occur as a result of lower limb malalignment and can cause elevated contact pressures and cartilage damage (Huberti and Hayes, 1984).

Figure 2.13 Contact area on the posterior surface of the patella during progressive angles of knee flexion as (adapted from Goodfellow et al. (1976)).

The contact area on the trochlea increases linearly until approximately 60° where it remains about constant until 90° flexion and then reduces in deeper flexion as the patella rests across the intercondylar notch (Goodfellow et al., 1976; Huberti and Hayes, 1984). At 30° flexion 19% of the patella bearing surface is in contact with the trochlea, this increases to 29% at 60° and is 28% at 90° (Matthews et al., 1977). The patella has a greater contact area on its lateral facet throughout flexion, with the area of contact with the lateral femoral groove surface 60% greater than the medial (Hehne, 1990; Salsich et al., 2003). This corresponds to its lateral translation with progressive flexion (Katchburian et al., 2003). Contact area is typically greater in weight bearing postures than non-weight bearing positions (Feller et al., 2007).

When the surfaces of the patella and trochlea are brought into contact during knee flexion, joint contact pressures and cartilage deformations can be measured. The patellofemoral joint contact pressures are termed patellofemoral contact mechanics, which arise as a direct result of the patellofemoral resultant force or the patellofemoral joint force produced from the quadriceps and patellar tendon forces (Figure 2.15).

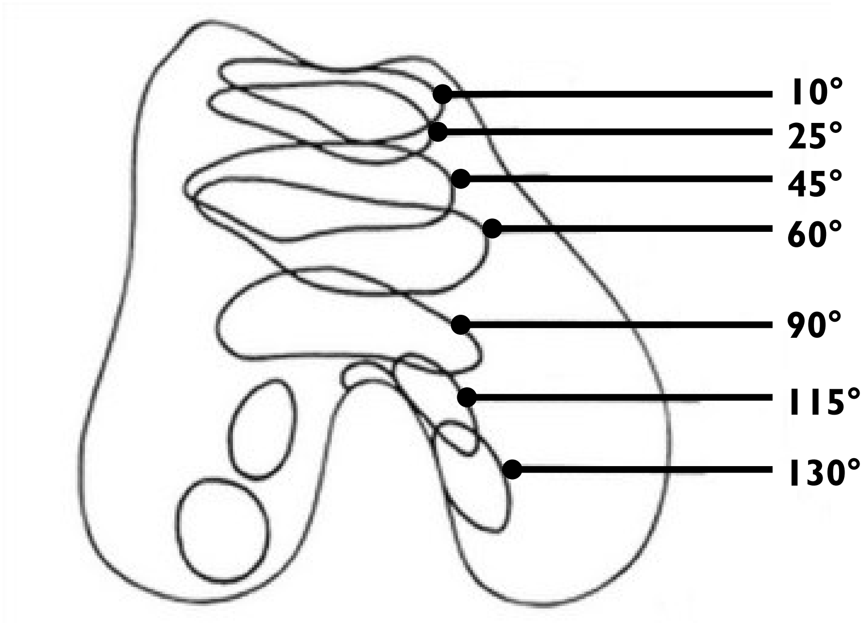

Figure 2.14 Trochlear contact area as during knee flexion from 10°-130° (as adapted from Goodfellow et al. (1976)).

2.4.3 Patellofemoral Joint Force

The patellofemoral joint is designed to tolerate high compressive stresses which, as previously discussed, is reflected in the thickness of the patellar cartilage (Fulkerson, 1997; Grelsamer and Weinstein, 2001). Although not initially apparent as a weight bearing joint, the stresses and forces imposed on the patellofemoral joint can exceed those of the tibiofemoral joint (Amis and Farahmand, 1996). A simplified model of the joint, as previously discussed, has been outlined to examine the forces acting on the patellofemoral joint in the sagittal plane (Amis and Farahmand, 1996; Sanchis-Alfonso et al., 2006) (Figure 2.15). Considering the movement of knee extension, three forces are taken to act on the knee: the quadriceps muscle tension (F_Q), the patellar tendon tension (F_{PT}) and the resultant force generated on the patellofemoral joint (F_{PFJ}). This model does not account for smaller forces such as those generated by the retinacular tensions and also assumes the patellofemoral articulation has low friction removing any significant shearing, resulting primarily in a compressive F_{PFJ}. When the knee is extended, F_Q and F_{PT} oppose one another resulting in a small joint force (Figure 2.15A). However, as the knee is flexed so the angle between F_Q and F_{PT} reduces meaning their tensions combine vectorially to result in a much larger F_{PFJ} pulling the patella onto the femur for the same F_{PT} (Figure 2.15B). Also, for the same F_{PT}, a much larger F_Q is required because the quadriceps tension is not pulling in the same direction as the patellar tendon tension when the knee is in flexion;

clinically this explains why elderly patients with weaker quadriceps may find it challenging to rise from lower chair heights, when the knee is flexed beyond 90º.

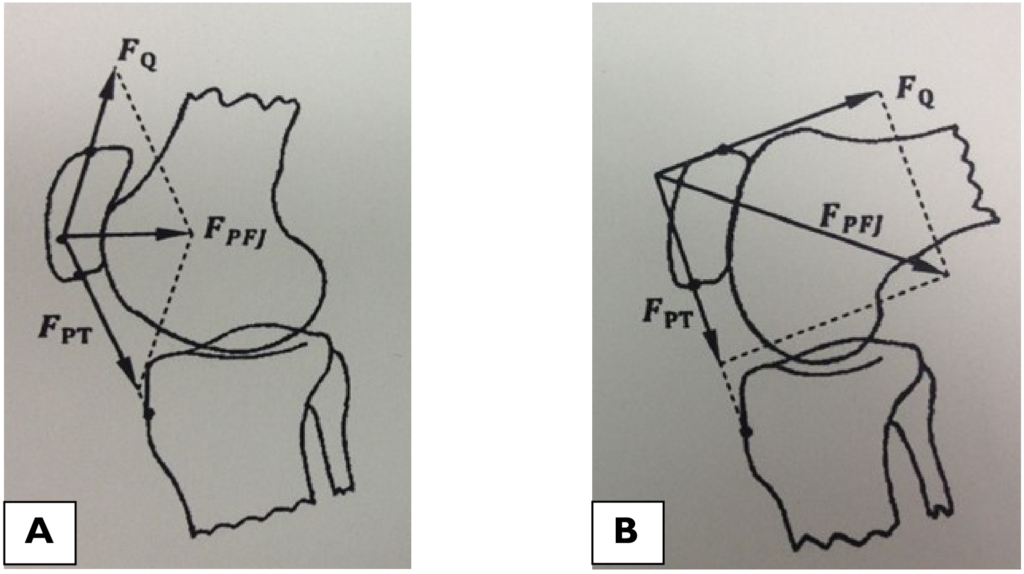

Figure 2.15 Calculation of the patellofemoral joint compressive force vector (F_{PFJ}). The image compares the F_{PFJ} in a knee close to full extension (A) with a knee in deeper flexion (B). The rise in F_{PFJ} as a result of increased knee flexion is evident (F_Q= quadriceps tendon tension, F_{PT}= patellar tendon tension and F_{PFJ} = joint force) (as adapted from Amis in Biedert (2005)).

During closed chain movements, the increase in F_{PFJ} in deeper flexion also occurs as a result of the increased lever arm of the knee as it is flexed, necessitating a more powerful quadriceps contraction to resist the flexion moment imposed by body weight (Figure 2.16). Calculation of the flexor moment is undertaken by multiplying the force that bends the joint (½ Body Weight) with the distance of its line of action (the perpendicular distance between the line of action of the force and instant centre of rotation of the knee (d_1 in Figure 2.16)). To be in equilibrium the knee must be acted upon by an equal and opposite extension moment, which arises from patellar tendon tension. The patellar tendon tension required to act on the knee is calculated as BW multiplied by distance d_1 divided by the perpendicular distance between the line of action of the patellar tendon and the instant centre of rotation of the knee (d_2 in Figure 2.16). It can be seen that d1 becomes much greater resulting in higher F_Q and F_{PT} forces as the knee is flexed (right image).

The maximum F_{PFJ} occurs between 70º - 80º knee flexion for most activities (Hehne, 1990; Huberti and Hayes, 1984). The F_{PFJ} is then mitigated by the tendofemoral contact point in deeper flexion, when the quadriceps wrap around the distal end of the femur, limiting further F_{PFJ} rises (Huberti and Hayes, 1984).

It is recognised that F_{PFJ} varies widely with different activities. Walking can result in forces up to 0.8 body weight (BW), whilst stair ascent and descent can cause forces of up to five times BW, with squatting resulting in increases of up to eight times BW (Mason et al., 2008; Trepczynski et al., 2012).

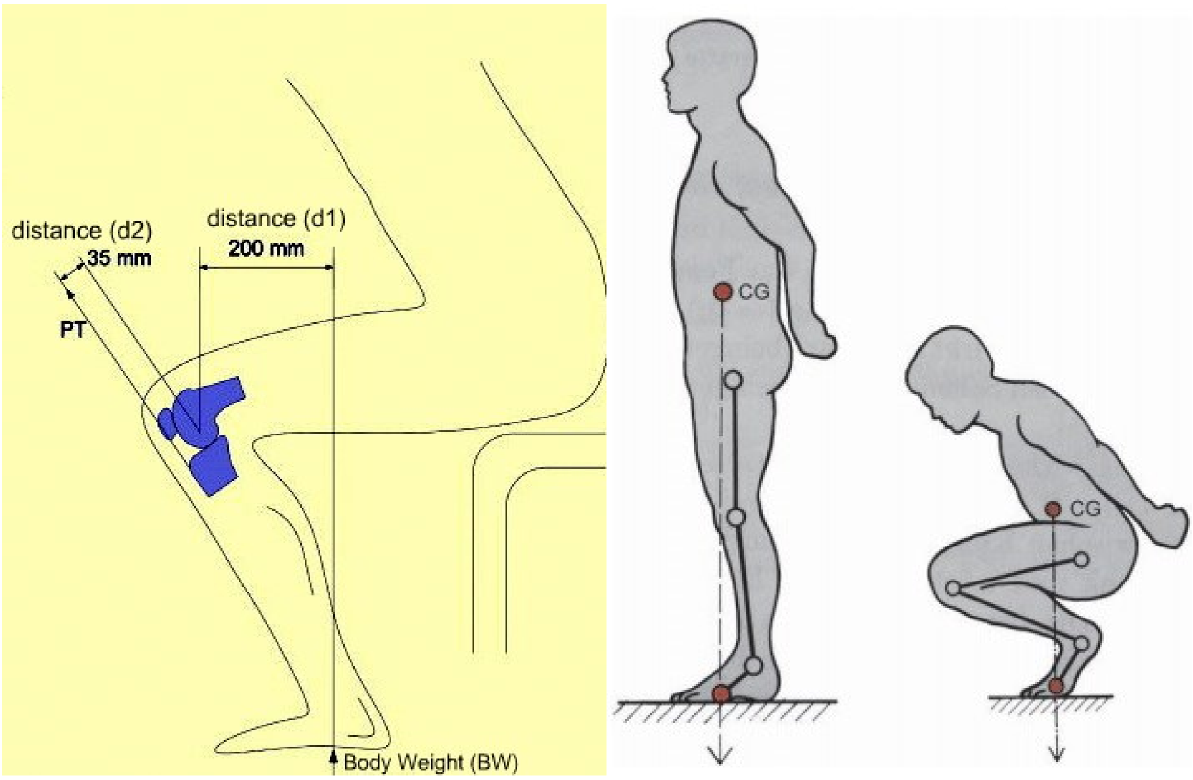

Figure 2.16 Free body diagram of a subject rising from a chair (closed chain motion). To enable equilibrium to be maintained: the patellar tendon (PT) must exert a moment equal and opposite to that caused by the body weight, so: (BW/2) x d1 = PT x d2 (as adapted from Masouros et al. (2010) (left image) and Bandi (1982) (right image)).

2.5 Patellofemoral Joint Pathology

Pain in the anterior region of the knee is one of the most common musculoskeletal disorders, particularly in active and athletic individuals, affecting both young and adult populations (Sanchis-Alfonso et al., 2006). It accounts for up to 40% of all knee problems presenting at sports medicine centres and will affect one in four of the population at some point in time (Bizzini et al., 2003; Ireland et al., 2003; McConnell, 1996). Consequently it has commanded much attention in orthopaedic literature. However despite its high prevalence the aetiology of patellofemoral joint pathology is not fully understood and treatment approaches remain controversial.

2.5.1 Patellofemoral Joint Pain

Patellofemoral joint pain is associated with an insidious onset of diffuse or local knee pain, typically made worse with activities causing patellofemoral joint compression. Frequently these are cited as prolonged sitting, going up and down stairs, kneeling, squatting, running and jumping (Crossley et al., 2002; Heintjes et al., 2005). Generally a combination of intrinsic and extrinsic variables are postulated to be predisposing factors of patellofemoral joint pain (Witvrouw et al., 2005). Included in the intrinsic factors are anatomical abnormalities (irregularities of patella, trochlea, increased Q-angle, muscular imbalance, quadriceps weakness, delayed hamstring activation, excessive pronation), and/or repetitive microtrauma (overuse) to the connective tissue (Dye, 2005; Witvrouw et al., 2005). External factors may include sudden increases in mileage or training intensity and poor equipment / training surface (Messier et al., 1991; Powers, 2000; Witvrouw et al., 2000). The foremost difficulty with this problem is the lack of one defined precursor that causes the onset and perpetuation of the disorder. Instead it seems likely that there are physiological, biochemical, biomechanical and anatomical properties working individually or collectively to cause pain (Dye, 1996; Powers et al., 2003). The majority of patients with this disorder will be treated conservatively with physiotherapy, and as such this population of patients will not be considered in detail in this thesis.

2.5.2 Patellar Dislocation

The epidemiology of patellofemoral joint dislocation has traditionally been difficult to quantify given its relatively low incidence (Atkin et al., 2000). However it has been reported to account for 3% of all knee injuries (Stefancin and Parker, 2007), with an incidence rate of between 29 and 43 individuals per 100, 000 reported (Fithian et al., 2004; Nietosvaara et al., 1994). Despite its low prevalence this topic has commanded much attention in orthopaedic literature, as a result of the unsatisfactory outcomes reported at short and longer term follow up in patients suffering the injury. One case series evaluating dislocation patients undergoing conservative management found a re-dislocation rate of 44% and a 19% incidence of pain or subluxation in the cohort who did not suffer recurrent dislocation at an average of 13 years follow up (Máenpáá et al., 1997). This is further substantiated by numerous reports highlighting the link of dislocation to on-going instability, pain and loss of function (Arendt et al., 2002; Beasley and Vidal, 2004; Cofield and Bryan, 1977) in knees which have been treated non-surgically.

Osteochondral bruising, lesions and fractures of the medial facet of the patella (Figure 2.17) and/or lateral femoral condyle are common findings reported from radiographic, MRI and arthroscopic investigations performed post dislocation (Arendt et al., 2002; Nomura and Inoue, 2005; Sallay et al., 1996). These findings are similar to those reporting osteochondral defects in the knee following anterior cruciate ligament (ACL) rupture, which have been linked to subsequent development of osteoarthritis and related to trauma at the time of injury (Keays et al., 2010; Øiestad et al., 2009). It is interesting in the context of reports of ongoing pain, recurrent instability, reduced functional levels and patellofemoral arthritis following patellar dislocation (Arendt et al., 2002; Atkin et al., 2000; Sallay et al., 1996), that the most common management following primary dislocation is conservative treatment (Bicos et al., 2007; Cofield and Bryan, 1977; Fithian et al., 2004; Stefancin and Parker, 2007).

Figure 2.17 Dorsal view of two right sided patellae. Left image showing intact and smooth articular cartilage and the right image depicting an osteochondral defect on the distal medial facet of the patella.

2.5.3 Predisposing Factors

Three systems are known to act in unison to ensure patellofemoral joint stability throughout the range of knee flexion (Senavongse et al., 2003). The bony geometry of the patella and trochlear groove provide static stabilisation (Senavongse and Amis, 2005), the quadriceps and to an extent the gluteal muscles function as active stabilisers (Powers et al., 2003), while the ligaments and retinacula provide passive joint restraint (Desio et al., 1998).

Wide individual subject variations in the geometry and articulations of the distal femur and patella have been identified in the literature. These can predispose individuals to patellar dislocation (Dejour, Walch et al. 1994, Staubli 1999). These often subtle differences have been investigated in detail, with correction of many providing the underlying rationale for numerous surgical interventions currently in practice for the treatment of patients suffering patellar dislocation. The Medial Patellofemoral Ligament (MPFL) has been identified as the most important patellofemoral joint stabiliser in early knee flexion from 0°-30°, contributing 50%-60% of the passive resistance to lateral patellar motion through this range (Desio et al., 1998; Panagiotopoulos et al., 2006; Senavongse and Amis, 2005). It is examined in depth in Chapter 3. Trochlear shape and tibial tuberosity position have also been identified as a factors contributing to increased incidence of patellar instability, as has patella alta (Dejour et al., 1994; Fithian et al., 2004; Simmons and Cameron, 1992). These pathologies are now each discussed in greater detail.

2.5.3.1 Tibial Tuberosity - Trochlear Groove Distance

In the anatomically normal knee joint the tibial tuberosity is positioned distal and slightly lateral to the femoral sulcus directing an inferolateral force vector to act on the patella during knee flexion. Consequently the presence of an excessively laterally aligned tibial tuberosity relative to the trochlea groove results in an increased lateral vector acting on the patella. This pathology has been identified in a large percentage of patients examined following patellar dislocation (Dejour et al., 1994).

The tibial tuberosity - trochlear groove distance (TT-TG) was first described as a measurement using axial X-rays taken in 30° flexion by Goutallier et al. (1978), enabling quantification of the coronal alignment of the extensor mechanism. Later adaptation of this method enhanced accuracy (Wagenaar et al., 2007) through the use of Computed Tomography (CT) scans, taking two slices: one through the proximal trochlea and one through the proximal part of the tibial tubercle, and then measuring the distance between the deepest point of the trochlear groove and the anterior tibial tuberosity (Dejour et al., 1994) (Figure 2.18). More recently MRI scans have been used to make these measurements, with the inclusion of cartilage proposed to make them more accurate than CT scans (Schoettle et al., 2006). Normal ranges for TT-TG distance have been inconsistently reported: Pandit et al. (2011) reported a range from 9-10mm, Wittstein et al. (2006) using MRI found an average distance of 9.4mm, Alemparte et al. (2007) investigating healthy volunteers identified a mean distance of 13.6mm and Dejour et al. (1994) reported in a widely cited paper normal distance to be around 13mm, defining pathological range as being in excess of 20mm to provide an indication for surgical intervention.

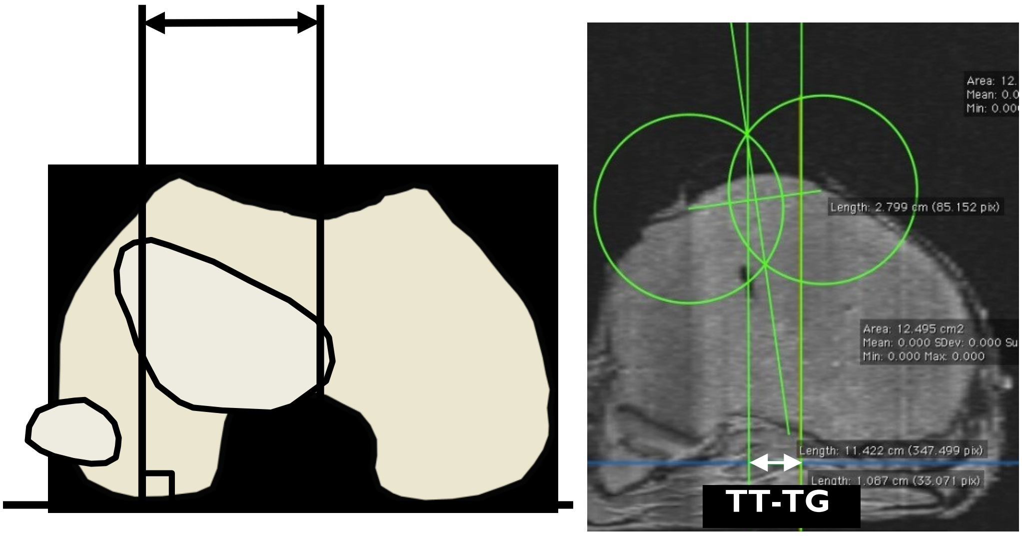

Figure 2.18 Tibial tuberosity – trochlear groove (TT-TG) distance: left side an image representing cross sectional CT slices from which to take TT-TG measurements and right side MRI sections to enable TT-TG measurement.

2.5.3.2 Trochlear Dysplasia

The lateral vector applied following quadriceps contraction is counteracted by the concave trochlea which, as previously outlined, is typically raised. The convex shape of the patella is contoured to track over the trochlear groove, providing a relatively uniform contact pressure across the patellar width (Hehne, 1990). Trochlear dysplasia is present when the trochlear groove has a flattened proximal articular zone and a shallow distal zone, the intercondylar groove is flattened or in the worst cases convex (Pfirrmann et al., 2000) (Figure 2.4). It is intuitive therefore that in patients with trochlear dysplasia, high levels of dislocation are common given the loss of lateral restraint to the laterally acting vector through knee flexion (Dejour et al., 1994). Despite initial underestimation of its presence (Dejour, 1990), intra-operative findings have confirmed the presence of flattened or convex intercondylar grooves, providing a rationale for operative intervention with trochleoplasty surgery (Donell et al., 2006; Von Knoch et al., 2006). Dysplasia is typically diagnosed with a positive crossing sign (when the floor of the trochlear groove crosses the anterior border of the two femoral condyles) identified from lateral x-rays taken at 30˚ knee flexion. A widely used classification of these problems is summarised in Figure 2.19 (Dejour et al., 1994). Unfortunately since this thesis focusses on cadaveric analysis, and as specimens with dysplastic knees are not easily accessible from tissue banks, this thesis does not examine this pathology in detail.

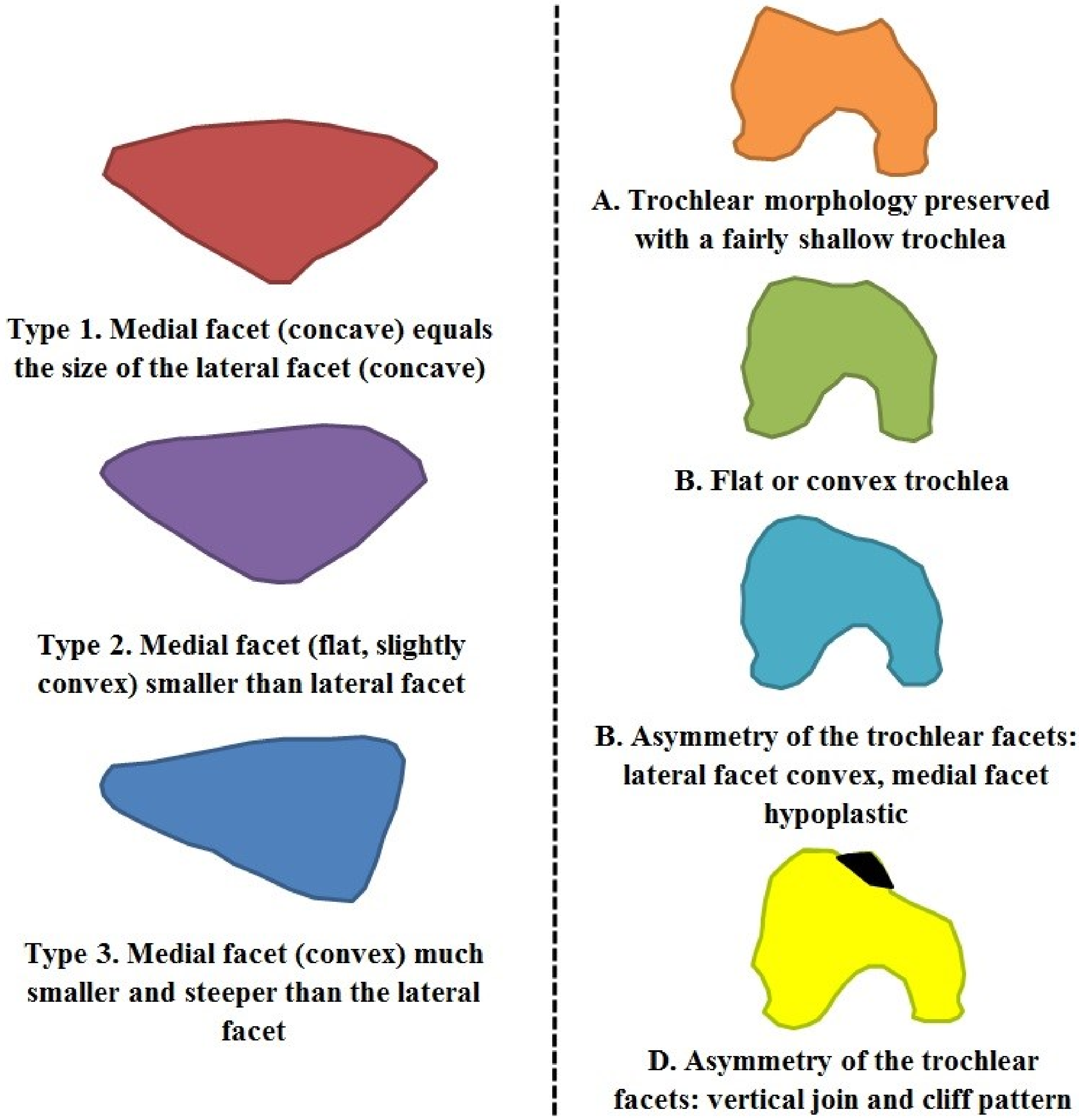

Figure 2.19 Wiberg classification of patellar morphology: Type I, Type II and Type III (left side), and on the right side Dejour's classification of the dysplastic trochlea: Grades A, B, C, and D.

2.5.3.3 Patella Height

2.5.3.3.1 Patella Alta

Patella alta exists where the patella lies proximal to its normal position, making the patella vulnerable to dislocation given the loss of trochlear restraint through early flexion (Figure 2.20). It often co-exists with abnormalities such as trochlear dysplasia or an overlengthened patellar tendon (Caton et al., 1990; Dejour, 1990) and is reported to be present in 24% of patients with patellar instability versus 3% of patients with no instability (Dejour et al., 1994).

Patella alta results in decreased contact between the patella and trochlea, resulting in elevated patellofemoral joint pressures and the potential for subsequent development of osteoarthritis (Luyckx et al., 2009; Stefanik et al., 2011; Ward and Powers, 2004). It results in reduced resistance to lateral translation of the patella, particularly in early knee flexion, given the loss of lateral bony constraint of the trochlear groove to the quadriceps pull acting on the patella (Singerman et al., 1994; Ward and Powers, 2004). Clinical studies report successful surgical treatment of recurrent patellar dislocation in this population with tibial tuberosity distalisation (Magnussen et al., 2013; Simmons and Cameron, 1992).

Patients suffering patellar dislocation from no/minimal trauma have been found more likely to have patella alta than those with traumatic dislocations (Geenen et al., 1989). The pathology is associated with recurrent dislocation of the patella, pain and joint effusion (Insall et al., 1976; Insall et al., 1972; Møller et al., 1986). Such clinical findings emphasise the critical need to assess patellar position during the clinical evaluation of the anatomical alignment of the knee, especially in patients with pain or instability.

2.5.3.3.2 Patella Infera

Patella baja occurs when the patella is situated lower than anatomically typical. It is usually classified as congenital, acquired or a combination of the two (Chonko et al., 2004). Acquired baja can be the result of trauma or surgery which cause changes in the quadriceps activation or joint mechanics, such as an elevation of the knee joint line in total knee replacement (Chonko et al., 2004). Meanwhile congenital baja occurs from an early age and can be the result of disease such as poliomyelitis (Giori and Lewallen, 2002). Symptoms of pain and loss of range of motion are common complaints in patients presenting with patella baja. It is not linked as a causative factor to patellar instability.

2.5.3.4 Patellar Height Measurement

The measurement of patellar height has been controversial, with a number of methods proposed in the literature. It is most commonly defined using a range of patella tibia indices. The most common of these are summarised overleaf and in Figure 2.20.

1. Insall-Salvati Ratio: Length of patellar tendon to longest sagittal diameter of the patella (Ratio > 1.2 indicative of patella alta) (Insall and Salvati, 1971).

2. Modified Insall-Salvati Ratio: Length of the patellar tendon to length of articular surface of patella (Ratio > 2 indicative of patella alta) (Grelsamer and Meadows, 1992).

3. Blackburne-Peele: The ratio of the length of the perpendicular line drawn from the inferior pole of the articular surface of the patella to the tangent of the tibial plateau and the length of articular surface of the patella (Ratio > 1.0 indicative of patella alta) (Blackburne and Peel, 1977).

4. Caton-Deschamps Ratio: Distance from the lower edge of the articular surface of the patella to the antero-superior angle of the tibia outline to the length of the articular surface of the patella (Ratio > 1.2 indicative of patella alta) (Caton, 1989).

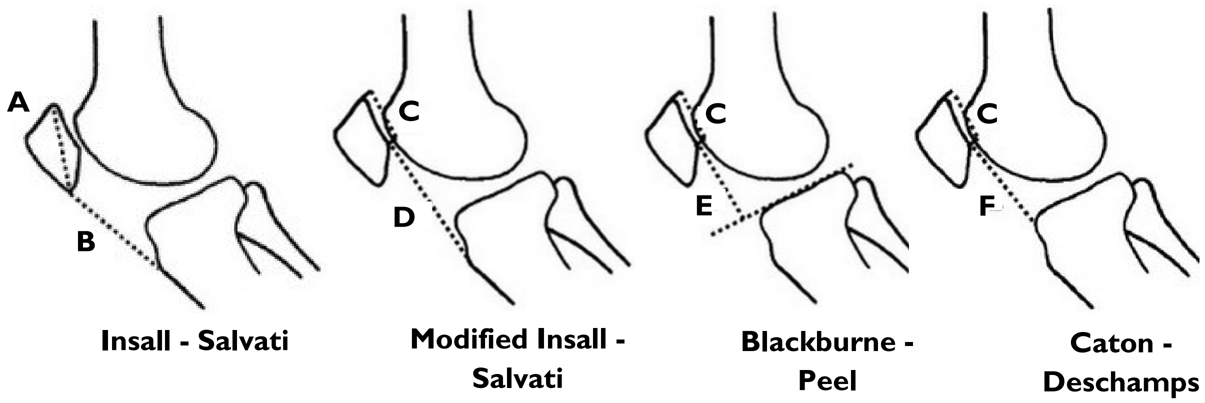

Figure 2.20 Indices of patellar height. **A** patellar length, **B** patellar tendon length, **C** articular surface of the patella, **D** distance from the inferior articular surface of the patella and the patellar tendon insertion, **E** height of the inferior pole of the patellar articular cartilage above the tibial plateau, **F** distance between the inferior articular surface of the patella and the anterior tibial plateau (as adapted from Shabshin et al. (2004)).

The range of assessment methods reported for the quantification of patellar height has led to a lack of agreement amongst surgeons in the diagnosis of patellofemoral disorders (Saggin et al., 2012). Patella tibia indices have been criticised as relating patellar position to the tibia and failing to define the patella in relation to the femoral trochlear groove, which is likely to be of greater clinical significance. Consequently a series of patella femoral / trochlear indicies have also been reported. These are summarised overleaf and in Figure 2.21.

1. Bernageau: a sagittal view X-ray with the quadriceps contracted and the knee fully extended. The distance (d) between the superior line of the trochlea (T) and the inferior edge of the articular surface of the patella (R) is measured (if R is > 6mm above T, authors report this is indicative of patella alta) (Bernageau and Goutallier, 1984) (Figure 2.21).

This measurement came under some criticism as a result of the difficulty encountered in clearly defining the necessary landmarks on radiographs. It has therefore subsequently been modified.

2. Patellotrochlear Index: this method uses MRI, thus accounting for the true articular relationship of the patellofemoral joint. Sagittal view MRI images are taken with the foot in 15º external rotation, the knee in 0º extension and the quadriceps muscle relaxed. The index is the ratio baseline trochlea (BLt) / baseline patella (BLp) reported as a percentage (values less than 12.5% are reported indicative of patella alta) (Biedert and Albrecht, 2006) (Figure 2.21).

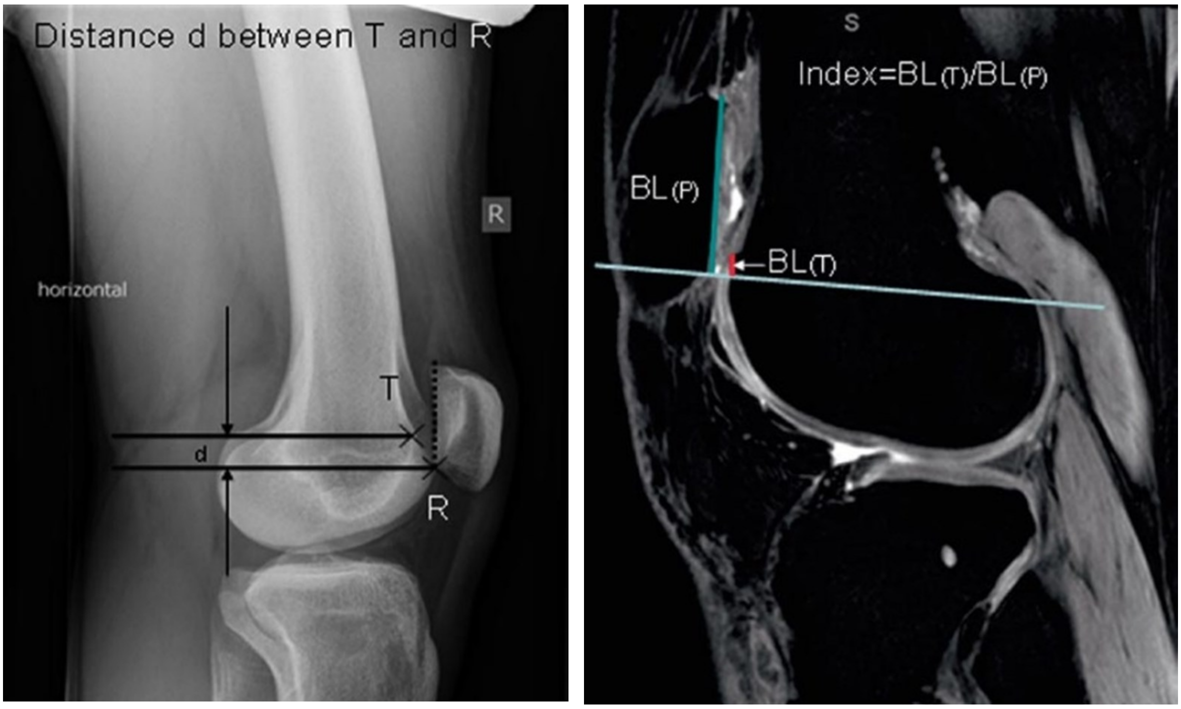

Figure 2.21 Bernageau Index (left image) and Biedert and Albrecht Index (right image) (as adapted from Zaffagnini et al. (2010)).

However, this improved method has also been criticised. Biedert and Albrecht (2006) examined ‘normal’ knees, using only one slice of MRI data and thus failed to account for cases of patellar dislocation, where axial alignment of the patella may result in it not lying directly anterior to the proximal trochlea on the MRI slice examined (Biedert and Albrecht, 2006; Dejour et al., 2013). This is highlighted in Figure 2.22, where the patella is shown to lie in a lateral position in relation to the trochlear groove. This may be a common occurrence in patients following patellar dislocation who would typify the population undergoing this measurement. Such alignment would not be accounted for by the patella femoral / trochlear indices previously defined.

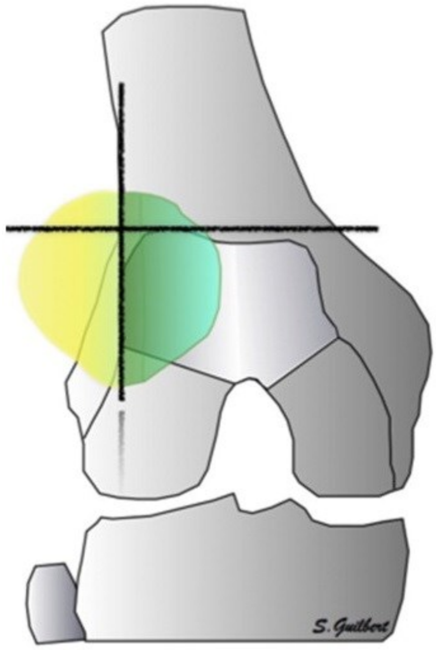

Figure 2.22 The patella is shown to lie lateral to the trochlear groove, which would be accounted for in the Saggital Patellofemoral Engagement but not by the Patellotrochlear Index (as adapted from Dejour et al. (2013)).

Recently therefore a third measurement has been proposed to address these issues.

3. Sagittal Patellofemoral Engagement (SPE): this involves measurement from MRI scan using two different image slices. The first slice is that where the longest patellar articular cartilage is measured (PL) (Figure 2.23 (left image)). The second slice is the sagittal section where the femoral trochlear cartilage extended most proximally (Figure 2.23 (right image)). The PL line is superimposed on the second MRI slice and a second line (TL) drawn parallel to the PL line. TL represents the distance from the most proximal edge of the trochlear groove to the distal end of PL. The SPE can then be calculated as the ratio between TL and PL (Dejour et al., 2013) (Figure 2.23).

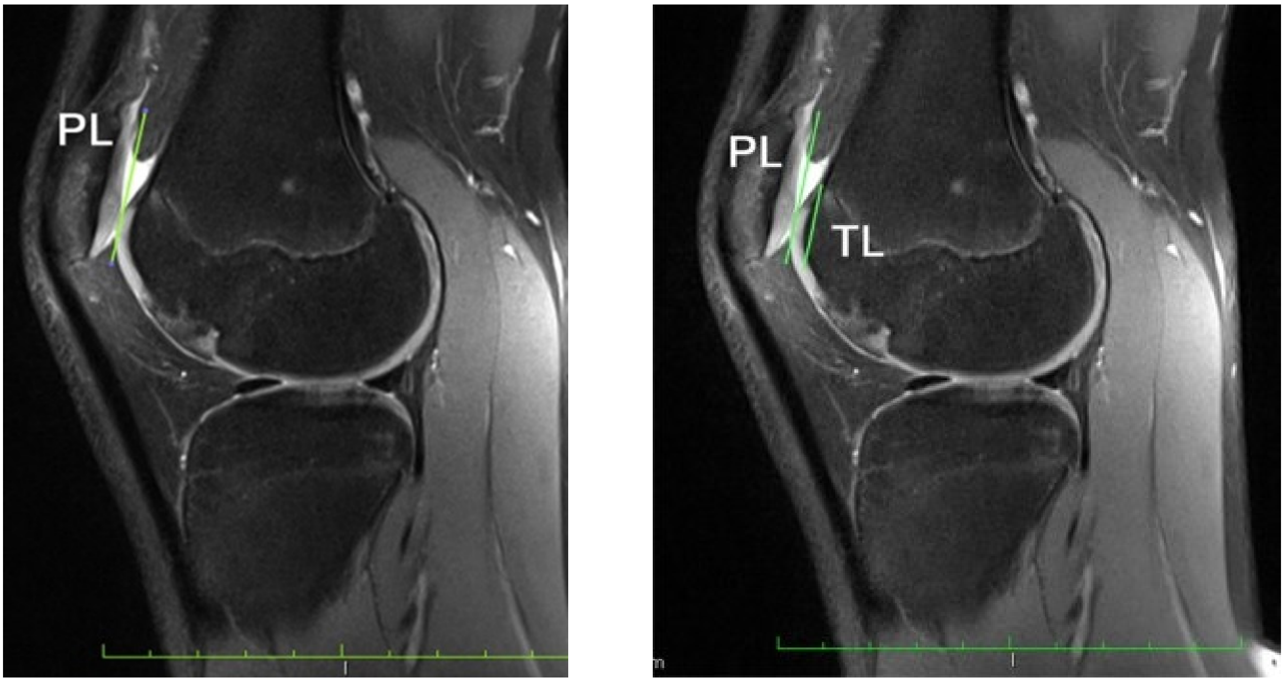

Figure 2.23 Sagittal Patellofemoral Engagement (SPE) measurement taken from MRI scans using Osirix freeware DICOM viewer. Standard non weight bearing scan, taken with the knee close to full extension (as adapted from Dejour et al. (2013)).

This index has only recently been defined, so it is yet to be seen whether it gains more general acceptance as a method for defining patellar height in relation to the trochlea, whilst addressing issues of patellar axial alignment. The orthopaedic community would benefit from the use of one consistent measure of patellar height to enable comparison of data across studies and aid in the standardisation of surgical intervention for this population of patients.

The ability to assess the biomechanics of patella alta *in vivo* and *in vitro* presents the researcher a challenge of either direct measurement or simulation of pathology in cadaveric specimens. One group simulated patella alta in ovine stifle joints by cutting and then progressively elongating the patellar tendon by 2mm increments using a custom made patellar tendon lengthening device (Bertollo et al., 2012). They found that even a 2mm increase in patellar tendon length resulted in significant changes to patellar tilt. Unfortunately, again owing to the lack of available specimens with this specific and unique pathology, patella alta is not examined in the cadaveric experiments covered by this thesis.

2.6 Conclusion

The patellofemoral joint is a synovial joint which forms part of the knee joint complex. It permits movement in six degrees of freedom and is stabilised by ligamentous, muscular and bony interaction. Deficiencies or anomalies in any one of these systems can increase an individual's susceptibility to developing patellofemoral pathology. Knowledge of contact mechanics and patellar kinematics can therefore aid in the diagnosis and subsequent management of patients with patellofemoral joint disorders.

CHAPTER 3

THE MEDIAL PATELLOFEMORAL LIGAMENT

3.1 Background

The role of the Medial Patellofemoral Ligament (MPFL) in patellofemoral joint stability has been outlined and is widely accepted. However current literature lacks consensus when describing its anatomy and length change behaviour, highlighting the need for further research into this intricate structure.

3.2 Anatomy

The MPFL is now widely acknowledged to be present in all knees (Baldwin, 2009). However its presence has been widely debated, with some studies only identifying it in 29%-88% of dissected knees (Aragão et al., 2008; Conlan et al., 1993) whilst others found it to be present in all examined knees (Amis et al., 2003; Baldwin, 2009; Nomura et al., 2000a; Tuxøe et al., 2002). This disagreement has been attributed to the complex anatomy in the medial region of the knee, wide inter-subject variations in the dimension and thickness of the ligament and the variety of dissection techniques employed to examine the area (Bicos et al., 2007). Discrepancy and debate as to the precise anatomical details of the ligament remain on-going in the literature (Baldwin, 2009).

The ligament fibres of the MPFL widen at both the femoral and patellar insertions, with the ligament approximately 53mm in length (range: 45-64mm) (Mochizuki et al., 2013; Tuxøe et al., 2002). Its femoral attachment width reportedly ranges from 8mm-25mm (Amis et al., 2003; Baldwin, 2009; Philippot et al., 2009; Weber-Spickschen et al., 2011), with the patellar attachment ranging between 16-40mm (Aragão et al., 2008; Baldwin, 2009; Mochizuki et al., 2013; Philippot et al., 2009).

Tissues covering the anteromedial aspect of the knee are arranged in three distinct layers (Stefancin and Parker, 2007), with the MPFL identified to lie in the second layer below the deep fascia, but superficial to the joint capsule (Feller et al., 1993). Here it shares a close relationship with the superficial and superior fibres of the medial collateral ligament (MCL). Significant decussation of the MPFL and MCL has been described in all knees dissected by Desio et al. (1998), Burks et al. (1998) and Baldwin (2009). However Nomura et al. (2000a) found this relationship present only in 10 out of 20 dissected specimens. The challenge of dissecting the anatomy in this area to individual distinct structures is likely to be the reason for conflicting reports (Bicos et al., 2007), with Panagiotopoulos et al (2006) noting that the fibres in this area were impossible to separate. Further studies have alluded to the close intermeshing of the MPFL and MCL, adding support to the case of the stabilising role of the

MPFL on the patellofemoral joint (Conlan et al., 1993; Feller et al., 1993; Smirk and Morris, 2003; Tuxøe et al., 2002).

3.2.1 Femoral Attachment

Anatomical descriptions have tended to define the femoral attachment of the MPFL in relation to landmarks of the medial epicondyle, medial collateral ligament and the adductor tubercle and indeed some reports use these points interchangeably (Panagiotopoulos et al., 2006; Tuxøe et al., 2002). Amis et al. (2003) concluded the MPFL originated from the medial epicondyle of the femur, whilst Davis (2002) described the MPFL to take its femoral origin from links to the adductor tubercle and medial epicondyle. Desio et al. (1998) described a wide attachment which is spread by crossing fibres attaching to both the superficial fibres of the MCL and the adductor tubercle, with more direct attachment to the epicondyle. In a comprehensive study undertaken by Baldwin (2009), it was determined that the adductor tubercle provided exclusive attachment for the adductor magnus tendon and the medial epicondyle exclusive attachment for the MCL. Between these two landmarks a groove was described from which the MPFL originated. Bicos et al (2007) defined a similar point in an anatomical review of the knee joint. This location has now gained more widespread acceptance as the femoral tunnel position recommended for anatomical MPFL reconstruction (Bicos et al., 2007; Philippot et al., 2009).

Numerous reports have attempted to quantify the location of the MPFL origin on the femur, aiming to assist surgeons in recognising its location during reconstruction of the ligament intra-operatively. Phillippott et al (2009) described the femoral attachment as situated 10mm posterior to the medial epicondyle and 10mm distal to the adductor tubercle, while others found it 2mm anterior and 4mm distal to the adductor tubercle (LaPrade et al., 2007). Nomura and Inoue (2005), found the centre of the anterior border of the MPFL femoral attachment on average 9.5mm proximal and 5mm posterior to the medial epicondyle. However the use of such precise measurements presents two main problems. Firstly they are of limited use intra-operatively and secondly given inter-patient variability in femoral sizes it is problematic to use one set of measurements interchangeably; the adductor tubercle has been reported as between 15mm to 25mm from the medial epicondyle depending on the specimen (Philippot et al., 2009). Schöttle et al (2007) proposed a radiographic point for determining the femoral MPFL insertion using anatomical reference lines. A lateral view x-ray, aligning the posterior condyles of the femur is taken and the MPFL femoral attachment described as shown in Figure 3.1.

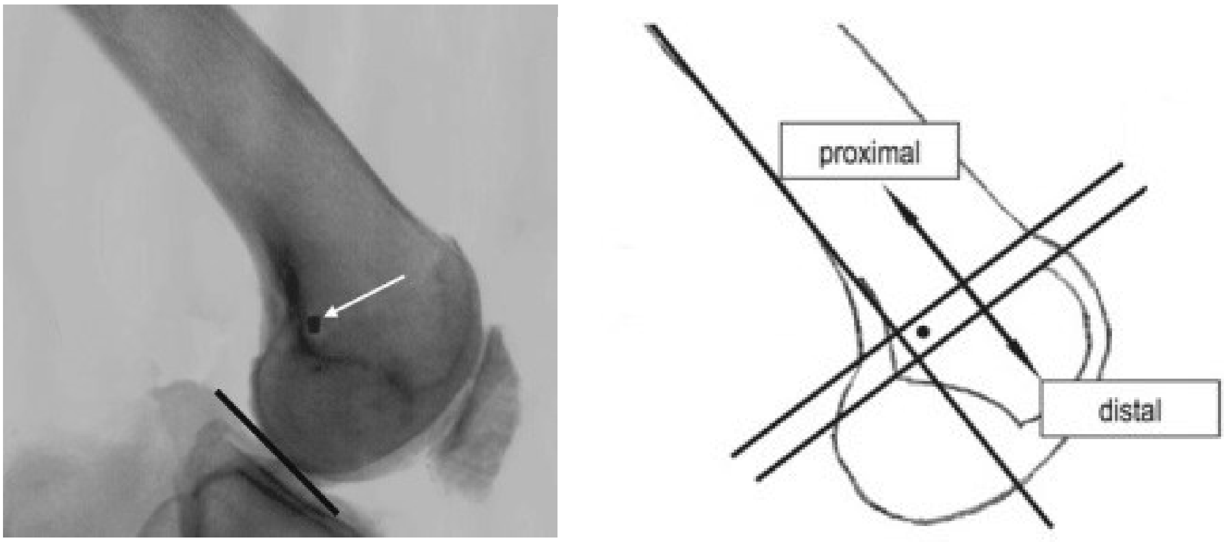

Figure 3.1 Left side image showing a true lateral view x-ray with the posterior femoral condyles aligned and a metal ball marking the femoral MPFL insertion (as adapted from Schöttle et al (2007)).

Right side image showing a line drawn as an extension to the posterior cortex. Two lines perpendicular to this line are drawn as shown; 1. The contact point of the medial condyle and posterior cortex and 2. Intersecting the most posterior point of Blumensaat line (as adapted from Schöttle et al (2007)).

This point has become widely referenced in clinical papers as a method to confirm anatomical femoral tunnel position during MPFL reconstruction (Servien et al., 2011). However its accuracy has been questioned; the radiographic guidelines were examined by Redfern et al (2010), and 5mm differences in the anterior-posterior and 7mm in the proximal-distal position around the actual attachments were determined. These differences likely occur as a result of the variability in extending a straight datum line from the curved outline of the posterior femoral cortex. Further investigations are merited to determine if a more accurate point can be determined for surgeons to use intra-operatively.

3.2.2 Patellar Attachment

The attachment of the MPFL to the medial patella and local soft tissues has been somewhat less contentious, although some discrepancies exist. Commonly described attachment points include the medial upper two thirds of the proximal patella (Nomura and Inoue, 2003), the proximal half of the patella (Conlan et al., 1993; Stefancin and Parker, 2007), and the superomedial aspect of the patella via the vastus medialis tendon (Amis et al., 2003). The length of MPFL attachment has been found to correspond closely with the length of articular cartilage along the medial patella border (Nomura et al., 2000a). Many studies generally accept that the MPFL attaches to the superomedial portion of the patella, and under the surface of the Vastus Medialis Obliquus tendon (VMO) (Smirk and Morris, 2003). However, despite frequent imaging and measurement studies, the relationship of the MPFL and the VMO tendon is not consistently defined. Indeed a recent report has instead reported the MPFL fibres mesh with the Vastus Intermedius (VI) muscle, whilst lying beneath the VMO fibres with little interlinking of the MPFL and VMO evident (Mochizuki et al., 2013). This finding has been previously alluded to in the literature (Aragão et al., 2008; Conlan et al., 1993) , but is not consistently reported. Again discrepancies are likely from the subjective challenge of outlining the MPFL borders given its close relationship with the fascia of the medial retinaculum. Radiographic landmarks at approximately the junction of the proximal 1/3 and distal 2/3 of the patella have been suggested for reference during reconstruction, however these have not been investigated in detail (Barnett et al., 2012).

3.2.3 Strength

Strength studies of the MPFL appear to provide a consistent basis for the stabilising role of the MPFL alongside its close anatomical position to the MCL and VMO. The MPFL has been found to have a mean failure load of 208N with the femur stabilised and patella distracted in an anterolateral direction until rupture (Mountney et al., 2005). The authors emphasised that these measurements likely underestimated its strength given the mean cadaver age used for testing was 70 years. Previously it has been reported the anterior cruciate ligament demonstrates approximately 2.5 times its strength in the third decade compared to the seventh (Woo et al., 1991). The apparent resilience of the ligament caused some surprise initially given its thin, transparent appearance and relatively small dimensions; however this work adds support to the role of the MPFL in patellofemoral joint stability.

3.2.4 Ligament Length Change Patterns

The biomechanical behaviour of the MPFL is not widely understood or agreed upon in current literature. It has previously been reported as close to isometric (Ghosh et al., 2009; Steensen et al., 2004; Tateishi et al., 2011) although this is disputed by others (Higuchi et al., 2010; Victor et al., 2009). Differing experimental methodologies account for these contrasting results. Smirk and Morris

(2003) first attempted to quantify the length change pattern of the MPFL and the effect of using differing femoral and patellar attachment points. They used embalmed cadavers with an axial 1.25kg load applied to rectus femoris. A string was attached between differing combinations of nails marking anatomical and non-anatomical MPFL attachments on the femur and patella to measure length changes through knee flexion range. A change of less than 5mm was considered isometric. The authors determined that the anatomical MPFL attachment points provided an isometric pattern from 0°-60° knee flexion. In contrast anteriorly positioned femoral attachments caused up to 12mm tightening of the MPFL. The authors acknowledged limitations of the study including the lack of a physiological patellofemoral joint load and the use of embalmed cadaveric tissue which failed to replicate *in vivo* soft tissue properties. Later studies have attempted more direct measurement of the ligament behaviour though knee flexion range. Authors have used progressive Magnetic Resonance Imaging (MRI) scans through knee flexion to enable subsequent 2D analysis and calculation of ligament length change patterns (Higuchi et al., 2010). Whilst a further study used CT scans to create a computerised knee model to derive ligament length change patterns via direct measurement (Victor et al., 2009). However both these methodologies are limited since both fail to account for the natural cam geometry of the medial condyle and also do not take account of the precise location of the origin and insertion of the MPFL.

MPFL isometry has been found to be most sensitive to the femoral attachment of the ligament, suggesting its importance in surgical outcome following reconstruction (Bicos et al., 2007; Smirk and Morris, 2003). This is similar to the literature reported in relation to ACL reconstruction where correct femoral tunnel positioning is now recognised as paramount to successful post-operative outcome (Amis and Jakob, 1998). Evidence suggests that failure to secure an anatomically accurate femoral attachment when undertaking MPFL reconstruction will result in adverse outcomes such as undesirable ligament tension and patellofemoral contact pressures (Bicos et al., 2007; Smirk and Morris, 2003; Thaunat and Erasmus, 2009). Numerous reports have outlined techniques for reconstruction, and radiographic landmarks, as outlined, have been suggested to guide anatomical femoral tunnel positioning, although these have not always been consistent (Redfern et al., 2010; Schöttle et al., 2007; Tateishi et al., 2011). Furthermore, little research has been published to suggest the consequence of an incorrectly positioned femoral tunnel (Smirk and Morris, 2003).

3.3 Clinical Relevance

This chapter presents data relevant to patient populations who have suffered patellar dislocation in the presence of otherwise normal anatomy in the knee. Typically this is a younger population of patients who suffer patellar dislocation as a result of trauma.

3.4 Aims

In summary, current literature about the MPFL has included differing femoral attachments and a lack of corroboration of studies of the length change patterns resulting from those attachments. The purpose of the first laboratory controlled study in this thesis was therefore threefold:

1. To determine the femoral attachment of the MPFL and recommend a reproducible femoral attachment site for undertaking anatomical MPFL reconstruction through the use of anatomical and radiographic methods.
2. To determine the length change pattern of the native MPFL through knee flexion range.
3. To investigate the effect of non-anatomical femoral attachment sites on length change patterns.
4. To investigate the effect of varying patellar attachment sites on length change patterns.

3.5 Materials and Methods

3.5.1 Specimen Preparation

Ethical approval for the study was obtained from the local Research Ethics Committee. Five male and three female right sided fresh frozen cadaveric knees of mean age 73.5 years (range 46-88), were obtained from a tissue bank for investigation. Throughout the experiment specimens were preserved in a polyethylene bag, stored in a secure freezer at -20˚C and thawed in a refrigerator at 2˚C for 36 hours prior to use. During use the specimens were kept moist with occasional water spraying.

The tibia and femur of each knee was measured to 15cm and 20cm respectively and the excess bone end sawn off. The head of the fibula was fixed to the tibia using two transversely positioned bone screws and the distal fibular bone then sawn off. The specimens were dissected to remove skin and subcutaneous fat with care taken to avoid insult to the medial and lateral retinacula (Figure 3.2). An identical dissection technique was used on all of the specimens to ensure standardisation.

Small metal pin heads were then used to mark the highest points of the medial epicondyle of the femur and the adductor tubercle. An additional point was then secured mid-way between these two points. This was a small metal eyelet mounted on a 1mm screw, large enough for a suture to run freely through (Figure 3.3). This mid-point was taken to define the centre of the femoral attachment of the MPFL and was consistently confirmed, by two researchers, during dissection for all knees as the anatomical MPFL femoral origin. This is in agreement with prior anatomical descriptions (Baldwin, 2009; Bicos et al., 2007).

Figure 3.2 The dissection technique used to remove skin and subcutaneous fat from specimens; the skin was removed carefully whilst ensuring the superficial fascial layer below was kept intact.

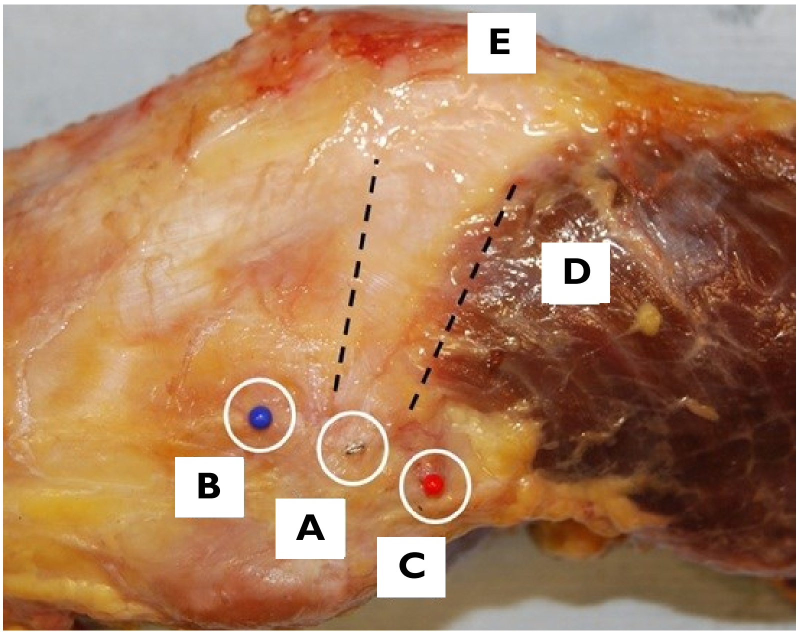

Figure 3.3 Location of the Medial Patellofemoral Ligament.
A: Eyelet marking the centre of the Medial Patellofemoral Ligament.
B: Pin marking the Medial Epicondyle of the Femur.
C: Pin marking the Adductor Tubercle.
D: Vastus Medialis Obliquus.
E: Patella.

A further four eyelets were then positioned 5mm proximal, distal, anterior and posterior to the central point; taken in reference to when the proximal-distal axis was oriented parallel to the line of the posterior femoral cortex (Figure 3.4). This reference axis was selected as it has previously been described in relation to the MPFL attachment and provided some standardisation for reference marking between the knees (Schöttle et al., 2007). Finally three further eyelets were used to mark the proximal half of the medial border of the patella proximally, centrally and distally (Figure 3.4).

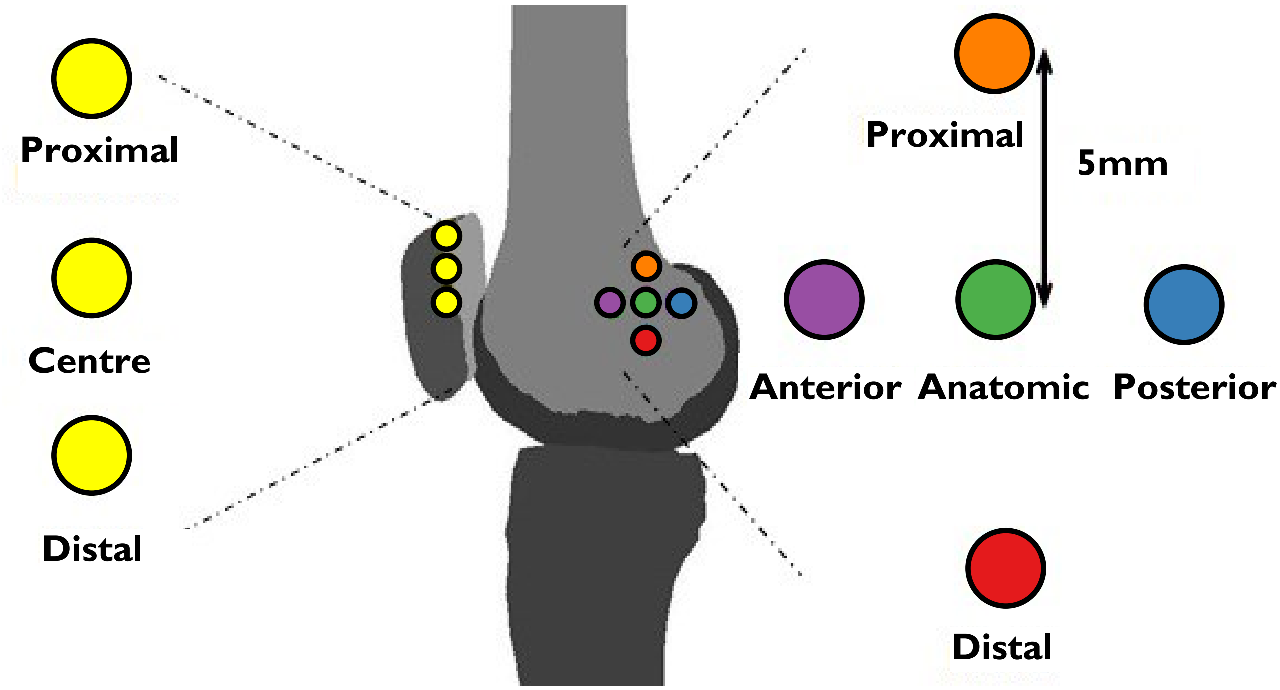

Figure 3.4 The selected sites for femoral tunnel position: Green = anatomical MPFL attachment as measured as the mid-point between the adductor tubercle and highest point of the medial epicondyle. Purple = 5mm anterior to the anatomical centre, Orange Circle = 5mm proximal to the anatomical centre, Red = 5mm distal to the anatomical centre and Blue = 5mm posterior to the anatomical centre, all oriented relative to the axis of the posterior femoral cortex. Patellar tunnel positions marked as: Proximal = at the mediosuperior border of the patella, Centre = central patellar attachment on the medial border and Distal = the mid patella level on the medial border.

The knees were prepared for X-rays, in order to define the anatomical position of the femoral attachment of the MPFL. The quadriceps muscle was stitched using a 2/0 monofilament suture (Ethilon 2/0, Ethicon Co., Somerville, NJ) and attached to a hole made in the anterior and proximal femoral bone (Figure 3.5). This applied a light tension to the patella, to simulate tension in the resting quadriceps muscle of specimens and to ensure standardised positioning of the patella in the trochlear groove during X-rays.

Figure 3.5 A suture sewn through the Rectus Femoris and Vastus Intermedius attaching to the proximal femur to provide some light resting tension in the muscle.

The knees were flexed 30˚ over a wooden block (Figure 3.6) and securely packaged in polyethene boxes, with a cross drawn in maker pen on the outside in line with the medial epicondyle of the femur to ensure consistency. This was necessary since the knee could not be removed from the transit box whilst in the clean hospital environment. Standardised lateral view X-rays were taken at a local hospital with the outlines of the posterior femoral condyles superimposed. This permitted examination of the femoral attachment points marked.

Figure 3.6 Knee positioned over a 30˚ wooden wedge to ensure standardisation of imaging.

Skyline view radiographs were also taken and used alongside the lateral x-rays to exclude any bony anatomy which could influence results (e.g. crossing-sign, severe osteoarthritis) (Dejour et al., 1994). Insall-Salvati ratio (Insall et al., 1972; Insall and Salvati, 1971) defining patellar height by the length of patella and length of the patellar tendon, and Sulcus Angle (Davies et al., 2000; Laurin et al., 1979) were calculated from the digital x-rays obtained for each specimen using Image J software as shown (Figure 3.7 and Figure 3.8).

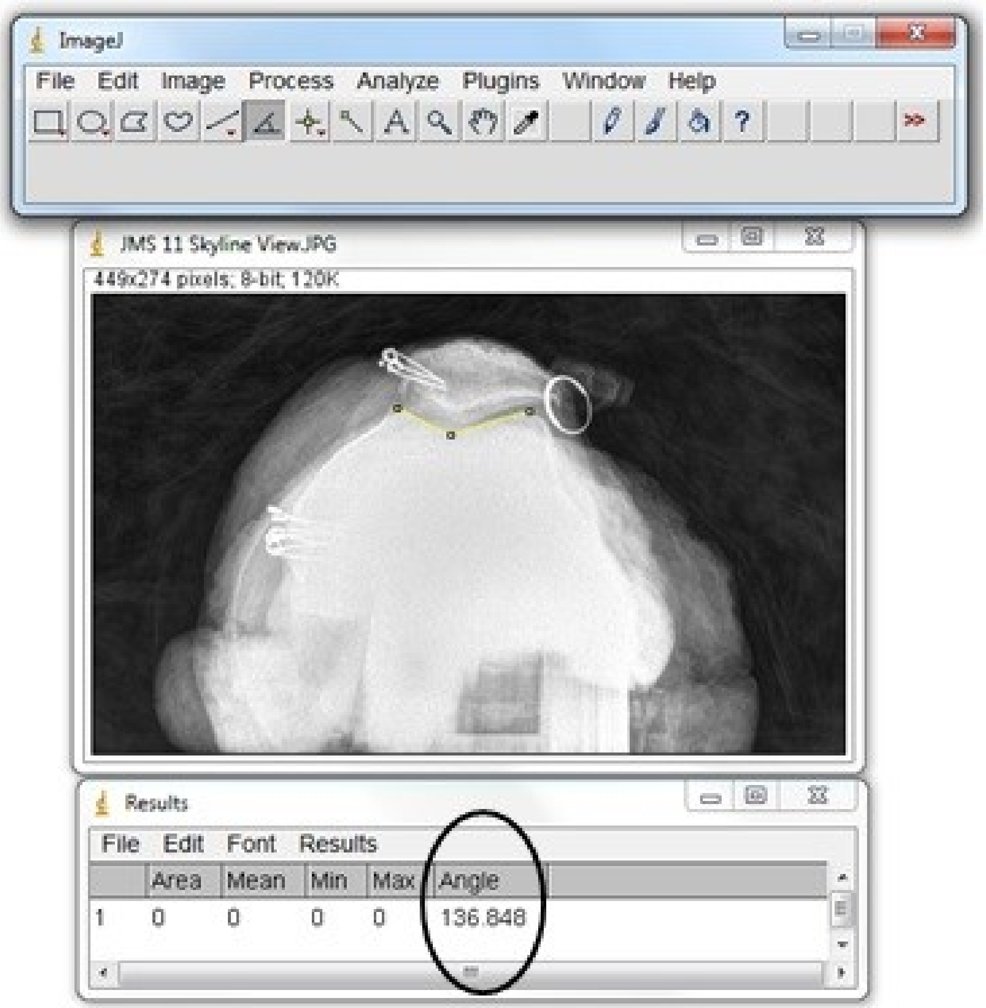

Figure 3.7 Measurement of the sulcus angle using Image J. This specimen was determined to have a sulcus angle of 137˚.

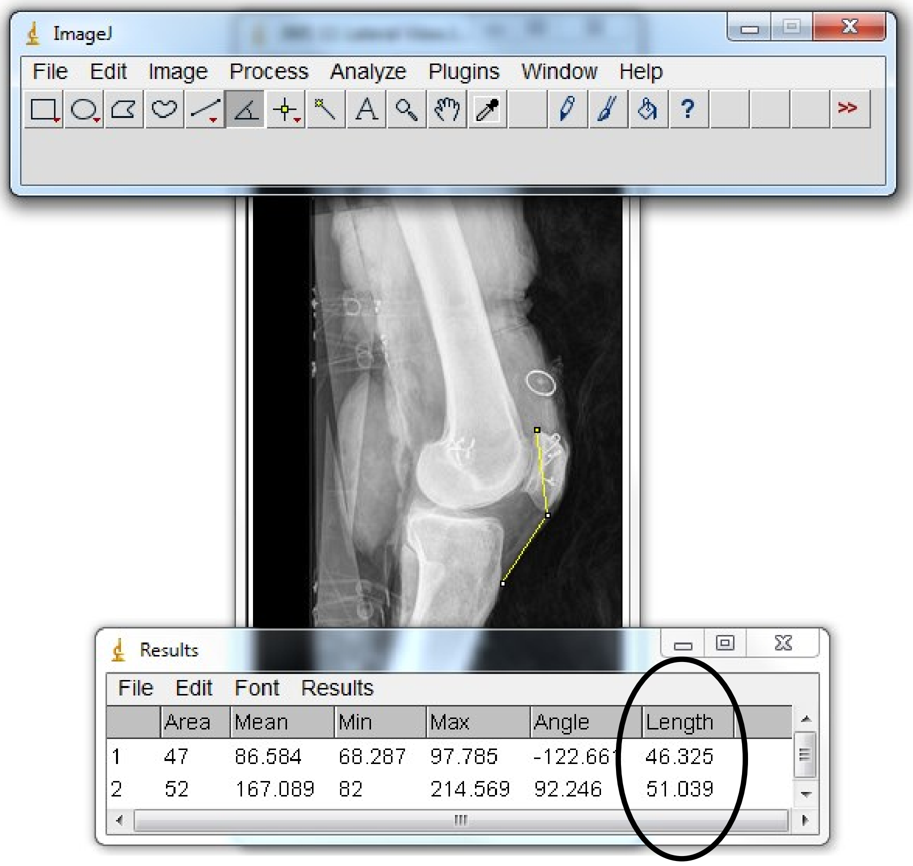

Figure 3.8 Measurement of the Insall-Salvati ratio using Image J. The black circle highlights the length of patella and the patellar tendon. This specimen was determined to have an Insall-Salvati ratio of 1.1.

Radiographic assessment of the patellar and femoral attachment points was achieved using the standardised lateral view x-rays superimposing the posterior femoral condyles. Anatomical reference lines as previously described (Redfern et al., 2010; Schöttle et al., 2007) were drawn on the radiographs to enable analysis (Figure 3.9): an extension of the posterior femoral cortex line, and two further anterior-posterior lines perpendicular to this: one at the most posterior part of Blumensaat's line and a second at the most proximal part of the posterior femoral condyle.

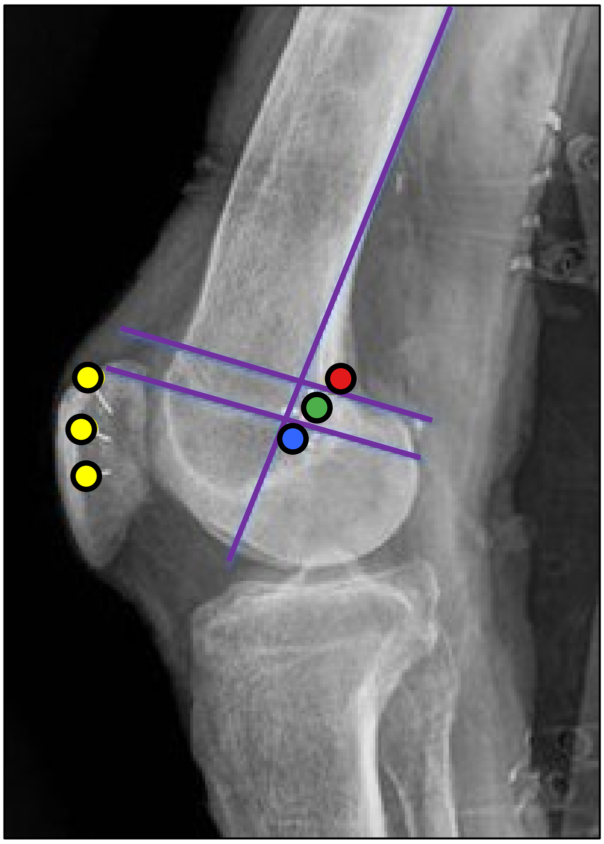

Figure 3.9 True lateral view x-ray of the femur with the posterior condyles aligned. Purple lines positioned as previously outlined (Schöttle et al., 2007). The green circle marks the centre of the anatomical MPFL attachment, the blue circle marks the medial epicondyle with the red circle marks the adductor tubercle. Three yellow circles mark the proximal, central and distal attachment sites on the medial border of the patella.

3.5.2 Specimen Loading

The knees were prepared to enable MPFL length changes to be recorded from 0º to 110º knee flexion. This was the movement permitted by the rig (Ghosh et al., 2009), and obtainable in all knees. Vastus Intermedius was elevated from the femur and the quadriceps separated into six components: Rectus Femoris (RF), Vastus Intermedius (VI), Vastus Lateralis Longus (VLL), Vastus Lateralis Obliquus (VLO), Vastus Medialis Longus (VML), and Vastus Medialis Obliquus (VMO) (Figure 3.10). Cloth material was securely stitched to the proximal end of each quadriceps muscle group and the Iliotibial Band (ITB) (Figure 3.10). This provided anchorage for the application of muscle load. The RF and VI muscles were grouped together to form a central muscle group. Distal tendinous fibres were left intact to ensure the physiological function of the quadriceps on the patella was maintained.

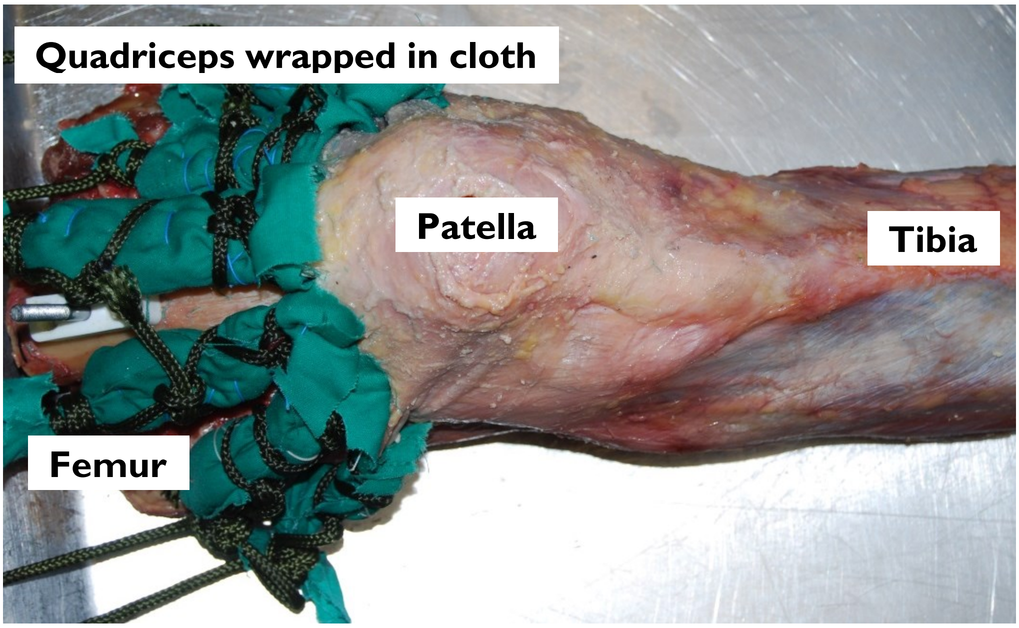

Figure 3.10 The quadriceps and iliotibial band dissected; each of the 6 components were wrapped and stitched in cloth with cord attached for loading the muscles in the test rig as shown.

Intramedullary rods were cemented into the femur and tibia. The knee was mounted in a testing rig using the femoral rod, with the anterior aspect uppermost (Figure 3.11). A total load of 175N was applied to the quadriceps and 30N to the ITB via a hanging weight and pulley system (Bull et al., 1999; Kwak et al., 2000). The ratio of quadriceps tensions were determined according to the mean physiological cross-sectional areas of the muscles: RF+VI 53%, VLL 33%, VLO 9%, VML 14%, and VMO 9% (Farahmand et al., 1998a; Senavongse and Amis, 2005). The muscle tensions were applied in their physiological directions in relation to the femoral axis (Farahmand et al., 1998a). The muscle tensions extended the knee against a restraining bar which controlled extension and allowed measurements to be taken at 10º intervals (Figure 3.11). This load is representative of an unloaded, open kinetic chain leg extension. Higher load levels were not used to avoid damaging the soft tissues across the large number of tests performed on each knee.

Prior to testing, ten loaded knee conditioning cycles were performed to avoid soft tissue hysteresis and settle the zero readings. The knee was flexed by pushing posteriorly against the tibial intramedullary rod to the count of five from 0º to 110º flexion and released to the count of five to full extension.

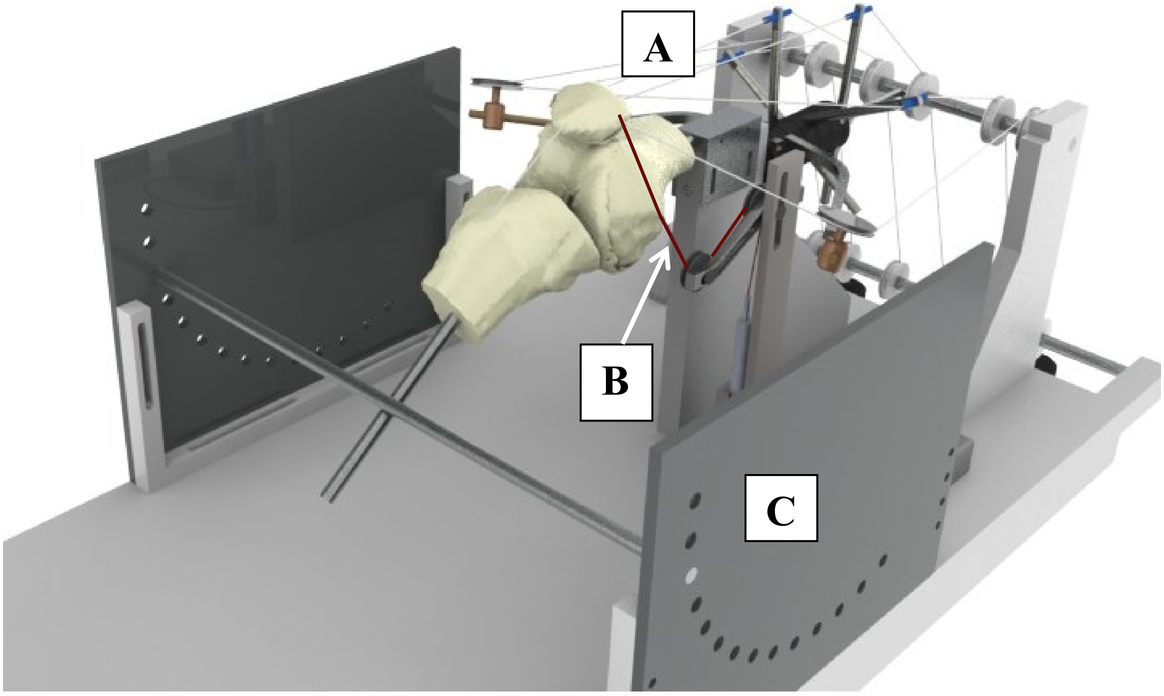

Figure 3.11 Knee kinematics test rig.

A: The muscle tensioning and weight system.

B: The suture used to measure the length changes between the patella and femur.

C: The slot and rest bar system for controlling the knee flexion angle.

3.5.3 Measurement of Ligament Length Change

MPFL length changes between different eyelet positions were measured using monofilament sutures (Ethilon 2/0, Ethicon Co., Somerville, NJ) attached to a linear variable displacement transducer (LVDTs – Solartron Metrology, Bognor Regis, UK) (Ghosh et al., 2009) (Figure 3.12). Because the suture slid freely between the tissues, it measured the changes in length between the eyelets at the femoral and patellar attachments. Thus, the suture itself was not stretched as it was tensed only by the weight of the sliding core of the LVDT (0.5N). The LVDT was confirmed by micrometer to be accurate to ± 0.01mm prior to testing. Length changes were recorded at 10º increments using Solartron 'Orbit' Excel software. The superficial fascial layer of the medial retinaculum was carefully separated from the second layer to enable the free movement of sutures between the layers. A suture was attached to each of the proximal, central and distal eyelets on the patella. Each patellar suture in succession was then passed between the layers of the medial retinaculum, and through a randomly selected femoral eyelet before being attached to the LVDT via a hook and pulley system to record the length changes (Figure 3.12). This was then repeated for the remaining four femoral eyelets, before the remaining two patellar sutures were assessed in a similar manner. Measurements were repeated for all 15 combinations of the 3 patellar eyelets with the 5 femoral eyelets. Ligament length changes were measured three times with the average from each combination used for analysis. This system had a mean test re-test difference of between 0mm±0mm – 1.4mm±0.5mm (mean ±standard deviation), with ICC of between 0.89 – 0.98 for all flexion angles (Section 3.8.3.4).

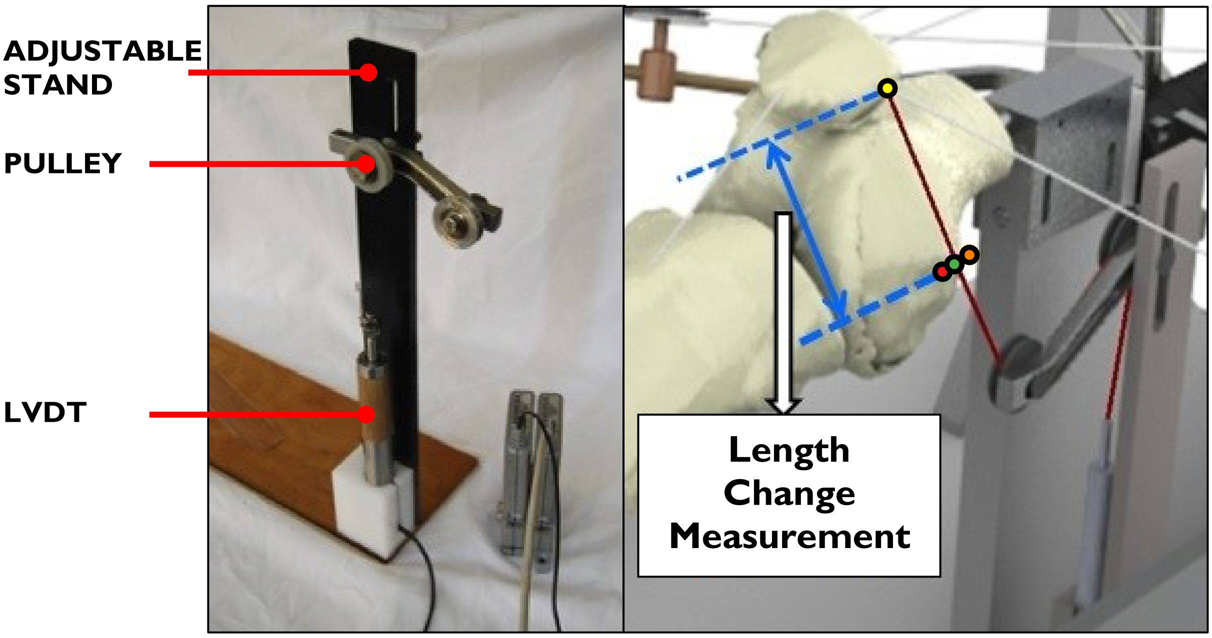

Figure 3.12 Left side: linear variable displacement transducer in a custom made stand, with pulley system. Right side: showing the length change measurement recorded for each combination of patellar and femoral attachments.

In order to determine the test-retest reliability of the ligament length change measurement method, the protocol was repeated two days later, using the central patellar and femoral points. This length change data was analysed versus the earlier data to determine the test-retest reliability using a Fishers Original ICC performed in SPSS (IBM Corp. Version 20.0. Armonk, NY, 2011).

3.5.4 Measurement of MPFL attachment position

The lateral view radiographs were used to provide a clinically relevant method for describing the MPFL attachments, and the radiographic attachment sites were correlated with photographic data to avoid magnification errors. All soft tissue was removed, the knees were clamped in a photographic stand, and a true lateral view photograph taken with the knee in full extension, using a spirit level to ensure true alignment to the camera axis and superimposition of the posterior and distal condyles (Figure 3.13).

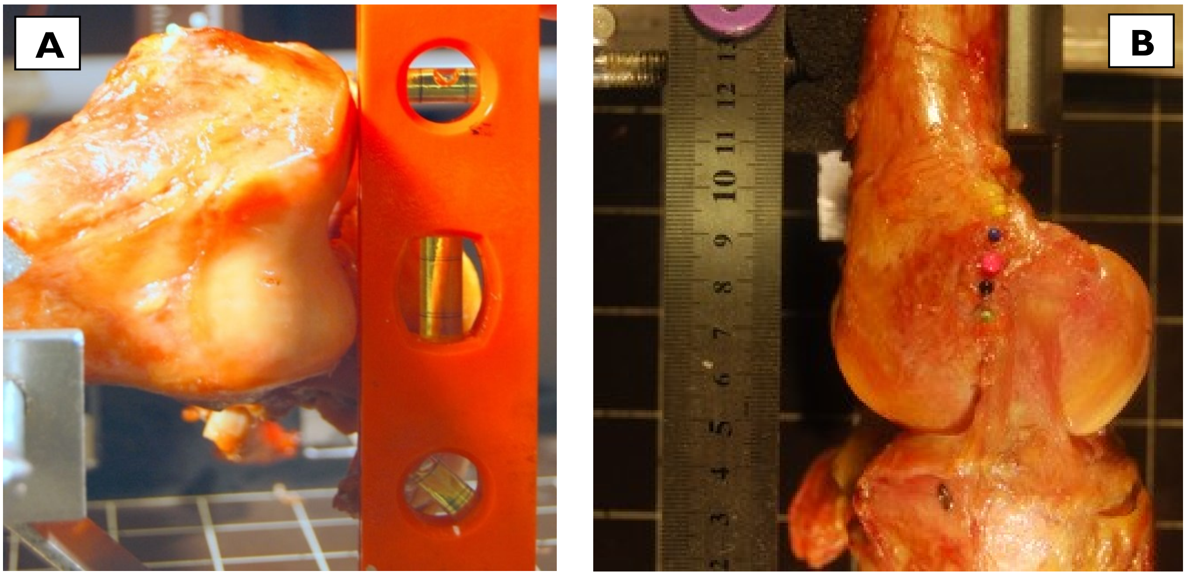

Figure 3.13 Spirit level measurement to ensure alignment of the distal femoral condyles (A), this was also assessed for the posterior femoral condyles before the femur was clamped in the photograph stand alongside a ruler (B).

A ruler in the photographs allowed correction of magnification and ImageJ (Maryland, USA) was used to make photographic measurements. ImageJ has previously been found to demonstrate high reliability to 0.01mm (Abramoff et al., 2004). The photographic A-P condylar diameter was defined as 100% and used to normalise measurements on the medial femoral condyle taken from the lateral view X-rays. This was to enable standardised radiographic templating measurements to be defined, applicable across the range of knee sizes. 2mm was subtracted from the photographic diameter value as this has previously been defined as the thickness of articular cartilage which is visible on the photograph but not on radiographs (Cohen et al., 1999). Measurements of the femoral diaphyseal diameter and the distance of the MPFL attachment from the anterior, posterior and distal femoral edges were taken from each radiograph to determine the normalised femoral MPFL attachment. Additional measurements were undertaken to measure the position of the MPFL attachment from the landmarks previously described by Schottle et al (2007). The proximal-distal line of Schottle et al is based on the posterior cortical outline of the femur. That can lead to inconsistency, because the outline is curved, this is shown in Figure 3.14. The proximal-distal position can be seen to remain level throughout, however the anterior-posterior position of the femoral attachment can be positioned either anterior, posterior or directly on the extension line of the femoral cortex, depending on where this line is drawn. Therefore the present study defined a fixed point along the femur from which the straight line could be extended; this was at one A-P condyle diameter from the distal end of the femur (Figure 3.15).

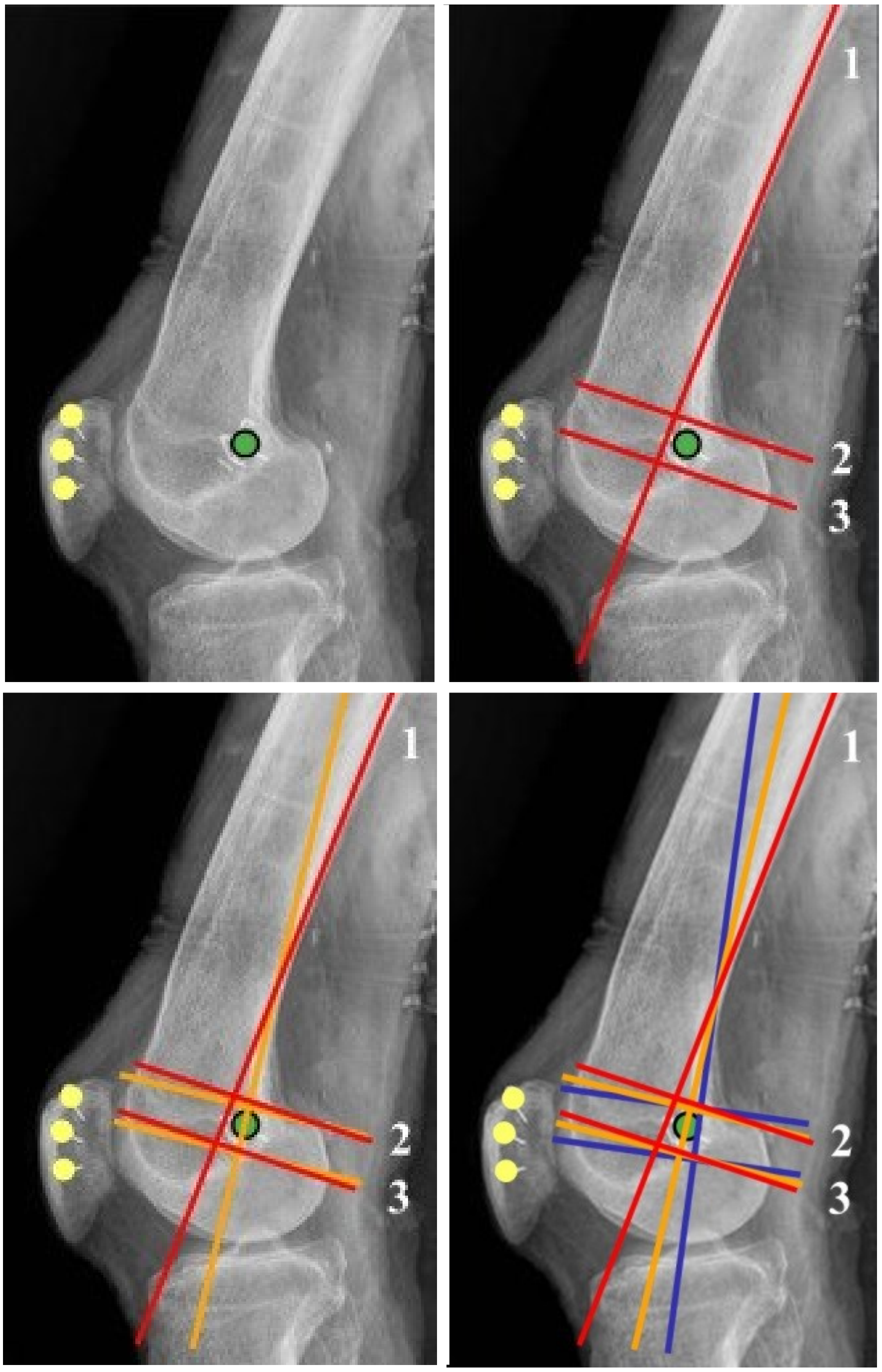

Figure 3.14 Schottle's method for determining the radiographic femoral MPFL origin (Schöttle et al., 2007). In all 8 specimens the anatomical MPFL eyelet was located between the two proximal-distal lines 2 and 3 (*top right image*), but can be seen to be; posterior (*top right image (red line)*), on (*bottom left image (orange line)*) or anterior (*bottom right image (blue line)*) to the extended line of the posterior femoral cortex (1), depending on where the line is extended from.

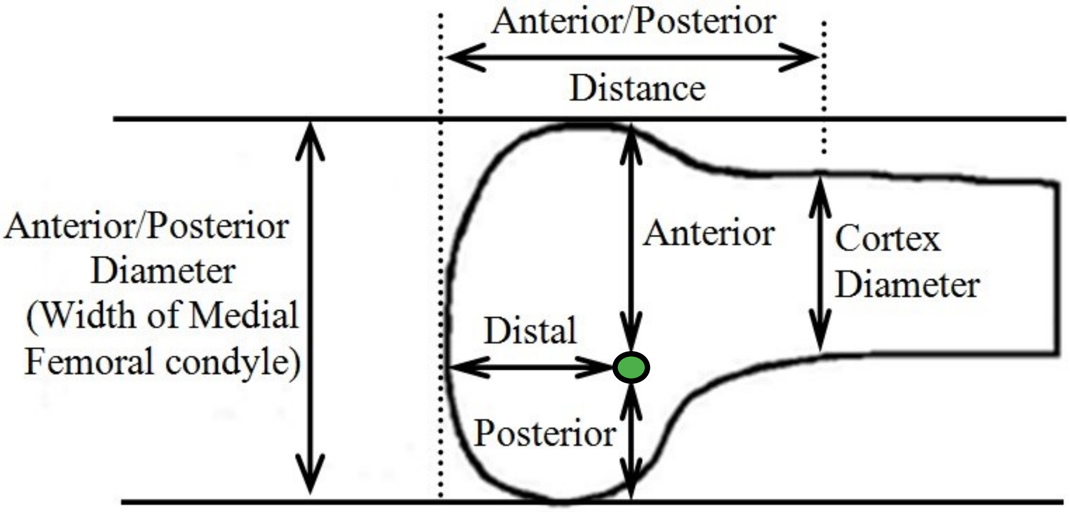

Figure 3.15 Measurements taken of distal medial femoral geometry (mm). The level of the femoral cortex diameter was standardised in reference to the A-P diameter of the medial femoral condyle as shown.

3.6 Analysis

Repeated-measures two-way ANOVA with Bonferonni post-hoc analysis were used to examine differences between length changes using the different patellar and femoral tunnel combinations, across the range of knee flexion. Significance was set at $\alpha = 0.05$ and analysis was performed in SPSS version 20. A Shapiro-Wilk test confirmed that the data sets were normally distributed. Femoral dimensions and distances were analysed for any relationships using a Pearson Correlation performed in SPSS with significance set at $P < 0.05$.

3.7 Results

All specimens were found to be within normal limits for both sulcus angle (range:134˚-142˚) and Insall-Salvati ratio (range: 0.9-1.1).

3.7.1 MPFL Length Change Patterns

The mean length change pattern of the MPFL using the central femoral and patellar points (PCent and FCent) was close to isometric, with a slackening of 2.1 ± 2.3mm from 0-40° knee flexion (Figure 3.16).

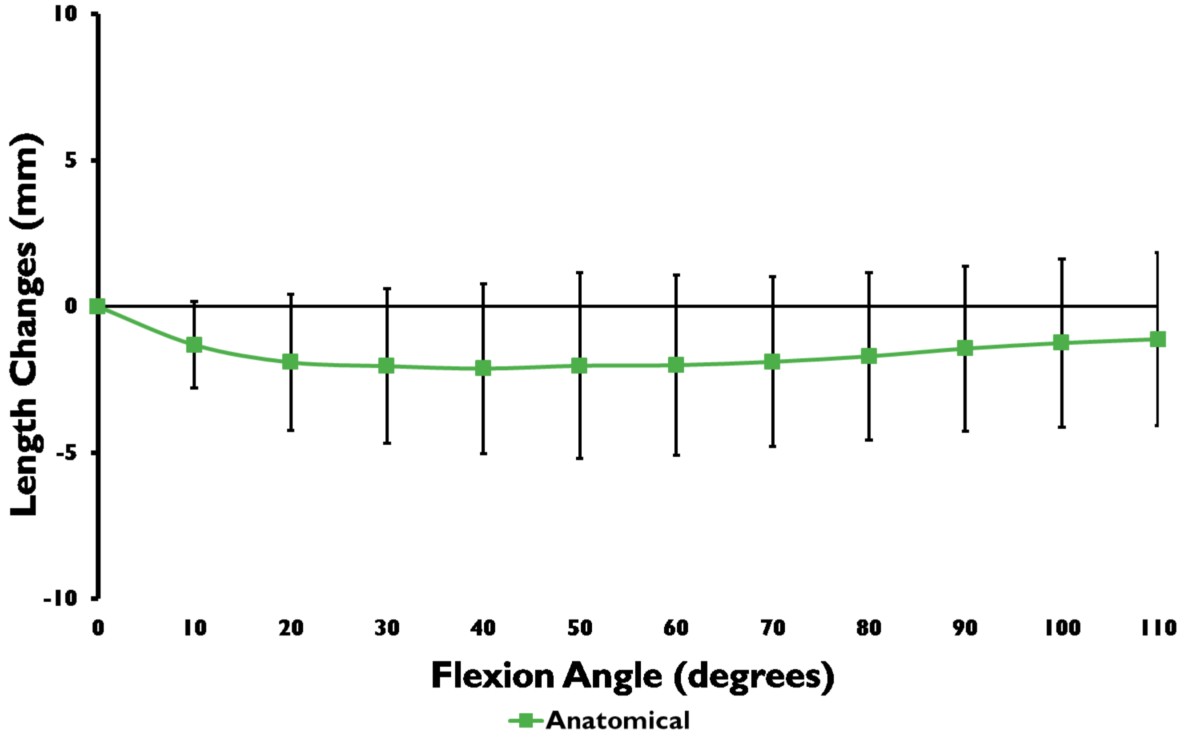

Figure 3.16 Length changes of the MPFL (mm) versus knee flexion (°) for the native MPFL (from central attachment points of the patella and femur). The negative length change means that the bone attachments approached each other, implying relative slackening of the MPFL.

There was a significant effect of femoral eyelet position on ligament length changes ($P<0.001$), and a significant but smaller effect of patellar eyelet position on ligament length changes ($P=0.016$). In addition a significant interaction was identified between flexion angle and femoral tunnel position ($P<0.001$) and flexion angle and patellar tunnel position ($P<0.001$) (Table 3.1a-c).

Table 3.1a-c MPFL length changes for combinations of femoral and patellar attachments through knee flexion range (mean and standard deviation (SD), n=8).

Flexion Angle (Degrees)	MPFL length changes with CENTRAL patella tunnel and different femoral Tunnel Positions (mm)									
	Centre (mm)		Distal (mm)		Proximal (mm)		Anterior (mm)		Posterior (mm)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
10	-1.3	1.5	-2.0	2.3	-0.5	1.3	-1.4	1.6	-1.6	1.9
20	-1.9	2.3	-3.4	3.3	-0.4	2.1	-2.0	2.6	-2.6	2.8
30	-2.0	2.7	-4.4	3.9	0.3	2.7	-1.9	3.1	-3.0	3.3
40	-2.1	2.9	-5.3	4.2	1.0	3.4	-1.7	3.5	-3.3	3.7
50	-2.0	3.2	-6.1	4.4	1.7	4.1	-1.4	3.8	-3.4	3.9
60	-2.0	3.1	-7.0	4.3	2.2	4.5	-1.3	3.9	-3.6	4.0
70	-1.9	2.9	-7.6	4.2	2.8	4.7	-0.9	4.2	-3.7	3.9
80	-1.7	2.9	-8.1	4.2	3.3	4.8	-0.5	4.5	-3.8	4.0
90	-1.4	2.8	-8.5	4.4	3.8	4.8	-0.1	4.9	-3.8	4.2
100	-1.3	2.9	-8.8	4.6	4.2	4.8	0.2	5.1	-3.8	4.2
110	-1.1	3.0	-9.1	4.9	4.5	5.0	0.4	5.5	-3.8	4.4

Flexion Angle (Degrees)	MPFL length changes with PROXIMAL patella tunnel and different femoral Tunnel Positions (mm)									
	Centre (mm)		Distal (mm)		Proximal (mm)		Anterior (mm)		Posterior (mm)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
10	-1.3	1.3	-2.2	1.3	-0.5	1.3	0.4	1.7	-1.4	1.5
20	-2.0	2.0	-3.9	2.1	-0.4	2.1	0.7	2.9	-2.4	2.4
30	-2.5	2.2	-5.3	2.4	0.3	2.7	0.8	3.9	-3.0	2.7
40	-2.6	2.5	-6.2	2.2	1.0	3.4	1.1	4.7	-3.4	2.7
50	-2.5	3.0	-7.1	2.1	1.7	4.1	1.4	5.6	-3.6	2.6
60	-2.6	3.2	-7.8	2.1	2.2	4.5	1.7	6.4	-3.9	2.6
70	-2.4	3.4	-8.3	2.0	2.8	4.7	2.0	7.1	-4.0	2.6
80	-2.2	3.5	-8.5	2.0	3.3	4.8	2.3	7.8	-4.1	2.5
90	-1.9	3.7	-8.7	2.0	3.8	4.8	2.7	8.5	-4.0	2.6
100	-1.5	3.7	-8.6	2.3	4.2	4.8	3.2	9.2	-3.8	2.5
110	-1.3	3.8	-8.6	2.6	4.5	5.0	3.7	9.9	-3.5	2.4

Flexion Angle (Degrees)	MPFL length changes with DISTAL patella tunnel and different femoral Tunnel Positions (mm)									
	Centre (mm)		Distal (mm)		Proximal (mm)		Anterior (mm)		Posterior (mm)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0	0.0
10	-0.7	1.7	-1.8	1.8	-0.1	1.8	-0.6	1.8	-1.0	1.7
20	-0.8	2.7	-2.8	2.7	0.6	2.5	-0.2	2.5	-1.3	2.4
30	-0.5	3.0	-3.4	3.0	1.6	3.0	0.3	2.9	-1.3	2.6
40	-0.2	3.4	-3.8	3.0	2.4	3.4	0.8	3.4	-1.3	2.7
50	0.0	3.8	-4.2	3.2	3.2	3.5	1.3	3.6	-1.3	2.8
60	0.1	4.1	-4.7	3.3	3.9	4.0	1.7	3.8	-1.5	2.8
70	0.1	4.4	-5.2	3.5	4.5	4.4	2.2	3.9	-1.8	2.9
80	0.3	4.6	-5.4	3.9	5.1	4.7	2.5	4.1	-2.1	2.9
90	0.3	4.8	-5.6	4.2	5.6	5.0	2.9	4.1	-2.2	3.1
100	0.5	4.9	-5.8	4.7	6.0	5.2	3.2	4.3	-2.3	3.3
110	0.5	5.0	-5.9	5.1	6.4	5.4	3.6	4.4	-2.4	3.5

Moving the femoral attachment point proximal-distal had a much larger effect on MPFL isometry than did anterior-posterior movements. Post-hoc tests revealed that, compared to the central eyelet, the eyelet positioned 5mm proximal caused a significant increase in MPFL length with knee flexion (P=0.005) and an eyelet 5mm distal resulted in significantly reduced length with knee flexion (P=0.001). An eyelet positioned 5mm anteriorly (P=0.794) or posteriorly (P=0.121) did not result in significant differences being found among the MPFL length change patterns in relation to the central attachment.

In relation to a specific femoral attachment, changing the patellar attachment site had a small but statistically significant effect (P=0.009) on the length change pattern. In relation to the distal patellar attachment, the proximal patellar attachment caused greater MPFL slackening in knee flexion (Figure 3.17a-c).

Figure 3.17a

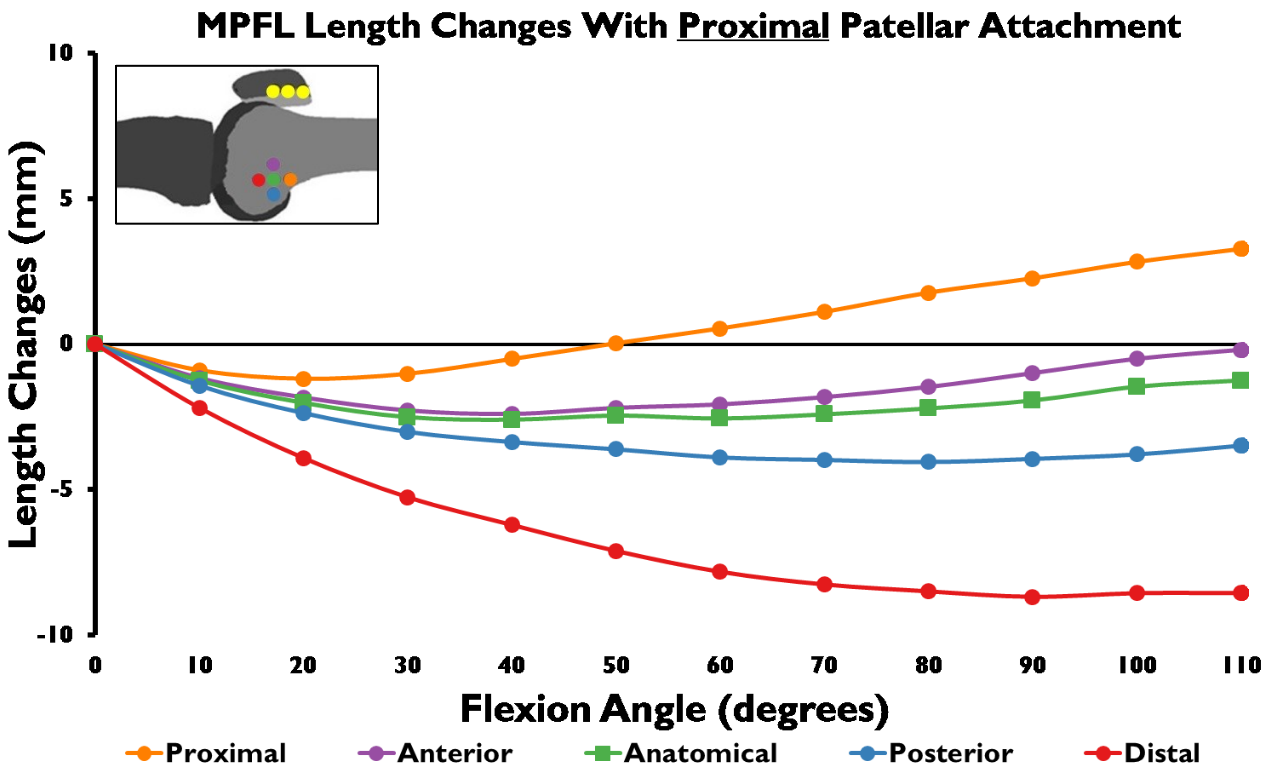

Figure 3.17b

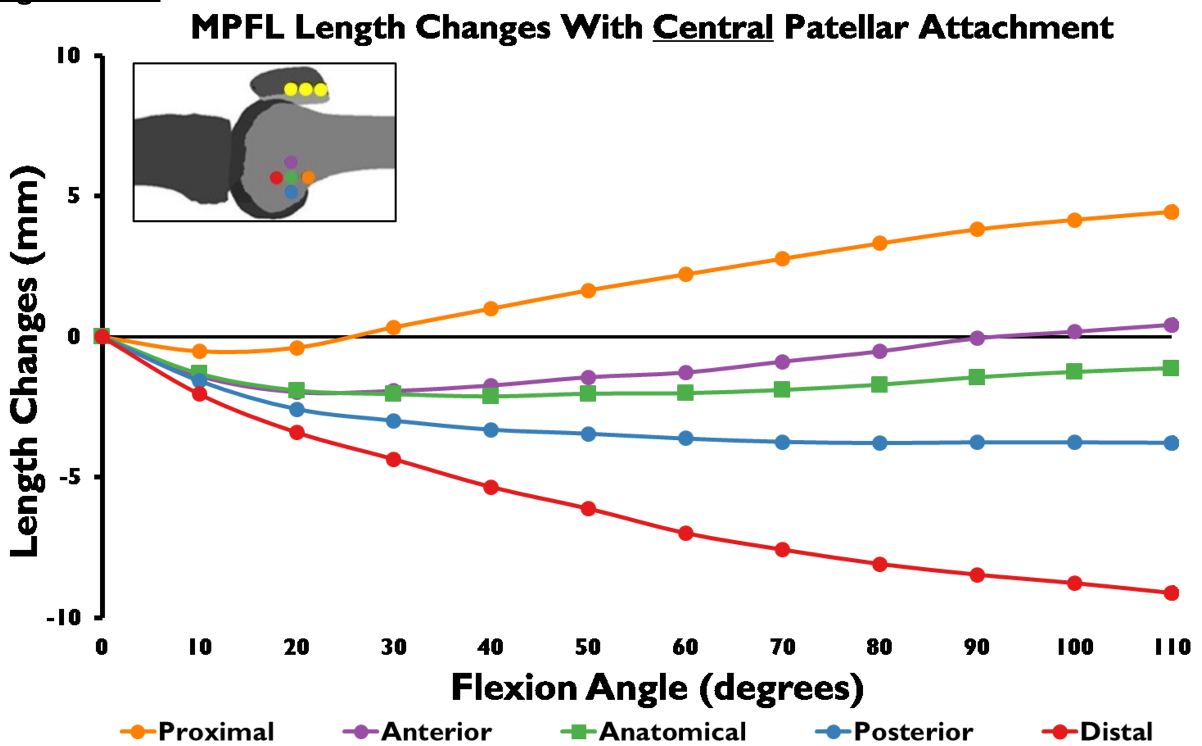

Figure 3.17c

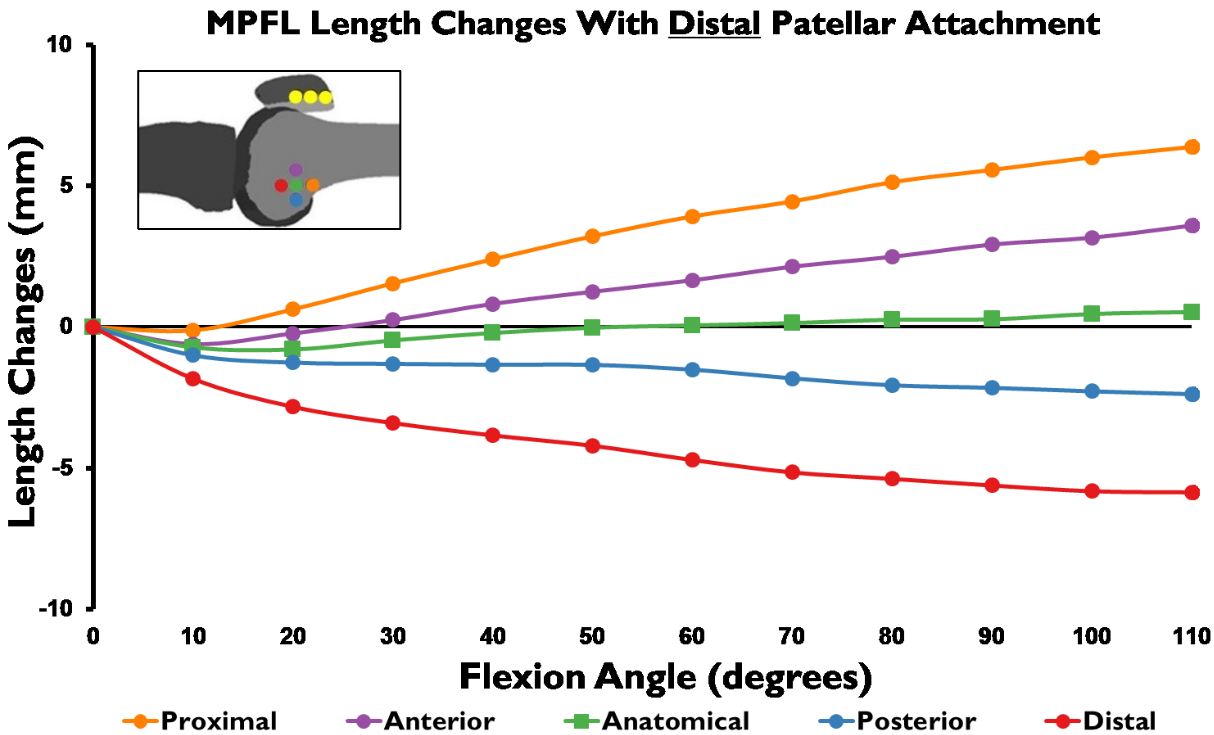

Figure 3.17a-c: Length changes in the MPFL (mm) versus knee flexion (°) with 5 different femoral tunnel positions (central, distal, proximal, anterior and posterior). 3.16a – depicts the proximal patellar attachment, 3.16b – depicts the central patellar attachment and 3.16c depicts the distal patellar attachment (Mean values; n=8. See Tables 1a-c for SD). The most isometric configuration was the central femoral point in relation to the most distal patellar point (Figure 3.17c).

Test-retest analysis of ligament length changes of the central femoral and central patellar points revealed high reliability, with a mean difference of 0mm – 1.4mm, with ICC of between 0.89 – 0.98 for all flexion angles (Section 3.9.3.4).

3.7.2 Femoral MPFL attachment site

The dissections confirmed that the centre of the femoral attachment of the MPFL was always located equidistant along the line from the medial epicondyle to the adductor tubercle, at the base of the groove between the bony prominences (Figure 3.18), as defined in the Method.

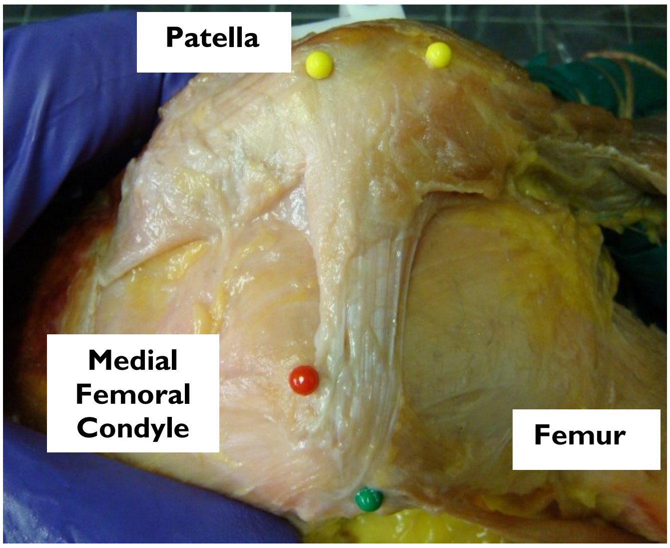

Figure 3.18 The MPFL dissected with pins marking the medial epicondyle (red pin), adductor tubercle (green pin) and MPFL attachments to the patella (yellow pins).

Proximal-Distal Positioning: In all 8 specimens the Central MPFL eyelet was located radiographically between the two previously-defined A-P lines (Figure 3.14). The two anterior-posterior lines were 5.8 ± 0.3 mm apart. The central eyelet was 3.2mm ± 0.7mm proximal to the line through the most posterior part of Blumensaat's line.

Anterior-Posterior Positioning: In relation to a line drawn as an extension to the posterior femoral cortex (one A-P condyle diameter from the distal end of the femur), the central MPFL attachment was 1.3 ± 0.7mm (Mean ± SD, range 0-2 mm) posterior to the extended posterior cortex line.

3.7.3 Anatomical Landmark Distances

The femoral dimensions and MPFL attachment sites are shown in Table 3.2. There was a significant correlation between the A-P diameters of the medial condyle and the femoral cortex (r=0.769: *P* = 0.026). The MPFL attachment site was significantly related to the femoral condyle diameter anteriorly (r=0.794: *P*=0.019), posteriorly (r=0.935: *P*=0.001) and distally (r=0.987: *P*<0.01). Additional relationships were identified between the MPFL attachment distally and anteriorly (r=0.867: *P*=0.005) and distally and posteriorly (r=0.874: *P*=0.004).

Table 3.2 Mean, Standard Deviation and Range of Medial Condyle Dimensions (mm; n=8).

Measure (mm)	MEAN	SD	RANGE
Medial Femoral Condyle A-P Diameter	**62**	**5.2**	**55 – 69**
Femoral Cortex Diameter	**34**	**5.2**	**29 – 43**
MPFL insertion to Posterior Femoral Condyle	**25**	**2.2**	**23 – 29**
MPFL insertion to Anterior Femoral Condyle	**37**	**3.7**	**32 – 41**
MPFL insertion to Distal Femoral Condyle	**31**	**3.2**	**27 – 36**

The normalised radiographic position of the centre of the femoral attachment of the MPFL was 41% ± 2 (37-43) (mean ± SD; range) from the posterior border of the medial condyle, 51% ± 1 (49-52) from the distal border, and 59% ± 2 (57-63) from the anterior border (Table 3.2). This radiographic point was 0 mm ± 0.7mm in the proximal / distal and 0.05mm ± 1.7mm anterior / posterior to the anatomical definition of the centre of the MPFL attachment in the photographs.

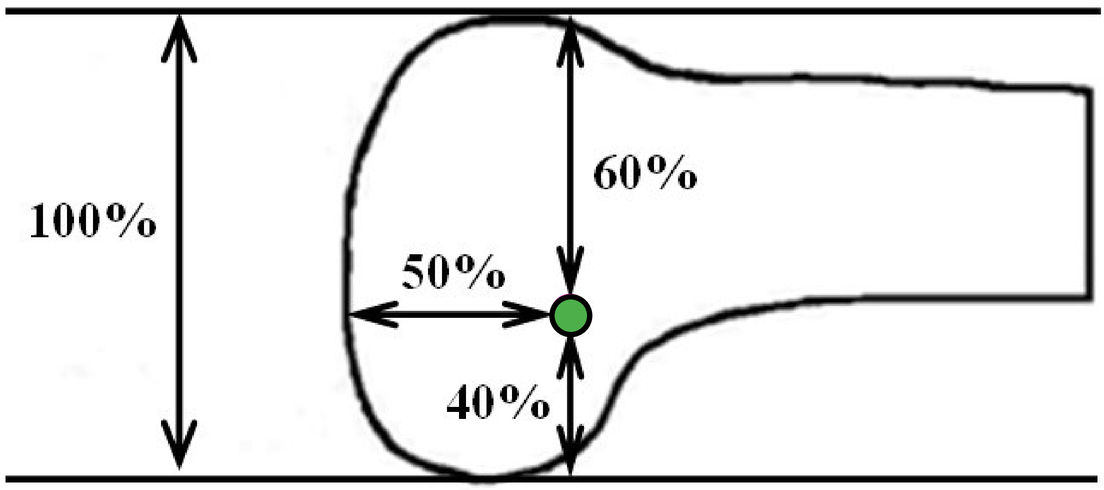

Figure 3.19 The MPFL attachment is defined in relation to the size of the medial femoral condyle: if A-P size was 100%, then the MPFL attachment was: 40% from the posterior, 50% from the distal and 60% from the anterior outline.

3.8 Discussion

This study has defined the most isometric position for patellar and femoral tunnel placement for MPFL reconstruction, discovered a simple normalising rule to define the femoral attachment position radiographically, and also shown the effects of straying away from those attachment points. The most-isometric points corresponded to the centre of the anatomical attachment on the femur and the centre/distal attachment to the patella. It was clear that malpositioning the femoral tunnel in the proximal-distal axis had much larger effects than moving it in the anterior-posterior axis, and that moving the attachment on the patella had a smaller effect.

The native MPFL was almost isometric through 0°-110° knee flexion, changing only a mean distance of 1.5 - 2mm depending on the patellar attachment examined. However, positioning an MPFL femoral attachment 5mm proximal or distal from its anatomical position resulted in a significantly non-isometric length change pattern. A proximal femoral attachment led to MPFL elongation with knee flexion, and slackening with extension; the opposite occurred with the distal femoral attachment: elongation with knee extension, and slackening in flexion (Figure 3.16). This was most notable when proximal and distal femoral attachments were combined with the proximal and central patellar points. The MPFL was found to be most isometric between the central femoral and most distal patellar attachments, with a mean change of only 1.1mm in length from 0°-90° knee flexion, in agreement with Steensen et al. (2004). This has been related to the cam formation of the medial femoral condyle (Farr and Schepsis, 2006). Moving from proximal to distal patellar attachments altered MPFL length changes by up to 2.5mm, meaning the patellar attachment is much less sensitive than the femoral, where moving 5mm proximal and distal changed the MPFL length measurements typically by 12mm. Noting the strength of MPFL reconstructions (Mountney et al., 2005), this implies high graft tension and elevated medial patellofemoral joint contact pressures. Non-isometric graft positioning may cause restricted patellar movement, loss of knee joint range of motion, increased medial patellofemoral joint contact pressures, graft failure or recurrent dislocation (Amis and Jakob, 1998; Bicos et al., 2007; Camp et al., 2010; Elias and Cosgarea, 2006; Tateishi et al., 2011). The native MPFL has previously been identified as “close to isometric” (Ghosh et al., 2009; Smirk and Morris, 2003), with the length changes most sensitive to the femoral attachment, suggesting its importance in outcome following reconstruction (Bicos et al., 2007). This is similar to literature relating to ACL reconstruction, where correct femoral tunnel positioning is paramount to successful outcome (Amis and Jakob, 1998). A non-anatomically positioned MPFL repair femoral attachment identified radiographically (Schöttle et al., 2007) was found to be the only statistically significant risk factor for unsuccessful surgery: 80% of patients with incorrectly positioned suture anchors suffered dislocation up to 4 years post-MPFL repair (Camp et al., 2010). If the graft is fixed with the knee in extension, then a proximal femoral tunnel would lead to increasing graft tension as the knee flexes. Conversely, if the graft is secured with the knee flexed a distally placed tunnel would lead to graft tension increasing when the knee extends. During the tensioning process the surgeon should ensure, as with distal ACL graft fixation, that the knee can be passively and comfortably brought to full extension.

The present study has defined the femoral attachment of the MPFL in relation to the size of the medial femoral condyle: if A-P size was 100%, then the MPFL attachment was 40% from the posterior, 50% from the distal and 60% from the anterior outline. This simple 40-50-60 percent rule may be used in radiographic templating when checking graft tunnel location. This point has been confirmed by later work examining the femoral MPFL insertion point using computer simulation (Oka et al., 2013). The current study confirmed that the proximal-distal position of the anatomical MPFL femoral attachment lies between the horizontal line through the most posterior part of Blumensaat’s line and that tangential to the most superior part of the posterior femoral condyle, as described by Schottle (2007) Accurate identification of this level appears paramount to securing a near isometric MPFL graft.

Anatomical studies have described the femoral attachment of the MPFL in relation to the medial epicondyle, medial collateral ligament and adductor tubercle, indeed these terms are used interchangeably in some reports (Panagiotopoulos et al., 2006; Tuxøe et al., 2002). More recently the adductor tubercle has been found to provide sole attachment for the adductor magnus tendon, with the medial epicondyle providing exclusive attachment for the medial collateral ligament, and between these two landmarks a palpable groove has been described, in which the MPFL has been defined to attach (Baldwin, 2009; Nomura and Inoue, 2006; Panagiotopoulos et al., 2006; Steensen et al., 2004; Steiner et al., 2006). This location is gaining acceptance as the femoral tunnel position for anatomical MPFL reconstruction (Philippot et al., 2009); it corresponds to the central attachment point defined in this study, with the closest behaviour to isometry of the MPFL. However, there remain conflicting findings. Phillippot et al (2009) reported that the attachment was 10mm posterior to the medial epicondyle and 10mm distal to the adductor tubercle, while LaPrade et al (2007) found it 2mm anterior and 4mm distal to the adductor tubercle. However, fixed measurements take no account of the range of sizes of knees: Philippot et al (2009) found that the medial epicondyle was 15mm to 25mm from the adductor tubercle. The radiographic guidelines from Schottle et al (2007) give a 5mm proximal – distal band for acceptable positions, but Redfern et al (2010), seeking to confirm Schottle's point, found differences of 5mm in the anterior-posterior and 7mm proximal-distal around the actual attachments. This study and others (Redfern et al., 2010; Tateishi et al., 2011) found the anatomical point was located posterior to an extension of the line of the posterior femoral cortex, as opposed to previous descriptions which placed it anterior to this line (Schöttle et al., 2007; Servien et al., 2011). The A-P positions differed because of variability in extending a straight datum line from the curved outline of the posterior femoral cortex. A fixed point along the cortex should be used, as in this study, for improved consistency. However, bony architecture varies as a consequence of weight bearing activity undertaken by individuals (Vainionpää et al., 2007). Therefore, the posterior femoral cortex may not represent a consistent anatomical landmark reliable for use in determination of the femoral tunnel. Thus, noting the limitations of prior methods, the use of normalised dimensions, such as the 40-50-60% of the A-P diameter rule found in this study, may be more appropriate, and was found to be very precise with a mean error of ±0.7mm in the proximal-distal direction.

It is well-known from previous work on the ACL that very small changes in femoral attachment have a large effect on the length-change pattern (Zavras et al., 2005), and this has proven to also be true for the MPFL. A change of only 5mm proximal-distal caused significant tightening or slackening with knee flexion. These findings demonstrate that the surgeon must be accurate in choosing the proximal-distal location of the femoral graft tunnel. Whilst this paper does not address reconstruction biomechanics, the accurate data relating to specific points of attachment would be expected to further expand our understanding of overall graft behaviour, with other fibers slackening or tightening relative to the central fibre. The key findings of this work allow the surgeon to peroperatively validate through the use of reliable anatomical landmarks, and radiographically, via the 40-50-60 percent rule, the most-isometric tunnel placement for MPFL reconstruction. It is hoped that this will be incorporated within the surgeons' armamentarium. Clinical studies are now required to confirm the clinical efficacy of this rationale.

3.9 Key Findings / Conclusion

Key findings from this work therefore are:

1. The femoral attachment of the MPFL was between the medial epicondyle and the adductor tubercle.
2. The central fibres of the MPFL were close to isometric.
3. A change of the femoral attachment of only 5mm in proximal or distal direction caused significant tightening or slackening with knee flexion, highlighting the need to be accurate in choosing the proximal-distal location of the femoral graft tunnel. Both anterior- posterior femoral and patellar positioning were not as sensitive.
4. A 40% / 50% / 60% rule has been proposed for use to aid in femoral attachment identification.

3.10 Addendum

3.10.1 Rig Set-Up

3.10.1.1 Muscle Loading

In an ideal environment it would be possible to simulate patellofemoral joint behaviour *in vitro* to directly replicate that present *in vivo*. However there are many restrictions to this, not least that measurement of patellofemoral joint muscle activity, kinematics and contact mechanics *in vivo* is fraught with challenges. It is vital to clearly define and standardise the variables surrounding muscle loading at the outset of cadaveric testing to ensure accurate interpretation of study outcomes. Therefore in the current study specific consideration was given to the following variables:

- Loading condition: Passive (unloaded) studies (typically examining knee range of motion and the contribution of soft tissue) versus loaded studies (Blankevoort et al., 1988; Goh et al., 1995). Open chain (e.g. seated knee extension with the foot off the ground) versus closed chain (e.g. a squat movement with the feet on the ground) motion (Escamilla et al., 1998).
- Type of muscle contraction: eccentric, concentric, isometric (Madeleine et al., 2001).
- Number of muscles ± soft tissue loading (Farahmand et al., 1998a; Merican and Amis, 2009).
- Load applied to derive muscle tensions (Farahmand et al., 1998a; Li et al., 1999).
- Line of action of the muscle (Powers et al., 1998).

3.10.1.1.1 Open Chain versus Closed Chain

Since this study was examining the patellofemoral joint, which is known to be highly influenced by muscle loading, it was decided to use a loaded study method (Huberti and Hayes, 1984). Following review of the literature it was determined that examination of closed chain knee flexion would demand the control of many variables. A closed chain motion is complex and the pattern of muscle activation during the movement is known to differ widely from subject to subject, making it challenging to standardise in a cadaveric model (Zeller et al., 2003). Loading the combination of any or all of the quadriceps, hamstrings, gluteal, hip flexor, gastrocnemius, soleus and popliteus muscles would need to be considered. There is not an established and validated method currently present to define the number of muscles required to be loaded to optimally replicate *in vivo* motion during closed chain cadaveric knee joint testing. In part this is due to the challenges of measuring muscle forces generated *in vivo*. To determine and control such variables would provide scope for more than a PhD within itself. Therefore, in the present experiment, it was decided to test the specimens during an open chain knee extension movement instead. This motion has been extensively examined previously during studies investigating the patellofemoral joint (Huberti and Hayes, 1984; Melegari et al., 2008). Although it could be argued that this ignores valuable variables pertaining to patellar motion, it was felt this compromise was justified given the ability to standardise the open chain testing. Furthermore during an open chain motion the tibia can be left unconstrained and free to rotate (Section 3.8.1.2).

3.10.1.1.2 Type of Muscle Contraction

Given the open chain movement considered during this experiment it was logical to examine a concentric quadriceps contraction (i.e. when the muscle shortens to move against resistance). Eccentric muscle contractions would be extremely challenging to directly simulate *in vitro* and even if achieved, would be unlikely to replicate *in vivo* situations given the absence of central motor control over the muscle. Individuals suffering patellofemoral joint dysfunction are known to suffer with symptoms during both concentric and eccentric contractions (stair ascent / descent), providing justification for the use of a concentric movement in the study (Crossley et al., 2004).

3.10.1.1.3 Muscle loading

The role of the quadriceps apparatus in patellar tracking is widely highlighted and recognised in the literature (Bull et al., 1998; Goh et al., 1995; Powers et al., 1998). These are the primary knee extensors and have a well-defined role in patellar stability (Blazevich et al., 2006; Senavongse and Amis, 2005). As their role is critical it was recognised they must be included in the experimental set-up. Review of the literature provided a clear method to determine the magnitude and direction of the loading of the six components of the muscle based on cross sectional area identified during prior cadaveric studies (Farahmand et al., 1998a). It would be possible to perform high resolution MRI scans of the specimens to determine individual specimen specific loading based on this method if the whole leg were available. However the specimens supplied only included about 20cm of femur, therefore this was not possible. Hence muscles were loaded in accordance with mean values previously reported in the literature (Farahmand et al., 1998a). This method has been shown to be superior to single axial loading and is a published and accepted method, therefore it was used for the present study (Ghosh et al., 2009; Powers et al., 1998). It incorporates multi-planar and oblique quadriceps loading, which are both known to be critical to simulating *in vivo* behaviour (Lieb and Perry, 1968).

The posterior muscles of the thigh, the hamstrings, consist of three individual muscles which are responsible for knee flexion. Semimembranosus and semitendinosus lie medially and biceps femoris lies laterally (Woodley and Mercer, 2005). A ratio of 3:2, lateral to medial cross sectional area has been reported in the hamstrings (Wickiewicz et al., 1984). However, cadaveric studies have commonly applied hamstring loading as a 50:50 ratio medial to lateral (Li et al., 1999; More et al., 1993). Very scant literature exists that relates the ratio of quadriceps to hamstrings load through knee flexion making it hard to dynamically load these muscles to simulate physiological conditions. It was therefore decided not to load the hamstring muscles during the present study since an inaccurate representation of *in vivo* hamstrings could have a significant and unpredictable effect on patellar joint contact mechanics and patellar kinematics.

It has been recognised in the literature that the ITB plays an important role in patellar tracking and contact mechanics, particularly from 60º-120º flexion (Kwak et al., 2000; Ostermeier et al., 2007b). The ITB has been identified to contribute up to 10%-20% of the restraint to lateral patellar motion (Christoforakis et al., 2006; Senavongse et al., 2003). Research has identified that biomechanical cadaveric studies should load the ITB when studying patellar kinematics and contact mechanics (Merican and Amis, 2009). It was therefore decided to load the ITB in the current study since it was directly influencing the patellofemoral joint. In prior studies which considered relative size of muscles it was determined that 30N tension on the ITB would be in the correct proportion to a quadriceps tension of 175N (Kwak et al., 2000, Merican et al., 2009).

3.10.1.1.4 Tibial Rotation

It is widely recognised that tibial rotation can influence patellar tracking (Nagamine et al., 1995). When torque was applied to induce tibial rotation, there was increased patellar tilting (Heegaard et al., 1994; van Kampen and Huiskes, 1990). However the precise effect of tibial rotation remains contentious (Nagamine et al., 1995). Attempts to constrain the tibia during cadaveric experiments could result in inadvertent rotation being introduced accidentally (Katchburian et al., 2003). Again during a closed chain motion, such as the stance phase of gait or a squat motion, when the femur rotates in relation to a fixed tibia it is very challenging to standardise tibial position. This is a result of the close relationship of the tibia to foot position and the challenge of positioning the foot *in vitro* when muscle tension is absent and there is little rigidity to support it (Redmond et al., 2006). Although this method has been attempted (Reider et al., 1981), the majority of studies examine a knee extension movement when the femur is fixed. The present study, examining this open chain motion, meant the tibia could be left unconstrained throughout testing. This allowed it to follow its natural path of motion and concerns over influencing tibial rotation were avoided.

3.10.1.1.5 Varus / Valgus

The varus-valgus position of the knee has been identified to alter during knee flexion and extension cycles (Blankevoort et al., 1996). However since the knees for this study were structurally intact, with no ligament or joint damage, during pilot testing we identified that there was very little valgus-varus motion permitted when the knees were mounted in the test rig and loaded with the quadriceps musculature, because the quadriceps tension compressed the tibia onto the femur. This observation is supported by prior work (Grood et al., 1981). Therefore during the study no attempt was made to restrict the natural varus-valgus motion of the knee.

3.10.1.2 Conclusion

Following review of the literature and pilot testing, as outlined, knees were experimented on whilst in a test rig permitting open chain knee extension motion with the quadriceps and ITB loaded (section 3.4.2). The tibia was left unconstrained and the femur fixed, no valgus – varus restrictions were imposed and the quadriceps and ITB were loaded in accordance with prior reports (Farahmand et al., 1998b; Merican and Amis, 2009).

3.10.2 Assessment of MPFL Length Change

3.10.2.1 Choice of Measurement Tool

A wide range of techniques have been reported to measure the length change behaviour of ligaments. Following a review of the literature four methods were identified which could be used to answer the current research question asking '*What is the length change pattern of the MPFL?*' The advantages and disadvantages of these were considered in turn to determine the optimal methodology for use in the present study. These are outlined below:

1. **LVDT** (Amis and Dawkins, 1991; Ghosh et al., 2009)
   - ✓ Direct measurement, enables the exact attachments of the ligament to be dissected and determined, high accuracy.
   - ✗ *In vitro* measurement.

2. **MRI / CT scan direct measurement** (Higuchi et al., 2010; Victor et al., 2009)
   - ✓ *In vivo* measurement (although indirect).
   - ✗ This method fails to account for the natural curved shape of the medial epicondyle, and makes it difficult to determine the exact attachments of the MPFL.
   - ✗ Unknown accuracy.

3. **Calliper** (Smirk and Morris, 2003)
   - ✓ Permits dissection and clear visualisation of the MPFL attachments.
   - ✗ Inaccurate (authors reported up to 5mm measurement error).

4. **Intraoperative** (Tateishi et al., 2011)

   ✓ *In vivo* direct measurement.

   × Requires a patient population and experienced surgeon to perform the testing. Challenging to determine the exact attachments of the ligament due to minimally invasive nature of the surgery.

Following consideration of all factors it was decided to use the LVDT method. This crucially enabled determination of the attachments of the ligament, permitted ligament length change measurement through full range of knee flexion and prior literature suggested it possessed good accuracy (Ghosh et al., 2009). It was planned to examine the knee in a test rig permitting physiological loading of the knee and it was anticipated this would help minimise the disadvantages associated with *in vitro* measurement.

3.10.2.2 Linear Variable Displacement Transducer

LVDT's are electrical transformers which enable measurement of linear displacement (Figure 3.20), and are commonly used in industry. They have a ferromagnetic chamber which produces a voltage change as the sliding core position is changed. The LVDT can be linked to device specific software ("Orbit" Excel, Solartron Metrology, UK) to produce an output in millimetres which corresponds to the position of the core in relation to the chamber, thus enabling length changes to be determined from its resting starting length.

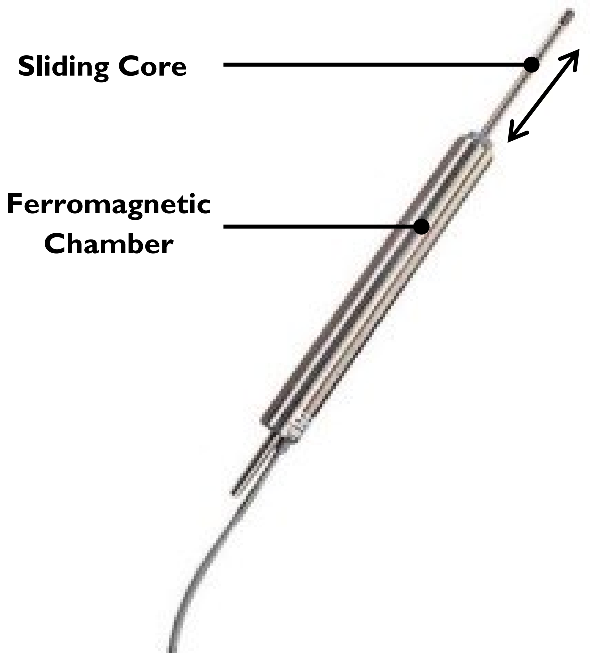

Figure 3.20 Linear Variable Displacement Transducer.

Prior literature examining soft tissue length change patterns of the native anterior cruciate ligament and the medial and lateral retinacula following total knee replacement have used the LVDT attached to a sliding suture to determine ligament length change patterns (Amis and Dawkins, 1991; Ghosh et al., 2009). This method is depicted in Figure 3.21.

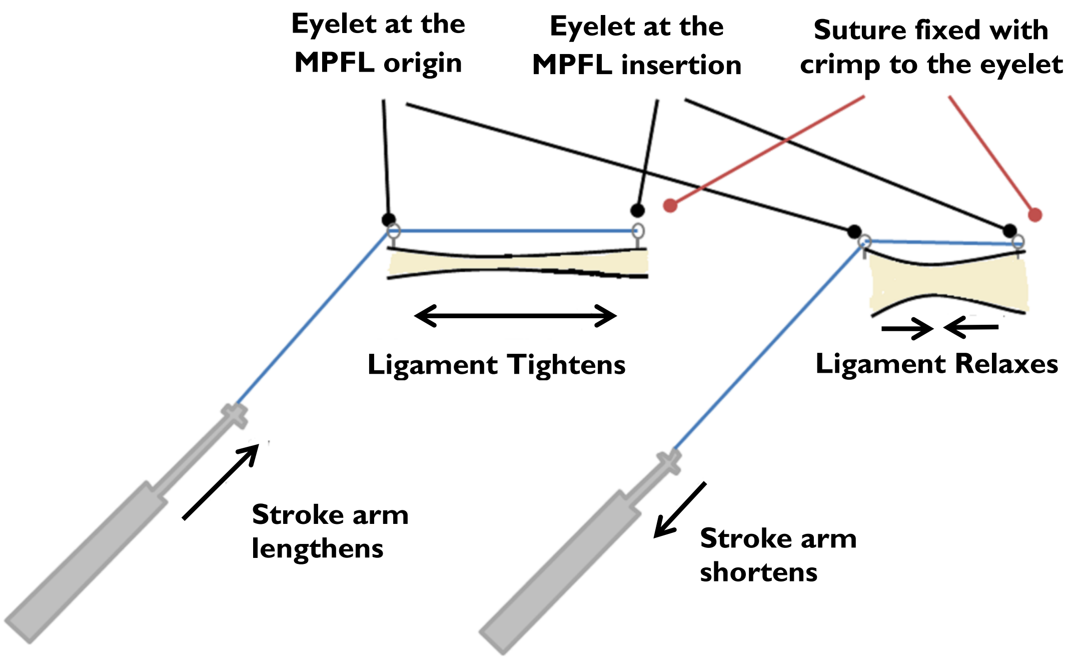

Figure 3.21 Method used for measurement of ligament length change using the LVDT. It can be seen that as the ligament becomes taut, the stroke arm of the LVDT is lengthened and a corresponding change is recorded on the local computer, with the opposite true as the ligament relaxes (adapted from Ghosh et al. (2009)).

3.10.3 Factors Affecting Experimental Design

3.10.3.1 Rig Set-Up

Careful consideration was given to where the LVDT should be positioned in the test rig. With the MPFL attaching from the medial patellar border to the medial femoral condyle, it was decided the LVDT was best placed in close proximity to one of these bones. Since the femur was fixed in the test rig and the tibia and patella were mobile, it was decided to fix the LVDT in relation to the femur. Two positions were therefore considered appropriate for the LVDT. These are shown in Figure 3.22 and were:

1. Fixed to the base of the rig so the probe was positioned next to the medial femoral condyle.

2. Fixed directly along the shaft of the femur.

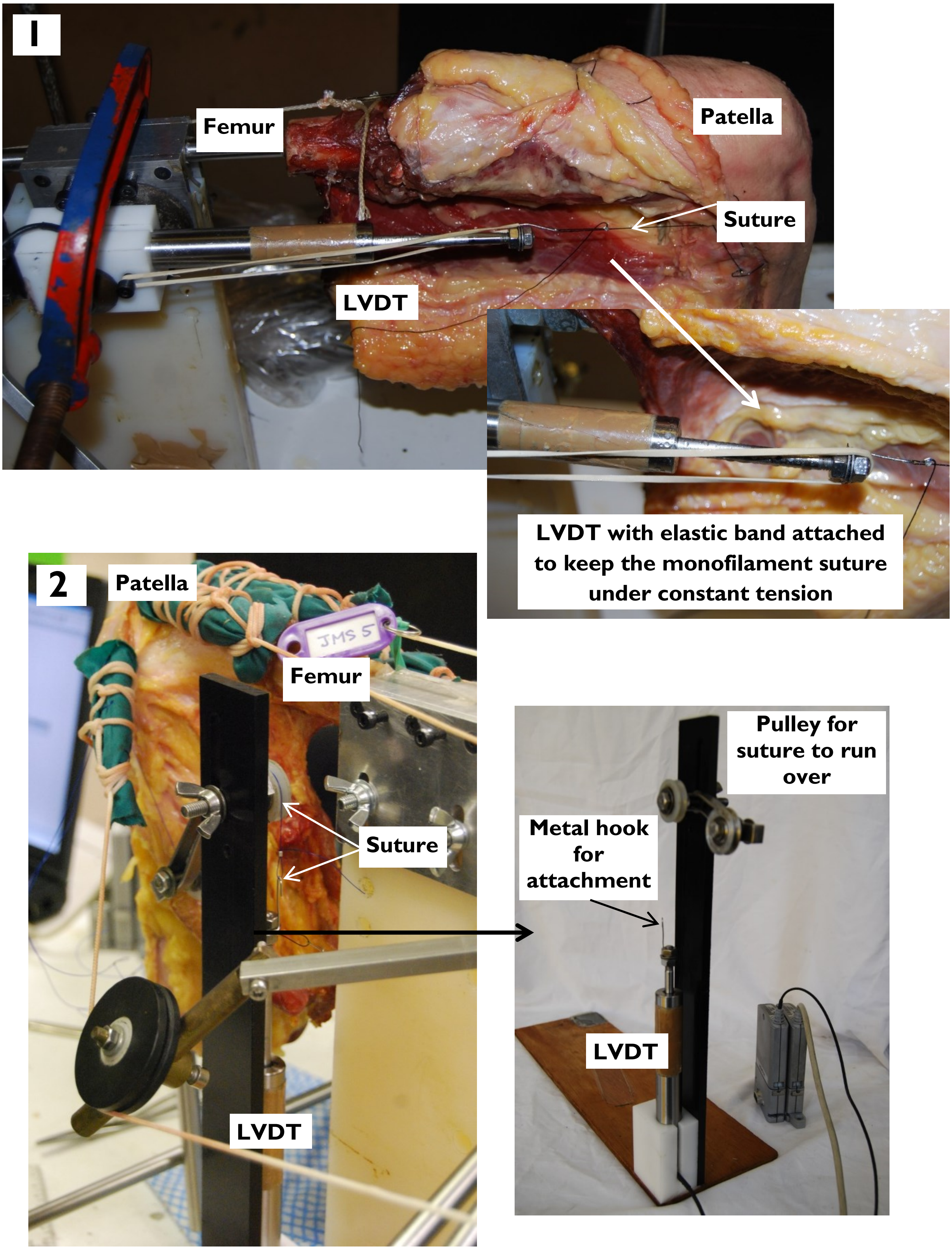

Figure 3.22 1. The first test position with the LVDT clamped parallel to the femoral shaft with an elastic band attached to the probe to provide tension to the suture. 2. With the LVDT positioned below the femoral epicondyle, by running the suture through a pulley system (the weight of the LVDT probe applies tension to the suture).

Following pilot testing of both positions it was determined that the second set up, with the LVDT perpendicular to the femoral shaft, was preferable. The test rig was adapted and an LVDT stand with the pulley system and hook attached to the probe arm developed as shown. This set up did not interfere with the positioning of the VMO muscle in the rig and permitted free movement of the suture throughout testing, unlike the first set up where the VMO muscle position in the test rig prevented free movement of the suture from the patella to the LVDT. Furthermore, with differing femoral sizes amongst the specimens, it was more difficult to standardise the position of the LVDT when it was mounted parallel to the femur compared to when it was fixed to the base of the test rig.

Prior work has used the LVDT with an elastic band fitted over the tip to impart a constant low tension to the LVDT core (as shown in Figure 3.22) (Ghosh et al., 2009). However a further advantage of the LVDT position in the second set up shown was that it was unnecessary to have the elastic band to apply tension. Since it was not known how the tension imposed by the elastic band would be altered as the sliding core was lengthened and shortened, it was instead decided to just use the weight of the probe to provide a constant tension to the suture.

3.10.3.2 Suture Used

In order to determine the optimal suture to run from the patellar attachment, through the femoral eyelet to attach to the hook at the end of the LVDT probe, a number of factors were considered:

1. Need to ensure the suture would not break.
2. The suture required to slide with little resistance over the local soft tissues.
3. Need to ensure the suture would not stretch.

In order to determine the suture most appropriate two factors were therefore considered:

1. Suture material.
2. Suture size.

A variety of suture materials were pilot tested, including Ethilon, Vicryl and Prolene (all Ethicon Co, Somerville, New Jersey). It was determined that Ethilon was the optimal to use since it allowed free movement of the suture over the local soft tissues (unlike Vicryl), and it did not stretch or ‘kink’ (unlike Prolene). Suture size using 1-0, 2-0 and 3-0 were examined. Size 2-0 was determined to be the thinnest material allowing the free movement of the suture along the path of the MPFL but was strong enough to resist breaking during the experimentation, unlike 3-0.

Initially the suture was positioned over Layer 1 of the medial retinaculum. However this resulted in the suture often failing to follow the path of the MPFL. Following experimentation it was found possible to carefully separate the first and second layers of the medial retinaculum. It was then possible to position the suture beneath the first layer of the medial retinaculum to follow along the fibres of the MPFL (Figure 3.23). This path was checked throughout the experiment to ensure that the suture was aligned with the MPFL and had not tethered or bent.

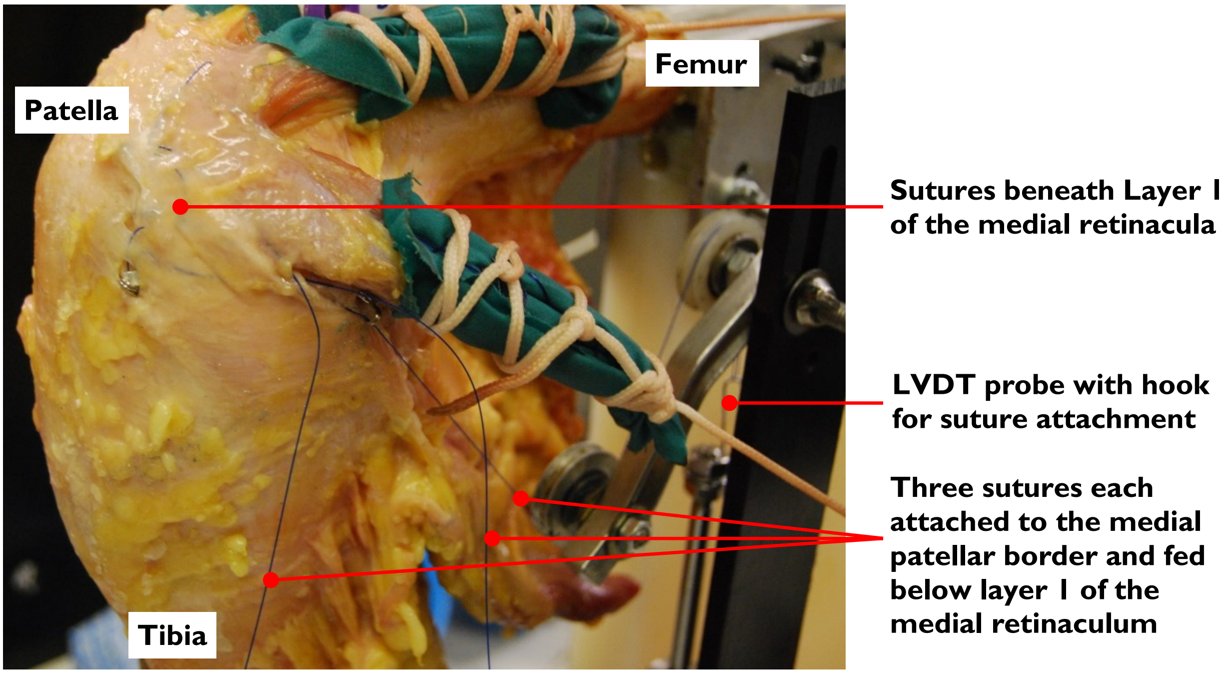

Figure 3.23 Knee in the test rig flexed to 90°. Three sutures are shown which were fixed to three different patellar eyelets attached along the superomedial patellar border. These were positioned deep to layer I of the medial retinaculum to lie over the MPFL fibres. One suture is seen to run from the patellar eyelet, through a femoral eyelet, along two pulleys and attached to the hook on the LVDT. This measurement was repeated for all combinations of the three patellar attachments with each of the five femoral attachments.

3.10.3.3 Points Tested

During the planning stage of this study we aimed to measure length change patterns of the native MPFL, and then examine 3 patellar attachments (proximal, central and distal positions along the superomedial patellar border) with 10 femoral attachments (5mm and 10mm proximal, distal, anterior and posterior to the anatomical attachment). However this had to be modified following pilot testing. With ten eyelets positioned to mark the femoral attachments it was impossible to permit free movement of the suture from the femoral attachment to the LVDT, since the suture typically got caught around some of the other eyelets in the area. It was therefore decided following pilot testing and preliminary results to test only 5 femoral attachment points 5mm proximal, distal, anterior and posterior (Figure 3.24).

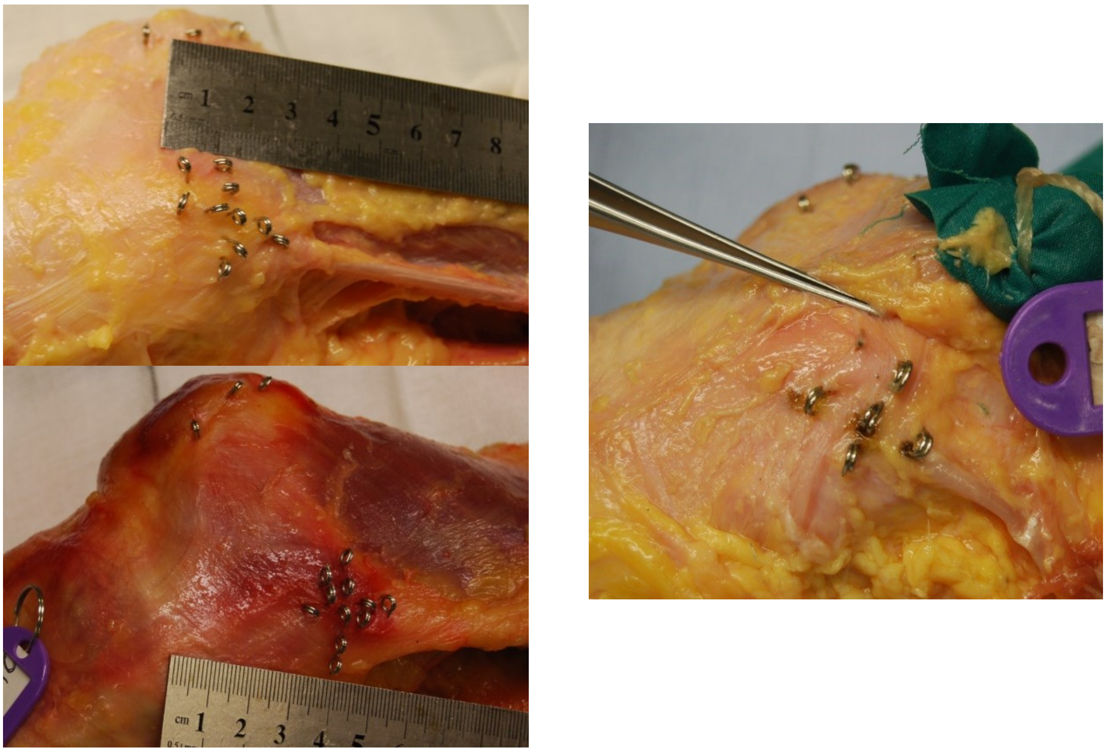

Figure 3.24 Left sided images showing 12 attachments marked on the femur; 10 femoral attachments (as previously outlined) and the media epicondyle and adductor tubercle. Following pilot testing it was determined it was impossible to allow free direct positioning of the suture between the desired femoral and patellar attachments when 10 attachments were present on the medial femoral condyle. Therefore it was decided to only measure 5 attachment points as shown in the right side image.

3.10.3.4 Ligament Length Change Pattern Reliability

Following refinement of the ligament length change pattern methodology as outlined it was necessary to establish the reliability of the system. In order to do this the measurement of the native MPFL (the length change pattern from the central patellar attachment to the anatomical femoral attachment) was performed on two separate days, one week apart using the methodology described. The sutures and eyelets were left in position between the testing days. The test was performed three times on each day and an average of the readings for each flexion angle taken. It was repeated on all 8 knees included in the testing.

Intraclass Correlation Coefficients (ICC) were then determined using SPSS. Results are summarised in Table 3.3.

Table 3.3 Showing the mean data (n=8) for and ICC values calculated from Test Re-Test measurements of the ligament length change patterns of the native MPFL measured using the LVDT.

Flexion Angle (°)	Mean Day 1	Mean Day 2	ICC
0	0.00	0.00	***1.00***
10	-1.31	-1.20	***0.98***
20	-1.91	-1.86	***0.96***
30	-2.04	-1.95	***0.94***
40	-2.12	-1.89	***0.89***
50	-2.02	-1.81	***0.90***
60	-2.00	-1.83	***0.91***
70	-1.89	-1.81	***0.89***
80	-1.70	-1.76	***0.89***
90	-1.43	-1.68	***0.89***
100	-1.25	-1.41	***0.91***
100	-1.11	-1.01	***0.95***

High ICC values of 0.89-1 were obtained across all angles of knee flexion. The high ICC values suggest excellent reliability (excellent ICC > 0.9) of the system and therefore it was decided the method was appropriate to use during the experiment to measure ligament length change patterns of the MPFL.

CHAPTER 4

MEDIAL PATELLOFEMORAL LIGAMENT SECTIONING

4.1 Background

Patellar dislocation as a consequence of trochlear dysplasia and patella alta has been previously well highlighted in the literature (Dejour et al., 1994; Fithian et al., 2004; Simmons and Cameron, 1992). However, a further subset of patients suffer patellar dislocation in the absence of bony abnormalities: these patients often suffer trauma through a direct blow to the patella or twisting movements of the knee resulting in patellar dislocation (Panni et al., 2011; Stefancin and Parker, 2007). Given its anatomical position, lateral patellar dislocation is not possible without damage to the MPFL (Mountney et al., 2005). Indeed this has been widely confirmed with study findings of MPFL rupture in 95-100% and injury in 100% of patients suffering patellofemoral dislocation, reported from MRI and surgical investigations (Sallay et al., 1996; Sillanpää et al., 2009b; Weber-Spickschen et al., 2011). In addition to MPFL injury, osteochondral bruising, lesions and fractures of the medial articular facet of the patella and / or lateral femoral condyle are common findings following dislocation (Arendt et al., 2002; Nomura and Inoue, 2005; Sallay et al., 1996). These findings are similar to those reporting osteochondral defects found in the knee following anterior cruciate ligament rupture, which have been linked to subsequent development of osteoarthritis and have been related to trauma at the time of injury (Keays et al., 2010; Øiestad et al., 2009).

Reports of recurrent instability rates greater than 50% with non-operative management (Cofield and Bryan, 1977) are likely to have contributed to the increase in primary management of patellar dislocation with operative MPFL repair (Ahmad et al., 2000; Arendt et al., 2002). However despite the high prevalence of re-dislocation with non-operative treatment and reports of ongoing pain, recurrent instability, reduced functional levels and patellofemoral arthritis following patellar dislocation (Arendt et al., 2002; Atkin et al., 2000; Sallay et al., 1996), the most common first choice management following primary dislocation remains conservative treatment (Bicos et al., 2007; Cofield and Bryan, 1977; Fithian et al., 2004; Stefancin and Parker, 2007).

There is no clear relationship established in the literature between MPFL rupture and subsequent identification of abnormal patellar kinematics, or contact mechanics. Indeed a portion of patients are known to make a good recovery without the need for any operative intervention following primary dislocation (Cofield and Bryan, 1977). Therefore it seems appropriate to investigate in further detail the consequences of MPFL transection to aid in the clinical reasoning process of the management of patients following patellar dislocation. Prior work has examined the effect of MPFL transection on patellofemoral joint mechanics and kinematics, but these have not been examined simultaneously or through the full range of knee flexion, and the reported results were inconsistent. Some authors have suggested that the MPFL has a significant effect on patellar kinematics when transected, resulting in up to 7.2mm increases in lateral patellar translation compared to the intact knee (Philippot et al., 2009). A further group reported 4mm increases of lateral translation (Ostermeier et al., 2007a) and others did not find a change in kinematics following sectioning (Sandmeier et al., 2000). These results are potentially a consequence of the different methods used to investigate the research question, with a range of quadriceps loading methods reported and differing measurement systems used to assess joint motion. Only one prior study has examined the effect of MPFL transection on joint contact mechanics (Beck et al., 2007), but this study only examined measurements at three different knee flexion angles, and crucially not from 0°-30° when the MPFL has its main role (Senavongse and Amis, 2005) (Chapter 3).

4.2 Aims

Given the high levels of patient dissatisfaction following patellar dislocation and the association of MPFL rupture with later joint degeneration, this study was designed to investigate the effects of MPFL transection on patellofemoral joint mechanics and kinematics. Noting clinical findings it was hypothesised that MPFL transection would:

1. Cause increased distance between the patellar and femoral MPFL attachments.
2. Cause increased lateral patellar tracking.
3. Cause increased lateral contact pressures.
4. Cause decreased medial contact pressures.

4.3 Clinical Relevance

This chapter presents data relevant to all patients who dislocate their knee cap, since it relates to the effect of injury to the MPFL following patellar dislocation. This is of particular relevance to patients suffering acute patellar dislocation, but also to those suffering low velocity patellar dislocations during activities of daily living, where the medial structures are also known to be damaged.

4.4 Materials and Methods

4.4.1 Specimen Preparation

Ethical approval for this study was granted from the local Research Ethics Committee. The testing for this experiment followed on from the prior experiment, and the same specimens described in the previous chapter were used for testing in this study. Thus knees were positioned in the same test rig, and the muscles dissected and loaded as outlined in Chapter 3.

4.4.2 Ligament Length Change Measurements

The adductor tubercle and the medial epicondyle were marked with metal pins. A small metal eyelet was positioned midway between these points, defining the femoral attachment of the MPFL (Baldwin, 2009; Philippot et al., 2009). This anatomical point was confirmed with true lateral view X-rays, as described in Chapter 2 and was consistent with prior reports (Schöttle et al., 2005). A further eyelet pin was positioned on the superomedial border of the patella, half way between the uppermost medial patellar border and the mid patellar body, marking the mid-point of the patellar attachment of the MPFL (Baldwin, 2009; Stephen et al., 2012). A monofilament suture (Ethilon 2/0, Ethicon, Somerville NJ) was secured to the patellar eyelet. The superficial and middle layers of the medial retinaculum were separated, and the suture passed between them and through the femoral eyelet and connected to a linear variable displacement transducer (Solartron Metrology, Bognor Regis, UK) (Ghosh et al., 2009; Stephen et al., 2012). This method is described in more detail in Chapter 3.

4.4.3 Tekscan Contact Pressure Measurements

A Tekscan 5051 pressure sensor (Tekscan, I-Scan ™, Boston, MA) was used to measure contact pressures between the patella and trochlea. It measured 55.9mm by 55.9mm, was 0.1mm thick, and comprised 1936 sensel measurement points (each $0.8mm^2$) with a saturation pressure of 3.48MPa.

Calibration of each sensor was undertaken using a materials testing machine (Instron 5585, High Wycombe, Bucks, UK). A sheet of 3mm thick silicone rubber, with a modulus of elasticity of approximately 0.7MPa, was placed on either side of the sensor during calibration to simulate the compliance provided by patellofemoral cartilage (Donzelli et al., 1999; Drewniak et al., 2007; Mow et al., 1980; Wilson et al., 2003). The sensor and rubber sheets were compressed between two metal plates to produce an area of uniform pressure (Elias et al., 2009) (Figure 4.1). Ten pre-conditioning cycles to 2MPa were performed, approximately 20% greater than anticipated pressures, based on pilot study testing. Each sensor was equilibrated to 10%, 50% and 90% of the maximal anticipated load to normalise the readings and ensure identical output from each sensing element with a uniform pressure applied. An identical two-point power-law calibration was performed on each sensor at loads approximately 20% and 80% of the expected maximum joint pressure using Tekscan software.

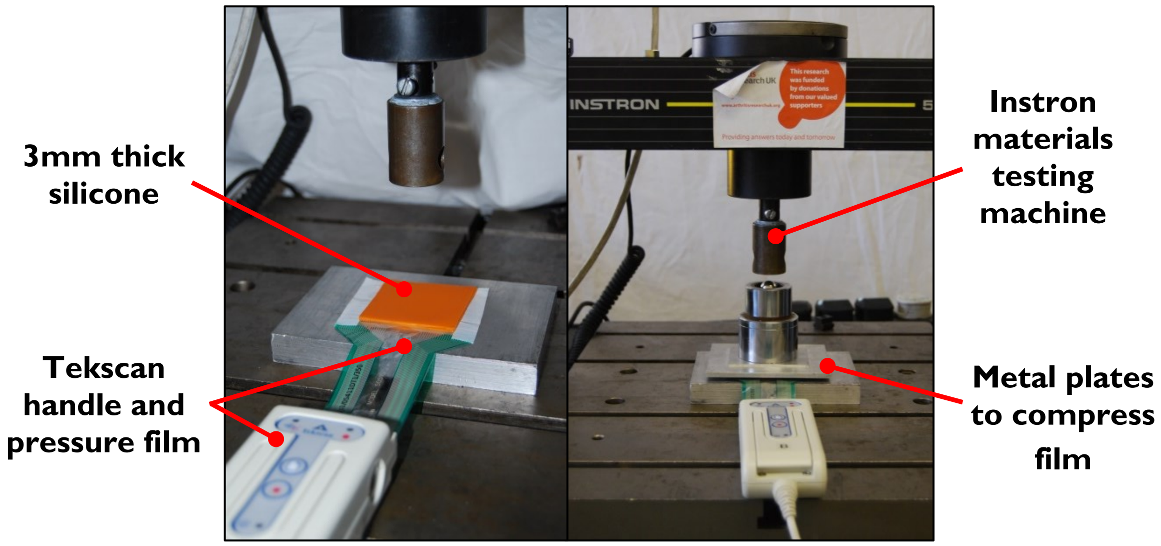

Figure 4.1 Tekscan calibration: left image showing the Tekscan sensor sandwiched between two rubber sheets, and the right image showing the sensor and rubber sheets positioned between two metal plates ready to be compressed by the Instron materials testing machine.

Care was taken to avoid disruption of the retinacula when inserting the Tekscan sensor into the joint (Figure 4.2). The proximal quadriceps were elevated and two plastic tubes inserted through the proximal patellofemoral synovial pouch and positioned along the medial and lateral ridges of the trochlea. The tubes were slid distally to contact the soft tissues on either side of the distal patellar tendon. The distance between the tubes was as wide as necessary to ensure accommodation of the sensor into the joint without causing it to crease. Two 15cm needles, attached to 2-0 Ethilon sutures (Ethicon, Somerville NJ), were pierced through the skin externally to internally and guided up through the tubes to emerge proximally. The needles were pierced through the outer edges of a 5051

Tekscan sensor, which had the outer edges reinforced with cloth tape to prevent damage, and then guided back through the tubes to pierce the skin and exit 5mm from their respective entrance points. The sensor was pulled into position using the sutures, and secured over the trochlea by stitching the sutures to the local soft tissues. The proximal end of the sensor was attached to a Tekscan Evolution handle with USB connection to enable data transfer to a local PC (Figure 4.2).

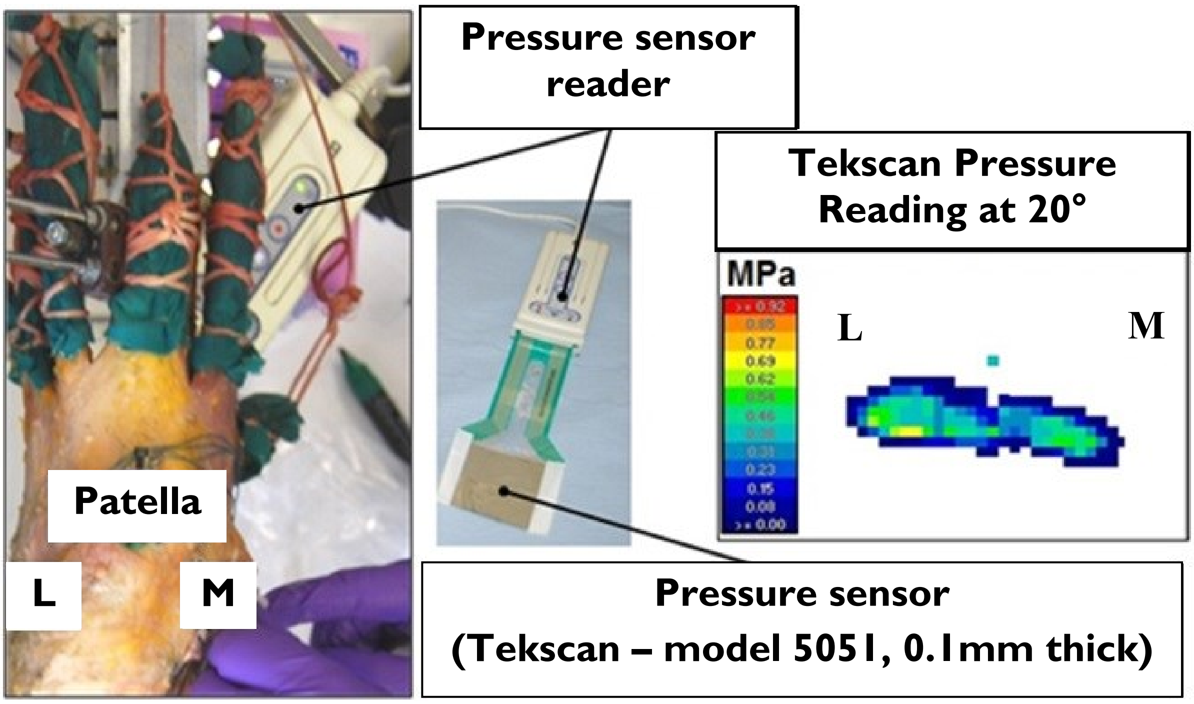

Figure 4.2 Tekscan pressure sensor positioned in the patellofemoral joint and connected to the evolution handle (left and centre images). The computer screen display in a 'normal' knee flexed to 20^0 (L= Lateral, M=Medial) is shown in the right image.

Patellofemoral contact pressures were analysed as medial and lateral facet data. A 1mm rod was inserted through the hole in the patella and pressed onto the Tekscan sensor, to separate the medial and lateral facets. For each flexion angle, a Tekscan file was saved and exported as an ASCII file to an Excel spreadsheet. The peak and mean medial and lateral pressures, and their position in relation to the trochlear groove, were calculated. This system had a mean test re-test difference of between 0.03MPa ± 0.02MPa (mean medial pressure) – 0.3MPa ± 0.3MPa (peak lateral pressure) (mean±standard deviation), in line with prior work (Wilson et al., 2003).

4.4.4 Patellofemoral Joint Tracking Measurement

A polaris optical tracking system and probe (Northern Digital Incorporated, Waterloo, Canada) were used with active optical trackers (Traxtal Technologies, Toronto, Canada), attached to the bones (Figure 4.3). This system had an overall volume root mean square (RMS) distance error of 0.35mm for a single marker (Wiles et al., 2004). The Traxtal tracker used for the patella had a reported accuracy of 0.04mm with a precision of 0.03mm (Merican and Amis, 2009).

The knee was positioned in the test rig, with the femoral co-ordinate system aligned to the long (anatomical) axis of the femur and the most-posterior points of the femoral condyles (Figure 4.3). The tibial intramedullary rod was unconstrained and therefore the tibia was free to rotate. The Traxtal probe was used to digitise sets of metal fiducial markers attached to the bones to aid later data analysis. Patellar motion was described in relation to the femur by a standard convention (Bull et al., 2002; Merican and Amis, 2009).

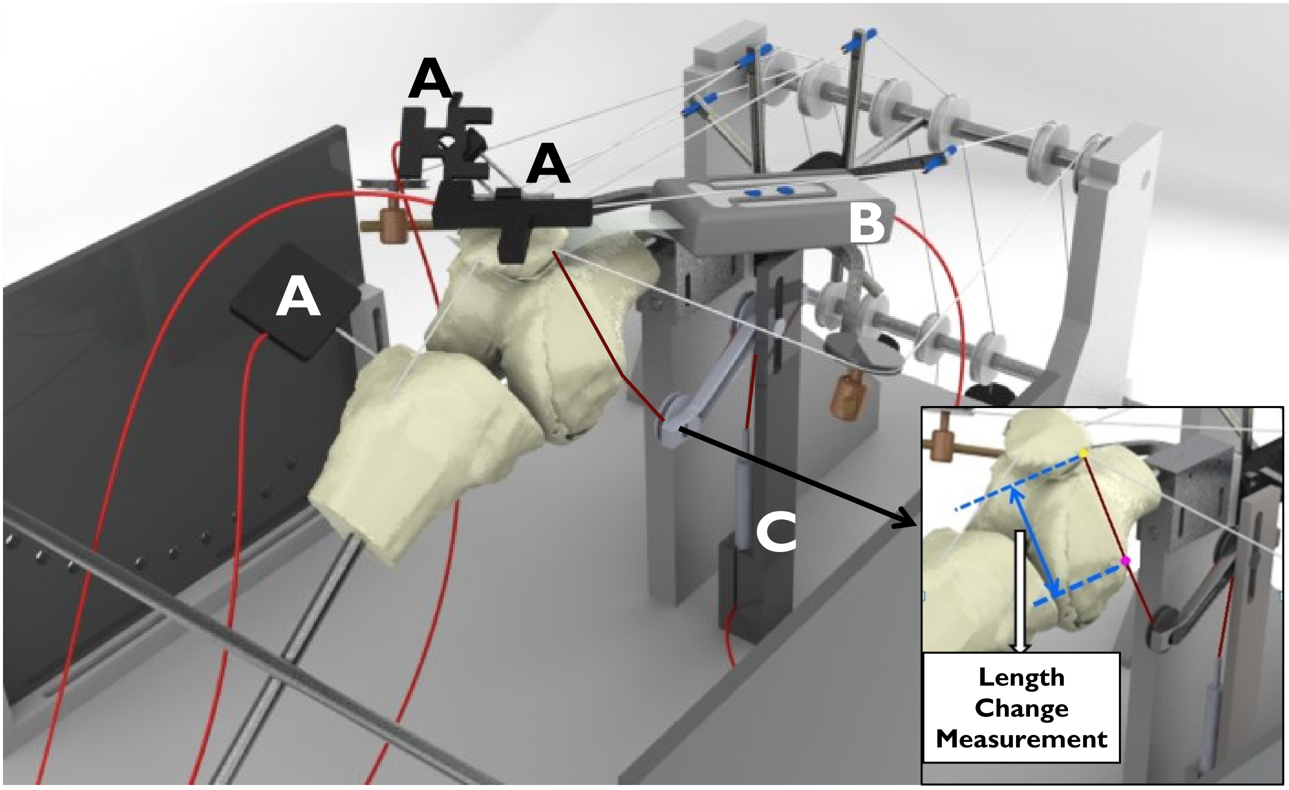

Figure 4.3 Knee joint rig, with attached muscle loading system.
A: Optical trackers.
B: Tekscan film and handle.
C: Linear variable displacement transducer: showing the suture passing along the path of the MPFL between attachment points on the patella and femur, before leading to the LVDT.

4.5 Testing Procedure

The knee was slowly flexed and extended from 0°-90° ten times. These 'conditioning' cycles enabled hysteresis to be minimised. Data were recorded at 10° intervals with the MPFL intact and following transection of the MPFL. Flexion angle order was randomised for the different specimens tested to avoid potential bias. The MPFL was transected near the femoral attachment to minimise trauma to the retinaculum and adjacent structures.

4.6 Analysis

The dependent variables of the study were: the length between the MPFL attachments, mean and peak medial and lateral facet articular contact pressures, and patellar kinematics, principally medial-lateral translation and tilt. Custom written MATLAB scripts (MATLAB 8.0, The MathWorks Inc., Natick, MA) calculated mean and peak contact pressures and patellar motion (Appendix A+B). Data was analysed in SPSS using a two-way repeated-measures analysis of variance, with flexion angle and ligament status (intact or transected) examined. A Shapiro-Wilk test confirmed that the data sets were normally distributed. A power calculation was undertaken, using GPower (Version 3.1, Baden-Württemberg, Germany, 2013), based on a 7mm mean change of patellar lateral translation (taken from a prior study) (Philippot et al., 2012a). It was determined that a sample size of 8 was required to detect a significant change with 80% power and 95% confidence. The tests compared the two conditions MPFL intact versus MPFL transected. Post-hoc paired t-tests with Bonferroni correction at individual flexion angles were applied when differences across test conditions were identified. Significance level was set a priori to $P<0.05$.

4.7 Results

4.7.1 Ligament Length Changes

The MPFL was almost isometric when it was intact, with a mean reduction in length of 1.9mm±2.3mm (Mean ±SD) from 0-20° knee flexion (Figure 4.4). Cutting the MPFL resulted in an increase in distance between its attachment points compared to the intact knee, of up to 4.9mm ± 2.5mm (Mean ±SD) at 20° of flexion. There was a significant effect of ligament transection on length change patterns between the femur and patella (P=0.003). Paired t-tests identified significant increases in length at all flexion angles from 0°-110° inclusive (P=0.003-0.007).

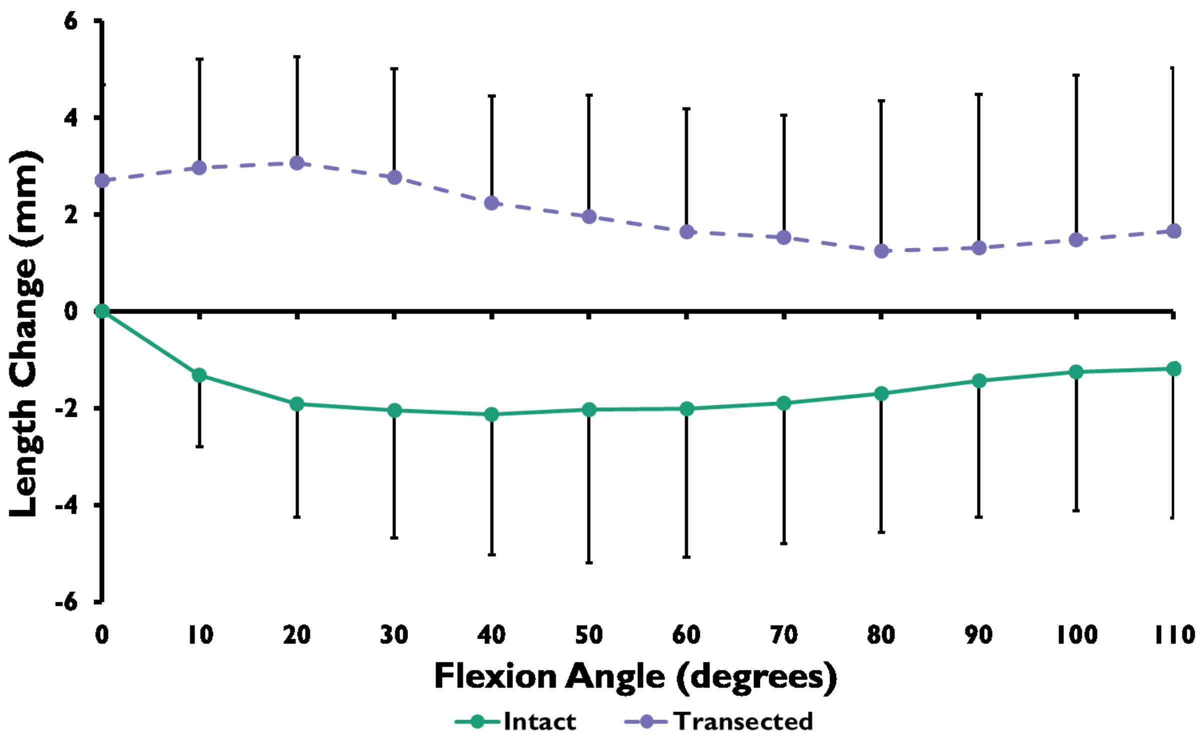

Figure 4.4 Length changes between the attachments of the MPFL versus knee flexion before and following transection of the MPFL (Mean ± SD, n=8). P<0.05 for all flexion angles.

4.7.2 Medial Patellofemoral Joint Contact Pressure

The peak pressure on the medial facet of the patellofemoral joint was reduced by cutting the MPFL (Figure 4.5) (P=0.02). This effect occurred towards knee extension: flexion angle had a significant effect (P<0.001). At 20° knee flexion, for example, the peak pressure reduced by 42% from 1.79±0.56MPa to 1.04±0.31MPa when the MPFL was cut (P=0.001).

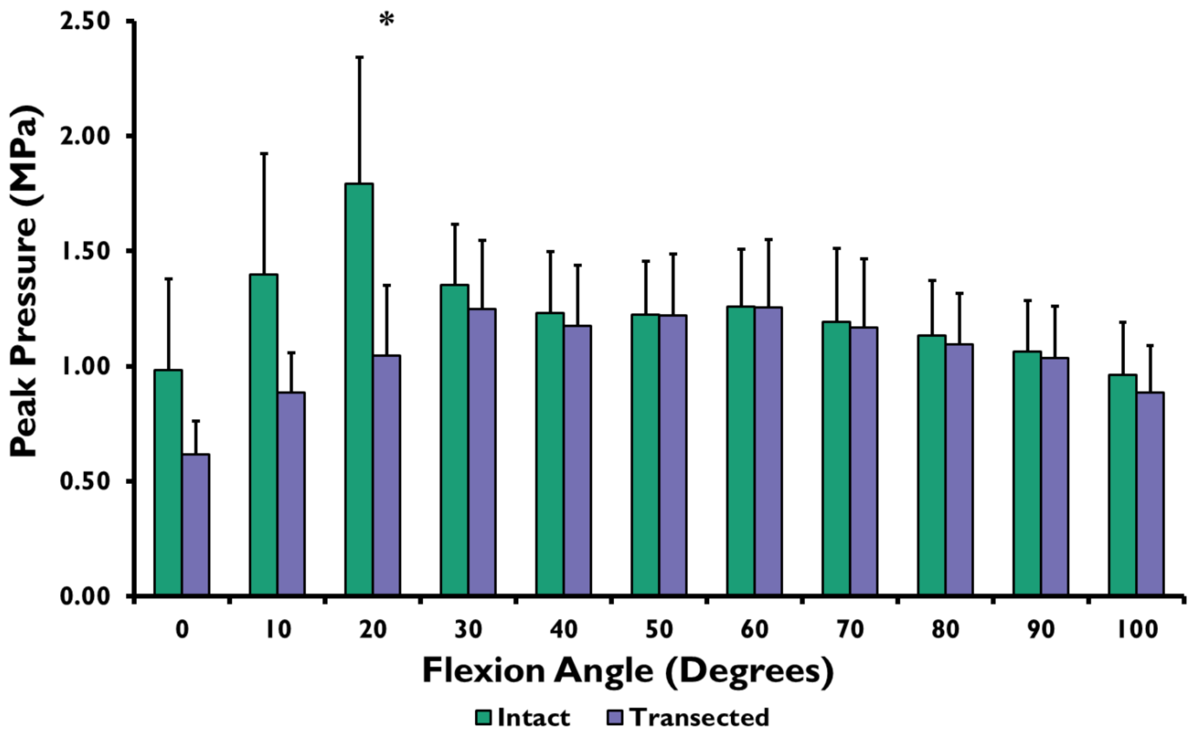

Figure 4.5 Peak medial patellofemoral joint contact pressures (MPa; mean + SD) of the intact knee and with the MPFL cut during knee flexion from 0°-100°. *P=0.001.

Similarly, the mean medial contact pressure was reduced by cutting the MPFL (*P*=0.016) (Figure 4.6). This effect was also strongest near knee extension (*P*=0.001) with significant reductions identified by the paired t-tests at 0° and 20° knee flexion.

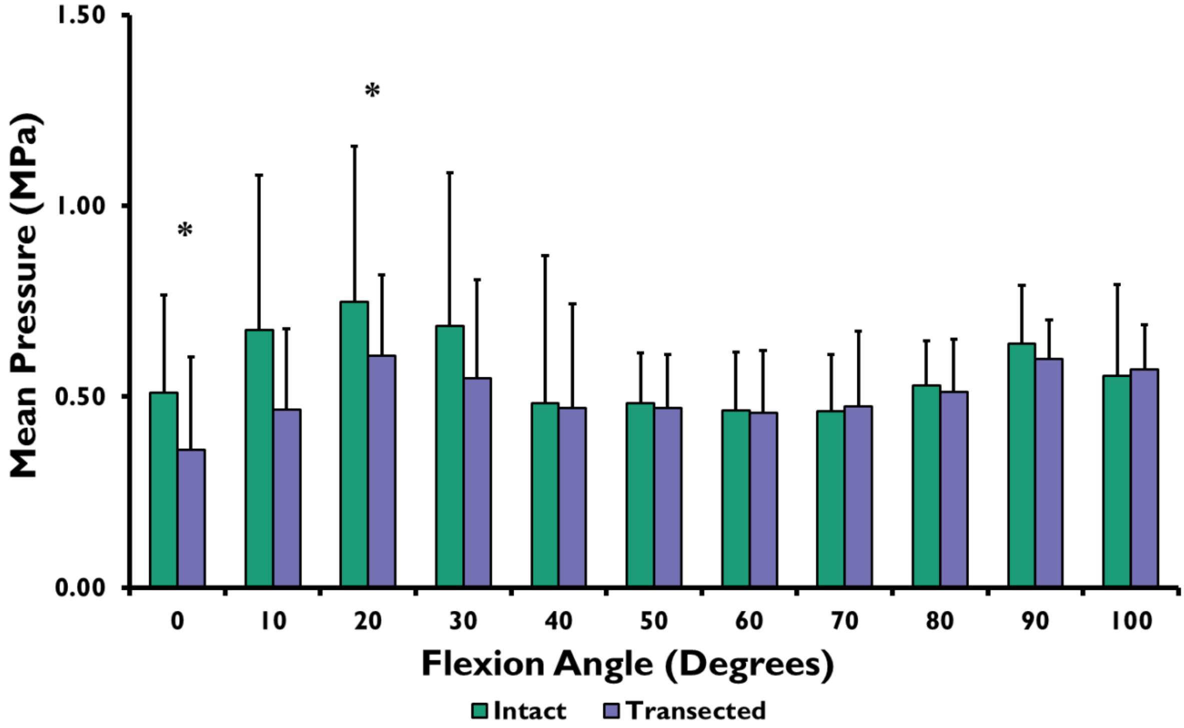

Figure 4.6 Mean medial patellofemoral joint contact pressures (MPa; mean + SD) of the intact knee and with the MPFL cut during knee flexion from 0°-100°. **P*<0.05.

4.7.3 Lateral Patellofemoral Joint Contact Pressure

Cutting the MPFL caused increases in lateral facet contact pressure (Figure 4.7). A significant difference of peak lateral contact pressure was not found overall (P=0.186) (Figure 4.8).

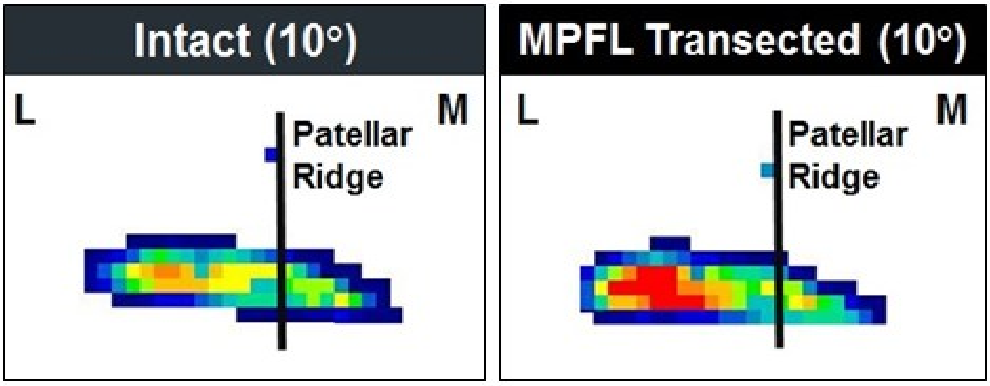

Figure 4.7 Screen Tekscan image of the pressure reading from one knee at 10 degrees of knee flexion, before and after MPFL transection. It is clear that contact shifted laterally and that the lateral facet contact pressure increased after MPFL transection. This was a typical pattern observed for all knees. (Scale: dark blue: close to zero pressure; mid-green: 0.45 MPa; Red: 0.9 MPa)

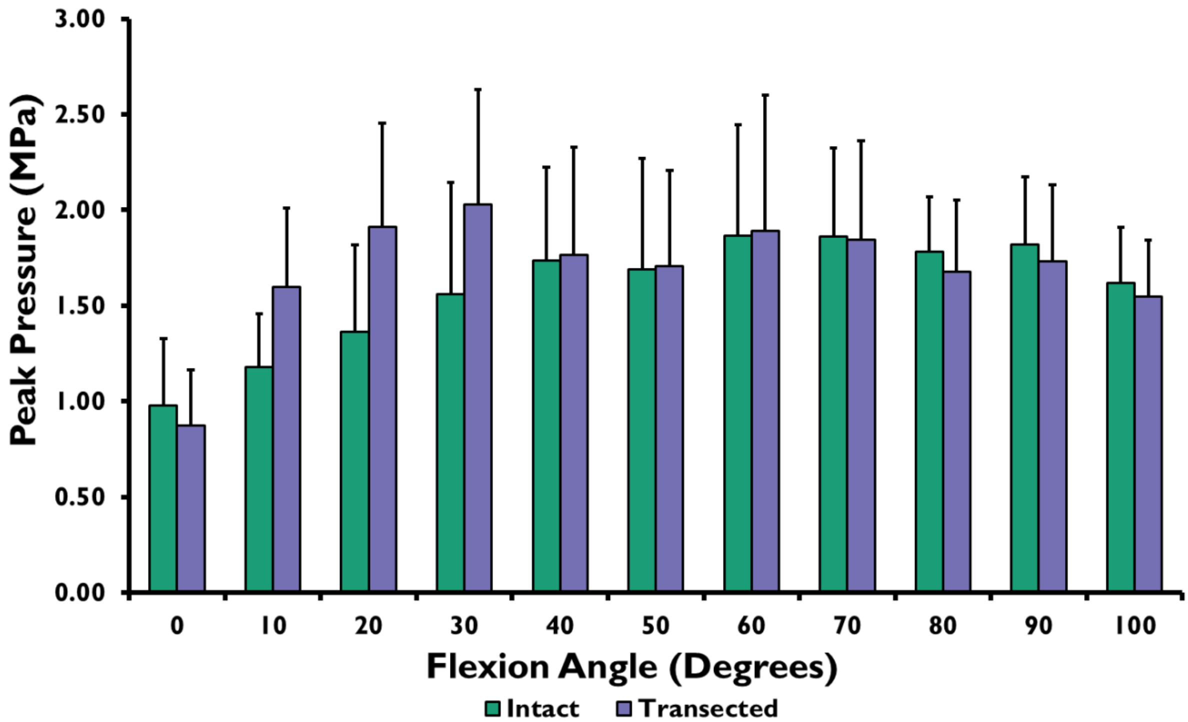

Figure 4.8 Peak lateral patellofemoral joint contact pressures (MPa; mean + SD) of the intact knee and with the MPFL cut during knee flexion from 0°-100°.

Mean lateral facet pressure was significantly increased after cutting the MPFL (P=0.009). This effect was strongest near knee extension (P<0.001), with paired t-tests identifying significant increases at 20° and 30° flexion (Figure 4.9).

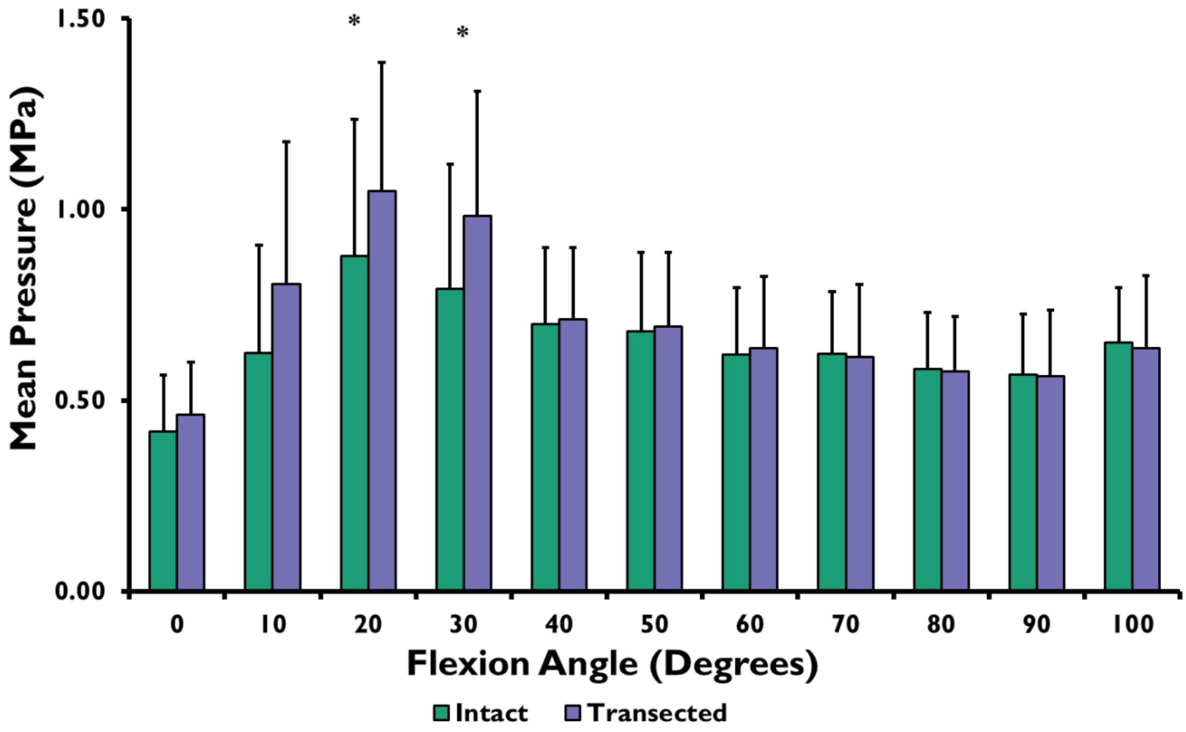

Figure 4.9 Mean lateral patellofemoral joint contact pressures (MPa; mean + SD) of the intact knee and with the MPFL cut during knee flexion from 0°-100°. *P<0.05.

4.7.4 Patellofemoral Joint Kinematics

4.7.4.1 Patellar Tilt

After cutting the MPFL, patellar lateral tilt increased by up to 1.1°±0.7 in early knee flexion. A significant difference in patellar tilt was found between intact and transected MPFL conditions (P=0.032). There was a significant effect of flexion angle (P<0.001), with paired t-tests identifying a significant increase at 20° (Figure 4.10).

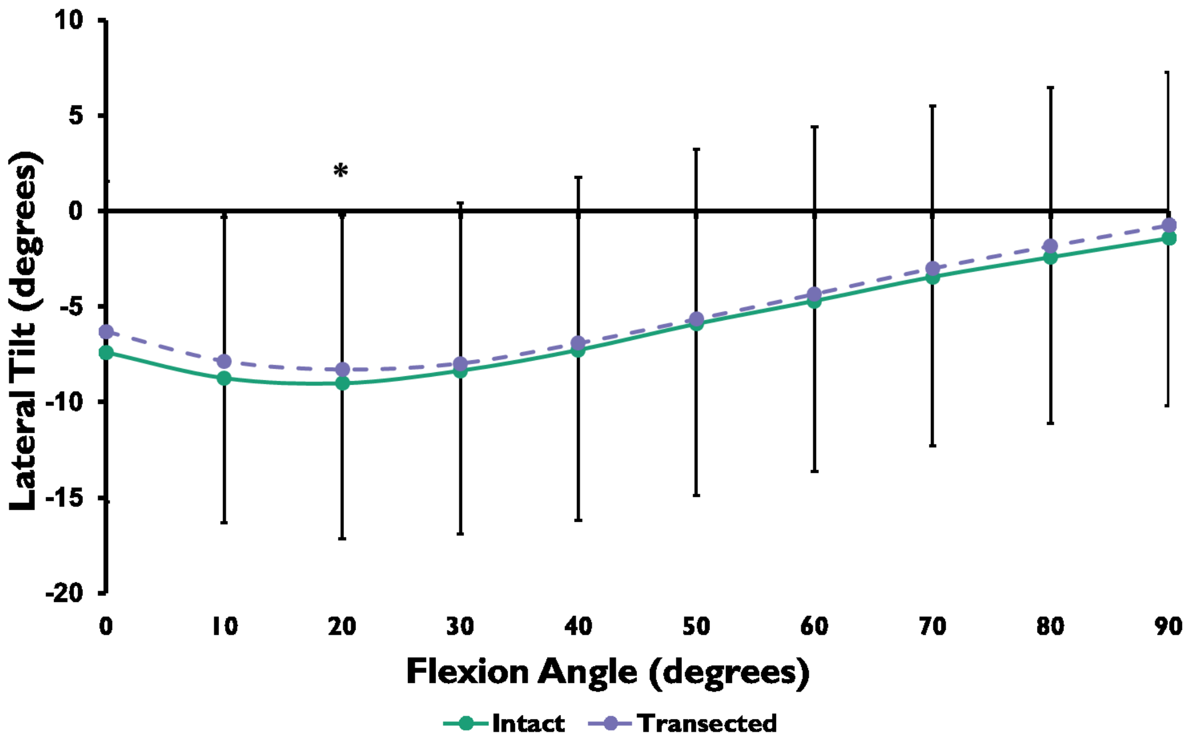

Figure 4.10 Patellar lateral tilt (°; mean ± SD, n=8) from 0°-90° knee flexion in the intact knee and with the MPFL cut. *P<0.05.

4.7.4.2 Medial-Lateral Translation (Shift)

As a result of cutting the MPFL, the patellar lateral translation increased by 1.1mm±0.3mm in early knee flexion (Figure 4.11). A significant difference in medial-lateral translation was detected between intact and cut MPFL (P=0.002). There was also a significant effect of flexion angle (P<0.001), with significant differences at 0°, 10°, and 30° knee flexion identified in post-hoc t-tests (P<0.05).

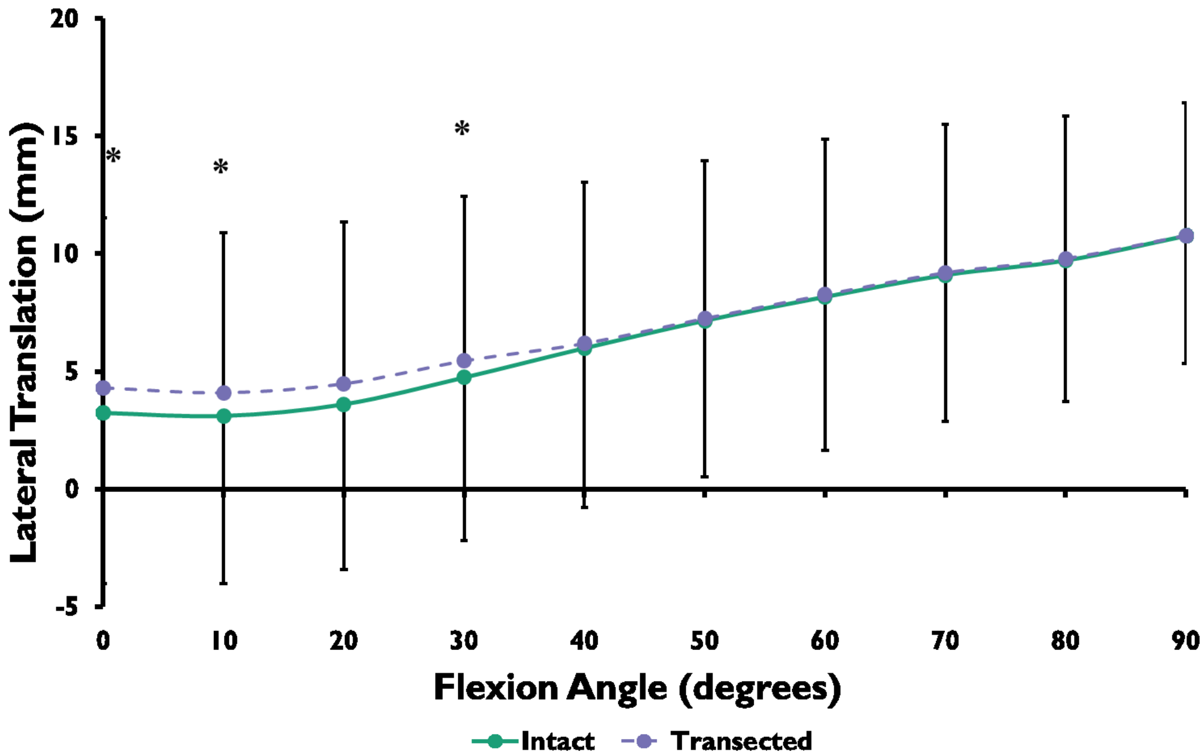

Figure 4.11 Patellar medial-lateral translation (mm; mean ± SD, n=8) from 0°-90° knee flexion in the intact knee and that with the MPFL cut. *P<0.05.

4.8 Discussion

Cutting the MPFL resulted in a significant increase in the distance between the attachment points on the patella and femur, a decrease in peak and mean medial patellofemoral joint contact pressures, an increase in peak and mean lateral patellofemoral joint contact pressures and increases in patellar lateral tilt and lateral translation in early knee flexion. These findings indicate lateralised motion of the patella in early knee flexion following MPFL transection, with a redistribution of loading away from the medial and onto the lateral facet. This confirmed the initial hypotheses. Prior reports have highlighted the role of the MPFL as a patellar stabiliser in early knee flexion (Senavongse and Amis, 2005; Sillanpää et al., 2009b). The tracking data and pressure maps suggest that, with MPFL deficiency, there may be a tendency for the distal-lateral aspect of the patella to collide with the prominent proximal-lateral edge of the trochlea in early knee flexion.

In the intact knee, the patella tracked and tilted medially in early flexion, while it was guided into the trochlear groove, and then laterally to 90°, similar to prior reports (Amis et al., 2006; Merican and Amis, 2009). This matched a rise in medial patellofemoral joint contact pressures in early flexion, before these decreased and stabilised in deeper flexion. Lateral patellofemoral contact pressures increased as the patella engaged in the trochlea to 30° knee flexion before reducing in greater knee flexion.

This study found a significant increase in the distance between the patellar and femoral attachments of the MPFL following transection in early flexion. This was supported by matching lateral patellar translation and tilt. These patellar motion changes are in agreement with a prior report, although of a smaller magnitude. Ostermeier et al. (2007a) identified increases up to 4mm in patellar lateral translation and 4.5° lateral tilt following MPFL rupture, but their test rig loaded the quadriceps axially and did not include the medial force vector of the vastus medialis. Philippot et al. (2012a) reported increases of up to 7.2mm and 7.6° in lateral patellar translation and tilt following MPFL section, but they used a very low quadriceps load of 10N. Another study concluded that MPFL rupture caused no change in patellar motion (Sandmeier et al., 2000), but no statistical analyses were performed and the raw data showed an increase of 2.62mm in lateral patellar tracking following MPFL rupture, hence their conclusions must be interpreted with caution. Differences in the magnitude of the changes of patellar motion following MPFL rupture are likely to be a consequence of different methodologies, including quadriceps tensioning (Farahmand et al., 1998a; Powers et al., 1998), failure to load the ITB (Merican and Amis, 2009) and differences in the co-ordinate systems used and reliability of equipment used to measure patellar motion. It was also noted that the change in distance between the MPFL attachments (4mm) was measured to be longer than expected from the movement of the patella, when the MPFL was cut (1mm translation and 1˚ tilt). It could be hypothesised that the altered mechanics affected other soft tissues around the MPFL. It may be expected that small movements of the patella will cause large changes in contact pressures because the articular surfaces will only be compressed by 1 or 2mm under the loads in these experiments. Therefore 1mm lateral translation plus 1˚ lateral tilt should be expected to increase the lateral contact pressures, and decrease them medially.

This study found significantly increased lateral joint pressures and decreased medial pressures near knee extension following MPFL transection. One other study (Beck et al., 2007) had similar outcomes, measuring contact pressures at 30°, 60° and 90°, but their pressures were somewhat different, possibly due to the lack of ITB tension, or the simplified muscle tension they applied, which causes underestimation of contact pressure at 0° knee flexion and overestimation at 90° (Powers et al., 1998). These findings support the literature outlining the role of the MPFL in early flexion (Baldwin, 2009; Bicos et al., 2007) and support the relationship between patellar tracking and pressure as suggested previously (Wünschel et al., 2011).

It is important to note that these results are likely to have underestimated the role of the MPFL, since in this study it was transected in isolation using a scalpel. This isolated the effect of the MPFL and was likely to have minimised the apparent consequence of MPFL rupture, due to a lack of associated damage to local soft tissue such as tearing of the vastus medialis obliquus which has been reported to occur clinically at the time of patellar dislocation (Panagiotopoulos et al., 2006; Sallay et al., 1996).

This experiment found that loss of MPFL function significantly increased lateral patellar tracking and lateral patellofemoral joint contact pressures, particularly in early knee flexion. This maltracking relates to the role of patellar engagement in patellar stability (Monk et al., 2011). These changes, when considered alongside osteochondral defects reported at the time of dislocation (Arendt et al., 2002; Nomura and Inoue, 2005), may have long term consequences for the articular cartilage and joint function. It is widely hypothesised that abnormal patellar tracking, causes increased contact stresses and subsequent cartilage wear (Dye et al., 1998, Sanchis-Alfonso et al., 2006). This suggests the need for intervention, following initial dislocation in order to reverse the changes in knee kinematics and contact mechanics found in this study. This intervention may be either conservative in the form of strengthening and motor control exercises or surgical, including MPFL reconstruction.

4.9 Key Findings / Conclusion

Key findings from this work therefore are:

1. MPFL transection results in increased distance between patellar and femoral MPFL attachments.

2. It causes increased lateral patellar tracking.

3. It causes increased lateral contact pressures.

4. It results in decreased medial contact pressures.

These kinematic and contact mechanics changes have potential long term consequences for the articular cartilage. These findings can therefore be used as a rationale to perform MPFL reconstruction in patients following patellar dislocation when the clinical presentation of the patient suggests this is indicated.

4.10 Addendum

4.10.1 Patellofemoral Joint Contact Pressure Measurements

4.10.1.1 Which Measurement Tool?

Review of the literature highlights a number of methods which have previously been used to measure articular joint contact pressures including: dyes, computer models, cement casts and pressure sensitive film (Aglietti et al., 1975; Goodfellow et al., 1976; Huberti and Hayes, 1984; Wiberg, 1941). Following consideration, it was decided to use pressure film to assess the joint contact pressures in the current study. This enabled testing to be minimally invasive to the local anatomy and was within both time and expense resources available for the experiment. Two measurement systems were identified as potentially appropriate to use; descriptions and the main advantages and disadvantages of each are described in greater detail below.

Fuji Film Prescale Pressure Measuring System (Fuji Photo Film Co. Ltd., Tokyo, Japan)

- Fuji pressure-sensitive film consists of two sheets. One sheet contains liquid dye filled micro-capsules of differing diameters which rupture at different stress states. The second sheet contains a developer substance which records the dye when the capsules are ruptured.

✓ The use of Fuji Film is a well evaluated technique with a broad range of literature existing to support its use (Bachus et al., 2006).

✓ It is appropriate to position in the patellofemoral joint and measure joint contact mechanics, as prior studies have shown.

✗ Cannot perform repeated measurements, single use film. Therefore for the current study ten sheets would be required just for testing one knee through full range of knee flexion. This would make the study both time consuming and expensive.

✗ The film permanently discolours if its surfaces are depressed. This can happen when the film is being inserted into smaller spaces and needs to be temporarily depressed (such as the synovial pouch of the patellofemoral joint).

✗ The film does not provide real time data during measurements.

✗ The system has low sensitivity and validity (Harris et al., 1999).

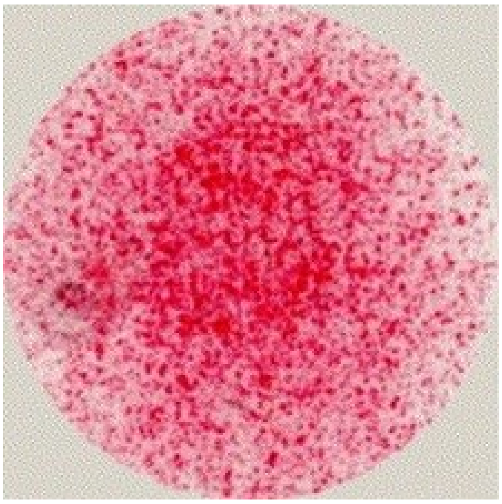

Figure 4.12 Example of the output from Fuji film following compression of the film in an Instron materials testing machine. The darker red areas correspond to greater pressures and the lighter areas correspond to areas of lesser pressure (as adapted from Bachus et al. (2006)).

Tekscan system (I-Scan™, Tekscan, Inc., Boston, MA)

- The Tekscan film consists of two polyester sheets which have patterns of electrical conductor deposited on them. The sheets are bonded together with a layer of semi conductive ink separating them. Sensing elements (sensels) are present where the line of electrical conductor on one sheet crosses that on the other. When force is applied to the sensels, the ink layer is compressed, resulting in a change in electrical resistance which is recorded with specialised real-time software via a connected handle.

✓ The system provides real time images on screen.

✓ The measurements are instantly repeatable, allowing for multiple tests to be carried out without the need for the sensor to be repositioned in between tests.

✓ The Tekscan system has been reported to have greater reliability and reproducibility of measurements reported, compared to Fujifilm (Bachus et al., 2006; Harris et al., 1999).

✗ The system lacks an explicit well tested outline of a technique to calibrate the sensors.

✗ This technique has been evaluated less than Fujifilm. It is more modern and has thus has only come into more common use in recent years.

✗ The sensors are easily damaged by shearing or creasing, and are expensive to replace.

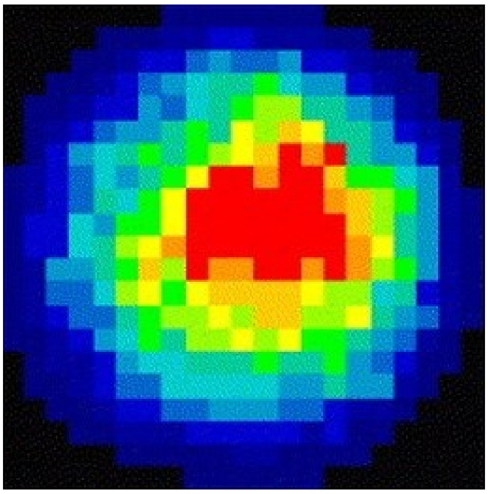

Figure 4.13 Example of the output from Tekscan film following compression in the Instron machine. The red area corresponds to a region of greater pressure, and other areas: yellow-green-blue correspond to progressively reduced pressures (as adapted from Bachus et al. (2006)).

Following review of the various pros and cons of each system it was decided to use the Tekscan system for the current experiment. This was primarily based on the ability of the Tekscan sensor to perform multiple measurements without the need to replace the sensor between individual tests. Given the number of different measurements taken in the current study this was critical. In addition its reported improved reliability and validity made it the logical choice for current study use.

4.10.1.2 Selection, insertion and fixation of the pressure film

The literature highlighted that two different Tekscan sensors had been used for previous experiments measuring patellofemoral joint contact pressures (Beck et al., 2007; Garretson et al., 2004). These were the 5051 sensor and the 4000 sensor (Figure 4.14). A discussion with a local representative from Tekscan and review of different sensor dimensions and shapes available from their website confirmed that these were the most appropriate selections to measure patellofemoral joint contact pressures. It was planned to pilot test both sensors in a test knee to determine which would be best suited to the experiment. However when first test specimen was examined, it was apparent that the 4000 sensor was not large enough to cover the surface area of the patella and trochlear groove. It was determined therefore that the 5051 sensor should be used for all testing instead, since it comfortably covered the surface area of the patella and trochlea.

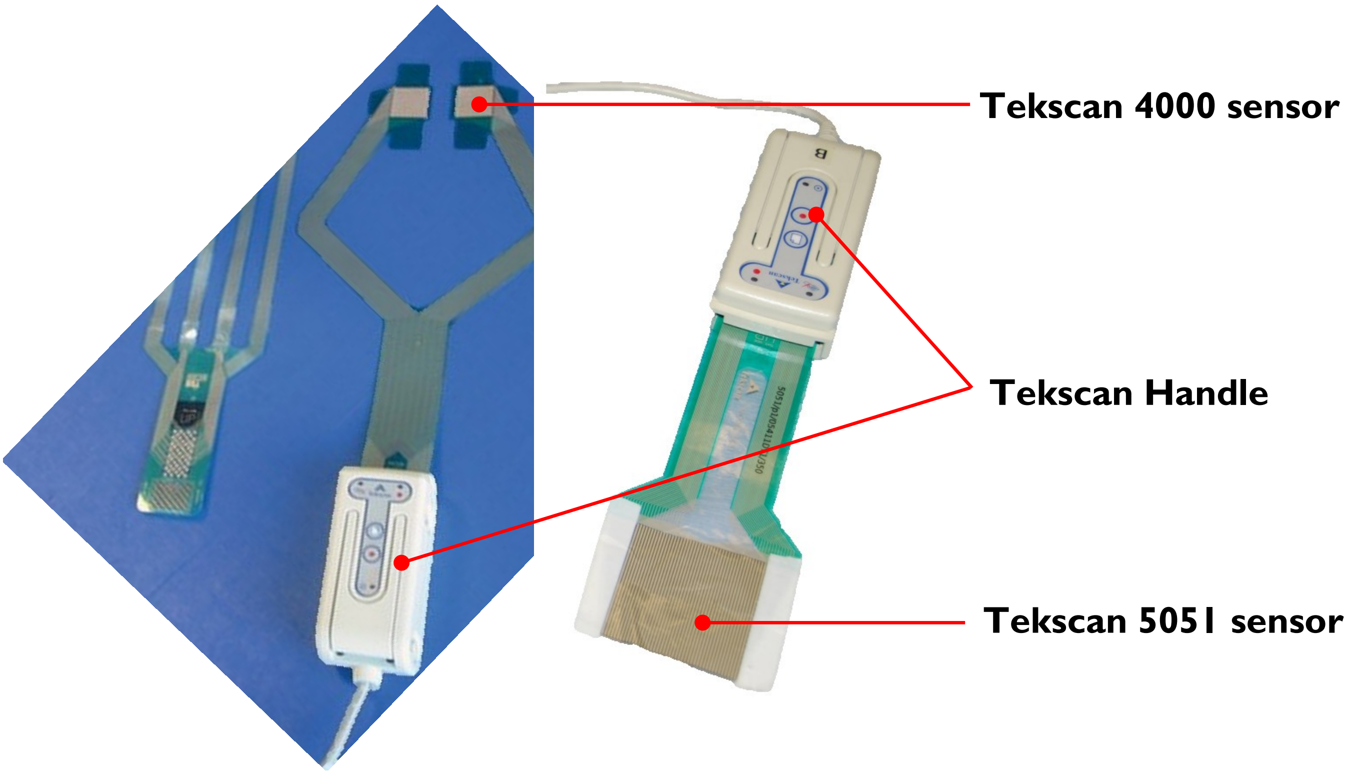

Figure 4.14 Tekscan 4000 sensor (left side) and 5051 sensor (right side) attached to the Tekscan handle.

Careful consideration was given to determining the method used to introduce the Tekscan sensor to the patellofemoral joint cavity. Unfortunately the nature of the task meant it was inevitable that some local soft tissues surrounding the patellofemoral joint would need to be cut. Logically there was the choice of three entrance points for the sensor into the joint:

1. Proximally; anterior to the femoral shaft, but deep to the quadriceps muscle.
2. Laterally; through an incision in the lateral retinaculum.
3. Medially; through an incision in the medial retinaculum.

Clearly the medial approach would require dissection of the MPFL to access the joint, therefore this was instantly discounted. This left either the lateral or proximal approach as options. The lateral approach had been performed and reported on previously (Beck et al., 2007; Garretson et al., 2004), nevertheless given the role of the lateral retinaculum in patellofemoral joint mechanics it was concerning that opening the joint in this area would require cutting the transverse fibres of the iliotibial band which would impact on patellar kinematics and contact mechanics (Merican and Amis, 2009). However the alternative of approaching the joint from proximally meant it would be hard to position the Tekscan handle, since the quadriceps were loaded proximal to the patella where the handle would be positioned. A pilot study was therefore undertaken to examine both methods and determine the better of the two.

As expected the lateral approach opening required to accommodate the sensor was substantial, and resulted in weakening of the muscle and fascial attachments of the VLO and ITB to the patella. This was concerning since the significant weakening of the soft tissues left them vulnerable to breaking when they were tensioned during the experiment. Alongside this, given the length of the opening required, it was evident on visual observation that patellar mechanics altered following the incision. Conversely the proximal approach meant only the proximal patellofemoral joint capsule had to be opened, since the vastus intermedius was elevated from the femur anyway to permit quadriceps loading. After a number of practice tests it was evident that if the Tekscan sensor was inserted at a slight angle, it was possible for the tail of the sensor and the Tekscan handle to exit the joint between the VMO and VML muscles (Figure 4.15). It was therefore decided to use this technique in order ensure any potential compromise to patellofemoral joint kinematics and mechanics was minimised throughout testing.

Two options were available to ensure secure fixation of the sensor once positioned in the patellofemoral joint cavity. One was to glue the sensor into position and the other was to suture it in position by stitching it to the local soft tissues. It was concerning that fixing the sensor to one of the joint surfaces with glue would result in shear stresses which could not be directly accounted for. In addition since the Tekscan does not conform to the surface geometry of the patella or femur, the sensor would be creased when glued to the surface, inevitably increasing measurement error. Therefore it was decided to suture the sensor into position using the technique outlined in section 4.3.3 in Chapter 4.

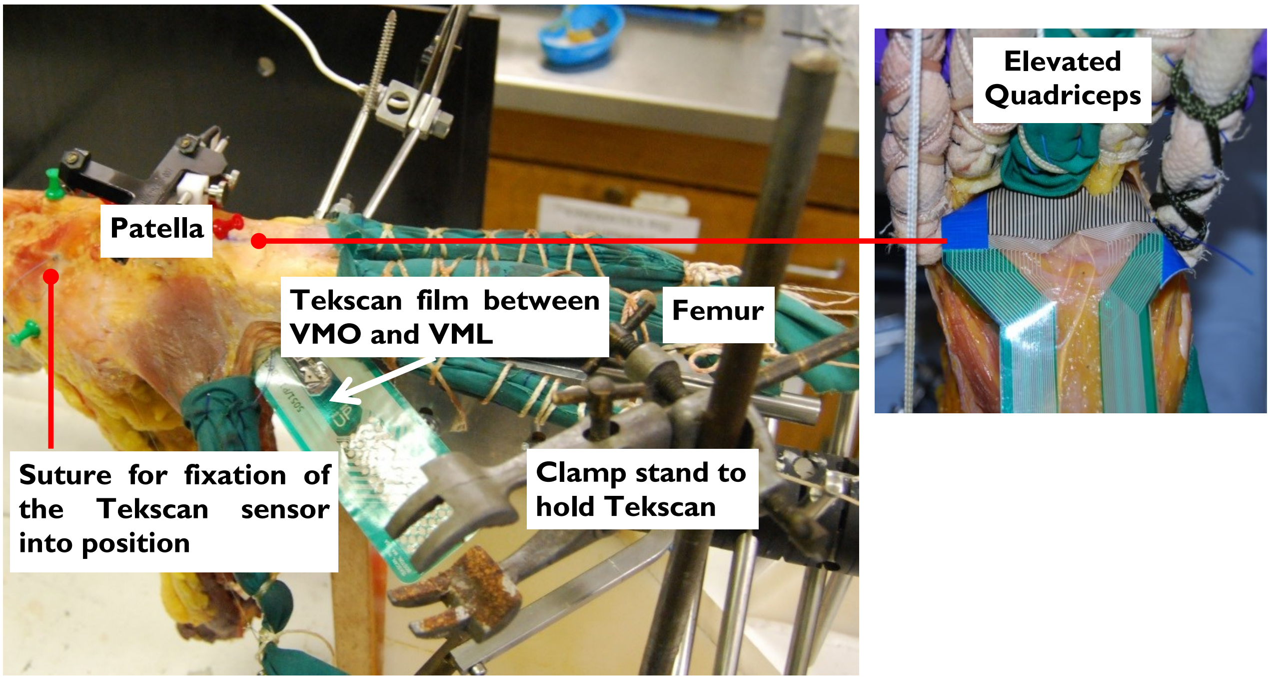

Figure 4.15 The Tekscan sensor was inserted into the patellofemoral joint through the proximal joint capsule to be situated between the quadriceps muscles and the anterior femur with the tail exiting the joint cavity between the VMO and VML muscles as shown in the left image.

4.10.1.3 Calibration and sensor range chosen

In order for the Tekscan software to convert resistance to pressure units, the sensors had to be equilibrated and calibrated prior to use. Tekscan offer thirteen different 5051 sensors reading a range from 0.48-172.38MPa (7 – 25,000 PSI). It was therefore necessary to have an estimation of the maximal pressure which was going to be measured during the experiment in order to determine the sensor most appropriate for use. Prior work reported maximal patellofemoral joint contact pressures in the region of 2MPa generated from cadaveric testing (Beck et al., 2007). Three sensors reading 150 PSI, 250 PSI and 350 PSI were therefore chosen to pilot test initially based on this pressure. The aim was to determine the most sensitive and specific set up for testing i.e. that in which even low levels of pressure between the patella and trochlea are detected and all pressures are accurately read.

The sensors were conditioned, equilibrated and calibrated using a materials testing machine (Instron 5585, High Wycombe, Bucks, UK), as described in Chapter 4 section 4.3.3, and in accordance with manufacturer guidelines. They were compressed between two sheets of 3mm thick silicone, with an elastic modulus of 0.7 MPa to simulate the compliance provided by patellofemoral joint cartilage (Donzelli et al., 1999; Drewniak et al., 2007). Ten pre-conditioning cycles loading the sensor to 3.2kN (creating a pressure 20% greater than the maximal 2MPa pressure anticipated) were performed initially on all sensors to condition them. The sensors were then equilibrated to 10%, 50% and 90% of the anticipated 2MPa pressure. Therefore 0.32 kN, 1.60kN and 2.85kN forces were applied to the sensors. Finally the sensors were calibrated using a two-point power-law calibration loading each sensor with 20% (0.64kN) and 80% (2.56kN) of the anticipated load.

Following conditioning, equilibration and calibration each of the three sensors were used to measure patellofemoral joint contact pressures, with the sensor inserted and fixed into position as described in section 4.8.1.2. Joint contact pressures were measured during knee flexion at 0°,10º, 30°, 60° and 90°. When readings obtained with the 350 PSI sensor were compared to those from the other sensors it was evident the contact area was much smaller indicating the 350 PSI sensor did not pick up lower pressure levels between the patella and trochlea. Between the other two sensors: the 150 PSI and 250 PSI, there was little subjective difference between the readings collected as on screen image captures. However analysis of the data indicated the 150 PSI sensor was more sensitive at detecting the lower patellofemoral joint pressures and was also not over-saturated by the peak pressure. Therefore it was decided to use 150 PSI sensors for the experiment.

During pilot testing it was identified that it was possible to vary the sensitivity of the sensors further. Within the Tekscan software the sensor could be set to be sensitive to read low pressure levels (Low-3, Low-2, Low-1), to read a default pressure, or to be less sensitive to reading lower pressures (High-1 High-2 and High-3). Unfortunately Tekscan do not provide clear guidelines as to exactly what effect altering these settings has on the readings provided by the sensors. Even following direct personal correspondence with the company the answers to questions raised regarding sensitivity settings were not clear. During the pilot study therefore with the sensor in place the effect of altering the sensitivity level from low to default to high was investigated. It was evident that setting the sensitivity to Low-1, 2 and 3 levels resulted in increased detection of the lower contact pressures at the outer borders of the patellofemoral joint, which were not detected when the sensitivity was set to High-1, 2 or 3. There was little difference between the Low 1, 2, or 3 settings, but it was evident that setting the sensitivity of the sensor to Low was appropriate for measurements involving smaller contact areas which required precise measurement (such as in the current study). Meanwhile setting the sensitivity to High would be more appropriate for more gross measurements when sensitivity is not so critical. For the purpose of the present study therefore the sensitivity of the sensors was set to Low-1, in order that changes in lower joint contact pressures could be detected.

4.10.1.3.1 Reliability

Following refinement of the Tekscan measurement protocol as outlined, the reliability of the system was examined using a test re-test methodology. A brand new 5051 Tekscan sensor was conditioned, equilibrated and calibrated as previously described. One knee was prepared for testing as outlined in section 4.3 and a Tekscan sensor inserted. The knee was then flexed from 0°-90° and joint contact pressures recorded at six flexion angles (0°, 10°, 20°, 30°, 60° and 90°) using the Tekscan pressure film. This was repeated three times. The film was then carefully removed and wiped dry, and then re-inserted and the measurements repeated. Data was analysed as outlined in section 4.4.1 using a custom written MATLAB script (full detail: Appendix A). Mean and peak medial and lateral joint contact pressure data from each experiment were compared across the six different angles of knee flexion. Unfortunately in order to preserve the sensors and due to financial restraints it was only possible to repeat this test in one knee rather than all eight knees.

The system was determined to have, at worst, a test re-test difference of between 0.03 ± 0.2MPa (for mean medial pressure) – 0.3 ± 0.3 (for peak lateral pressure). These findings were within limits of prior reported work in the area and determined to be acceptable for the present study (Wilson et al., 2003).

4.10.2 Patellar Tracking Analysis

4.10.2.1 Measurement System

Patellar tracking refers to the motion of the patella in relation to the femur as the knee flexes and extends. In clinical research, non-invasive assessment of joint kinematics is commonly undertaken with motion capture systems such as Vicon (Oxford Metrics, Oxford, UK) (Troje, 2002). However, due to issues of adipose tissue and estimation of the joint centre these systems can demonstrate low reliability (Pohl et al., 2010). More recently measurement techniques using fluoroscopic and MRI measurements have been developed and implemented to permit more valid *in vivo* measurement of lower limb kinematics (Logan et al., 2004; Tashman, 2008). These systems are growing in popularity; however *in vitro* experimentation has advantages when measuring subtle movements of the patella, the main one being direct attachment of measurement trackers to the patella, femur and tibia. These measures can also be undertaken *in vivo* however they are invasive procedures and require the patient to be anaesthetised, therefore inhibiting quadriceps function and limiting the assessment of patellar tracking.

In the laboratory at Imperial College London two systems were available which were appropriate to measure patellar kinematics in the context of the present study. These were an electromagnetic system: the Flock of Birds (Ascension Technology, Burlington, VT) and an optical tracking system: Polaris Tracking (Northern Digital Incorporated, Waterloo, Canada) with active optical trackers (Traxtal Technologies, Toronto, Canada). A summary of the advantages and disadvantages of each system is listed below:

Polaris Optical Tracking (Northern Digital Incorporated, Waterloo, Canada) (Khadem et al., 2000; Merican et al., 2011)

- ✓ The system has previously been used by other investigators in the same test rig as that planned to be used for the current system. The system has been reported as a valid and reliable measure of patellar tracking (Merican et al., 2011).
- × The trackers need to be in sight of the tracker camera through full range of knee flexion. During crowded experiments, when measuring multiple outcomes simultaneously this can be challenging.

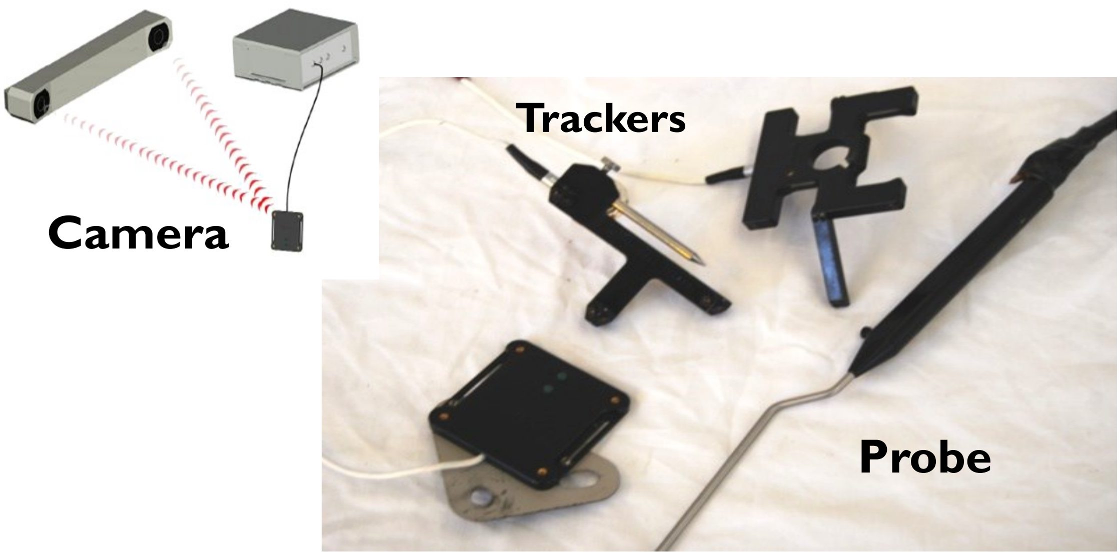

Figure 4.16 Polaris optical tracking system (left image) alongside probe (Northern Digital Incorporated, Waterloo, Canada) with active optical trackers (Traxtal Technologies, Toronto, Canada) (right image).

Flock of Birds (Ascension Technology, Burlington, VT) (Amis et al., 2006)

- ✓ The system has been used by this research group previously for investigations looking at patellofemoral joint tracking.
- × The system is susceptible to distortion from nearby metal sources and has lower accuracy reported compared to optical tracking systems (Koivukangas et al., 2013).

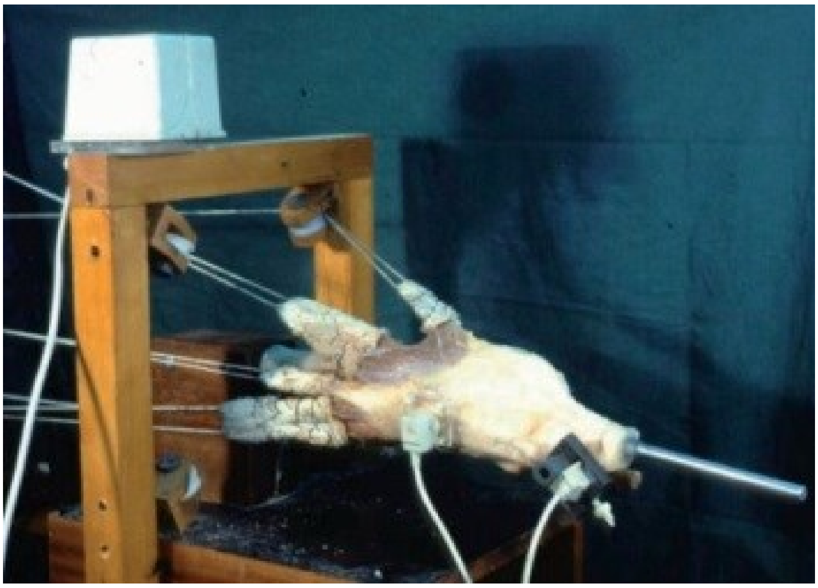

Figure 4.17 Right sided knee mounted on its side, with an electromagnetic transmitter shown top left and motion sensors attached to the patella and distal tibia (as adapted from Amis et al. (2006)).

Following review of each system in the laboratory and considering the pros and cons outlined for each, it was decided to use the optical tracking system to assess patellar kinematics during the present experiment. This was primarily based on the fact that part of the test rig planned to be used for the experiment was made of metal which could interfere with the electromagnetic measurements if the Flock of Birds system was used. Furthermore optical tracking was widely supported in the literature with its main restriction related to line of sight problems. However this was not noted as an issue in a prior study using optical tracking to measure patellar motion in the same test rig, and so this was not thought a significant disadvantage (Merican et al., 2011).

4.10.2.2 Optical Trackers

Three optical trackers were used to track the motion of the patella, tibia and femur. It was necessary to fix a tracker to each bone and so three different mounts were devised, one for each tracker. The mounts ensured no movement was permitted between the tracker and bone or the tracker mount and tracker. Mounts were specifically designed to mount the patella and femoral trackers (Figure 4.18 & Figure 4.19). The tibial tracker was mounted using Brainlab fixation (Brainlab AG, Feldkirchen, Germany) (Figure 4.20). The mounts allowed some flexibility in their location when attaching them to bone. This meant they could be adjusted initially until it was ensured that they were visible through the full range of knee flexion. It was confirmed during fixation that no movement was permitted between the tracker mount and bone surfaces or the tracker and tracker mount. In addition the mounts standardised the tracker position in relation to the bone, thus enabling the trackers to be removed from the specimens during the experimentation and ensuring no change in the tracker position when they were reattached.

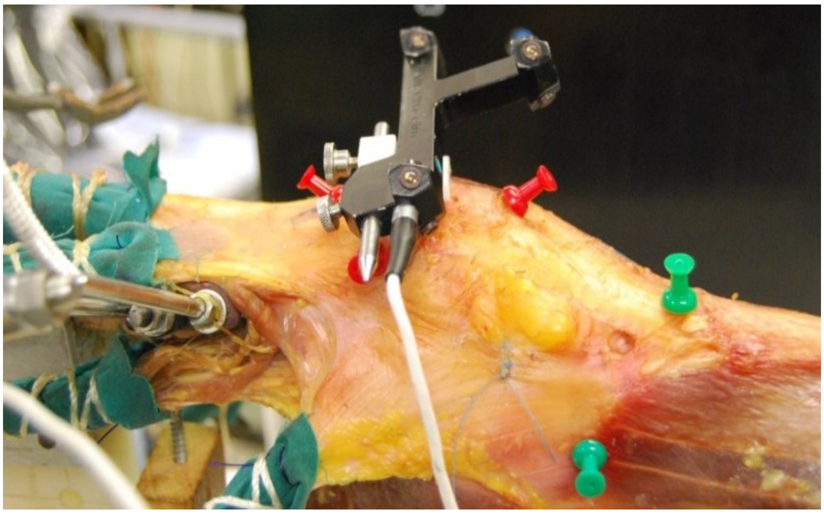

Figure 4.18 The tracker used to record patellar motion fixed with a custom made mount to the patella.

The tibial tracker was used to measure knee flexion, with zero defined as the knee in full extension, with no hyperextension of the tibia. The femoral tracker was used as a reference, in case the moving loads during knee flexion-extension caused bending deflection of the femur.

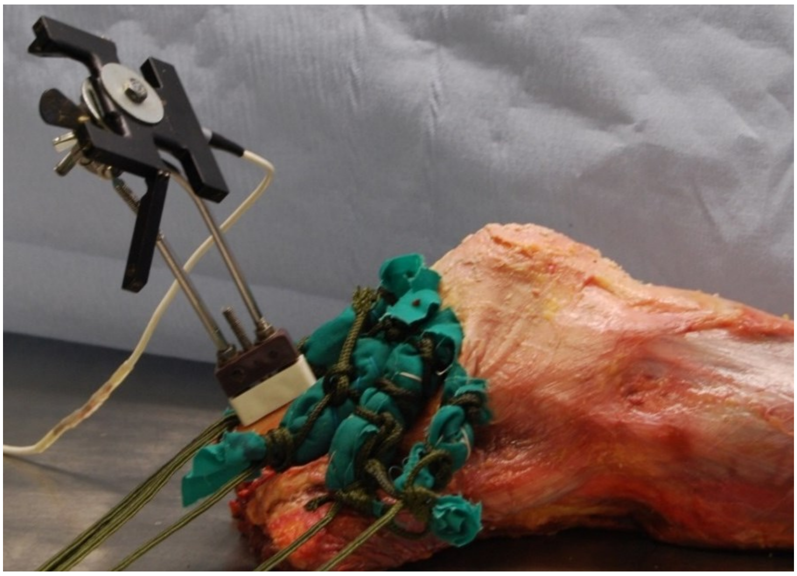

Figure 4.19 The tracker used to record femoral motion fixed with a custom made mount to the femur.

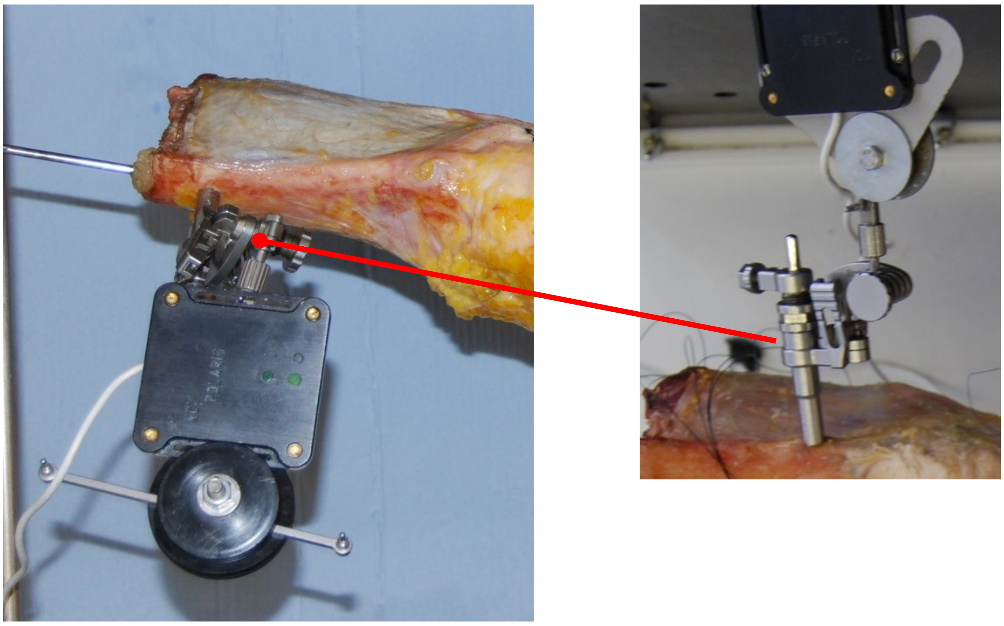

Figure 4.20 The tracker used to record tibial motion, fixed using adapted Brainlab attachments to the tibia (left image). Right image shows the Brainlab mounting in greater detail.

4.10.2.3 Patellar Motion

In order to describe 3D motion of the patella, its movements must be referenced to another bone. This is typically the femur, although the tibia has been used in previous studies (Reider et al., 1981; Stein et al., 1993). However use of the tibia as a reference has been criticised as inaccurate due to tibial internal rotation that occurs in relation to the femur in terminal knee extension (Katchburian et al., 2003). The purpose of this study was to examine patellofemoral joint kinematics so it was decided that it was appropriate to define the motion of the patella in relation to the femur. The axes and origins of the coordinate systems of both the patella and femur therefore required to be defined. Clear description of the coordinate system of the bone is critical to ensure correct interpretation of study results. Researchers have conducted the same experiment using different coordinate systems and identified widely different tracking patterns from each system (Blankevoort et al., 1996; Bull et al., 2002).

The proximal-distal axis of the femur has been defined as each of; the anatomical, mechanical and trochlear groove axis by different researchers (see detail Chapter 2), with the use of each offering differing advantages and disadvantages. The choice of this axis is crucial since it can influence the magnitude and perceived direction of motion (Bull et al., 2002). Clearly the femoral groove axis as a reference for patellar motion is useful since it defines a true spatial relationship of the patellofemoral joint. However there are challenges to access the trochlear groove in the laboratory setting, without causing soft tissue trauma, since MRI imaging is not available. Deriving the groove axis from the anatomical axis has been found to be imprecise since their relationship is variable (Feinstein et al., 1996). Furthermore the patellar groove can be abnormal (Dejour et al., 1994) and is not linear, limiting its use as a reference axis (Barink et al., 2006). Therefore the anatomical axis was chosen for use in the present experiment since the trochlear axis could not be exposed without interfering with local soft tissues and the mechanical axis could not be accurately defined since specimens did not have the femoral head and neck attached. The medial-lateral femoral axis meanwhile can be defined by the line that is parallel along the most posterior aspect of both femoral condyles (Amis et al., 2006), the line that joins the lateral and medial femoral condyles (Pennock and Clark, 1990) or in reference to a line connecting the two spheres fitted to each condyle (Kwak et al., 2000). Since the shape of the femoral trochlea is often abnormal in patellofemoral patients (Dejour et al., 1994) the posterior condyles were chosen for use in this experiment as they were thought to be the least susceptible to variation, thus increasing the reliability of femoral positioning (Fulkerson et al., 1987).

The patellar axes have not been examined in the literature to the depth of the femoral axes. This is a likely consequence of the challenges in identifying standardised anatomical landmarks on the patella for reference. Prior researchers have used a combination of the most prominent medial / lateral /

proximal or distal edges of the patella to define the axes (Ahmed et al., 1999; Goh et al., 1995; Reider et al., 1981). An alternative method described has been to determine the centre of the patella and then defining a coordinate system for it based on the femoral position with the knee in full extension (Koh et al., 1992; van Kampen and Huiskes, 1990). The former method of determining landmarks has the advantage of being anatomical; however there are issues in the subjective identification of points on the patella in the absence of imaging and distinctive landmarks. The patella is small, meaning that even minor inaccuracies can result in relatively large errors. Furthermore the points have to be located through skin and do not necessarily have a direct relationship with articular geometry. The latter method of defining the patellar coordinate system based on the femoral position also has disadvantages. Mainly it zeroes the angular relationship between the patella and femur in extension. Thus it will not detect patellofemoral joint pathologies in terminal extension since it assumes the patellofemoral joint is in correct alignment in this position, thus variation of tilt from knee to knee cannot be appreciated nor can the normal anatomical alignment of the knee. Furthermore, since the initial position of the patella is highly dependent on the loading applied to the knee, it does not permit direct comparison of results from studies using different patellar loading methods. Alternative methods using x-rays and tantalum markers to define anatomical landmarks are defined in the literature, but these again are reliant on subjective estimation of patellar landmarks (Veress et al., 1979). After consideration, it was decided to use the first option described and define the patellar axes in relation to defined anatomical landmarks on the patella. This was determined since the MFPL was known to have its most important role during the terminal thirty degrees of extension and thus it was critical to have information relating to the patella in this position. Additionally it was recognised this this would make comparison of present study results to other literature easier in the lack of an accepted gold standard of muscle loading.

4.10.2.4 Description of Patellar Tracking

Following definition of the femoral and patellar axes, patellar tracking can be determined during knee motion. This is possible using two different mathematical methods. Firstly motion can be defined in reference to the femoral body fixed axis. Clinicians can find this terminology confusing. For example in knee extension, rotation of the patella using this axis would be termed patellar tilt in the clinical setting, but then at 90º flexion the same movement about the femur would be clinically defined as rotation. To help prevent confusion, a second system has instead been proposed. This refers to motion in relation to a combination of the femoral and patellar body fixed axes. This is known as the Joint Coordinate System (Grood and Suntay, 1983) and has been widely used to define tibiofemoral motion and more recently adapted for use with the patellofemoral joint (Bull et al., 2002). This system defines patellar tilt as the spinning motion of the patella around its own long axis. Meanwhile patellar translation is defined in reference to the medial-lateral femoral axis, as is patellar flexion. This latter reference system was used during the analysis and reporting of present study findings.

Based on recommendations in the literature the direction of patellar motion must be clearly defined to ensure correct interpretation of data (Bull et al., 2002). For the purpose of all analysis in this thesis, lateral patellar tilt, translation and rotation were taken to be positive and patellar position is related to flexion angle throughout (Figure 4.21).

4.10.2.4.1 Static or Dynamic

Two methods can be used to measure patellar motion; static and dynamic. Each method has advantages and disadvantages, with no consensus regarding the gold standard agreed. Dynamic measurements are collected continuously at a set frequency throughout full range of knee flexion. They offer the clear advantage of ensuring any disturbances in short arcs of flexion are not missed (Katchburian et al., 2003). However data from such experiments can be complex to analyse and relate to exact angles of knee flexion. Further problems arise when multiple outcome measures are collected from a study simultaneously, such as the pressure and LVDT readings in the present study, since the systems are independent and therefore relating the measurements to set time periods can lead to inaccuracies. Instead measurements therefore were taken at progressive 10º intervals in the present study. The main criticism of this method is that measures of the static patella can be misleading. However since the knee was loaded in the rig throughout testing this was not a problem in the present study, and these concerns are generally more relevant to clinical settings where static measurements may fail to account for loss of muscle tension in unloaded positions (Brossmann et al., 1993).

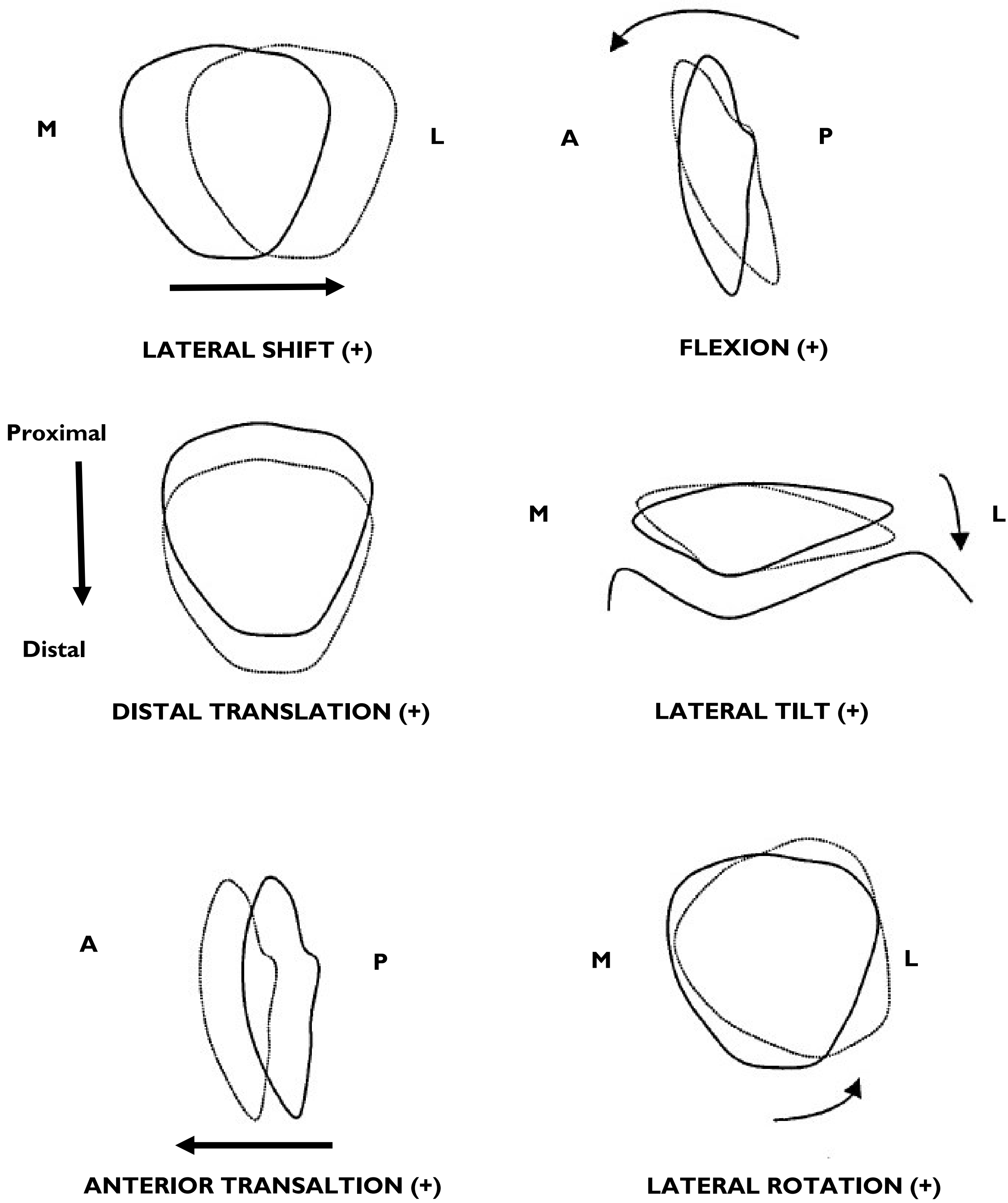

Figure 4.21 Definition of patellar motion for the current thesis (based on a left sided knee).

4.10.3 Optical Tracking Set-Up

Metal screws were fixed to the following bony landmarks to enable standardised digitising of local bony landmarks across all specimens. Three landmarks on each bone were digitised using the Traxtal probe stylus:

1. **Femur**: anatomical medial epicondyle, lateral epicondyle and proximal end of the femoral rod (the centre of this was drilled with a small hole large enough for the probe tip to lock into, prior to the rod being cemented into the femur).

2. **Patella**: centre of the medial patellar border, centre of the lateral patellar border and distal pole of the patella.

3. **Tibia**: centre of the medial tibial joint line, centre of the lateral tibial joint line and distal end of the tibial metal rod (a pre drilled hole was positioned in this, similar to that described for the femur).

The end of the digitising probe fitted tightly into the top of the metal screws. A custom written MATLAB script, available from a current PhD student's work, was then used to process the recorded anatomical landmarks and raw tracking data to obtain patellar motion in relation to the femur. The patellar and femoral body fixed axes were defined from the anatomical landmarks, hence, the origins of the axes systems were at about the central point of the patella and mid epicondylar axis, respectively. The axes systems and raw tracking data were then processed using series of transformation matrices (full detail: Appendix B). This processing method had been previously verified against measurements recorded from a 'wooden knee' positioned and measured in the experimental test rig used in the present study.

4.10.3.1 Reliability

Once the kinematic measurement protocol was finalised, the reliability of the system was examined using a test re-test methodology. One knee was prepared for testing as outlined in section 4.3. Once it was positioned in the test rig, the trackers were attached to the patella, tibia and femur and bony landmarks digitised as described in section 4.9.3. Kinematic data was then recorded from 0°-90° knee flexion (measurements taken at 0°, 10°, 20°, 30°, 60° and 90°). The measurements were repeated three times to allow the mean to be calculated. The knee was removed from the test rig and the trackers removed. Exactly the same test was then repeated, with the knee positioned back in the test rig, the trackers again attached to the tibia, femur and patella and the bony landmarks digitised. A second set of kinematic data was then recorded from 0°-90° knee flexion, with measurements repeated three times. This whole process was undertaken on two different knees in turn.

Data was analysed as outlined in section 4.5, using the MATLAB script (Appendix B). Patellar translation and tilt data recorded from the two different testing sessions were then analysed using a Fishers Original ICC test in SPSS. Patellar tilt was found to have an ICC of 0.89 and patella translation an ICC of 0.9. The high ICC values suggest excellent reliability (excellent ICC > 0.9) of the system and are in line with previous reports (Wiles et al., 2004).

4.10.4 Conclusion

This study utilised a Tekscan 5051 sensor inserted via a proximal incision in the patellofemoral joint capsule to assess joint contact pressures. It was fixed into position with sutures to the local soft tissues to prevent movement. Sensors were conditioned, equilibrated and calibrated prior to use. Patellar motion was examined using an optical tracking system and patellar motion was defined in relation to the femoral and patellar axes, with bony landmarks digitised to set up coordinate systems of each bone as defined. Patellar lateral tracking, tilt and rotation were all taken to be positive.

CHAPTER 5

MEDIAL PATELLOFEMORAL LIGAMENT RECONSTRUCTION

5.1 Background

The anatomy and length change behaviour of the MPFL examined in Chapter 3 has highlighted attributes which support its role in patellar stability. Prior work adds weight to these deductions, with the MPFL recognised to be a primary patellar stabiliser in early knee flexion (Senavongse and Amis, 2005). The MPFL is known to be injured or ruptured during every patellar dislocation (Guerrero, 2009) and the previous chapter highlights the consequence of MPFL transection in terms of altering patellar contact mechanics and kinematics, findings supported in the literature (Philippot et al., 2009; Steiner et al., 2006). Consequently, MPFL reconstruction following patellar dislocation is becoming a popular surgery undertaken worldwide. Clinical outcomes following MPFL reconstruction have generally been positive at short and midterm follow up, with low re-dislocation rates and good functional outcomes reported (Drez Jr et al., 2001; Ellera Gomes et al., 2004; Nomura et al., 2007; Schöttle et al., 2005). However, there are reports of patients who suffer recurrent dislocation, pain and poor function following reconstruction, and later require revision surgery (Bollier et al., 2011; Wünschel et al., 2011). Adverse outcomes may result from non-anatomical femoral tunnel positioning (Camp et al., 2010; Shah et al., 2012) or grafts being over-tensioned during surgery (Elias and Cosgarea, 2006; Thaunat and Erasmus, 2009; Wünschel et al., 2011). Both of these complications are hypothesised to alter patellar kinematics and joint contact pressures.

The location of the femoral attachment of the MPFL has been widely debated (Baldwin, 2009), however an anatomical femoral tunnel position, confirmed radiographically, has been proposed in the literature for use during MPFL reconstruction (Redfern et al., 2010; Schöttle et al., 2007; Stephen et al., 2012). Despite this, determining the location of the optimal femoral tunnel position continues to be a challenge. Servien et al. (2011) examined twenty-nine MPFL reconstructions post-operatively and determined nineteen to be correctly positioned in accordance with Schöttle et al. (2007), while ten were either proximal or anterior to the anatomical position. A recent retrospective report meanwhile found 64% of femoral tunnels in a population of fifty MPFL reconstruction patients were non-anatomically positioned (McCarthy et al., 2013). Non-anatomical femoral attachment position during MPFL repair has been shown to be the only significant risk to surgical failure (Camp et al., 2010). Symptoms of ongoing pain, recurrent dislocation and joint degeneration were reported in a case series of five patients, who each required revision surgery as a consequence of non-anatomical femoral tunnel position in MPFL reconstruction (Bollier et al., 2011). This finding is substantiated by reports of ligament length change patterns from cadaveric studies which have shown proximally positioned tunnels to result in MPFL elongation (or stretching) in knee flexion, with distal tunnels resulting in ligament elongation in knee extension (Smirk and Morris, 2003; Stephen et al., 2012).

Presently there is limited data to suggest a graft tensioning protocol for use during MPFL reconstruction surgery. Prior research has suggested 2N (Beck et al., 2007) to provide sufficient tension to restore patellofemoral joint contact pressures, whilst patellofemoral joint tracking was found to be restored with 10N tension by others (Philippot et al., 2012b). Reconstruction should find a compromise between too slack a graft, potentially permitting patellar subluxation, versus over-tensioning which may increase medial joint contact pressures through increased forces in the graft as a result of excessive graft shortening (Sherman et al., 2012). The MPFL slackens in early flexion and is almost isometric through knee range of motion once the patella has engaged with the trochlea (Ghosh et al., 2009; Stephen et al., 2012), suggesting it to possess minimal tension. Therefore it would be anticipated, and has been suggested in prior investigations, that over-tensioning could lead to increased medial patellofemoral joint contact pressures (Beck et al., 2007; Elias and Cosgarea, 2006). The optimal knee flexion angle for graft fixation has not been reported previously, with technical notes in the literature suggesting a range of angles are used by surgeons currently (Mikashima and Kobayashi, 2006). It could be hypothesised that graft fixation with the knee in a position where the patella is constrained in the trochlear groove, with the MPFL in a lengthened position, would be optimal to restore normal joint kinematics, although there is no direct evidence to support this theory.

5.2 Aims

Numerous studies outline techniques for MPFL reconstruction (Nomura and Inoue, 2003; Schöttle et al., 2005; Steiner et al., 2006), however there remains a lack of rigorous scientific evidence to support these. The aim of the present study was to determine the optimal technique to reconstruct the MPFL to replicate intact ligament behaviour, considering three variables. Noting previous findings it was hypothesised:

1. That MPFL reconstruction would restore patellar tracking and contact pressures to intact values.

2. That non-anatomical femoral attachments would fail to restore intact joint contact mechanics and kinematics.

3. That graft over tensioning would lead to increased medial patellofemoral joint contact pressures, decreased lateral patellofemoral joint pressures and increased medial patellar translation and tilt.

4. That graft fixation at terminal extension would have a significant effect on patellar mechanics and contact mechanics.

5.3 Clinical Relevance

This chapter contains findings most relevant to patients suffering acute, traumatic patellar dislocation in the presence of otherwise normal anatomy in the knee. These findings do not relate to the group of patients who have abnormal anatomy in the paterllofemoral joint, as discussed in Chapter 2, including trochlear dysplasia, patella alta or elevated TT-TG distances. These are examined in greater detail in Chapters 7 and 8.

5.4 Materials and Methods

5.4.1 Specimen Preparation

Ethical approval for this study was granted from the local Research Ethics Committee. In order to conserve resources, in particular cadaveric specimens, the same specimens described in the previous chapter were used for testing in the present study. This experiment directly followed on from the prior work in Chapter 3. Therefore knees were already positioned in the test rig, with the muscles dissected and loaded as outlined in section 3.4.2. The Tekscan sensor and optical trackers were also in position and ready for testing to commence as described in Chapter 4 and shown in Figure 5.1 below.

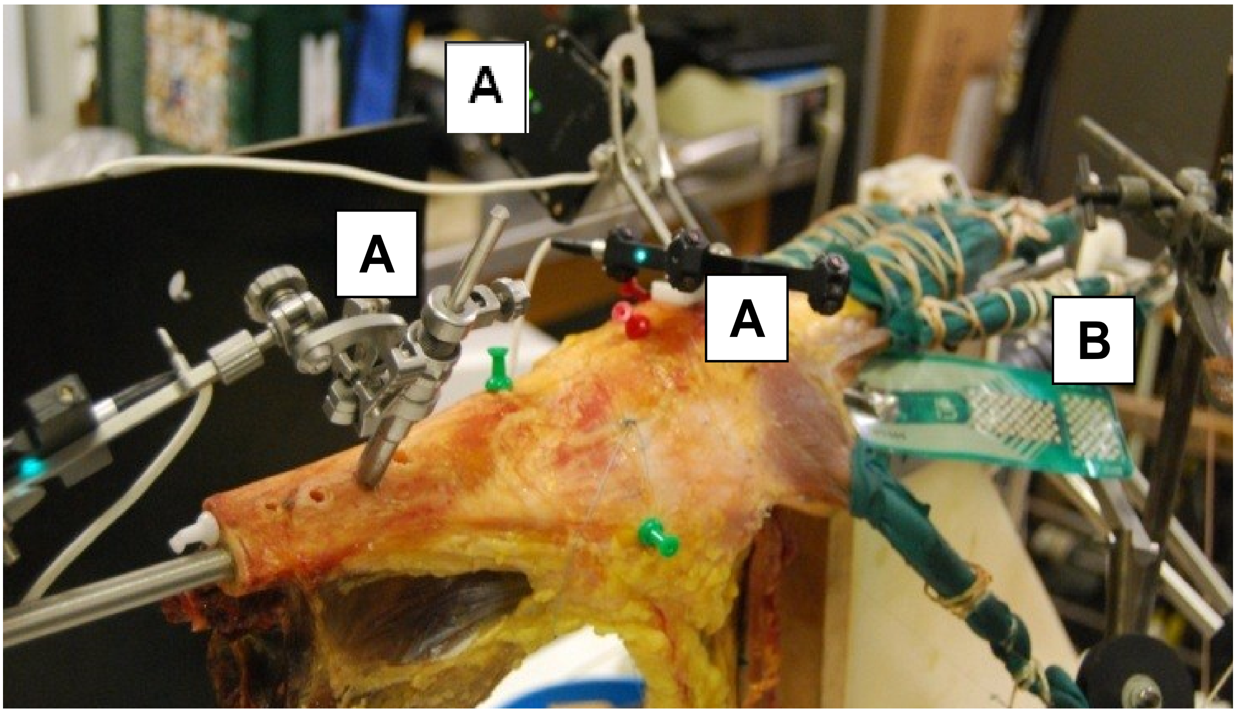

Figure 5.1 Photograph of knee joint rig, with attached muscle loading system.

A: Optical trackers attached to the femur, patella and tibia.

B: Tekscan pressure film, ready to be attached to the Tekscan handle.

5.5 Testing Procedure

Ten 'conditioning' cycles of knee flexion-extension were performed through 0°-90°, this ensured correct operation of all equipment and minimised any hysteresis. Kinematic and pressure measurements were taken on the intact knee and repeated after the MPFL was transected (Chapter 4). The MPFL reconstruction was then undertaken by one experienced Consultant Orthopaedic Surgeon.

The gracilis tendon was harvested and at least 20cm usable tendon obtained from each knee. The tendon was debrided of muscle tissue, both ends whip stitched for 30mm with monofilament suture Ethilon 2-0 (Ethicon Co., Somerville, NJ) and stored in a moist swab. The patella was approached through a 20mm incision to expose the patellar border from the superomedial corner to just above the midpoint of the medial patellar margin. A rongeur created a 20mm long groove 5mm wide to enable the graft to be embedded in the superomedial border of the patella (Figure 5.2). Two 2.4mm guide wires were drilled transversely through the patella from medial to lateral, from the proximal and distal ends of the groove. Two Ethibond 2 sutures (Ethicon Co., Somerville, NJ) were looped round the mid part of the gracilis graft and passed through the patellar tunnels. The sutures were then fully tensioned and tied at the lateral aspect of the patella, securing the mid length of the graft in the patellar groove (Figure 5.2).

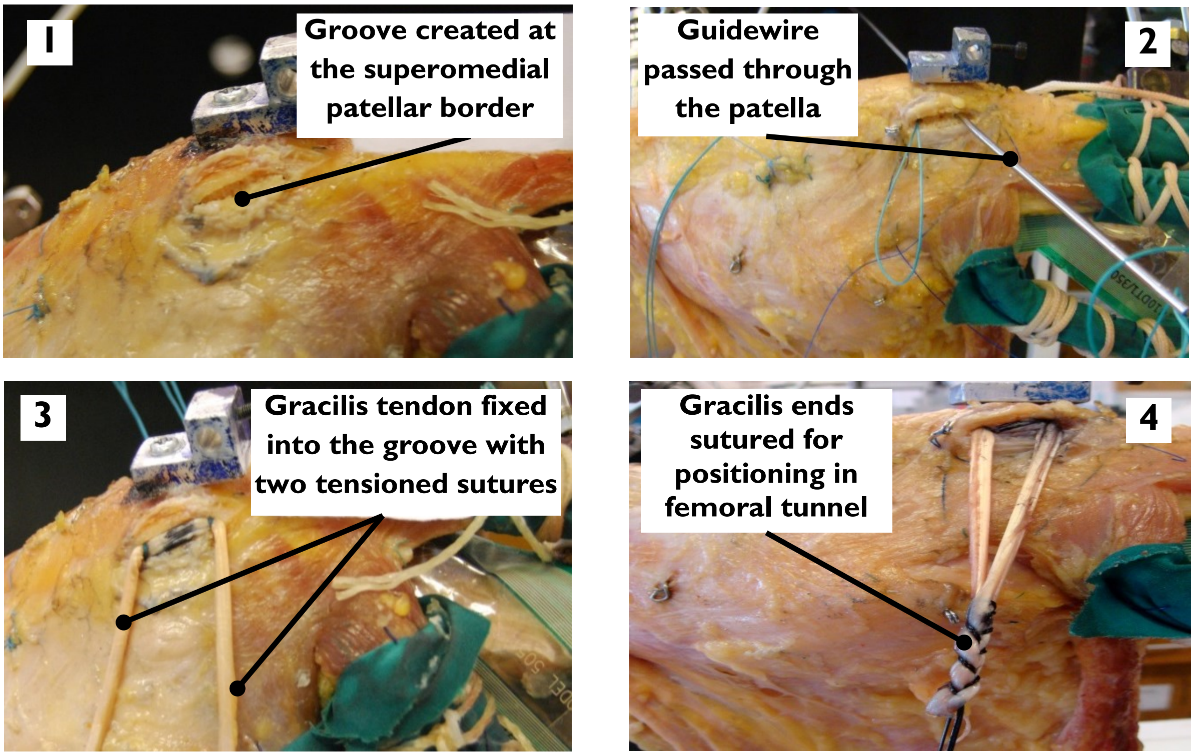

Figure 5.2 Medial aspect of one test knee: showing step by step method (1-4) for fixation of the gracilis tendon to the superomedial patellar border.

The femoral tunnel was drilled at the position of the previously marked anatomical centre of the MPFL attachment, using a 2.4mm guide wire from medial to lateral, and enlarged using a 6mm reamer to a 30mm depth. The second and third layers of the medial retinaculum were separated using thin curved scissors passed from the patella to the femoral attachment of the MPFL (Figure 5.3). Suture loops were used to pull the ends of the graft between the retinacula and into the femoral tunnel and the sutures pulled out laterally at the femur. Once passed mediolaterally, the sutures were led over a pulley where a hook was attached to the ends, enabling a graft tension of 2N, 10N or 30N to be applied. They were then clamped at the tunnel entrance to prevent movement during knee flexion when the knee was in the correct angle of flexion for fixation to be measured.

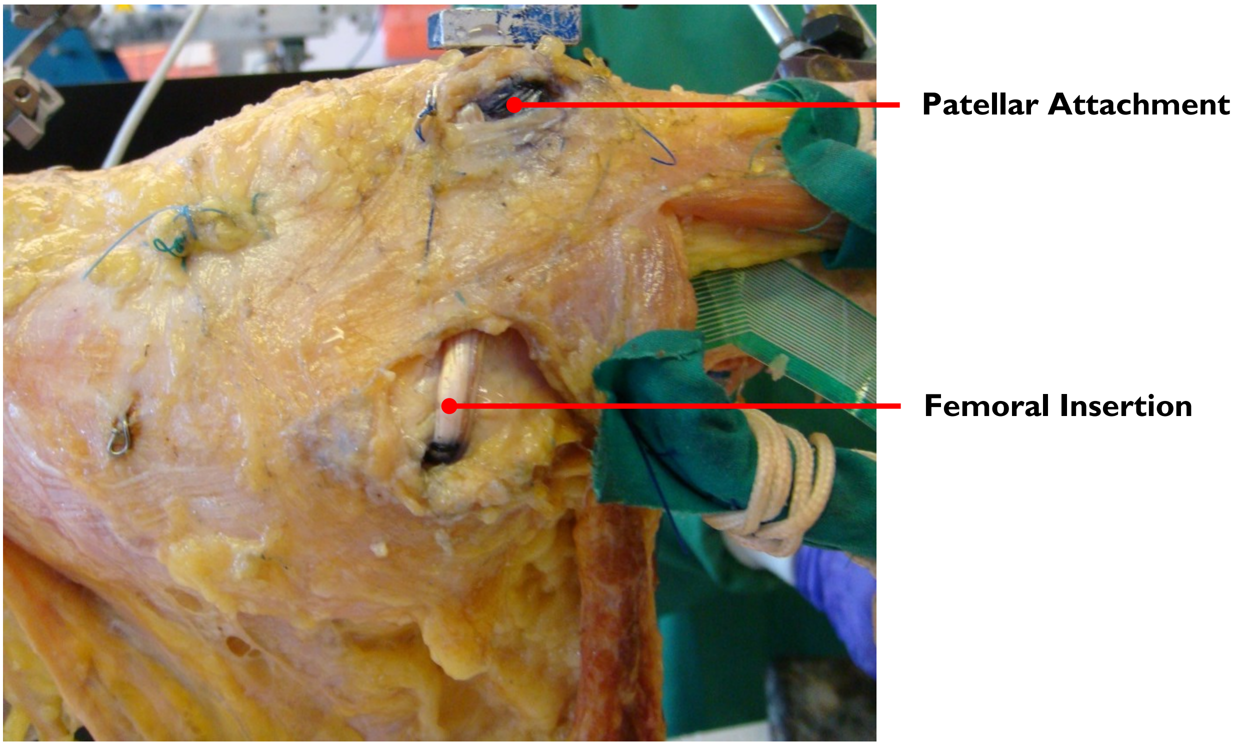

Figure 5.3 Medial view of a knee in the test rig following MPFL reconstruction with patellar and femoral attachment and insertion sites visible.

5.6 Experimental Protocol

Measurements were taken with the knee intact and then in nine different reconstructed conditions, at six different flexion angles (0, 10º, 20º, 30º, 60º and 90º). Because of concerns over damage to the soft tissues observed during pilot testing when multiple variables were examined, it was decided to limit testing to combinations of three tunnel positions (Figure 5.4), three graft tensions and three fixation angles. Since the MPFL is close to isometric across the range of knee flexion-extension (Stephen et al., 2012) and possesses very little tension in early flexion and no tension in deeper knee flexion, the authors hypothesised that a graft fixed with tension sufficient to hold it taut in the tunnel, which did not cause elongation, may be optimal to reproduce intact mechanics. From pilot testing 2N tension was determined to be sufficient to simulate this. Prior research has suggested both 2N (Beck et al., 2007) and 10N (Philippot et al., 2012b) tensions may be appropriate to restore patellofemoral mechanics so these were compared. Lastly 30N was examined to ascertain the consequence of over-tensioning, whilst avoiding damage caused to soft tissues and articular cartilage found using higher loads during pilot testing. The angle of fixation was based on a literature search of the most commonly reported practices and the experience of the surgeons conducting the research (Mikashima and Kobayashi, 2006). The hypothesis was that fixation would be most optimal at knee flexions greater than 30˚ because the patella would be centred on the trochlear groove. Finally the proximal and distal tunnel positions were decided based on prior ligament length change patterns (Chapter 3) (Stephen et al., 2012).

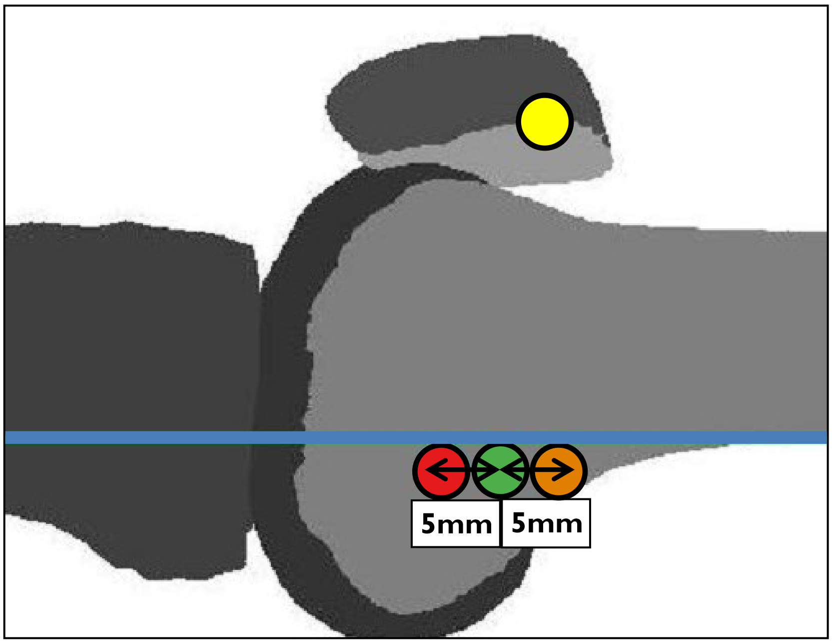

Figure 5.4 Position of the anatomical femoral tunnel and the proximal and distal tunnel positions in relation to it.

Graft tensions of 2N, 10N and 30N were applied with fixation of the graft undertaken at 0°, 30° and 60° of knee flexion using a secure clamp at the lateral aspect of the femur. Measurements were taken with the knee intact, and then in 9 different reconstructed conditions. Patellofemoral contact pressures and kinematics were measured for each condition at 0°, 10°, 20°, 30°, 60° and 90° knee flexion. The graft was then removed from the femoral tunnel and the tunnel filled-in with bone cement. A further two tunnels were drilled, 5mm proximal and 5mm distal to the anatomical tunnel (Figure 5.4). Measurements were then repeated in a randomised order, at the previous flexion angles, with the graft in the proximal and distal tunnels, tensioned with both 2N and 10N. Thirty Newtons was not applied in combination with the non-anatomical tunnels, following earlier observations of abnormal behaviour with this tension and concerns of inflicting soft tissue damage.

5.7 Analysis

Custom written MATLAB scripts calculated mean and peak contact pressures and patellar motion (Appendix A+B). The co-ordinate system of each bone was set up so lateral patellar translation and tilt were taken to be positive (Chapter 4).

Data was analysed in SPSS. A Shapiro-Wilk test confirmed that the data sets were normally distributed. The primary factors were tunnel position (anatomical, proximal, distal), flexion angle (0˚, 10˚, 20˚, 30˚, 60˚, 90˚), graft tension (2N, 10N, 30N) and angle of fixation (0˚, 30˚, 60˚). The dependent variables were: mean and peak medial and lateral facet articular contact pressures, and patellar tilt and lateral translation. Two-way repeated -measures analyses of variance (RM ANOVA) were performed comparing the effects of the primary factors on the dependent variables across the arc of knee flexion as follows:

1. Three analyses were performed for fixation angle comparing 0˚, 30˚ and 60˚ fixation, with the intact knee, keeping graft tension constant at 2N, 10N and 30N in turn.
2. Three analyses were undertaken for graft tension comparing 2N, 10N and 30N tensions, with the intact knee and data from each fixation angle in turn.
3. Two analyses were undertaken for tunnel position and the intact knee: one with a 2N graft tension and one with a 10N tension applied.

Post-hoc one way ANOVA and paired t-tests with Bonferroni correction were applied when differences across test conditions were found.

5.8 Results

5.8.1 Graft Fixation Angle

The Two-way RM ANOVA did not identify any significant differences caused by changing the angle of flexion among 0°, 30° or 60° when 2N tension was applied to the graft for each of the variables considered: patellar tilt (P=0.875), patellar translation (P=0.909), peak (P=0.175) and mean (P=0.542) medial contact pressures or peak (P=0.345) and mean (P=0.182) lateral contact pressures.

When 10N tension was applied to the graft, significant differences were found between reconstructions with the graft fixed at different angles of flexion and when compared to the intact knee, for mean (P=0.036) and peak (P<0.001) medial and mean (P<0.001) lateral articular contact pressures and tilt (P=0.039). Significant effects were not found on translation (P=0.160) or peak lateral pressure (P=0.149). The effect of graft fixation was found to vary with flexion angle (P<0.01) with graft tensioning at 0˚ resulting in increased medial contact pressures in early knee flexion compared to the intact knee or graft tensioning at 30˚ or 60˚.

Significant differences were identified among all dependent variables after a graft tensed to 30N had been fixed at 0˚, 30˚ or 60˚: patellar tilt (P<0.001), patellar translation (P=0.014), peak (P<0.001) and mean (P=0.003) medial contact pressures or peak (P=0.005) and mean (P<0.001) lateral contact pressures. Again the effect of graft fixation was found to vary with flexion angle (P<0.01) with increased medial contact pressures after fixation at 0˚ compared to the intact knee or tensioning at 30˚ or 60˚.

For the anatomically reconstructed and fully extended knee (0˚), a one-way ANOVA was used to compare fixation angle at 0˚, 30˚ and 60˚, when the graft was tensioned with 10N for each of the significantly different dependent variables. This identified increased mean (P=0.001) and peak (P<0.001) medial contact pressures and a decrease in mean lateral pressure (P=0.024), but no significant difference in patellar tilt (P=0.589) when the graft was tensed at 0° flexion, compared with grafts fixed at 30° or 60° (Figure 5.5). This analysis was not repeated with 2N graft tension, because no significant differences had been found, nor with 30N tension, because deleterious effects had already been identified, so it was not used here.

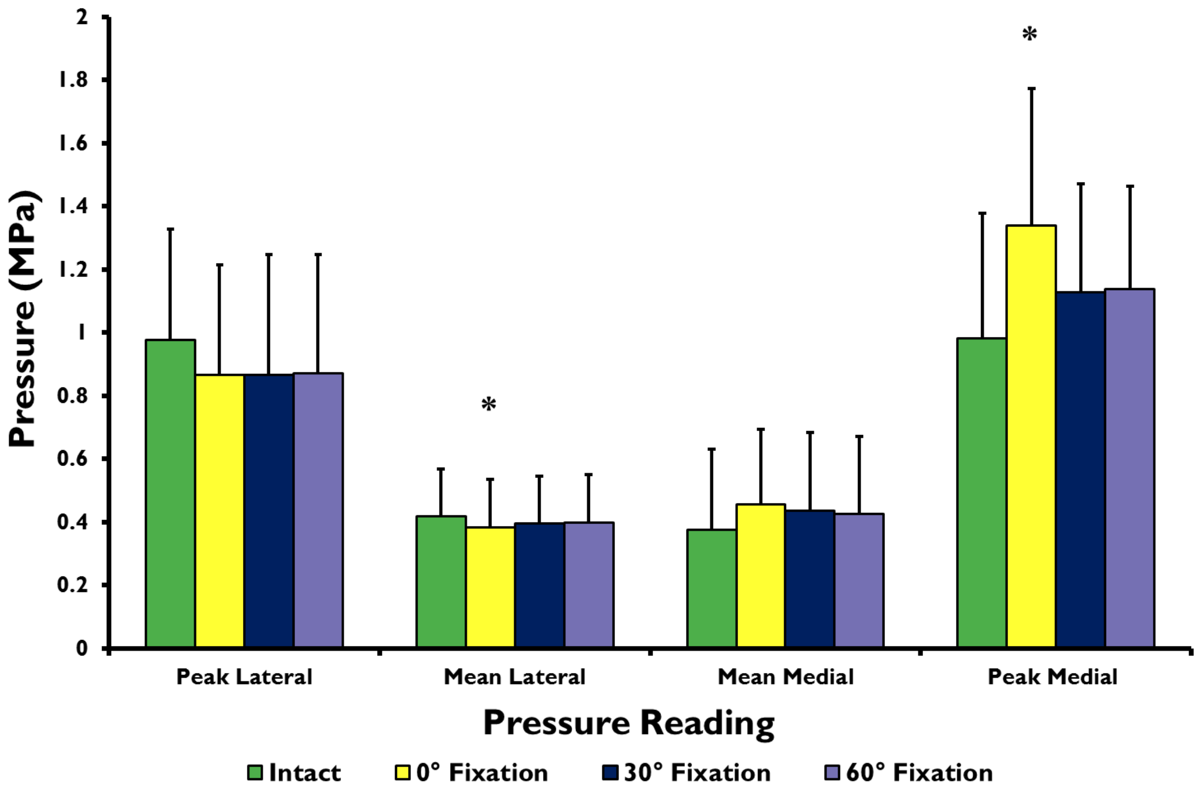

Figure 5.5 Peak and mean medial and lateral patellofemoral joint contact pressures (MPa; mean + SD, n=8) at 0° knee flexion in the intact knee and following anatomical reconstruction with graft fixation at 0°, 30° and 60° with 10N of graft tension. *$P<0.05$.

5.8.2 Graft tension effects with anatomical femoral tunnel

The graft tension had significant effects on all the variables measured, whether it was fixed at 0˚, 30˚ or 60˚ knee flexion (Figure 5.6). In general, increasing graft tension from 2N to 10N to 30N caused the following changes from the values in the intact knee, across the range of knee flexion: increased mean medial contact pressures (P=0.003, P=0.006, P=0.008, for grafts fixed at 0˚, 30˚, 60˚ respectively, Figure 5.7); and peak medial contact pressures (P<0.001, P<0.001, P<0.001, Figure 5.8). Whilst decreases in mean lateral contact pressures (P<0.001, P<0.001 P<0.001, Figure 5.9); and peak lateral contact pressures (P<0.001, P<0.007, P<0.002, Figure 5.10) were identified. Medial tilting of the patella (P<0.001, P<0.001, P<0.001, Figure 5.11); and medial translation of the patella (P<0.004, P<0.001, P<0.00, Figure 5.12) were also significantly affected.

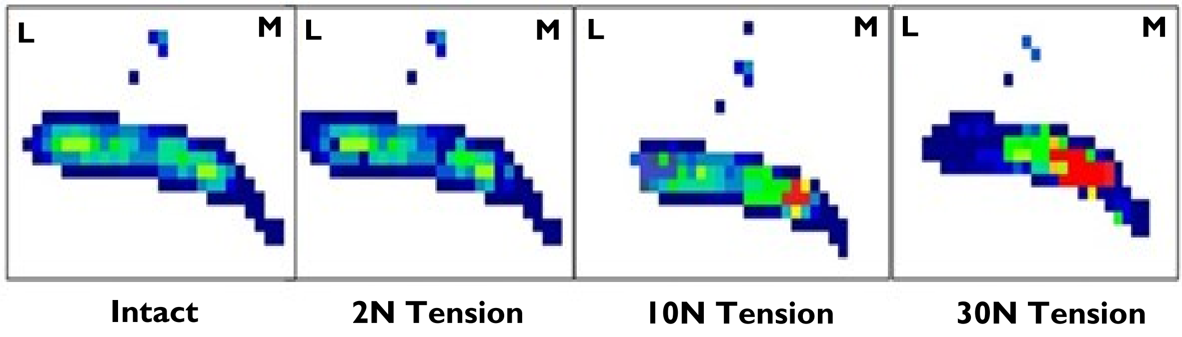

Figure 5.6 Screen Tekscan image of the pressure reading from one knee at 10 degrees of knee flexion, showing the intact state, the graft tensioned with 2N, 10N and 30N. It is clear that with increasing graft tension the contact pressure increased and shifted medially. This was a typical pattern observed for all knees.

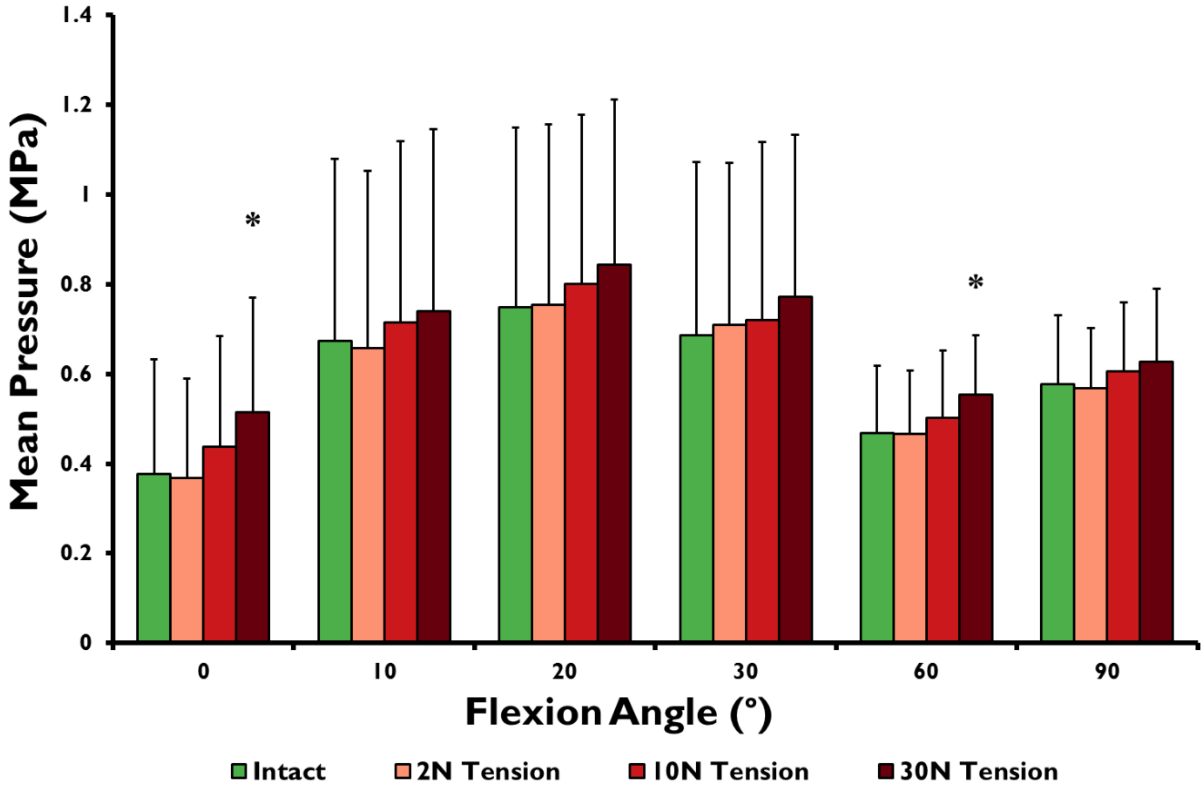

Figure 5.7 Mean medial patellofemoral joint contact pressures (MPa; mean + SD, n=8) of the intact knee and following anatomical reconstruction with graft fixation at 30° and graft tension with 2N, 10N and 30N. **P*<0.05.

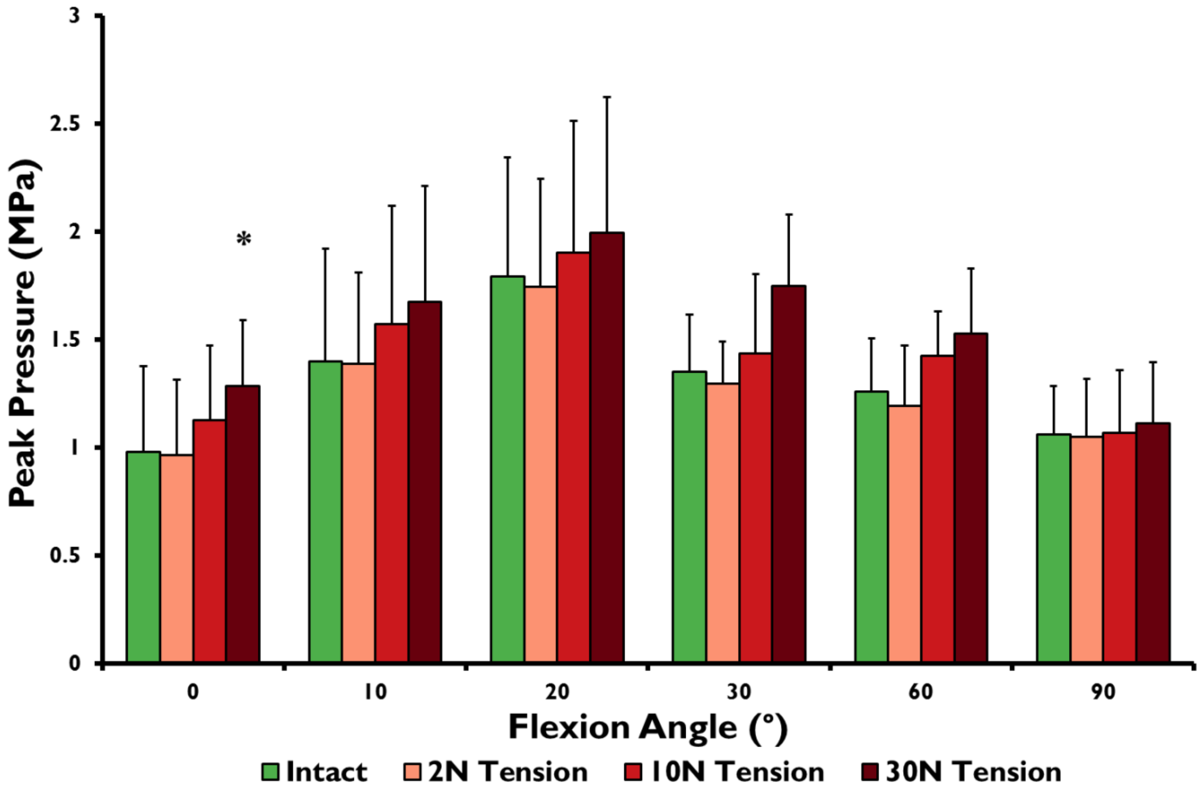

Figure 5.8 Peak medial patellofemoral joint contact pressures (MPa; mean + SD, n=8) of the intact knee and following anatomical reconstruction with graft fixation at 30° and graft tension with 2N, 10N and 30N. *P<0.05.

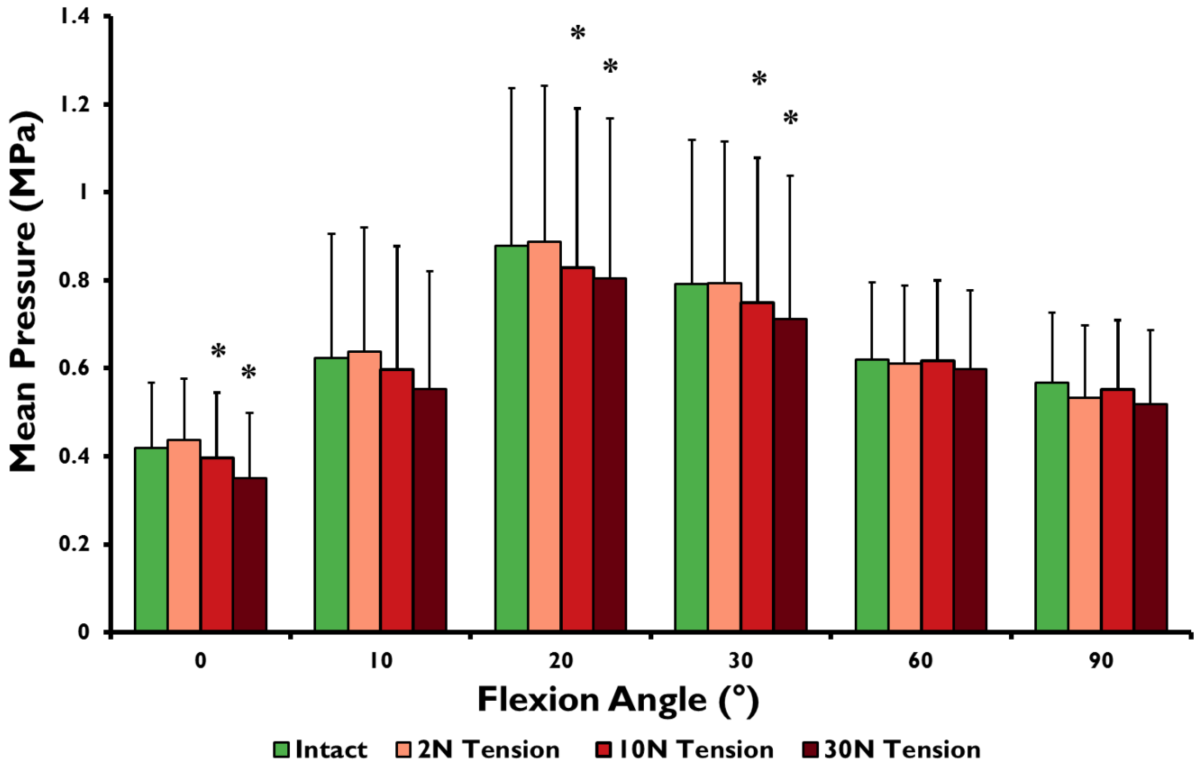

Figure 5.9 Mean lateral patellofemoral joint contact pressures (MPa; mean + SD, n=8) of the intact knee and following anatomical reconstruction with graft fixation at 30° and graft tension with 2N, 10N and 30N. *P<0.05.

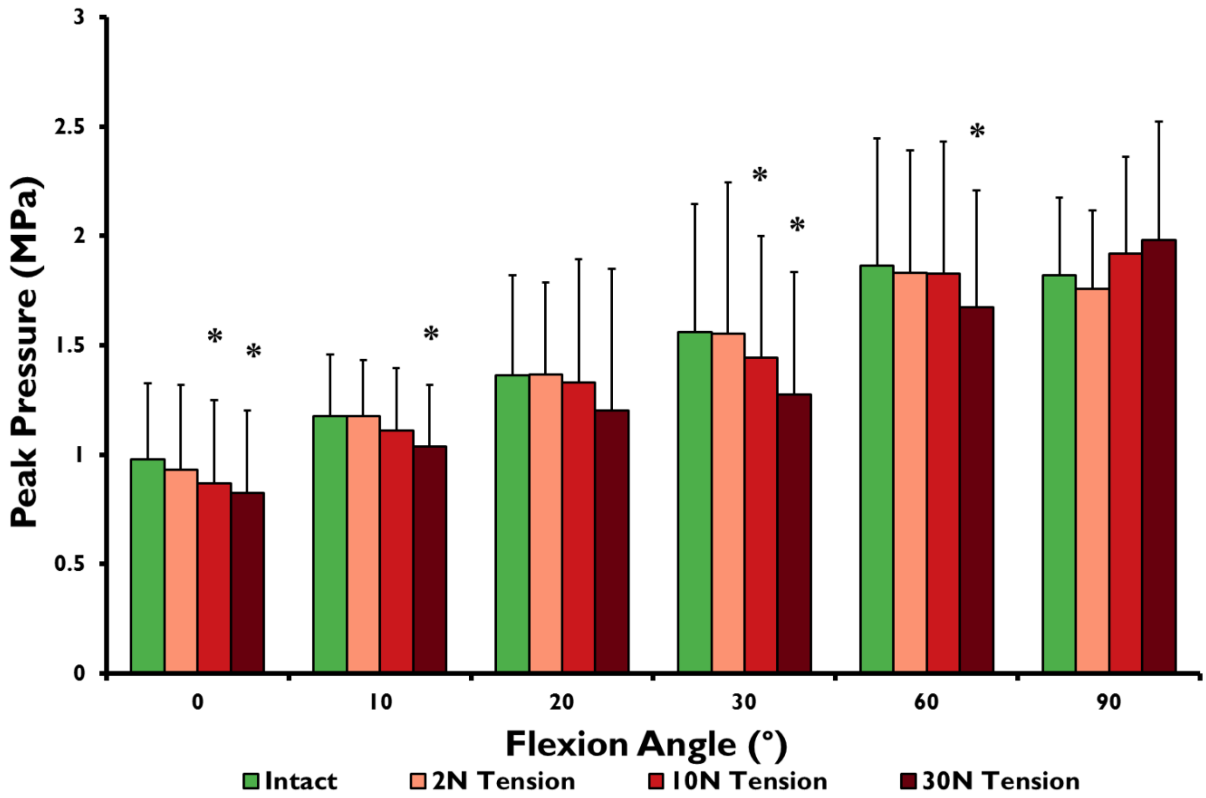

Figure 5.10 Peak lateral patellofemoral joint contact pressures (MPa; mean + SD, n=8) of the intact knee and following anatomical reconstruction with graft fixation at 30° and graft tension with 2N, 10N and 30N. *$P<0.05$.

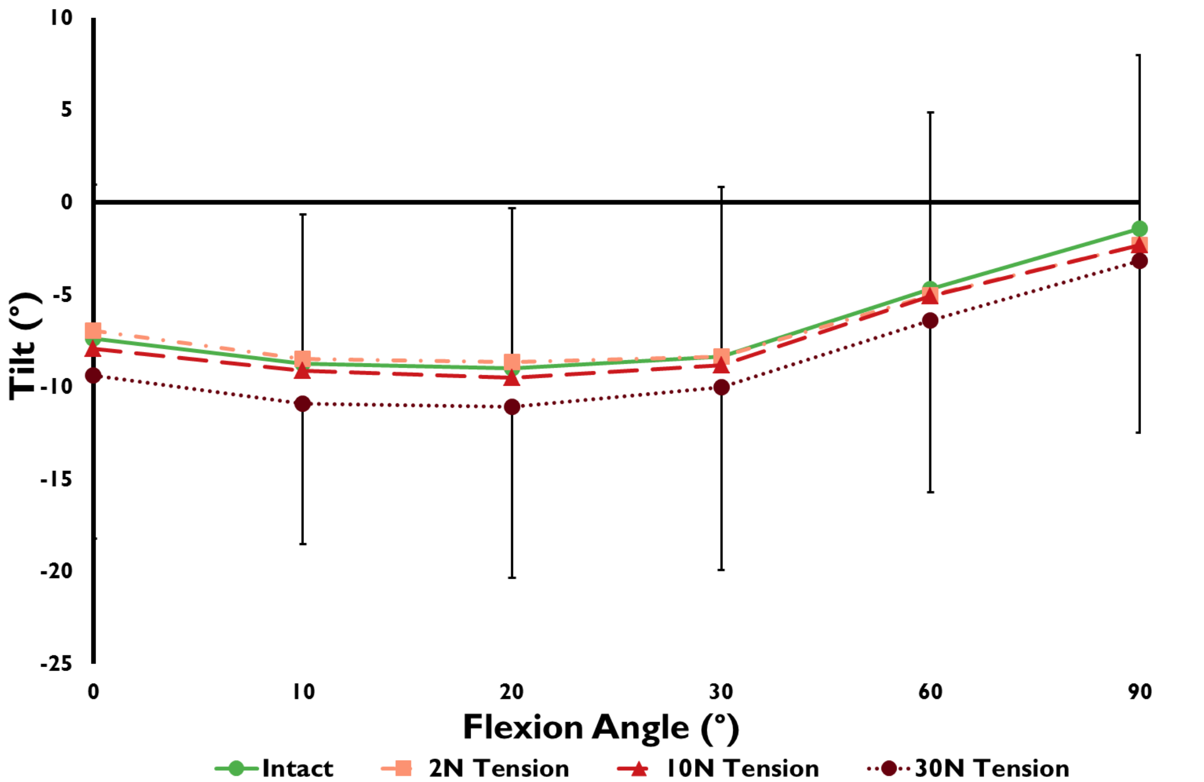

Figure 5.11 Patellar lateral tilt (°; mean ± SD, n=8) from 0°-90° knee flexion in the intact knee and following anatomical reconstruction with graft fixation at 30° and graft tension with 2N, 10N and 30N.

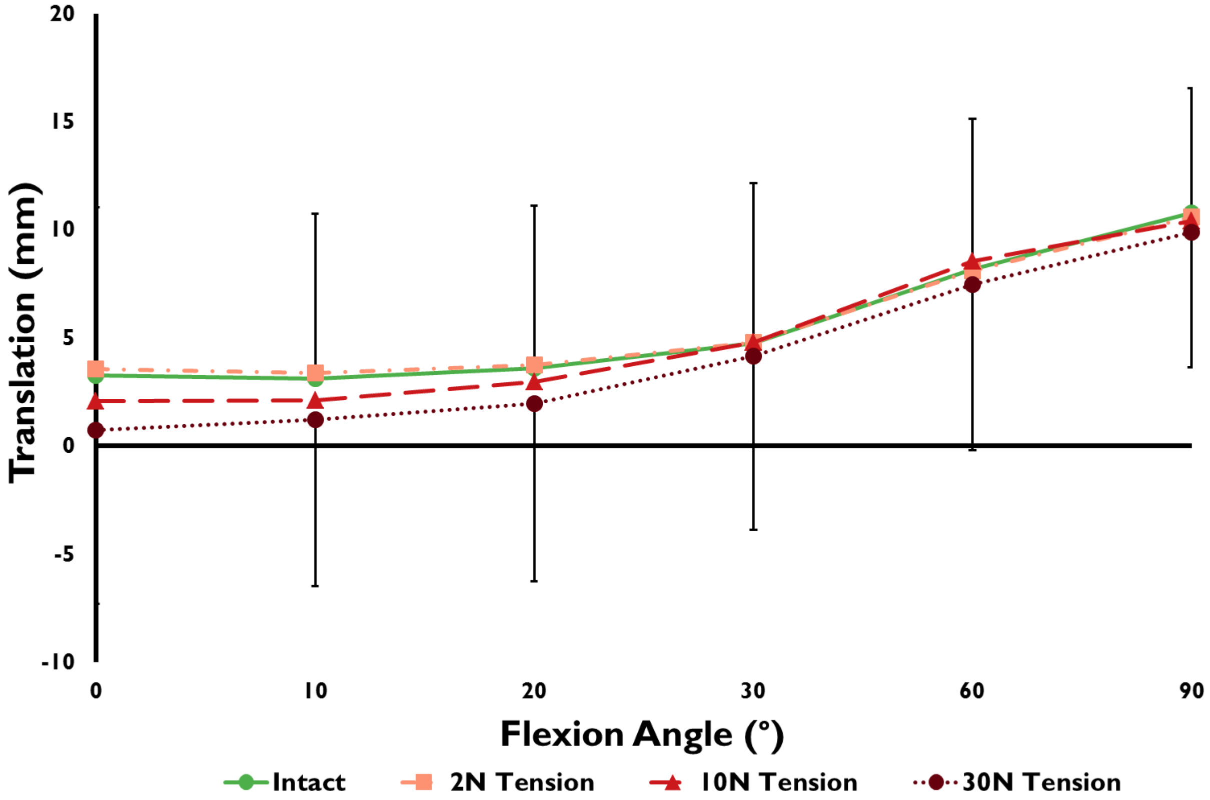

Figure 5.12 Patellar lateral tracking (mm; mean ± SD, n=8) from 0°-90° knee flexion in the intact knee and following anatomical reconstruction with graft fixation at 30˚ and graft tension with 2N, 10N and 30N.

Post hoc testing identified significant differences in patellar tilt at 0° when 10N was applied and at 0°, 10º and 20º when 30N was applied. Significant differences in patellar tracking were identified at 0° when 10N was applied and 0º, 10º and 20º when 30N was applied.

5.8.3 Femoral Tunnel Position

Non-anatomical femoral tunnel positions with graft tensioning of 2N had a significant effect on mean (P=0.008) (Figure 5.13) and peak (P=0.003) (Figure 5.14) medial patellofemoral joint contact pressures, but no significant effect on mean (P=0.371) and peak (P=0.146) lateral contact pressures, patellar tilt (P=0.479) or patellar translation (P=0.132), when compared to the intact knee and that reconstructed using the anatomical tunnel. The effect of tunnel position was found to vary with flexion angle (P<0.001), with the distal tunnel resulting in increased medial contact pressures near to full knee extension, and the proximal tunnel resulting in increased medial contact pressures in deeper flexion.

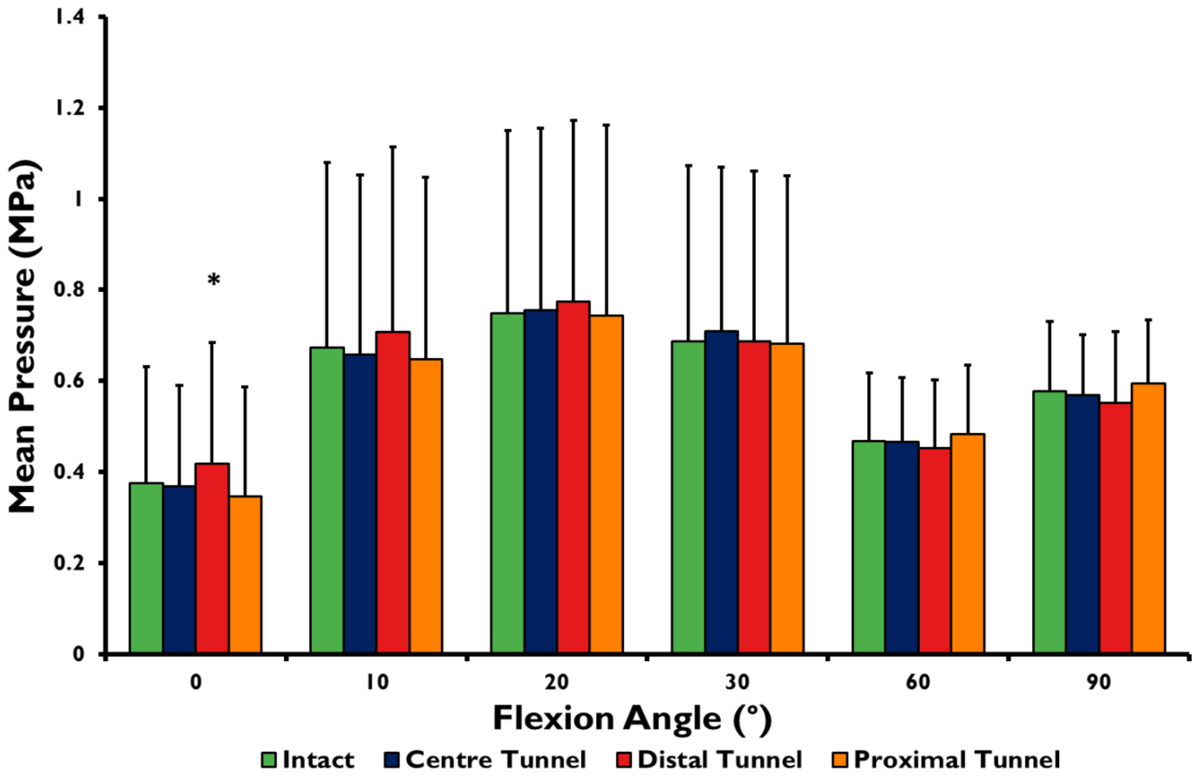

Figure 5.13 Mean medial patellofemoral joint contact pressures (MPa; mean ± SD, n=8) of the intact knee and following reconstruction with graft fixation at 30˚, **2N** graft tension and central, proximal and distal tunnel position. **P*<0.05.

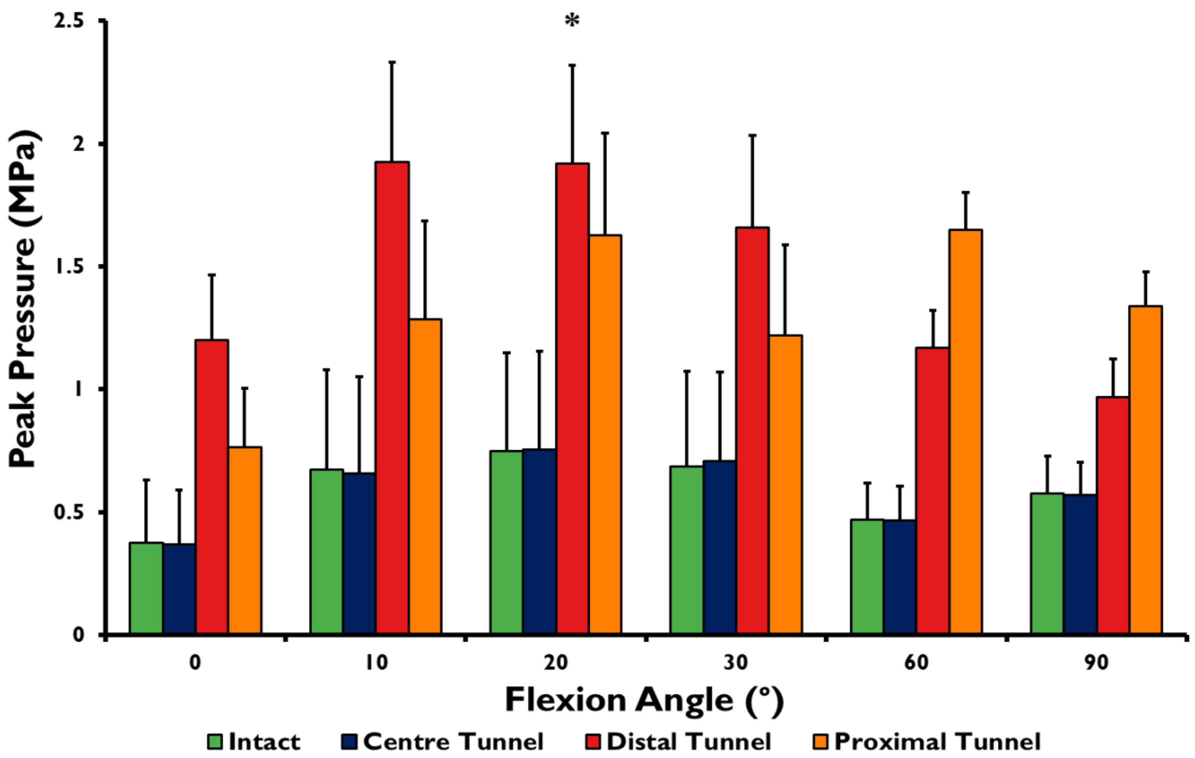

Figure 5.14 Peak medial patellofemoral joint contact pressures (MPa; mean ± SD, n=8) of the intact knee and following reconstruction with graft fixation at 30˚, **2N** graft tension and central, proximal and distal tunnel position. **P*<0.05.

When the graft was tensioned to 10N, tunnel position had a significant effect on mean (P=0.027) (Figure 5.15) and peak (P=0.002) (Figure 5.16) medial contact pressures and peak (P<0.001) (Figure 5.17) lateral contact pressures and patellar tilt (P=0.008) (Figure 5.18). However, mean lateral contact pressures (P=0.316) (Figure 5.19) and patellar translation (P=0.100) (Figure 5.20) were not found to be significantly different. Again flexion angle had a significant effect on all variables (P<0.001).

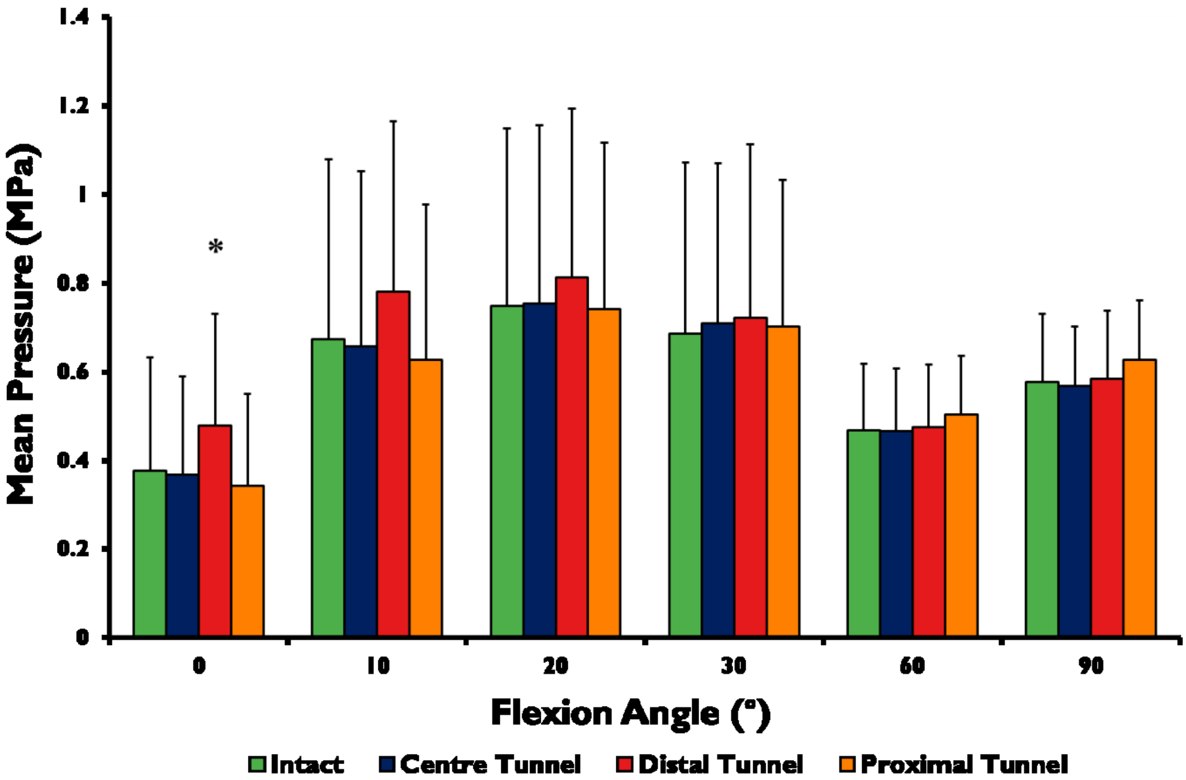

Figure 5.15 Mean medial patellofemoral joint contact pressures (MPa; mean ± SD, n=8) of the intact knee and following reconstruction with graft fixation at 30°, **10N** graft tension and central, proximal and distal tunnel position. *P<0.05.

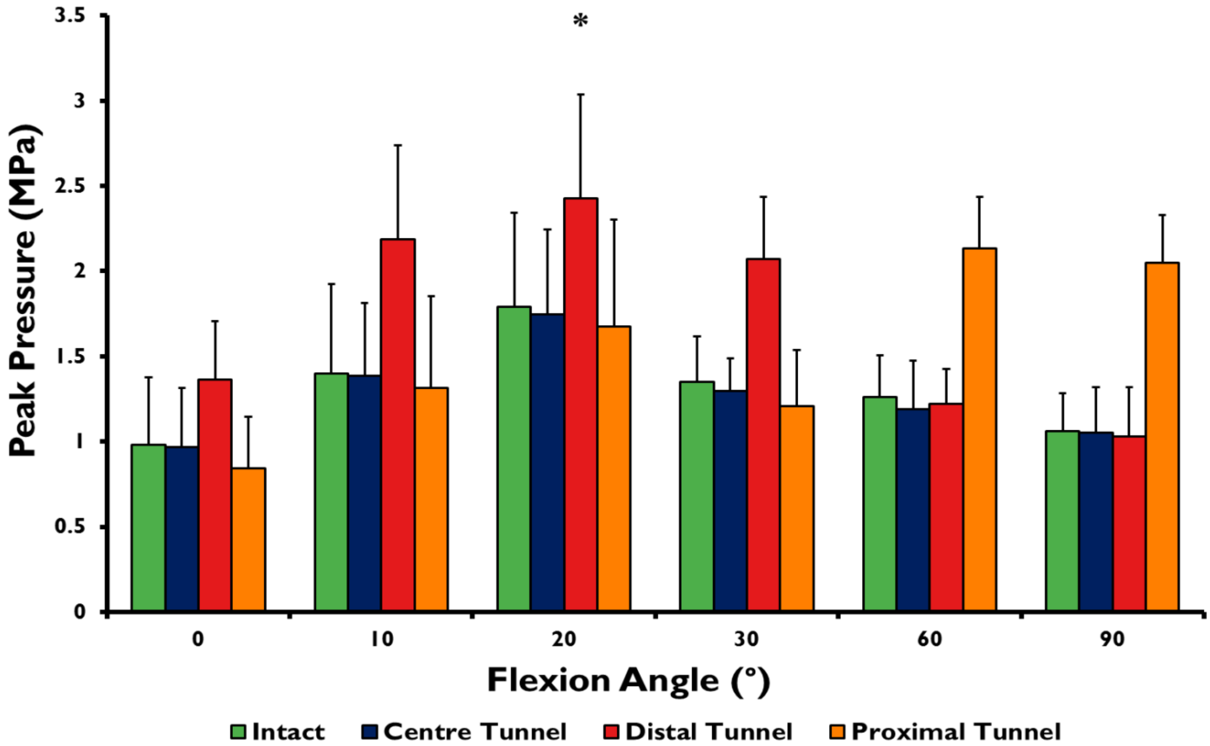

Figure 5.16 Peak medial patellofemoral joint contact pressures (MPa; mean ± SD, n=8) of the intact knee and following reconstruction with graft fixation at 30°, **10N** graft tension and central, proximal and distal tunnel position. *P<0.05.

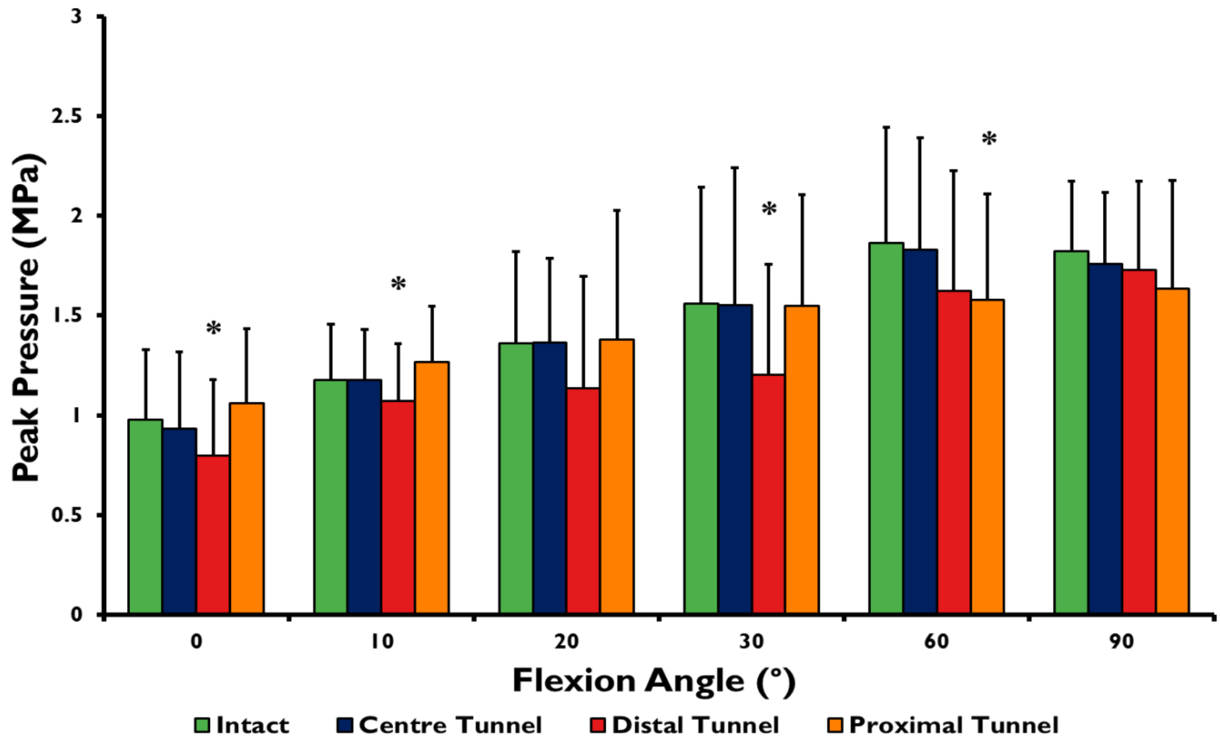

Figure 5.17 Peak lateral patellofemoral joint contact pressures (MPa; mean ± SD, n=8) of the intact knee and following reconstruction with graft fixation at 30°, **10N** graft tension and central, proximal and distal tunnel position. *P<0.05.

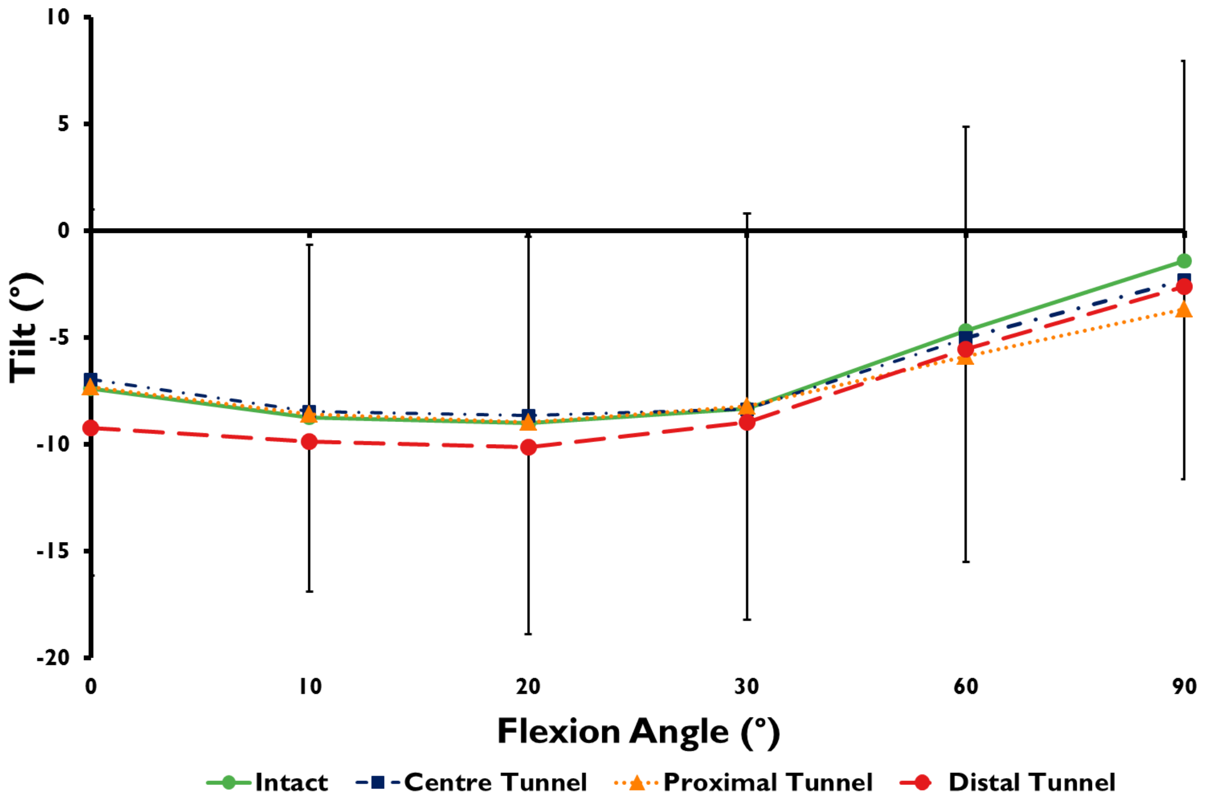

Figure 5.18 Patellar lateral tilt (°; mean ± SD, n=8) from 0°-90° knee flexion in the intact knee and following reconstruction with graft fixation at 30˚ and central, proximal and distal tunnel position with **10N** graft tension.

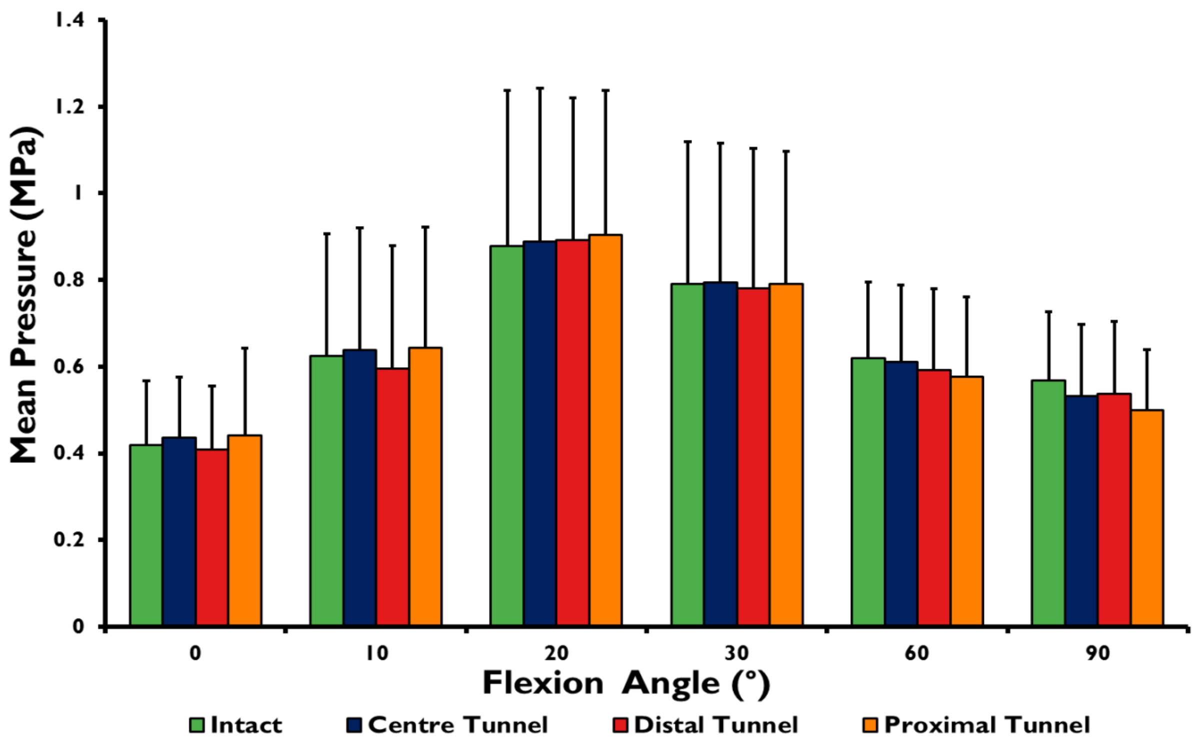

Figure 5.19 Mean lateral patellofemoral joint contact pressures (MPa; mean ± SD, n=8) of the intact knee and following reconstruction with graft fixation at 30˚, **10N** graft tension and central, proximal and distal tunnel position.

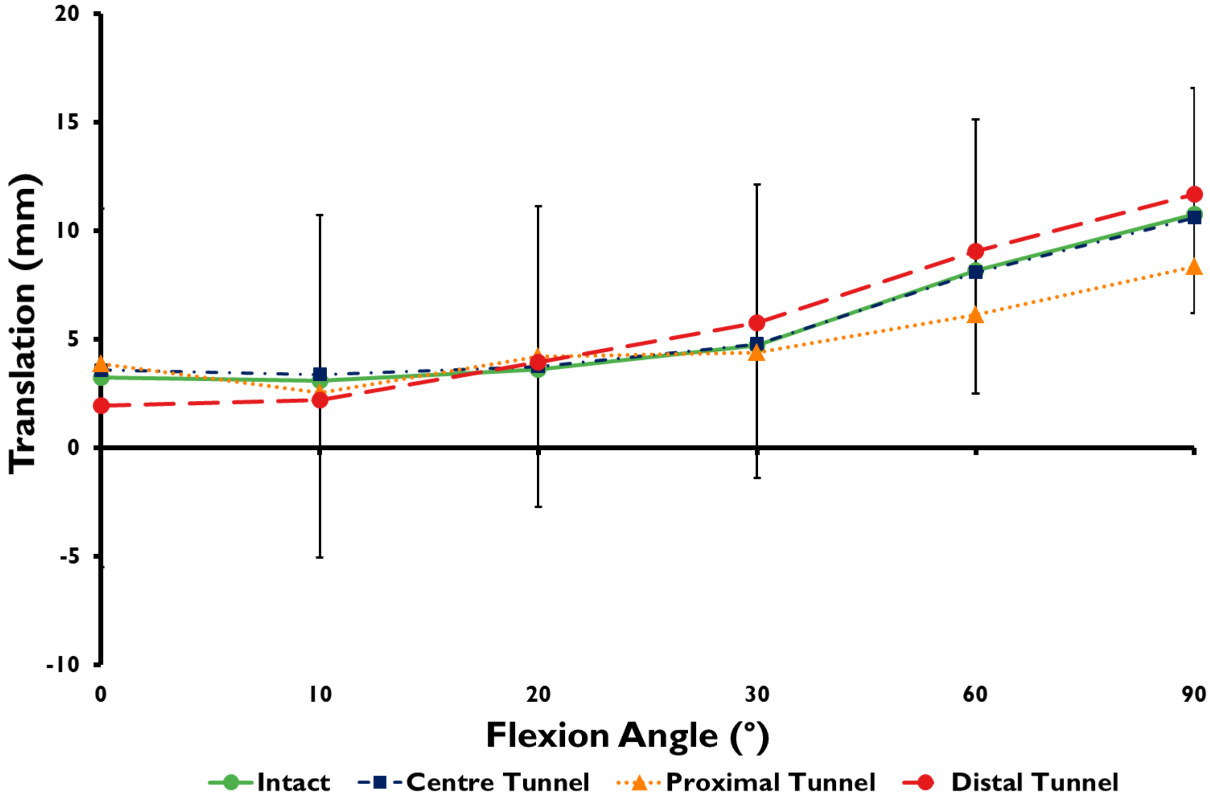

Figure 5.20 Patellar lateral translation (mm; mean ± SD, n=8) from 0°-90° knee flexion in the intact knee and following reconstruction with graft fixation at 30˚ and central, proximal and distal tunnel position with **10N** graft tension.

Post hoc testing identified significant differences in patellar tilt at 0° when a distal tunnel position was used with 2N tension (P<0.05) and at 0º, 20º, 30º and 90º when a distal tunnel was used with 10N tension (P<0.05). 10N tension also caused a significant effect on patellar tilt at 90º (P<0.05) with the proximal tunnel. Post hoc testing was not performed on patellar tracking or mean lateral pressure data due to the lack of an initial significant RM ANOVA. The effect of tunnel position was not investigated with 30N graft tension, following earlier observations of abnormal behaviour with that tension.

5.9 Discussion

This cadaveric study found that an anatomically positioned MPFL reconstruction tensioned to 2N and fixed at 0^o, 30^o or 60^o knee flexion restored patellar tracking and patellofemoral joint contact pressures to the intact state. However, grafts situated 5mm proximal or distal to the anatomical femoral MPFL attachment caused significant increases in peak medial pressures and medial patellar tilt during knee flexion or extension respectively, and thus potentially have implications for long term joint health. Over-tensioning to 10N or 30N caused significantly elevated medial contact pressures and reduced lateral pressures, along with medial tilt and translation. This is concerning when considered alongside the known medial patellar facet osteochondral defects which are often present post dislocation (Nomura and Inoue, 2005; Nomura et al., 2003). As long as the graft was tensed only to 2N there was no significant effect of the flexion angle where it was tensed. However, for 10N and 30N tensions, fixation of graft tension at terminal knee extension resulted in increased medial contact pressures and reduced lateral pressures in early flexion, compared to grafts fixed at 30° or 60° flexion. These findings highlight the accuracy desirable in MPFL reconstruction and the influence of graft tension and tunnel position on post-operative patellofemoral joint mechanics.

Graft tensioning during MPFL reconstruction presents the challenge to obliterate instability but avoid over-tightening. Tension in the graft occurs when it is stretched beyond its resting length. This is the first laboratory study to examine the effect of MPFL reconstruction tension on patellar kinematics and contact pressures simultaneously. Results demonstrate that a graft tensioned to 2N was sufficient to restore patellofemoral joint mechanics to the intact condition, whilst tensions of 10N or more caused increased medial contact pressures and medial patellar tilt. The contact pressure findings are in agreement with those of Elias and Cosgarea (2006) via computational modelling and Beck et al (2007) who used cadaveric knees to demonstrate increased medial pressures resulting from over-tensioned MPFL grafts. The tracking outcomes of the present study did not match those of Phillipot et al (2012b), who suggested 10N graft tension to restore patellofemoral joint mechanics. This difference likely relates to differing quadriceps tensioning protocols, with Phillippot et al (2012b) applying only 10N of quadriceps load to test specimens, compared to the 205N used in the present study. Clinical studies have reported the adverse consequences of over-tensioning, resulting in post-operative pain and necessitating later revision surgery (Thaunat and Erasmus, 2009), and the present contact pressure findings confirm the increased articular cartilage contact pressures as a consequence of an over-tensioned graft. This study has demonstrated the minimal graft tension required to restore patellar mechanics and should serve as a caution against over-tensioning during MPFL reconstruction. Findings support the concept that the MPFL acts only as a guide or ‘check rein’ on the patella, with high tensions resulting only during maltracking events (Bicos et al., 2007). When the MPFL is examined by MRI, it is seen to act more in a posterior direction than medially, and so tensioning the MPFL reconstruction will tend to load the medial facet of the patellofemoral joint, rather than medialising the patella.

In previous surgical studies the MPFL graft has been tensed and fixed at a range of knee flexion angles from 0°-90° (Drez Jr et al., 2001; Mikashima and Kobayashi, 2006; Nomura and Inoue, 2003), but there has been no objective evidence to identify the best angle to do this. There has been no literature to suggest the consequence of choosing the wrong angle of flexion when tensing the MPFL graft. However, the MPFL is close to isometric (Ghosh et al., 2009), which suggests that the result should not depend on the flexion angle, and this was found in the present study for the 2N graft tension. Failure to identify further effects of the fixation angle at the time of fixation may relate to the experimental design, since fixation occurred whilst the knee was loaded with 205N, potentially limiting patellar motion. This method was used because the tension in the quadriceps during surgery is not known, but will be likely to rise with increasing flexion angle. Therefore to enable result comparison and eliminate a type 1 error (false positive) the present methodology was used. If the clinical population includes knees with some trochlear dysplasia (unlike the normal knees in this study), they would have been less stable, and so would have shown larger changes resulting from MPFL tension. It is suggested that because the patella is not centred in the trochlear groove at full extension it is difficult to estimate the length of the MPFL, so fixation of the graft may be better undertaken with knee flexion between 30°-60°, when the patella is stabilised in the trochlea.

This study found that the mechanics of the patellofemoral joint were very sensitive to the site of the femoral graft tunnel. A graft positioned anatomically restored joint contact pressures and patellar tracking, whereas tunnels only 5mm proximally or distally resulted in significant increases in peak medial pressures and medial patellar tilt during knee flexion or extension, respectively. This correlates with previously reported ligament length change data which has identified the critical need to use an anatomical attachment to ensure ligament isometry and optimal outcome (Smirk and Morris, 2003; Stephen et al., 2012). Melegari et al. (2008) did not find an effect on patellofemoral contact pressures, and did not measure the kinematics. However, they used a quadriceps tendon autograft and only axial quadriceps loading, leaving the ITB unloaded, which may have influenced patellar motion. Furthermore, they used a small sensor which did not allow measurement from the entire patellofemoral joint surface area (Kwak et al., 2000), and measurements were only recorded from 30° onwards, possibly missing valuable information, since the main role of the MPFL is reported in early knee flexion from 0°-30° (Senavongse and Amis, 2005). A computational study investigating proximal tunnel malpositioning found similar results (Elias and Cosgarea, 2006).

Current findings are supported by clinical studies which have reported pain, osteoarthritis development and revision surgery necessary in patients with malpositioned femoral tunnels having undergone MPFL reconstruction (Bollier et al., 2011). This is important given a recent report suggesting that, in a sample of 29 patients, 35% had non-anatomical femoral tunnels (Servien et al., 2011). However, a recent clinical study has reported no adverse effects from femoral tunnel malpositioning evident in their clinical population (McCarthy et al., 2013). These results suggest there is no need for caution when determining anatomical landmarks during MPFL reconstruction, unlike the present study findings. Although these must be taken with caution since outcomes are only reported at follow up periods of 2-5years, a time frame likely to be too short to detect any significant degenerative joint changes. Furthermore the authors found that only 36% of tunnels in their population of 50 patients were anatomically positioned (with anatomical defined as within 9mm of the isometric point defined in the literature (Schöttle et al., 2007)). Therefore the very low change in pre and post-operative Knee injury and Osteoarthritis Outcome Scores (KOOS) reported by the study may be an overall reflection of unsatisfactory surgery as a result of inadequate femoral tunnel positioning. Prior work highlights MPFL reconstruction to be a successful intervention with good clinical outcome improvements using patient questionnaires (Christiansen et al., 2008; Schöttle et al., 2005).

There are increasing numbers of MPFL reconstructions and with recent findings suggesting its superiority in treatment of patients suffering patellar dislocation when compared to non-operative treatment (Bitar et al., 2011), its popularity is likely to rise further. Case reports have suggested adverse effects from this procedure as a consequence of surgical error (Bollier et al., 2011; Thaunat and Erasmus, 2009). The findings of the current study emphasise the technical challenges of undertaking this procedure to restore pre-injury patellofemoral joint mechanics. As with other ligaments the importance of an anatomical femoral tunnel has been demonstrated and the minimal tension necessary to correct patellar kinematics and contact mechanics to the intact state highlighted.

5.10 Key Findings / Conclusion

Key findings from this work therefore highlight the variables required to be controlled to undertake optimal MPFL reconstruction:

1. **Graft tension:**

    - ✓ 2N tension: restored patellar tracking and joint contact pressures.

- ✗ 10N tension: increased medial pressures, patellar tilt and decreased lateral contact pressures.
- ✗ 30N tension: increased medial pressures, patellar tilt, translation and decreased lateral contact pressures.

2. Tunnel Position

- ✓ Anatomical MPFL Reconstruction: restored intact joint mechanics.
- ✗ 5mm proximal tunnel: increased medial contact pressures and medial patellar tilt during knee flexion.
- ✗ 5mm distal tunnel: increased medial contact pressures and medial patellar tilt during knee extension.

3. Fixation Angle

- ✓ 30°, 60° fixation restored intact joint mechanics.
- ✗ 0° fixation: significant increases in medial joint contact pressures.

Anatomically positioned reconstructions with 2N tension fixed at 30° or 60° knee flexion restored joint contact pressures and tracking. However, graft over-tensioning, or femoral tunnels positioned too proximal or distal caused significantly elevated medial joint contact pressures and increased medial patellar tilting. The importance of correct femoral tunnel position and graft tensioning in restoring normal patellofemoral joint kinematics and articular cartilage contact stresses is evident.

5.11 Addendum

5.11.1 MPFL Reconstruction Method

The aim of this section of the thesis was to define the optimal method to reconstruct the MPFL and it was of utmost importance that the techniques undertaken were directly transferable and applicable to clinical practice. In order to do this a number of factors required consideration regarding the reconstruction method and technique. This section presents an outline of the factors considered and a summary of the decision making regarding the rationale for the final methodology.

5.11.1.1 Graft Selection

Clinically graft selection will often depend on a range of factors including; the experience of the operating surgeon, patient preference, the activity level and demands of the patient, and tissue availability (Cohen et al., 2009). Clearly the ideal graft for use in medial patellofemoral ligament reconstruction should have structural and biomechanical properties similar to those of the native ligament. The MPFL has been found to have a strength failure of 208N and stiffness of 8N/mm (Amis et al., 2003; Mountney et al., 2005), and is known to be a relatively thin with an often transparent structure (Baldwin, 2009; Bicos et al., 2007) (Chapter 3). Clinical and cadaveric studies have reported a range of tendon grafts selected for use during MPFL reconstruction:

- Gracilis (Christiansen et al., 2008; Schöttle, 2009).
- Semitendinosus (Schöttle et al., 2005; Xie et al., 2012).
- Semitendinosus and Gracilis combined (Drez Jr et al., 2001).
- Quadriceps (Goyal, 2013).
- Patellar (Bitar et al., 2011; Steiner et al., 2006).
- Synthetic Ligament (Ellera Gomes, 1992; Nomura et al., 2000b).

Unfortunately there is no gold standard randomised controlled clinical trial investigating the clinical outcome in patients undergoing MPFL reconstruction with different graft types. Furthermore there is a lack of longer term studies to establish any differences between individual groups of graft type patients. Therefore it was challenging to determine the optimal graft to use for reconstructing the

MPFL in the present study. Basic science studies were therefore examined to determine the graft which most closely replicates the properties of the MPFL previously described. Literature reports the failure strength and stiffness of individual grafts. These are outlined below (Mean ultimate load (N) and Mean Stiffness (Newtons / mm)):

- Gracilis single strand: **837 N / 160 N/mm** (Hamner et al., 1999).
- Gracilis double strand: **1550 N / 336 N/mm** (Hamner et al., 1999).
- Semitendinosus single strand: **1060 N / 213N/mm** (Hamner et al., 1999).
- Semitendinosus double strand: **2330 N / 469N/mm** (Hamner et al., 1999).
- Quadriceps: **1075 N** / **stiffness unknown** (Harris et al., 1997).
- Patellar: **2977 N / 455 N/mm** (Race et al., 2000).
- Synthetic: **unknown** (since the graft is not clearly defined in the original paper).

The gracilis and quadriceps tendons provided the closest properties to replicate the MPFL. It was decided therefore to use one of these grafts for reconstruction. Review of the quadriceps method described by Goyal, (2013) highlighted that since the tendon is left attached at the patella, no additional patellar fixation is required. This is an advantage since patellar fixation is known to be much more vulnerable to failure than the graft (Mountney et al., 2005). Clinical outcomes reported were similar at short term to other reconstruction techniques (Goyal, 2013). The method used is shown in Figure 5.21 below.

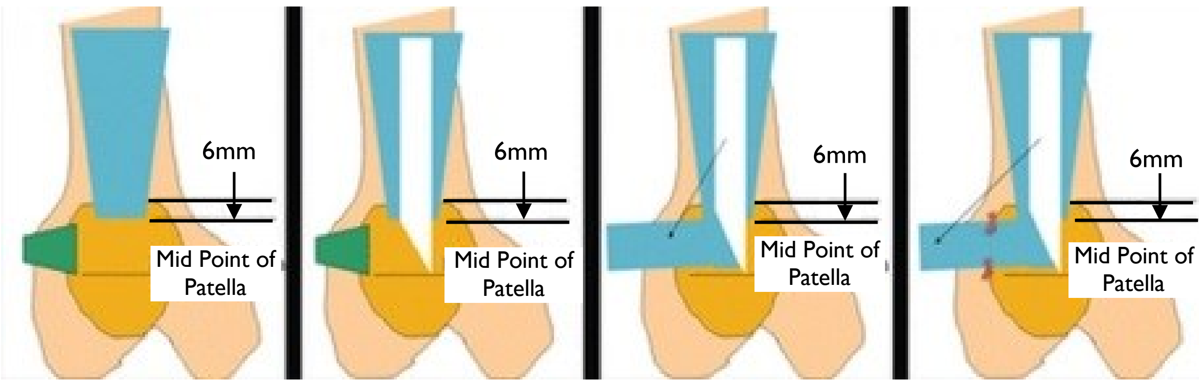

Figure 5.21 Quadriceps reconstruction method: MPFL (dark green), quadriceps tendon (light blue), and surgical plane of separation and dissection of the graft (white). Left to right: 1. normal anatomy of the MPFL and quadriceps tendon 2. the plane and dissection points for graft harvest 3. the graft is rotated and turned medially to mimic the anatomical position of the native MPFL 4. stay sutures at the patella provide anatomical patellar fixation. (as adapted and reported from Goyal (2013)).

Following review of this technique it was concerning that since the quadriceps tendon originated from the centre point of the patella, and was rotated to lie over half the body of the patella, that it would inevitably have an impact on patellar kinematics. The native MPFL originates from the medial patellar border, therefore the reconstruction described by Goyal (2013) is not anatomical. The technique was trialed in the laboratory using a cadaveric knee specimen available from a colleague's earlier completed work. It was apparent the quadriceps tendon overlying the patella had an effect on patellar tilt when the free end was tensioned and the knee flexed. This is clearly something which would merit future investigation to quantify in relation to other proposed methods, but unfortunately, due to limited resources, was beyond the scope of the present study.

Alternatively the gracilis tendon enables anatomical reconstruction of the MPFL with direct fixation to the patella and femur, whilst replicating similar structural properties to the native MPFL. Short and mid-term reports have again reported positive clinical outcomes using this method (Christiansen et al., 2008; Schöttle, 2009). However to reconstruct the fan shape of the MPFL (Kang et al., 2010) and ensure a wide attachment of the graft along the medial patellar border, a double strand gracilis graft has more commonly been utilised in clinical trials. This has the disadvantage of increasing both the graft stiffness and failure strength (Hamner et al., 1999); however clinical follow up found superior clinical results in patients with double bundle gracilis MPFL reconstructions compared to those with single bundle grafts (Wang et al., 2013).

Data reported in Chapter 3 showed that the MPFL is close to isometric across the range of knee flexion-extension. The effect of altering its length change pattern through non-isometric positioning of the anatomical markers was outlined. Since the purpose of this study was to investigate the effect of non-anatomical reconstruction, it was crucial to reconstruct the MPFL in as anatomically accurate a manner as possible. A double strand gracilis tendon graft reconstruction was therefore used in the present study. In the absence of any literature comparing the different reconstructions, this decision was based primarily on the structural properties of the tendon and its ability to replicate native MPFL anatomy. Cadaveric and clinical research investigating outcomes using different MPFL reconstruction methods is an area which may benefit from future work.

5.11.1.2 Patellar Fixation

Review of the literature highlights five main methods described to attach the MPFL graft to the patella when undertaking a double bundle reconstruction. These are outlined as follows:

1. The through patella technique: two transpatellar tunnels with the graft fed out laterally and back in through the patella (Christiansen et al., 2008) (Figure 5.22).

2. Bone bridge: creating a tunnel in the patella for the graft to lie in, using two corresponding drill holes (Ellera Gomes, 1992; Mikashima and Kobayashi, 2006) (Figure 5.22).

3. Two interference screws: two separate tunnels drilled into the patella to fix the ends of the graft to the patella (Hapa et al., 2012; Schöttle et al., 2010) (Figure 5.22).

4. Suture anchors: a groove is created along the superomedial border of the patella for the graft to lie in and graft is fixed with two suture anchors (Song et al., 2013) (Figure 5.22).

5. Transpatellar suture fixation: fixation of the graft to the superomedial patellar border into a pre-cut groove, fixed with transosseous sutures (Servien et al., 2011) (Figure 5.23).

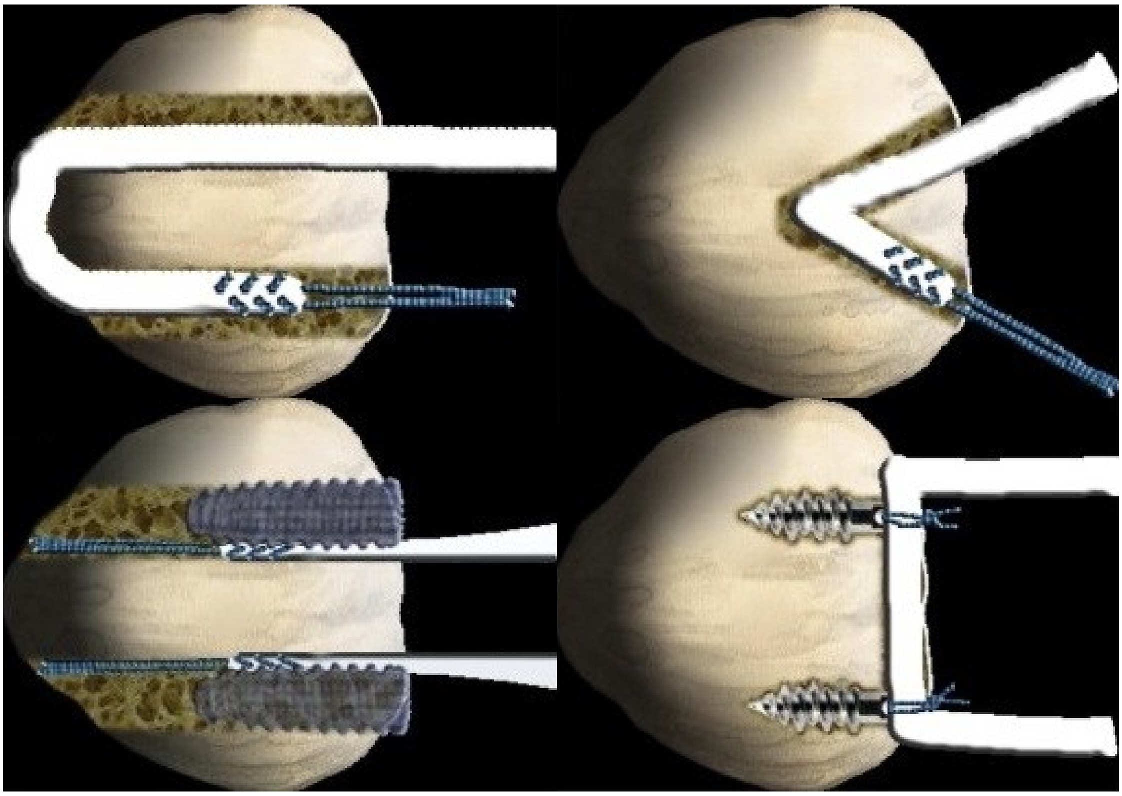

Figure 5.22 Methods for graft fixation to the patella. The **through patellar approach** (top left): **bone bridge** (top right), **interference screws** (bottom left) and **suture anchor fixation** (bottom right) (as adapted from Lenschow et al. (2013)).

Again review of the literature provided an array of procedures with a lack of consensus regarding the gold standard method. A pilot test was conducted in the laboratory, practicing each of the techniques on some used specimens which had intact patellae. It was quickly established that the bone bridge and trans-patellar graft techniques would be extremely challenging to undertake *in vitro*. The patellar bone was very hard and it was difficult to physically drill into the patella with the larger diameter drill ends. One patella cracked when it was drilled into to make the second part of the bone bridge. It was therefore determined this was a risky method to undertake in the current study, since if the patella fractured there would be no way to repair it and continue testing, thus valuable data could be lost. Review of the literature highlighted a number of reported cases where patellar fracture had occurred clinically in association with the transpatellar and bone bridge techniques (Ellera Gomes, 1992; Mikashima and Kobayashi, 2006), providing support for this decision.

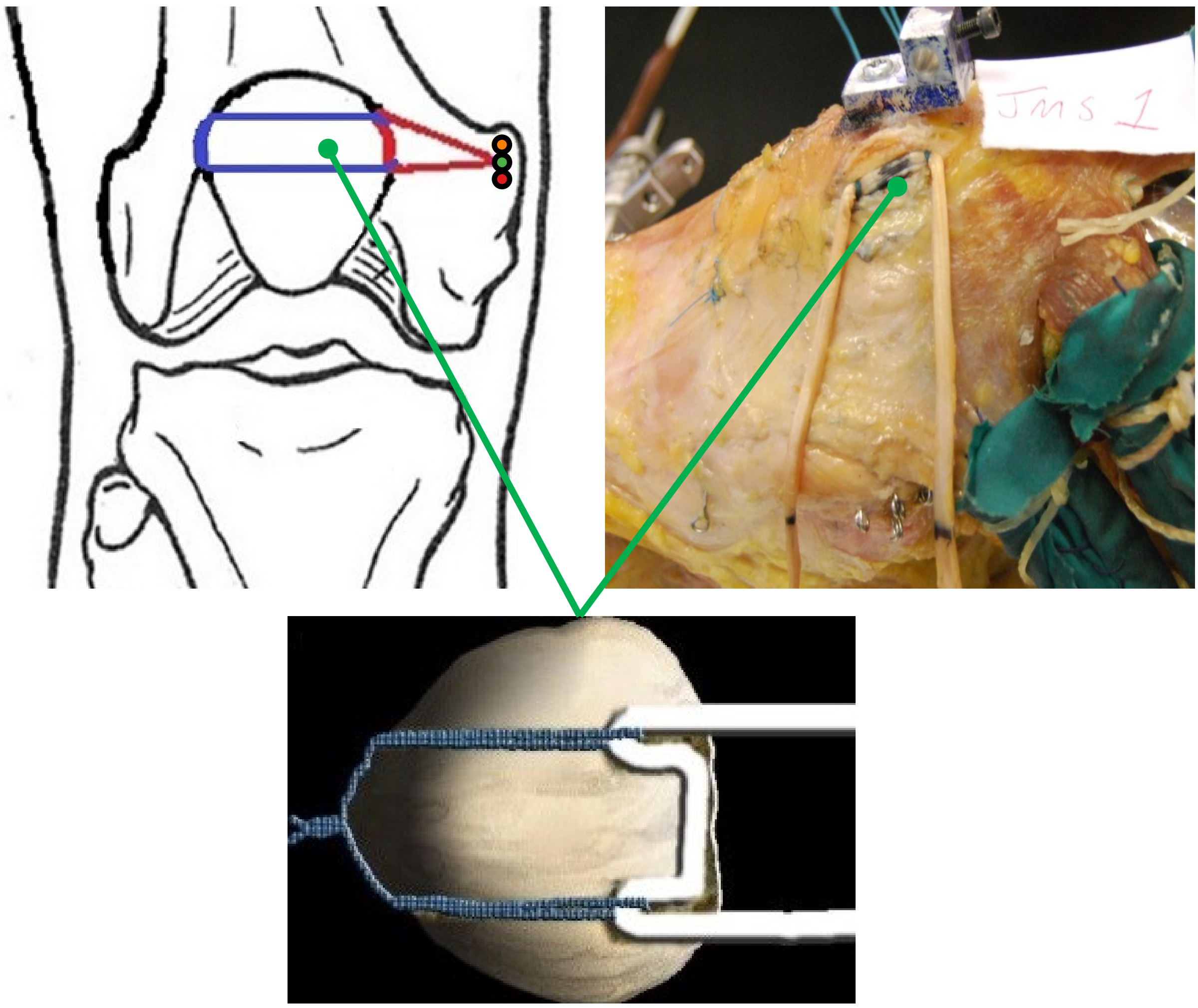

Figure 5.23 Transpatellar suture fixation of the patellar attachment. Schematic drawing (top left), *in vitro* fixation in the laboratory (top right) and below (as adapted from Lenschow et al. (2013)).

The suture and interference screw fixation were also practiced in the laboratory. A major drawback of using these techniques was the cost of getting the fixation anchors and screws which were priced at over £80 each. There were a few spare anchors and interference screws in the laboratory left over from prior work to practice with. Both these fixation methods proved very challenging to satisfactorily obtain good graft fixation. This was largely down to the poor bone quality of the specimens used. When the graft was sutured to the patella with these methods, and the femoral end of the graft tensed, the fixation failed in two out of the three practice specimens. Bone fracture with interference screw fixation (Parikh et al., 2013) and suture anchor failure during MPFL reconstruction (Dhinsa et al., 2013) are both reported side effects of MPFL reconstruction.

It was therefore decided to trial the final method outlined: transpatellar suture fixation (Figure 5.23). This has the advantage of requiring only very small transpatellar tunnels to be drilled through the patella. It also means graft tensioning can be permitted at either the patellar or femoral ends, whilst permitting anatomical reconstruction of the MPFL (Servien et al., 2011). This technique was practiced (as outlined in section 5.4) in the laboratory. It was found to be much easier to drill through the patella with small diameter drill ends (none of the practice specimens fractured). With a groove created at the superomedial patellar border and the graft secured with Ethibond 2 sutures (Ethicon Co., Somerville, NJ) the fixation was not affected when tension was applied to the graft. It was therefore decided to use this method for fixation in the study. Section 5.4 outlines this method in greater detail.

The precise position of the patellar tunnels was not found in Chapter 3 to have a significant effect on ligament length change pattern. The literature appears to widely accept that tunnel positions drilled to permit the graft to be sited along the superomedial border of the patella are commonly used with good clinical outcomes (Nomura and Inoue, 2006; Schöttle et al., 2010; Wang et al., 2013). Therefore the tunnels for the graft fixation to the patella were drilled where the most proximal and distal markers were positioned in Chapter 3.

5.11.1.3 Tension

A method for applying load to the gracilis graft was required in order to examine the effect of graft tension on study outcomes. Intraoperatively surgeons can use tension isometers, (commonly used during ACL surgery), to measure the tension they apply to grafts. Unfortunately there was not one of these available in the laboratory. Therefore it was decided to apply a known load to the graft in order to quantify the tension. Prior cadaveric work has used a load cell and actuator to apply a continual tension to the graft throughout full range of knee motion (Beck et al., 2007). This equipment was available in the laboratory but proved to be quite temperamental when it was set up. Therefore it was decided instead to use a weight and pulley system (Figure 5.24). Thus the graft tension could be easily changed by altering the load applied to the sutures at the lateral knee or patella. The method was piloted using two methods:

- Fixing the femoral attachment and applying tension to the sutures at the lateral patella.
- Fixing the patellar sutures and applying tension to the sutures at the lateral femur.

Following pilot tests of both methods, it was decided to apply tension at the femoral end of the graft. This was decided for several reasons:

- The patella had the optical tracker attached directly to it and the Tekscan underneath it. It was challenging to satisfactorily position the loading stand to apply tension to the graft without disrupting other testing equipment in the area. Furthermore it was difficult to find something to fix the sutures with at the patella when they were tensioned in order to avoid influencing patellar motion or the pressure and kinematic data recording.
- Most commonly, intraoperatively, surgeons tension the femoral graft end during MPFL reconstruction so this was the best replication of the clinical situation. This information was based on review of the literature and the experience of the surgeons involved with the present study (Mikashima and Kobayashi, 2006; Nomura and Inoue, 2003; Schöttle, 2009).

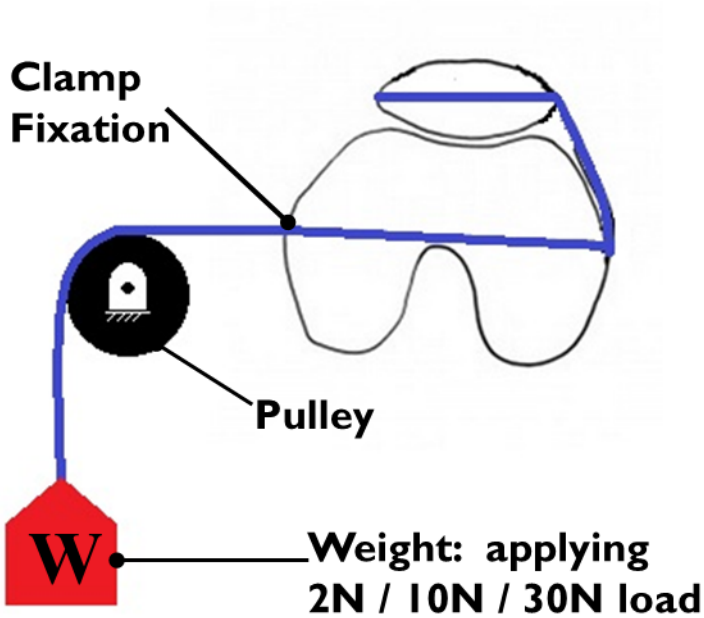

Figure 5.24 Method of tension application to the MPFL graft. Image shows a cross sectional view of the patella and femur, with the blue line representing the sutures and gracilis tendon and the red W pentagon indicating the load applied to tension the graft.

The literature review (section 5.1) highlights the lack of evidence to support a graft tensioning protocol for surgeons to follow during MPFL reconstruction surgery. In clinical commentaries graft tension is often qualitatively estimated and described. The primary aim of this part of the study was therefore to determine the optimal tension to apply during MPFL reconstruction to restore joint kinematics and contact mechanics. A secondary aim was to determine if there were any adverse effects of over tensioning the graft. It was established in Chapter 3 that the MPFL is close to isometric and how, in its natural state, the MPFL is not under tension (Shah et al., 2012). Based on this it was determined that very low tensions would replicate the function of the MPFL. Using the experimental set up described three different loads were investigated to apply tension to the graft: 0.5N, 1N and 2N. It was found that 2N resulted in the graft and sutures being held taut, whereas 0.5N and 1N were not sufficient to achieve this. Therefore 2N was the first tension selected to test.

Prior work used 10N tension and found that it satisfactorily restored patellar tracking, therefore it was logical to also investigate this tension (Philippot et al., 2012b). Finally to investigate the effect of over tensioning 30N was determined to be optimal. Thirty Newton and 40N were both pilot tested, based on prior literature (Beck et al., 2007). However it was found that over multiple tests, 40N caused damage to the Tekscan sensors and it was concerning the effect this was having on the integrity of the articular surfaces and local soft tissue of the retinacula.

In an ideal situation further tensions could have been tested. However given the multiple variables under examination in the present study it was decided to limit the investigation to just three tensions: 2N, 10N and 30N. This avoided any potential soft tissue or articular surface changes which may occur as a result of excessive testing (these were observed in some of the pilot test knees following multiple tension and tunnel position experimentation) which may have distorted results.

5.11.1.4 Fixation Angle

A second factor which could affect joint contact mechanics or joint kinematics post-operatively is the angle of knee flexion in which the knee is positioned when tension is applied and the graft is fixed during surgery (Sherman et al., 2012). Review of the literature again suggests a lack of consensus as to the position of the knee fixation, with the literature highlighting angles ranging from 0°-90° (Mikashima and Kobayashi, 2006; Schöttle et al., 2010; Shah et al., 2012). Logically it could be hypothesised that graft tension with the knee in a position where the patella is constrained in the trochlear groove, and the MPFL in a lengthened position, would be optimal to restore normal joint kinematics. However this has not previously been investigated. It was decided therefore to add this variable into the present study.

Direct discussion with the two experienced orthopaedic consultant surgeons advising on this project indicated one tensed the graft typically at 30º and the other at 60º knee flexion. These flexion angles were therefore investigated, since they were also widely supported by literature describing current clinical practice (Drez Jr et al., 2001; Ellera Gomes, 1992; Nomura and Inoue, 2003; Schöttle et al., 2010). Alongside this 0º was selected to test. It was hypothesised that since the patella is not centred in the trochlear groove in full extension that this could negatively impact on patellar kinematics and contact mechanics.

Pilot testing did not identify any clear trends resulting from fixation at each of these different flexion angles. It did, however, highlight several problems with the method:

- The need to fix the sutures at the lateral femur following the application of tension.
- The load required to be applied to the quadriceps to replicate *in vivo* behaviour of an anaesthetised patient, as would occur clinically.

Firstly, the issue of graft fixation. Although prior work had applied a constant tension to the graft throughout testing avoiding fixation (Beck et al., 2007), it was clear that this did not replicate what occurs intraoperatively when an interference screw would commonly be used at the medial femoral tunnel to fix the graft into position (Schöttle, 2009). However there were two problems identified with the use of an interference screw during this experiment:

- It would be difficult to keep removing and inserting a screw fixation throughout the full experiment since this would be both time consuming and likely to compromise the integrity of the bone and the graft.
- It would be unknown if any additional tension was being applied to the graft if it was caught around the screw as it was fixed into the femur.

Therefore following some pilot trials of different methods it was decided to simply clamp the sutures at the lateral aspect of the femoral condyle as they exited the femur. It was ensured that the sutures did not stretch under tension (they were tested up to a 30N load) and that the suture could not move or slide through the clamp when the knee was flexed and extended.

Secondly, the issue of quadriceps loading proved very challenging to determine. An anaesthetised patient would be unlikely to have 205N tension in their quadriceps and ITB. However, since it is not known how much resting tension is created by the quadriceps nor how this would be likely to rise as the knee flexes, it was decided to leave the 205N load on the specimens during graft fixation rather than make any inaccurate predictions. The reasoning behind this was to enable direct result comparison and thus permit rigorous statistical analysis, with the elimination of the potential for the occurrence of a type 1 error (false positive).

5.11.1.5 Tunnel Position

Data from Chapter 3 outlined the effect of anterior, posterior, proximal and distal femoral attachments on ligament length changes. The most significant changes occurred with too-proximal or too-distally positioned tunnels. Initially during pilot testing it was attempted to test all five points described in Chapter 3. However due to loss of bone integrity this was impossible to standardise and following pilot testing on one specimen this method was determined to be unrealistic and this idea abandoned.

A second pilot test was conducted with a protocol which investigated tunnels positioned anatomically, 5mm proximally and 5mm distally. Through trial and error it was determined that if one tunnel was tested and then cemented-in prior to drilling the other two tunnels, then the bone was able to stand the testing without breaking down. It was therefore decided to test just three tunnel positions, to avoid any compromise to data obtained.

5.11.1.6 Testing Protocol

Further investigation showed that the integrity of the local soft tissues, the articular joint surfaces and the Tekscan film were compromised when all three tunnel positions, all three graft tensions and all three fixation angles were investigated in combination, and tested at all flexion angles from 0º-90º (at 10º intervals).

Results from pilot testing showed similar trends from 30º-60º and 60º-90º ranges during the different trials. Therefore it was decided to take measurements at: 0º, 10º, 20º, 30º, 60º and 90º and not at 40º, 50º, 70º and 80º. Fixation angle did not appear to show any strong trend either; therefore it was decided to fix the non-anatomical femoral tunnels at just 30º. Thirty degrees was decided upon based on the hypothesis stated earlier and the fact that it was the most commonly reported angle in the literature. Finally, as previously discussed, the non-anatomical tunnels were not measured in combination with 30N tension.

5.11.2 Conclusion

Following extensive pilot study testing based on prior literature, hypothesis forming on the basis of prior knowledge, trial and error and expert opinion, it was determined to investigate the following variables:

- Three tunnel positions (5mm proximal, anatomical and 5mm distal).
- Three graft tensions (2N, 10 and 30N).
- Three fixation angles (0º, 30º and 60º).

Concerns over soft tissue and articular cartilage damage, highlighted during pilot testing, meant that the following combinations of trials were not performed:

- 5mm proximal or 5mm distal graft position with 30N graft tension.
- 5mm proximal or 5mm distal graft position with graft fixation at 0º or 60º.

All other variable combinations were examined at six different knee flexion angles. The in depth rationale for each of these decisions is summarised in section 5.10.1.

CHAPTER 6

VASTUS MEDIALIS WEAKNESS

6.1 Background

The management of patients following primary patellofemoral dislocation remains a controversial topic which lacks rigorous clinical evidence to guide treatment. An algorithm has been proposed to classify the treatment of patients following initial injury depending on their symptoms. It advocates surgical intervention only in those patients with a loose osteochondral fragment present in their knee. For all other patients, knee immobilisation in a brace and physiotherapy has been recommended (Jain et al., 2011).

More than 50% of patients are known to have ongoing symptoms following first time dislocation (Mäenpää and Lehto, 1997a). This highlights a lack of good understanding as to the needs of these patients. One prospective randomised trial on the treatment of patellar dislocation found surgical treatment was superior to non-surgical treatment in terms of reduced recurrence rates (Sillanpää et al., 2009a). The literature tends to support this, with re-dislocation rates generally varying from 13%-52% for conservative treatment (Cash and Hughston, 1988; Cofield and Bryan, 1977; Larsen and Lauridsen, 1982; Nikku et al., 2005) and 10%-30% for surgical intervention (Arendt et al., 2002; Harilainen et al., 1988; Hawkins et al., 1986; Nikku et al., 2005; Vainionpää et al., 1990). However these studies must be interpreted with caution since they represent a self-selecting portion of the population of patellar dislocators, consisting of those more severely affected patients who seek medical advice and intervention following injury. There is likely to be a further subsection of the population who dislocate their patella, find it relocates and do not seek any medical intervention. These are probably those least affected by injury and since there is no method to track these patients, they will not be included in figures from clinical studies (Fithian et al., 2004).

There are notable challenges in comparing all patellar dislocation patients together. More recent literature suggests there may be some benefit in classifying patients who dislocate their patella into two sub groups:

- Low Risk
    - Dislocation during high velocity activities.
    - Suffer direct trauma.
    - Active sports individuals / involvement in contact sports.

- High Risk
    - Dislocation during low velocity activities.
    - Family history of dislocation.
    - Younger patients.
    - Pre-existing deformities such as dysplasia or patella alta which reduce constraint.

It is suggested that those low risk patients will typically do well with conservative treatment, whereas those in the high risk group will commonly require surgical management as they are at greater risk of recurrent dislocation (Fithian et al., 2004; Hiemstra et al., 2013; Stefancin and Parker, 2007).

A second population who do not suffer frank dislocation of their patella often report pain in the region of the patella. Indeed, patellofemoral joint complaints account for 25-40% of all knee related problems presenting to sports medicine centres and represent the most common sporting knee injury in the under 50 population (Almeida et al., 1999; Bizzini et al., 2003; Devereaux and Lachmann, 1984; Taunton et al., 2002). Despite their high prevalence, the aetiology and treatment of these problems remains a challenge and consequently patients often develop chronic symptoms (Mihalko et al., 2007).

Insufficiency of the Vastus Medialis Muscles (VMM), particularly the Vastus Medialis Obliquus (VMO) has long been considered a contributing factor to the initiation or aggravation of patellar pain and instability, due to the reduction in the medial force acting on the patella (Fulkerson, 2002). Prior cadaveric work has examined the effect of reducing or removing VMO tension on patellofemoral joint contact pressures (Elias, 2009; Goh et al., 1995; Sakai et al., 2000; Wünschel et al., 2011). Findings highlight the important role the VMO plays in maintaining patellar tracking and patellofemoral joint mechanics. However, the role of the VMM in patellar tracking remains the subject of some debate (Mihalko et al., 2007; Powers, 1998), with literature often focusing on the importance of core stabilisers (gluteals, abdominals) rather than the local stabilisers (quadriceps) (Bolgla, 2008; Powers, 2003).

In the clinical setting the presence of the VMO remains controversial (Smith et al., 2009). Reduced VMO muscle bulk measured using MRI scans and reduced electromyography magnitude and timing have been identified in patients suffering patellofemoral pain (Cowan et al., 2001; Pattyn et al., 2011; Souza, 1991). Clinically, rehabilitation programmes directed at strengthening the VMO have reported successful clinical outcomes in these populations (Cowan, 2003; Crossley et al., 2002). However the ability of patients to recruit the VMO in isolation from the other quadriceps muscles has been questioned (Kohn et al., 2013). This is supported by anatomical research, in the fact that the VMO and VML are both supplied by the femoral nerve. Without independent innervation, authors report it is impossible to recruit the VMO independent of the Vastus Lateralis (VL) (Hubbard et al., 1997) (Chapter 2). Studies finding differences between activation timing of VMO and VML have been postulated to occur as a result of poor electrode placement, noise from another muscle (often the adductors) or the presence of pain (Kohn et al., 2013; Smith et al., 2009).

There is certainly confusion and disagreement around this topic. Dynamic structures have been identified to have less of a role in patellar stability than local ligaments and bony anatomy (Senavongse and Amis, 2005). The role of the medial quadriceps in providing mechanical stability to the patella (patellar resistance to lateral displacement) has not previously been closely examined in a physiologically loaded knee with the ITB tensioned. Such an investigation would help establish the role of quadriceps rehabilitation programs for post patellar dislocation patients or those patients with patellofemoral pain.

6.2 Aims

Numerous studies have investigated the role of the VMO in patellar contact mechanics and patellar tracking (Elias et al., 2009; Goh et al., 1995; Lee et al., 2002; Sakai et al., 2000). However these have not addressed the effect of the medial vastus muscles on patellar stability. Based on prior research and current knowledge it was hypothesised that relaxation of the VMO and VML would result in:

1. Increased lateral patellar tracking.
2. Increased lateral contact pressures.
3. Decreased medial contact pressure.
4. Reduced patellar stability.

6.3 Clinical Relevance

Findings presented in this chapter primarily relate to patients who have suffered traumatic patellar dislocation. In extreme cases this could lead to rupture of some VMO fibres from the patella.

6.4 Materials and Methods

6.4.1 Specimen Preparation

A new set of cadavers were used for this study. The specimens were made up of four male and five female right sided, fresh frozen knees of mean age 64 years (range: 48-92). They had no history of knee surgery or disease and were obtained from a tissue bank following approval from the local research ethics committee. MRI scans were taken of all knees and Tibial Tuberosity - Trochlear Groove (TT-TG) distance (Dejour et al., 1994) (range: 11 mm-19 mm), Sulcus Angle (Davies et al., 2000) (range: 131˚ -144˚) and Insall-Salvati ratio (Grelsamer and Meadows, 1992) (range: 0.8-1.1) were each found to be within reported normal limits for all specimens.

The specimens were stored, dissected and prepared for placement in the test rig using the method described in section 3.4.1. The patella was measured and the proximal-distal centre of the median patellar ridge determined. This was termed the 'functional centre point' of the patella (CP) (Merican et al., 2009). In preparation for patella stability testing, a 10 mm diameter cavity was drilled 10 mm deep from the anterior surface of the patella, ensuring the cartilaginous layer of the patella was not broken, at the CP and a polyethylene socket secured into the hole with four 13 mm long screws (Figure 6.1) (section 6.9.2.2).

6.4.2 Test Rig

The muscles and ITB were loaded with 205N, which represented an unloaded open kinetic chain leg extension. Higher loads were not applied to avoid damaging the soft tissues (Farahmand et al., 1998b; Merican and Amis, 2009). The knee was mounted in the test rig as previously described (Chapter 4) (Figure 6.1).

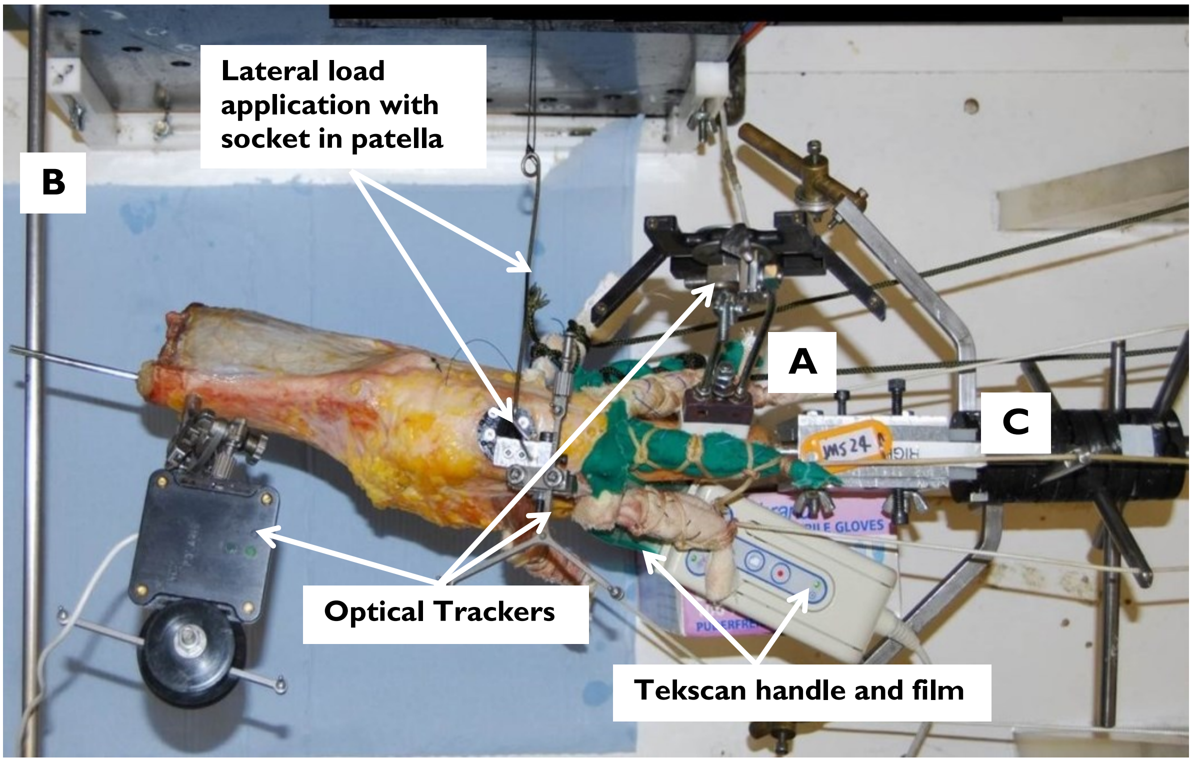

Figure 6.1 Aerial image of the knee rig used for testing. The femur was fixed into the rig via an intramedullary rod with the anterior aspect of the knee facing upwards and the posterior condyles aligned horizontally (A). The tibia was free to move under the influence of the quadriceps and ITB tensions, and held at each angle of knee flexion by a bar across the distal intramedullary rod (B). The muscle loading cables pass over pulleys at the right side of the diagram, to hanging weights (C).

6.4.3 Outcome Measures

6.4.3.1 Contact Pressures and Patellar Tracking

Patellofemoral articular joint pressures were measured using a Tekscan sensor as described in section 4.3.3. Patellar tracking was measured using the Polaris optical tracking system with Toolviewer software, with metal fiducial markers attached to the patella, femur and tibia as described in section 4.3.4.

6.4.3.2 Stability Testing

Patellar stability was measured by applying a 10 N lateral displacing force to the patella using a hook, attached to a weight over a pulley, which fitted into the polyethylene socket at the centre of the patella, pulling it laterally. The application of the lateral load did not inhibit the natural rotations of the patella, enabling all 6 degrees of freedom of patellar motion to be assessed. Patellar motion was described in relation to the femur by standard convention (Bull et al., 2002; Merican and Amis, 2009).

6.5 Experimental Protocol

Knees underwent ten ‘conditioning cycles’ from 0˚-90˚ flexion, in order to minimise hysteresis. Kinematic and pressure data were collected at 0˚, 10˚, 20˚, 30˚, 60˚ and 90˚ of knee flexion. The order of testing was randomised to avoid bias. When the VMO and VML were unloaded, the load which would have been applied to them was redistributed amongst the other quadriceps portions in proportion to their physiological loading (Table 6.1). The ITB was loaded with 30N throughout testing, therefore a constant 205N was applied. Measurements were taken for each condition both with and without an external lateral 10N displacing force applied.

Table 6.1 Loading Conditions.

Knee	VMO Load (N)	VML Load (N)	RF+VI Load (N)	VLL Load (N)	VLO Load (N)	ITB Load (N)	Total Load (N)
Physiological	16	24	61	58	16	30	205
No VMO	0	27	67	63	18	30	205
No VMO or VML	0	0	80	75	20	30	205

6.6 Analysis

Custom written MATLAB scripts calculated mean and peak contact pressures and patellar motion from the raw data produced by the Iscan and Toolviewer software (Appendix A+B). A power calculation, performed using GPower (Version 3.1, Baden-Württemberg, Germany, 2013) determined a sample size of 9 necessary to detect a significant change with 80% power and 95% confidence (Elias et al., 2009). The dependent variables were: mean and peak, medial and lateral facet articular contact pressures, and patellar translation and tilt. Data was analysed in SPSS. A Shapiro-Wilk test confirmed data sets were normally distributed.

Three different effects (Table 6.2) were investigated. Two-way RM ANOVA were performed with the three muscle load conditions (physiologically loaded, no VMO and no VMO or VML) and flexion angle (0˚, 10˚, 20˚, 30˚, 60˚ and 90˚) for each of the dependent variables examined. When muscle load significantly influenced the output, post-hoc paired t-tests with Bonferroni correction were applied comparing the physiologically loaded knee with each of the other conditions at individual flexion angles.

Table 6.2 Summary of Analysis Undertaken.

(n condition =physiologically loaded knee, physiologically loaded knee – VMO and physiologically loaded knee – VMO and VML)

Section	Comparison	Measurement
6.6.1	**The three different muscle load conditions compared***	Values of the physiologically loaded, physiologically loaded knee with no VMO and physiologically loaded knee with no VMO or VML compared at each angle of knee flexion. *n condition
6.6.2	**The combined effect of muscle load and 10N lateral load application****	The 'physiological condition with no load' values were subtracted from each of the three conditions with external load applied (i.e. physiological with 10N lateral load - physiologically loaded knee)** at each angle of knee flexion and compared. **((n condition + 10N load) – physiological condition with no load)
6.6.3	**The lone effect of the 10N load application*****	The values of each condition with no load applied were subtracted from the same condition with the 10N load applied*** at each angle of knee flexion and compared. ***((n condition +10N load) – (n condition))

6.7 Results

6.7.1 Effect of altering medial quadriceps muscle tensions

Removal of the load applied to the VMO, and the VMO and VML muscles together, resulted in an overall increase in lateral motion of the patella (Figure 6.2) throughout knee flexion. With the VMO and VML unloaded, lateral patellar translation (or 'shift') and tilt significantly increased by a mean of 4mm ($P<0.0001$) (Figure 6.3) and 2.8˚ ($P=0.012$) (Figure 6.4); mean medial articular joint pressure reduced from 0.48MPa to 0.14MPa at 0˚ ($P=0.003$) (Figure 6.5). There was a significant increase in mean ($P=0.01$) and peak ($P=0.004$) lateral contact pressure and a significant decrease in peak medial ($P<0.012$) contact pressure. Knee flexion angle had a significant effect on patellar translation ($P=0.002$), mean ($P=0.046$) and peak medial ($P=0.044$) contact pressures, with muscle loading having a greater effect in early knee flexion, before the patella engaged with the trochlear groove. No significant effect of knee flexion angle was found with any of the other variables: peak ($P=0.145$) and mean ($P=0.172$) lateral contact pressures or patellar tilt ($P=0.496$). Figures 6.3-6.6 highlight significant post-hoc test findings.

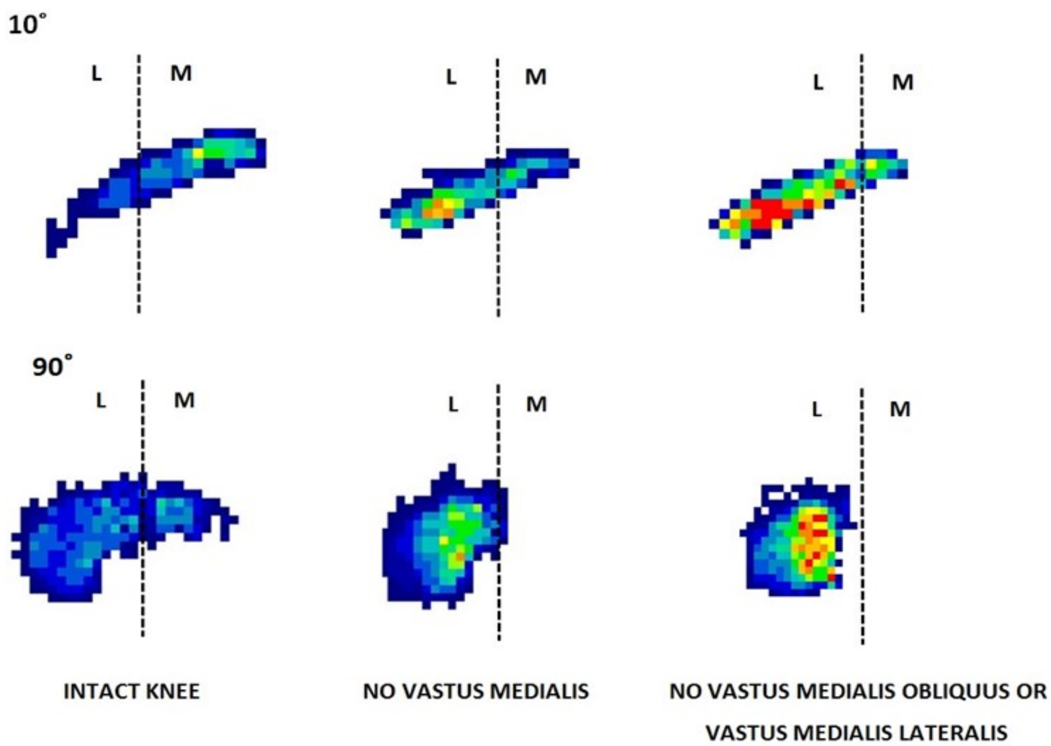

Figure 6.2 Screen Tekscan images of the pressure readings from two different knee specimens one with the knee flexed to 10˚ (top row) and one with the knee flexed to 90˚ (bottom row). It is clear that the contact pressure shifts progressively laterally as the medial vasti muscles are unloaded. This was a typical pattern observed for all knees (scale: dark blue: 0.2MPa; mid-green: 0.5MPa; Yellow: 1.0MPa, Red: 1.5MPa).

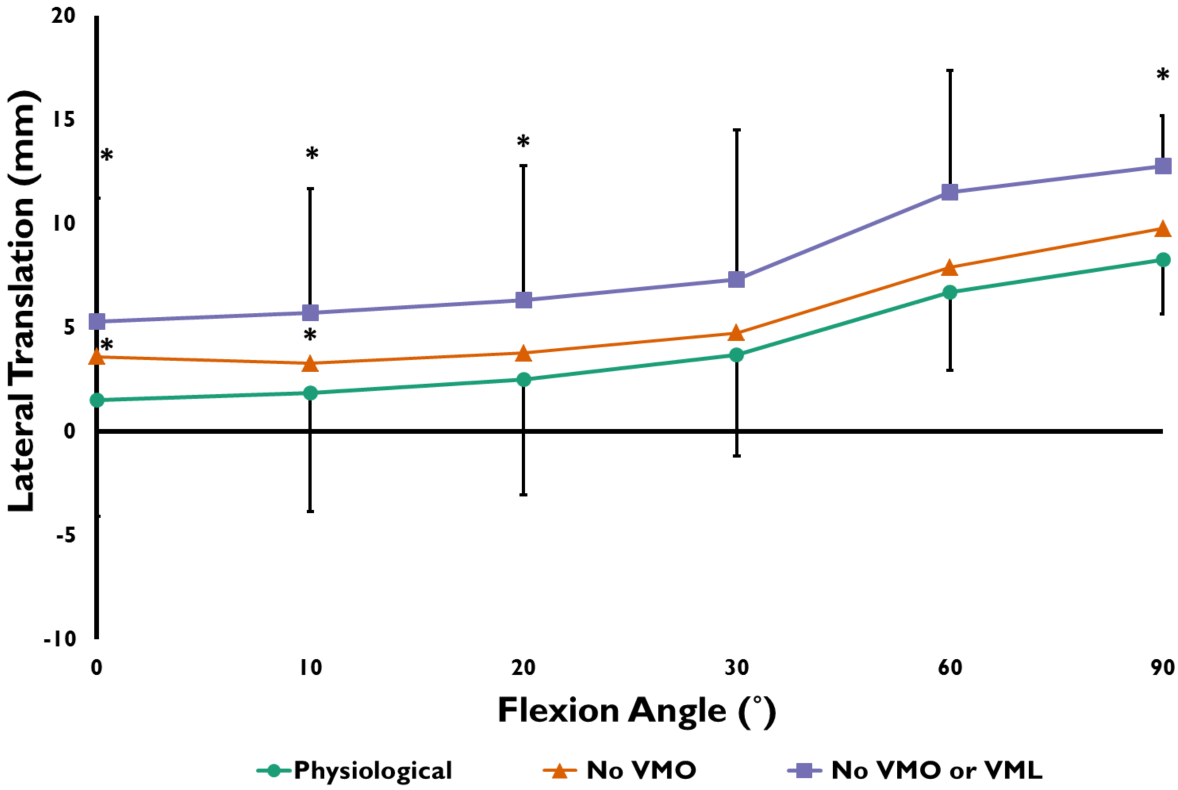

Figure 6.3 Patellar medial-lateral translation (mm; mean ± SD, n = 9) from 0˚ to 90˚ knee flexion with a physiological loaded share across the quadriceps, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML.

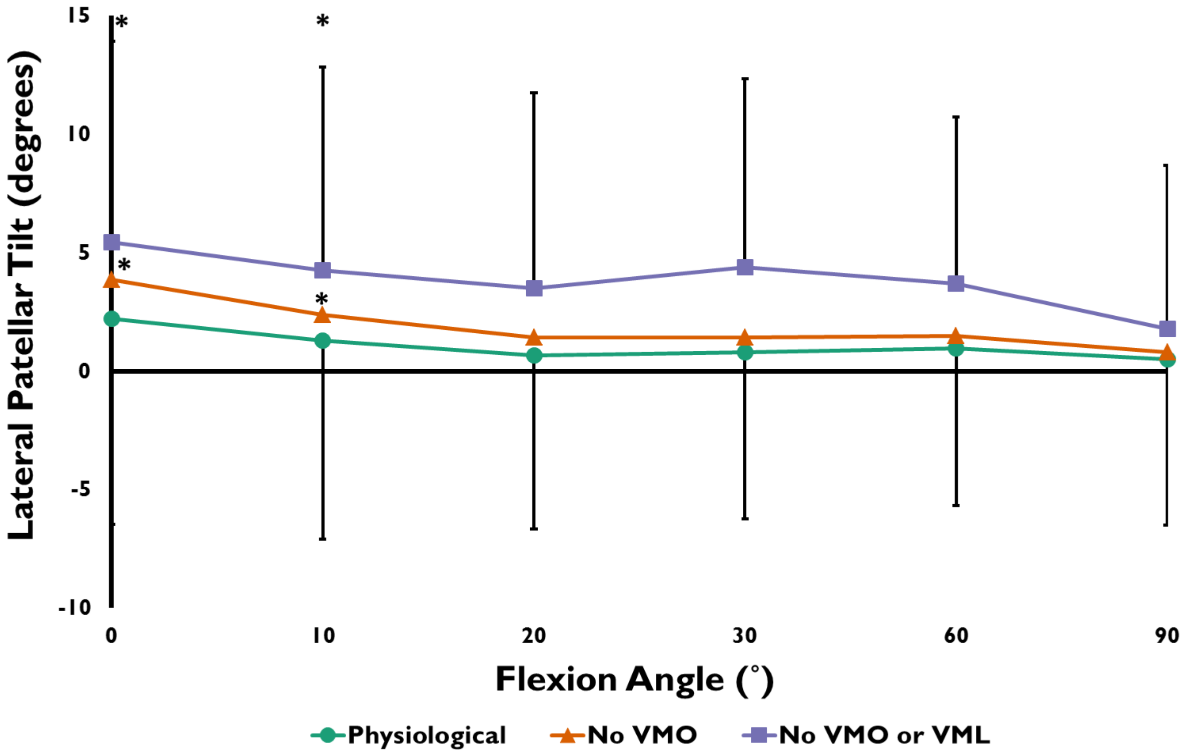

Figure 6.4 Patellar lateral tilt (˚; mean ± SD, n = 9) from 0˚ to 90˚ knee flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML.

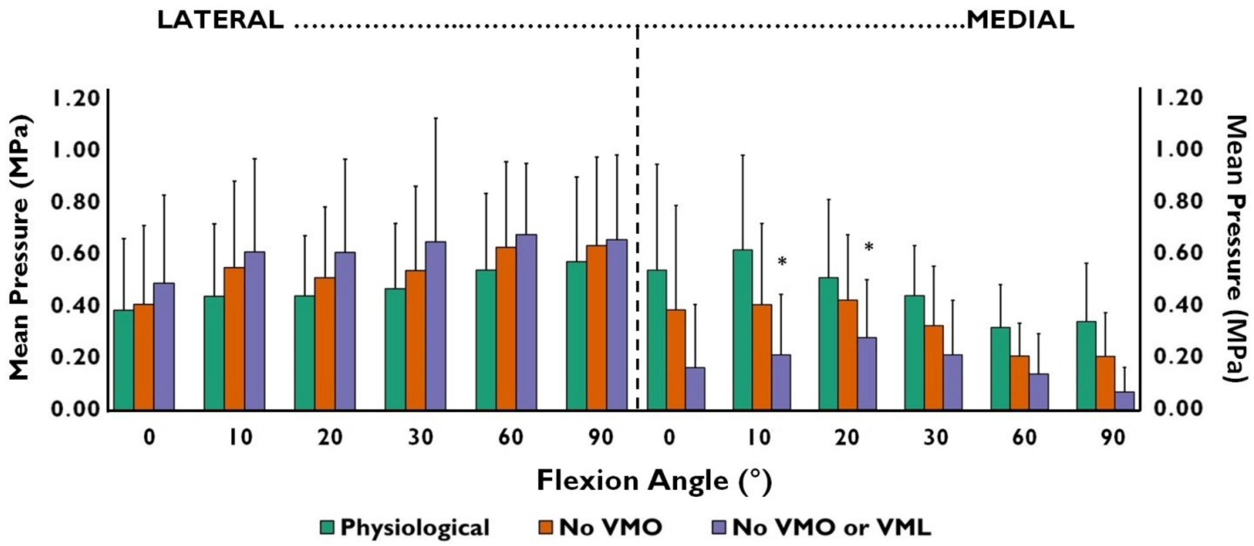

Figure 6.5 Mean lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean + SD, n=9) from 0˚-90˚ flexion for the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML. **P*<0.05.

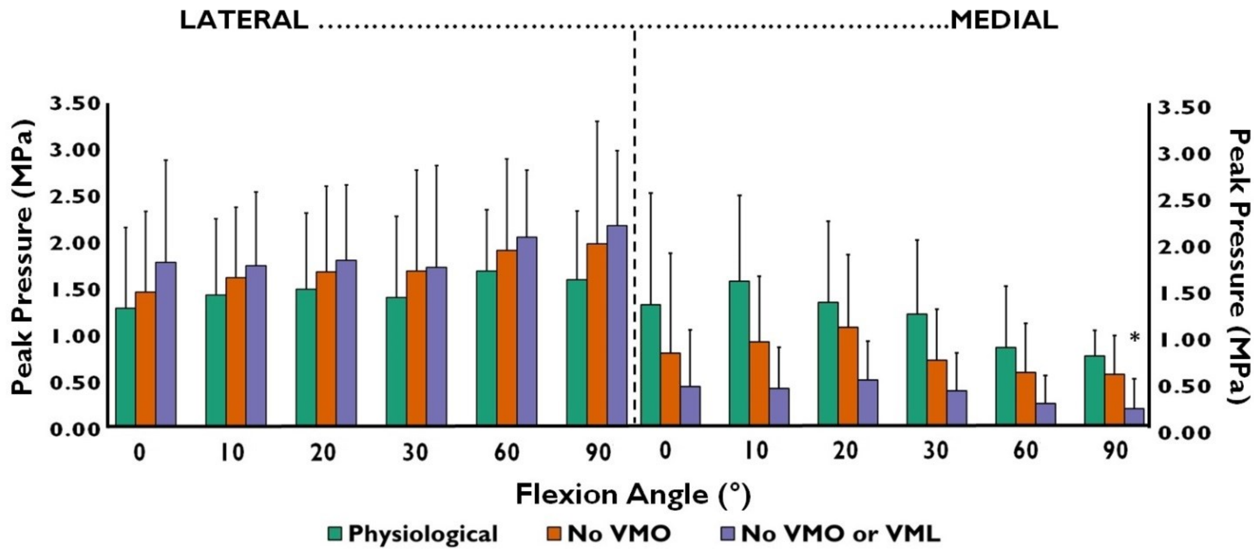

Figure 6.6 Peak lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean + SD, n=9) from 0˚-90˚ flexion for the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML. **P*<0.05.

6.7.2 Combined effect of muscle load + 10N lateral displacing force

The effect of the 10N laterally directed force increased significantly as the VMO and VMO and VML were unloaded. In the physiologically loaded knee the patella was 1.5mm lateral in full extension (Figure 6.3). The 10N lateral force increased lateral patellar translation by 0.9mm (Figure 6.7) to 2.4mm lateral. Unloading the VMO and VML resulted in the patella lateralising 3.8mm, to 5.3mm lateral (Figure 6.3), which was increased a further 1.2mm to 6.5mm lateral following the application of a 10N lateral load. Therefore the combined effect of relaxed VM and 10N lateral load equaled an extra 5mm subluxation (3.8mm + 1.2mm) (Figure 6.7).

There was a significant effect of the lateral load application on all dependent variables investigated: patellar translation (P<0.001) (Figure 6.7) and tilt (P=0.001) (Figure 6.8) and mean medial (P=0.004) and lateral (P=0.006) contact pressures (Figure 6.9) and peak medial (P=0.022) and lateral (P=0.004) contact pressures (Figure 6.10). The increase in lateral patellar tilt was dependent on knee flexion (P=0.021), with the effect of altering muscle load reducing as the knee was flexed deeper. No other variables were affected by angle of knee flexion; peak (P=0.180) or mean (P=0.117) medial or peak (P=0.514) or mean (P=0.510) lateral joint contact pressures or patellar translation (P=0.525). Figures 6.7-6.10 highlight significant post-hoc test findings.

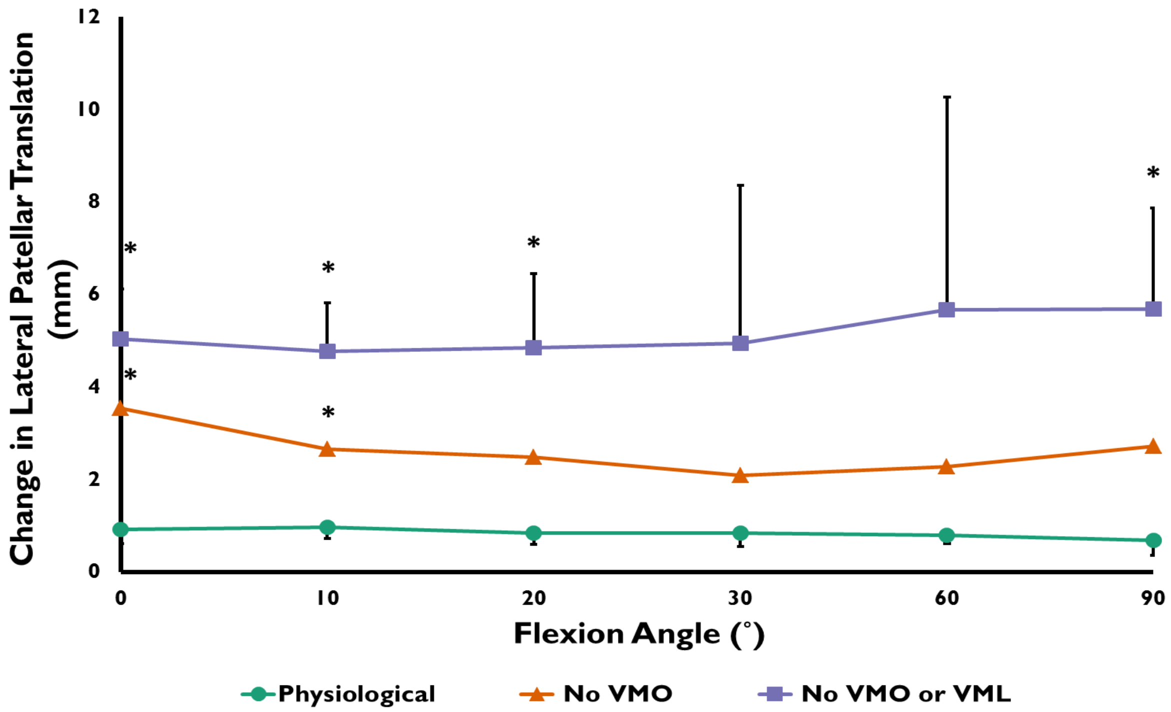

Figure 6.7 Change in patellar medial-lateral translation (mm; mean ± SD, n = 9) caused by the combination of the application of a 10N lateral displacing force and muscle unloading from 0˚ to 90˚ knee flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML. *P<0.05

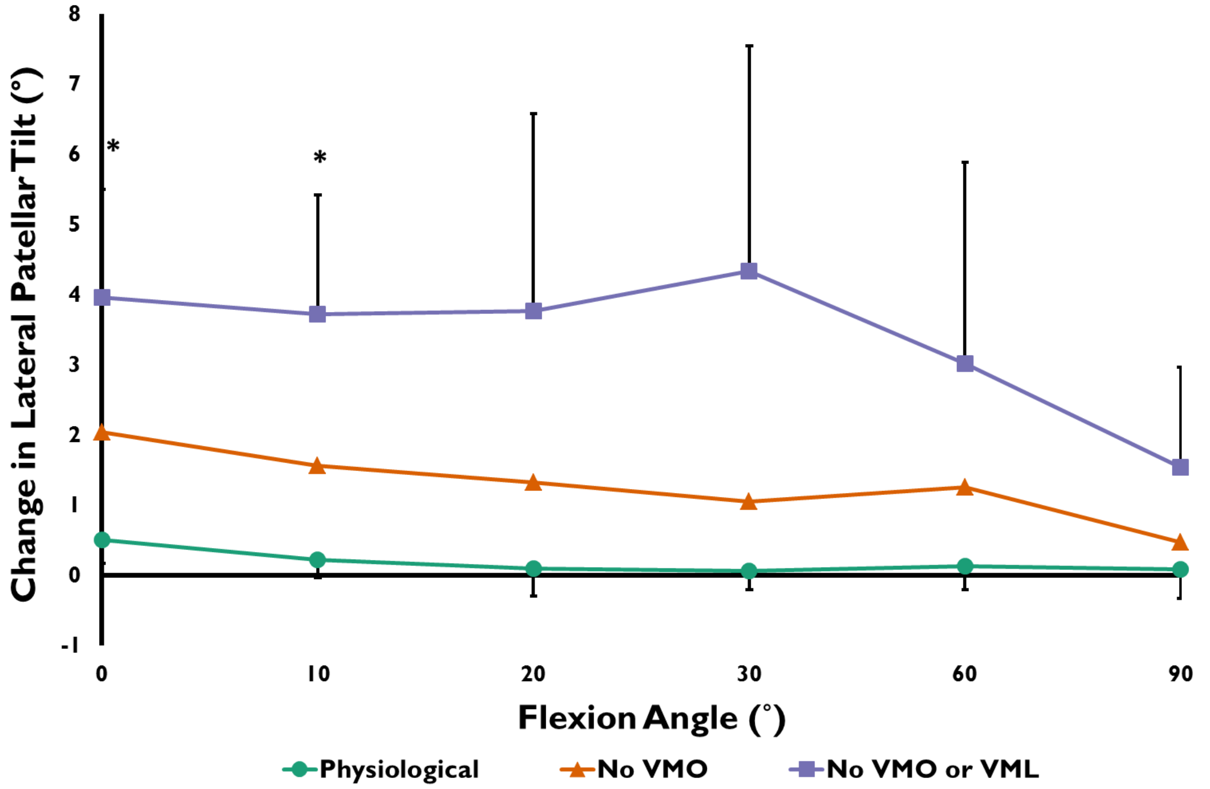

Figure 6.8 Change in patellar lateral tilt (˚; mean ± SD, n = 9) caused by the combination of the application of a 10N lateral displacing force and muscle unloading from 0˚ to 90˚ knee flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML. *P<0.05

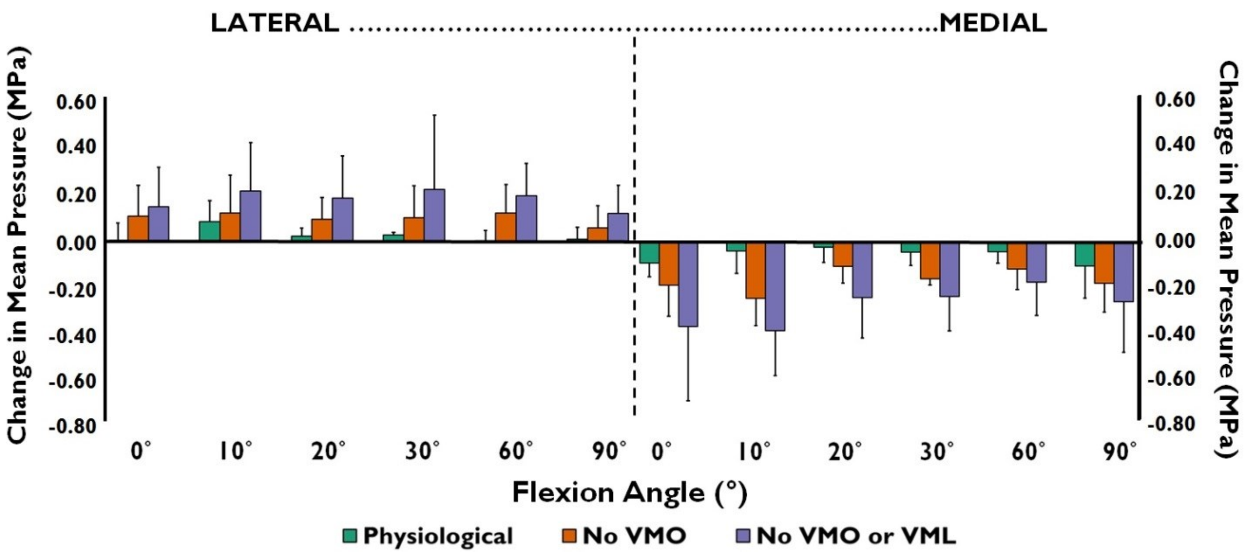

Figure 6.9 Change in mean lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean ± SD, n=9) caused by the combination of the application of a 10N lateral displacing force and muscle unloading from 0˚ to 90˚ knee flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML.

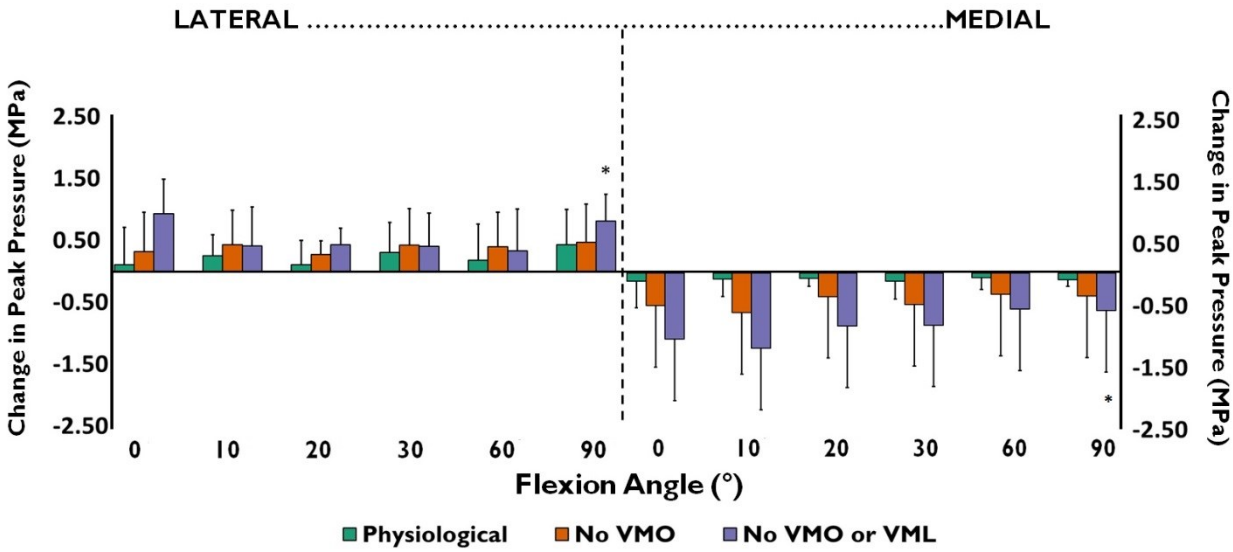

Figure 6.10 Change in peak lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean + SD, n=9) caused by the combination of the application of a 10N lateral displacing force and muscle unloading from 0° to 90° knee flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML. *P<0.05.

6.7.3 Effect of the lateral displacing force on each muscle loading condition

The lone effect of the 10N lateral displacing force was found to be minimal. At 20º with the VMO and VML unloaded, the patella tilted 3.5º lateral (Figure 6.4). The 10N lateral load increased lateral patellar tilt to 4.4º, indicating a change of only 0.9º as a direct result of the 10N load (Figure 6.8). In comparison in the physiologically loaded knee at 20º knee flexion the patella was 0.6° laterally tilted (Figure 6.4). This increased to 0.8º lateral following 10N load application, making a 0.2º change. Therefore the effect of lateral patellar load was extremely small across all measurements (Figure 6.11).

The change in patellar motion following the application of an external load for each of the muscle loading conditions was not found to have a significant effect compared to the condition with no 10N load applied for: patellar translation (P=0.210), peak (P=0.651) or mean (P=0.275) lateral or peak (P=0.280) medial patellofemoral joint contact pressures. Significant effects on patellar tilt (P=0.002) (Figure 6.11) and mean medial (P=0.006) (Figure 6.12) patellofemoral joint pressures were identified. However, as described, although these were reasonably consistent findings with lateral tilt generally increasing further as the quadriceps were progressively unloaded; these changes were very subtle. The results should be interpreted with caution since they are close to the zone of accuracy limits for the measurement tools. Significant post-hoc test findings are shown in Figures 6.11-6.12.

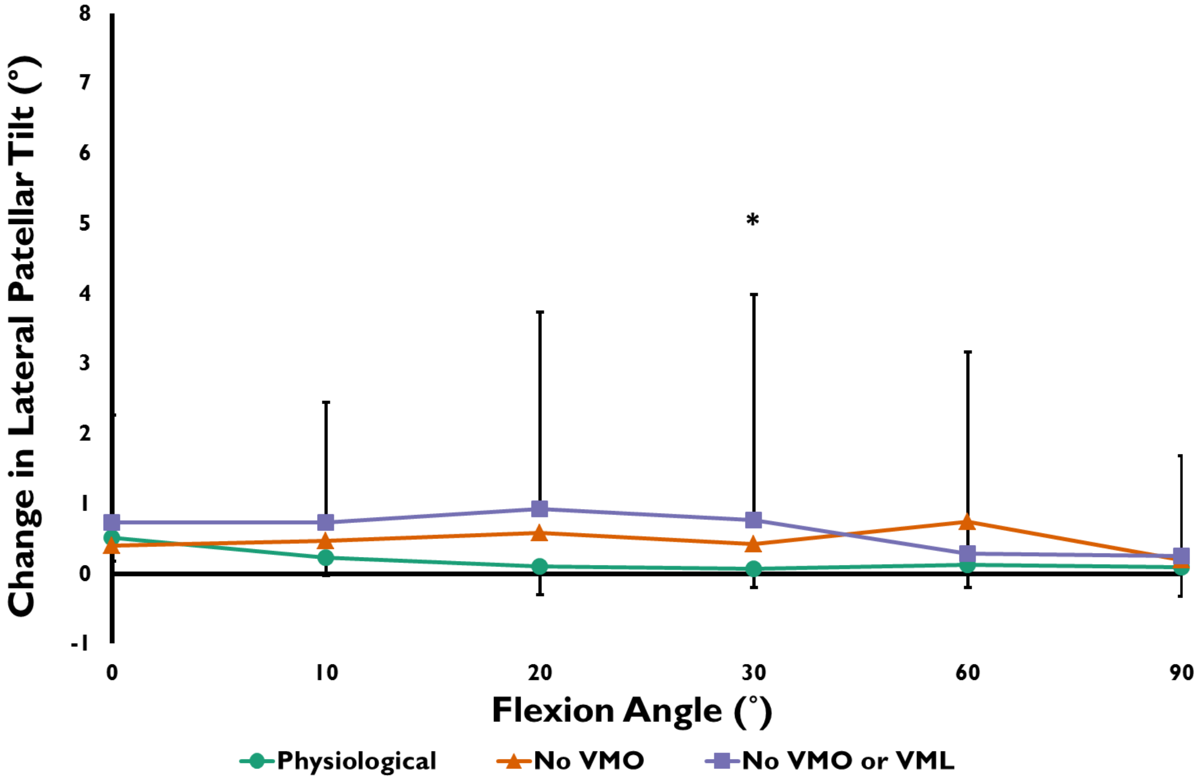

Figure 6.11 Change in patellar lateral tilt (˚; mean ± SD, n = 9) caused by the lone effect of the application of a 10 N lateral displacing force onto the patella from 0˚ to 90˚ knee flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML. **P*<0.05

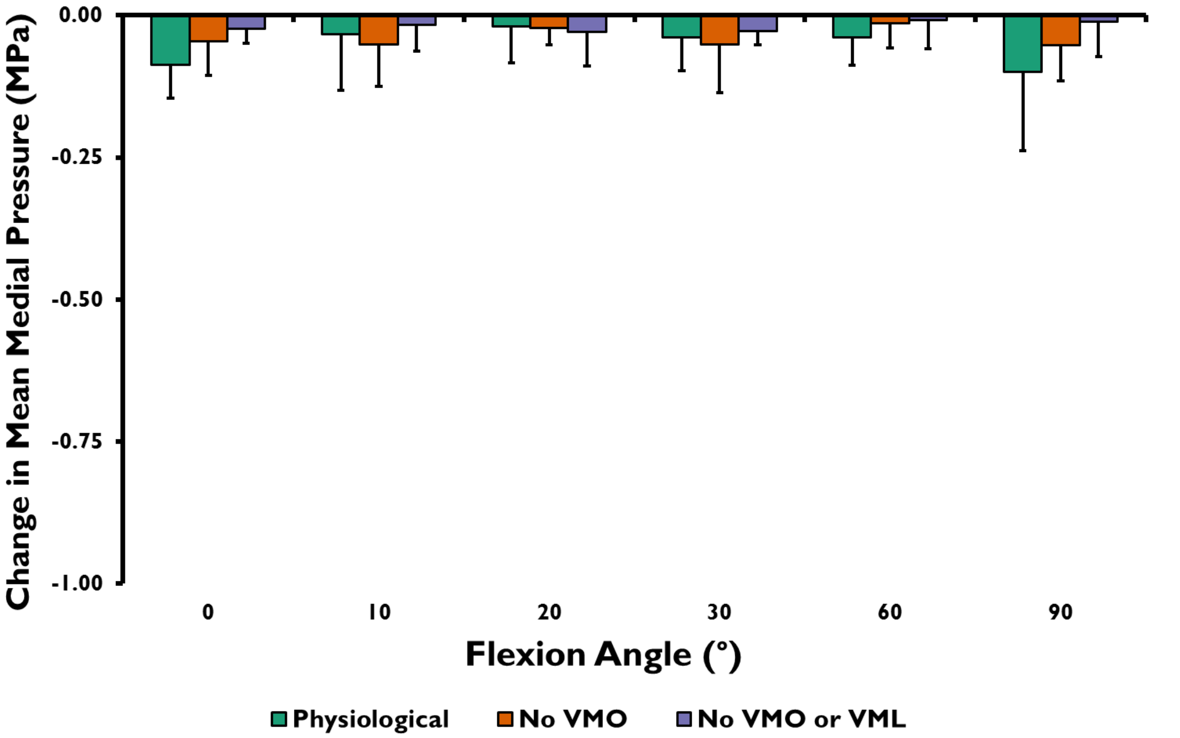

Figure 6.12 Change in mean medial patellofemoral joint contact pressures (MPa; mean ± SD, n=9) caused by the lone effect of the application of a 10 N lateral displacing force onto the patella from 0˚-90˚ flexion in the physiologically loaded knee, the physiologically loaded knee with no VMO and the physiologically loaded knee with no VMO or VML.

6.8 Discussion

Removing tension from the VMO and VML muscles resulted in increased lateral displacing of the patella with significant increases in lateral patellar translation, tilt and mean and peak lateral patellofemoral joint contact pressures alongside reductions in mean and peak medial contact pressures. The effect of a 10N laterally-directed displacing force on the patella was significantly increased following medial quadriceps unloading, highlighting a reduction in patellar stability. The results emphasise that a lack of VMM function increases lateral compartment patellofemoral joint pressure throughout the full range of knee flexion, permitting a more lateralised position of the patella resulting in increased vulnerability to dislocation. Clinical evidence suggests that this may also be related to patellofemoral joint pain (Pattyn et al., 2011; Souza, 1991). Controversy remains regarding the existence of the VMO (Smith et al., 2009). Current findings support its role in patellar stability and are similar to those highlighting the stabilising role of the MPFL (Senavongse and Amis, 2005; Stephen et al., 2013a). VMO atrophy has been described in patients with patellofemoral disorders (Pattyn et al., 2011) and this, considered alongside current findings, provides a rationale for the implementation of quadriceps rehabilitation programmes in patient populations with patellofemoral pathology.

Prior work has found an effect on patellofemoral contact pressure and patellar tracking from altering quadriceps loading (Elias et al., 2009; Goh et al., 1995; Lee et al., 2002; Lieb and Perry, 1968; Powers et al., 1998). However this is the first study to examine these effects on medial and lateral contact pressures and patellar motion simultaneously, through full range of knee motion without having damaged either medial or lateral retinacula during pressure film insertion. Both inhibition and atrophy of the VMO have been identified clinically in patients with patellofemoral joint pain (Cowan, 2003; Crossley et al., 2002). The present findings highlight that removal of VMO tension results in lateralised patellofemoral joint contact pressures and patellar motion throughout full knee flexion. Patellofemoral pain patients commonly report symptoms during activities necessitating quadriceps contraction such as ascending and descending stairs and standing up. It could therefore be hypothesised that lack of VMO function results in elevated contact pressures on the lateral facet of the patellofemoral joint as a consequence of reduced patellar–trochlear groove contact area (Figure 6.2). Future clinical studies could examine this relationship in order to obtain data to support the implementation of quadriceps strengthening programmes and modalities such as muscle stimulators to improve quadriceps activation in patellofemoral pain populations.

Earlier work examining patellofemoral stability has identified the MPFL to be the most important stabiliser through early knee flexion, with bony geometry the most critical factor in deeper flexion (Senavongse and Amis, 2005). However the VMO has generally been considered to have a role, albeit a lesser role, in patellofemoral stability. Prior work in Chapter 4 reported increases of up to 2 mm in lateral patellar translation and 1° lateral tilt following MPFL transection (Stephen et al., 2013a). This study found a similar effect of VMO unloading with increased lateral translation and tilt up to 2.1 mm and 1.7° permitted respectively throughout range. The relationship of the VMO and MPFL has not been consistently defined (Bicos et al., 2007) but it has been suggested that the fibres of each inter-link with one another (Smirk and Morris, 2003), with both structures contributing to patellar stability. Current study findings support the role of the vastus medialis fibres as stabilisers, since removal of applied muscle tension and application of a 10 N lateral load resulted in significantly increased lateral patellar motion and lateral patellofemoral joint contact pressures and reduced medial contact pressures. These findings add some support to the implementation of a VMO and VML based strengthening programme for patellar dislocation patients, which would be less invasive than MPFL reconstruction.

The final part of this study investigated the effect of applying a 10N laterally-directed load to the patella in the physiologically loaded knee and following relaxation of the VMO and both the VMO and the VML. It was anticipated that there would be an increased lateral patellar movement following VMO and VML relaxation compared to the effect of the lateral force on the physiologically loaded knee and that without the VMO. Although this was identified for patellar tilt and mean medial pressure, the effect was not significant across the other dependent variables measured. This could suggest there is indeed no effect of this. In part this may have resulted because the quadriceps were elevated from the femur, which may have resulted in a mild loss of stability. Alternatively it could suggest that 10N was an insufficient load to simulate *in vivo* forces. However the load was kept constant throughout testing, was chosen to avoid causing any damage to the restraining passive soft tissues in the absence of the normal quadriceps action (section 6.9.2), has been used previously in stability testing (Nomura et al., 2000a) and was comparative to the quadriceps loads used in the study (205N) as opposed to *in vivo* loads. Future studies may be merited to investigate a similar hypothesis with a greater lateral load applied.

The existence of the VMO remains contentious. A common clinical observation in patellofemoral patients is a loss of muscle bulk in the area where the oblique fibres of the VMO typically insert to the patella. However this atrophy can commonly be reflective of a loss of bulk throughout the whole of the quadriceps muscle overall, rather than just an isolated loss of VMO bulk. The anatomy in this area is known to demonstrate wide inter subject variability and anatomical studies have consistently failed to distinguish VL and VMO (Blazevich et al., 2006; Nozic et al., 1997). Throughout this commentary therefore it has been recommended that the current study findings support the use of quadriceps strengthening exercises in the management of these patients rather than VMO directed exercise, since the evidence would suggest it is impossible to independently contract the two (Smith et al., 2008). The present study outcomes highlight the significant contribution of the medial vasti muscles to patellofemoral joint tracking and stability. Loss of VMO and VMO+VML function resulted in significantly increased lateral patellar tracking, with increased lateral patellofemoral joint contact pressures and reduced medial joint contact pressures and patellofemoral joint stability. VMO atrophy has been described in patients with patellofemoral disorders and taken together with this study's findings, suggests a role for quadriceps strengthening rehabilitation programmes. The consequence of VMO unloading was found to be similar to that of MPFL sectioning previously reported. This provides some support for the implementation of conservative rehabilitation programmes for patients post patellar dislocation.

6.9 Key Findings / Conclusions

Key findings from this work therefore are that weakening of the VMO and VML muscles causes:

1. Increased lateral patellar translation and tilt.
2. Reduced medial patellofemoral joint contact pressures.
3. Elevated lateral patellofemoral joint contact pressures.
4. Reduced patellar stability.

6.10 Addendum

6.10.1 Quantification of quadriceps weakness

Following on from the surgical intervention reported in Chapter 5, this section of the thesis was undertaken to investigate the effect of conservative management on a population of patients with patellofemoral pain or patellar instability. Review of the literature determined this to be an extensively examined area, a reflection of the controversy present around the function of the VMM. However there appeared a gap in the literature that no prior cadaveric work had looked at the effect of quadriceps weakness on patellar stability whilst the ITB was loaded. Furthermore the VMO and VML had not been examined in depth previously. It was therefore decided to investigate this in greater detail.

Firstly the loading applied to the knees during the different test conditions required consideration. There were two main ways this could be investigated:

- Weakening the VMM.
- Complete removal of any tension applied to the VMM.

Initially it was assumed that removal of the full tension from the medial quadriceps muscles would be an unrealistic representation of what occurs *in vivo*. However following review of the literature it was determined impossible to quantify the change in muscle activity clinically in the VMM as a result of patellofemoral pain or instability. This was a consequence of the limitations of electromyography data and also the wide inter subject presentations evident in the different sub groups of patellofemoral pathology populations (Hiemstra et al., 2013; Smith et al., 2008).

It was therefore determined that an ideal solution would be to test multiple conditions of VMO and VML weakness and inhibition. A pilot study was therefore conducted to examine progressive VMO and VML tension reduction of 25%, 50%, 75% to 100% in turn. The weights removed from the medial quadriceps were re-distributed to the central and lateral quadriceps in accordance with the physiological cross sectional area of the muscles (as described in section 6.3). During pilot testing of the first specimen, the data identified a clear tend; progressively increased lateralisation of patellar motion and elevated lateral patellofemoral joint contact pressures as the VMO and VML were progressively further unloaded.

This testing protocol required the knees to undergo considerable soft tissue loading, particularly on the VLO muscle which in some specimens was quite weak and comprised mainly fatty tissue. In order to apply a constant 205N to the knee to enable comparison of results, the VLO had to be tensioned with more than the 16N previously reported (Farahmand et al., 1998a). Five test conditions were conducted each with the VMO and then the VMO and VML weakened by 25%, 50%, 75% and 100%. At the conclusion of the first pilot study, there was significant damage and tearing of some of the muscle fibres at the VLO evident. Unfortunately during the second pilot test in the second knee tested, the VLO muscle tore away from the rest of the quadriceps, as it was not strong enough to withstand the increased load applied to it. Therefore to minimise the repetitive demands placed on the soft tissues and instead of investigating all 5 conditions of VMM weakness, the investigation was limited to just one test. It was chosen to replicate the worst possible clinical scenario; that of removal of the VMO and VMO and VML tension completely. This enabled the results of this experiment to be compared with the previous results investigating the effect of MPFL transection on patellar tracking and patellofemoral contact mechanics.

6.10.2 Measurement of patellar stability

6.10.2.1 Background

Objective patellofemoral instability is clinically the term given to define patellar maltracking, subluxation of the patella relative to the femur and, on occasion documented episodes of dislocation. “Instability” also covers the subjective description given by patients to a sensation they get when the patella feels as if it is going to dislocate. In the laboratory setting a reduction in patellar stability can be measured in changes to kinematics or contact areas and patterns. The direct and invasive testing permitted in the laboratory mean stability can be examined rigorously, unlike in the clinical settling where it is extremely challenging to control and quantify motion of the patella in six degrees of freedom. The disadvantage of laboratory testing is inevitably related to the previously discussed challenges of simulating *in vivo* patellofemoral joint mechanics and kinematics.

A method for examining patellar stability was described and devised by a two former PhD students in the biomechanics group (Merican et al., 2009; Senavongse and Amis, 2005). This method permitted six degrees of freedom of patellar motion to be examined without interference of the load application. This was something prior authors had found challenging to achieve (Huberti and Hayes, 1984; Panagiotopoulos et al., 2006). It was therefore decided to use the method described by Senavongse and Amis (2005) and Merican et al (2009) to examine stability in the present study.

6.10.2.2 Method

An ACL jig (Arthrex part number AR1875, Naples, FL, USA) was positioned perpendicular to the anterior face of the patella and clamped to the edges of the patella (Figure 6.13). Using the jig a drill was centred over the proximal-distal centre of the median ridge of the patella. The median ridge was viewed clearly and measured from the anterior patellofemoral joint capsule incision made previously to insert the Tekscan sensor, thus retinacular tissue was not disrupted. The patella was measured and the proximal-distal centre of the median patellar ridge determined. This was termed the 'functional centre point' of the patella (CP). The patella was then firmly held and a 2mm hole drilled 10 mm deep from the anterior surface of the patella ensuring the cartilaginous layer of the patella was not broken. This was then progressively enlarged until it was 10mm. A specially made polyethylene socket was then secured into the hole with four 13 mm long screws (Figure 6.13). A wire hook was specifically made (Figure 6.14) to fit securely into the socket attached to the patella. This was then attached to a piece of string and run over a low-friction pulley with a weight hanging on the end of the cord to apply the lateral displacing force.

6.10.2.3 Load

The second question to be answered was to determine the optimal displacing force to pull the patella laterally. The following methods had been used previously to assess patellar stability:

- Application of a 10N lateral load (Nomura et al., 2000a).
- Application of a 100N lateral load (Ostermeier et al., 2007a).
- Measurement of the magnitude of forces resisting 10mm lateral displacement (Merican et al., 2009; Senavongse et al., 2003).

Due to the rig set up in the laboratory, it was not possible to operate the optical tracking system, Tekscan system and Instron machine simultaneously whilst testing the knee in the rig described. Therefore it was not possible to replicate the method of Merican et al (2009). Instead a number of pilot studies were performed to establish the optimal load to impart a lateral displacing force to the patella. One hundred, fifty, twenty and ten Newtons were all tested, based on prior research. It was found that 100N and 50N caused significant strains on the local soft tissues following early pilot testing. Therefore these were decided to be inappropriate for use.

Further testing considered 20N and 10N loads and found very little difference between them. It was decided ultimately to use a 10N displacing load since this had been used in prior work, and it was thought this would minimise as far as possible any soft tissue damage from the imposed load. This was important not only for the present study, but also since the same specimens would be used for two further experiments after completion of this one. Therefore avoiding soft tissue damage was of vital importance.

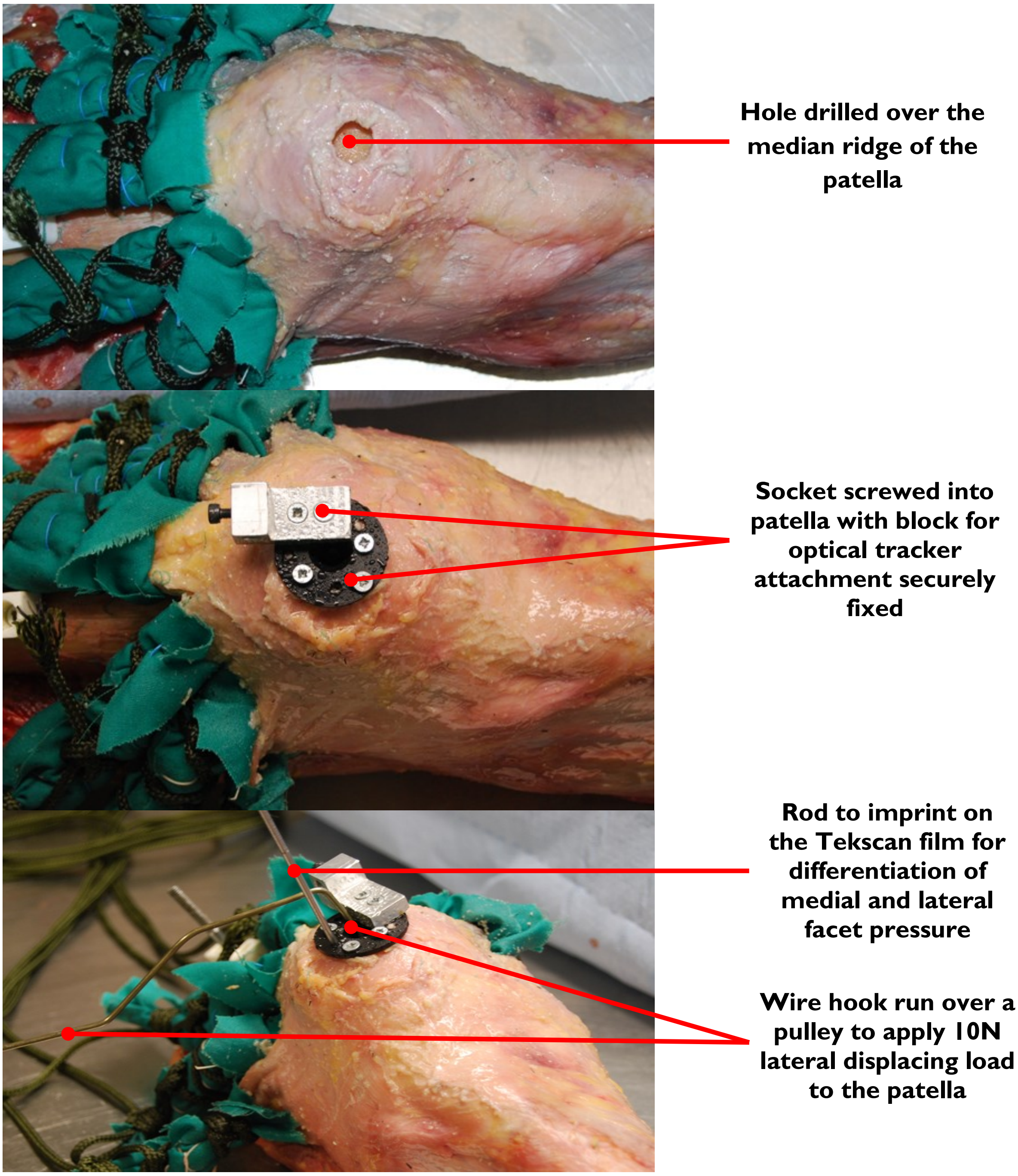

Figure 6.13 Insertion and fixation of the socket into the patella for application of a lateral displacing force to measure patellar stability (steps 1-3 from top to bottom).

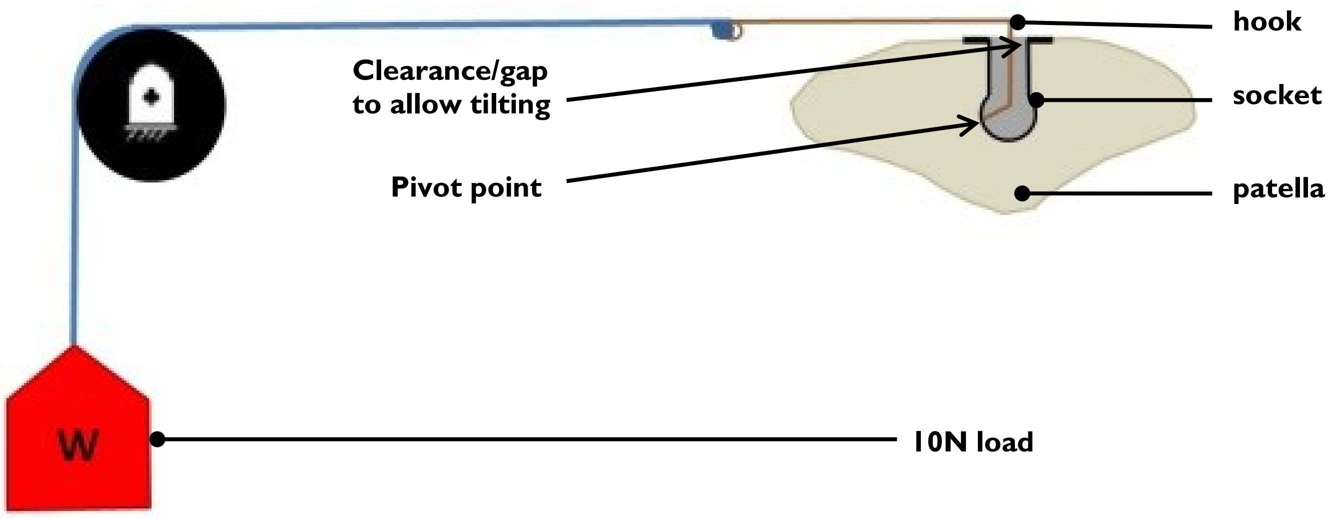

Figure 6.14 Method for 10N lateral load application to the patella.

6.10.3 Conclusion

Pilot testing identified concerns relating to soft tissue damage in the lateral muscles as a result of undertaking multiple test conditions examining the effect of unloading the medial quadriceps muscles. Therefore the present study examined only one condition: that with the VMO and the VMO and VML completely unloaded.

Patellar stability was examined with a socket inserted to the patella along the centre of its median ridge. A 10N lateral displacing load was applied to the patella. This was determined to be the optimal weight to detect change following load application, but avoid any local soft tissue damage caused by testing.

CHAPTER 7

TIBIAL TUBEROSITY – TROCHLEAR GROOVE DISTANCE

7.1 Background

Prior chapters in this thesis have highlighted the challenges faced by surgeons treating patients suffering patellar dislocation or pain and it is recognised that when conservative measures fail surgical intervention is often appropriate (Nomura et al., 2000a). The indications for MPFL reconstruction have been highlighted in earlier chapters (Chapters 3-5). However in a subsection of patients suffering dislocation, there will be additional anatomical anomalies present, alongside MPFL rupture, which contribute to their initial injury and could potentially cause perpetuation of their symptoms, resulting in re-dislocation if they are not adequately addressed (Dejour et al., 1994). Such factors may also impact on the decision process of whether to manage the patient conservatively or surgically.

Elevated Tibial Tuberosity-Trochlear Groove (TT-TG) distance has been identified as a risk factor for patellar dislocation and linked to recurrent dislocation (Chapter 3) (Pennock et al., 2013). In the adult population an early study identified a sub section of those patients with a history of patellar dislocation which had TT-TG distances of 20mm, compared to 13mm in controls (Dejour et al., 1994). Subsequent work has investigated younger athletes, finding a similar trend where patients suffering instability had mean TT-TG distances of 14.6mm, compared to 10.3mm in one study and 16.3mm versus 11.7mm respectively in another (Balcarek et al., 2010; Pennock et al., 2013). The purpose of the final section of this thesis is therefore to investigate the relationship of the MPFL and the TT-TG distance.

Tibial tuberosity (TT) transfer surgery to reduce an elevated TT-TG distance has been practised over the last century and has gained popularity in recent years for the treatment of patients suffering patellar dislocation. The surgery is only appropriate for those patients with an excessively lateralised TT in relation to the position of the deepest part of their trochlear groove (Figure 7.1) The surgical rationale is correction of the underlying extensor mechanism malalignment which is thought to cause maltracking of the patella in the trochlear groove, resulting in unbalanced cartilage loads, joint incongruity and potentially patellofemoral pain and joint instability as a consequence of the increased lateralised vector acting on the patella (Koëter et al., 2007) (Chapter 2: section 2.5.3.1). Clinical outcomes reported in case series of patients having undergone TT transfer are variable and inconsistent. Worsened post-operative pain, accelerated osteoarthritis and deteriorated functional outcomes are amongst some of the findings reported during follow up of these patients (Aglietti et al., 1994; Arnbjornsson et al., 1992; Crosby and Insall, 1976; Juliusson, 1984; Mäenpää and Lehto, 1997b). However, positive outcomes of reduced pain and re-dislocation rates and improved function have also been reported (Koëter et al., 2007; Pritsch et al., 2007). This highlights that although the procedure is aggressive, it can provide symptomatic relief for patients. An underlying problem however, is that it is difficult for the surgeon to know how much to reposition the tuberosity, in order to obtain the best result.

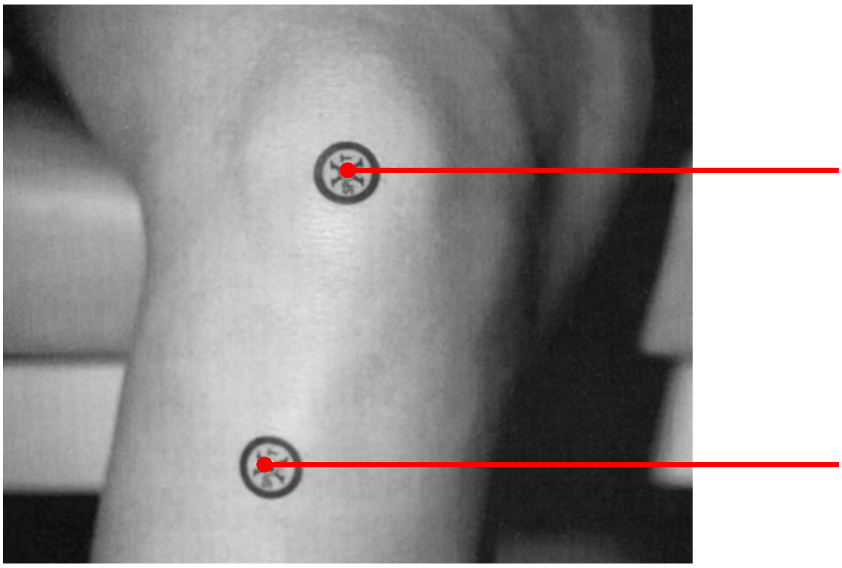

Figure 7.1 The right knee of a patient with an elevated tibial tuberosity-trochlear groove distance (measured with CT and found to be 22mm). The tibial tuberosity is markedly lateral to the inferior pole of the patella (as adapted from Sanchis-Alfonso et al. (2006)).

Given the complications which follow from incorrect positioning of the TT, pre-operative surgical planning appears to be crucial to optimise post-operative outcome. The procedure entails displacement of the tuberosity from its position on the tibia to enable realignment of the patellar tendon, typically to a medialised position. Prior cadaveric research has found that over-medialisation can result in elevated medial joint contact pressures (Kuroda et al., 2001). Currently limited guidelines are available to define the threshold of TT malalignment at which pathology is confirmed and intervention deemed appropriate / necessary. Normal ranges for TT-TG distance have been inconsistently reported: one study using MRI reported a range from 9-10mm (Pandit et al., 2011), a further MRI study found an average of 9.4mm (Wittstein et al., 2006) whilst a study investigating healthy volunteers using CT scans identified a mean TT-TG of 13.6mm (Alemparte et al., 2007). TT-TG distances in excess of 15mm have been highlighted to be pathological in populations of patellar dislocators (Balcarek et al., 2011; Dejour et al., 1994; Pennock et al., 2013). However case series are present describing tibial tubercle osteotomies in skeletally mature patients with patellar instability and TT-TG distances close to 10mm (Tecklenburg et al., 2010) and 15mm (Diks et al., 2003; Koëter et al., 2007). Evidently this is an area which would benefit from further investigation, and requires clarification before the role of TT-TG distance and MPFL state is more closely examined (Chapter 8).

Laboratory based cadaveric experiments have induced patellofemoral malalignment and found this results in adverse patellofemoral joint contact mechanics and kinematics (Kuroda et al., 2001;

Ramappa et al., 2006; Saranathan et al., 2012). These reports, alongside evidence of inferior clinical outcomes, have led to surgeons being warned over the consequence of over-medialised tibial osteotomies, leading to caution over the use of tibial transfer surgeries (Aglietti et al., 1994; Mäenpää and Lehto, 1997a). However, crucially, laboratory studies have not previously been performed on cadavers with the ITB loaded. It was hypothesised that this would lessen the adverse effects of TT medialisation and therefore was appropriate to examine in greater detail. Furthermore the effect of changing TT position on patellar stability has not previously been investigated.

7.2 Aims

The aims of the current study were therefore to determine the effect of progressive 5mm, 10mm and 15mm medialisation and lateralisation of the TT on patellar stability and patellofemoral contact mechanics and tracking. It was hypothesised that over medialisation would result in:

1. Increased medial patellofemoral joint pressures.
2. Reduced lateral patellofemoral joint contact pressures.
3. Increased medial patellar translation.
4. Increased medial patellar tilt.

Over lateralisation, meanwhile was hypothesised to result in:

1. Reduced patellar lateral stability.
2. Elevated lateral patellofemoral joint contact pressures.
3. Reduced medial patellofemoral joint contact pressures.
4. Increased lateral translation.
5. Increased lateral patellar tilt.

7.3 Clinical Relevance

This chapter considers a different population of patients compared to the prior chapters. The results relate primarily to patients who suffer recurrent, chronic patellar dislocations as a result of abnormal alignment of their extensor mechanism.

7.4 Materials and Methods

7.4.1 Specimen preparation

The local Research Ethics Committee granted approval for this study to be conducted. The experiment follows directly on from the investigation in Chapter 6, with testing performed on the same set of specimens. The knees were already located in the test rig, with the muscles dissected and loaded as outlined in section 3.4.2. The Tekscan sensor, optical tracking markers and polyethylene socket were also in position and ready for testing to commence. This experimental design ensured the cadaveric resources obtained were responsibly used.

At the conclusion of testing in Chapter 6, unfortunately the fibres of the VLO muscle in one of the nine knees failed and broke away from its distal insertion site. For this study therefore, only eight knees were available to test; five female and three male specimens of mean age = 61 (range 48-79). None had any history of disease or knee surgery. Patellar height (range= 0.9-1.1), sulcus angle (range = 131˚ -140˚) and TT-TG distance (mean=10.4mm; range; 8.1mm-13.5mm) were all determined to be within previously reported normal ranges (Dejour et al., 1994; Grelsamer and Meadows, 1992; Koskinen et al., 1993; Pennock et al., 2013). Throughout testing, as in prior experiments, specimens were kept moist with occasional water spraying.

7.4.2 Measurements

The teskcan sensor measuring patellofemoral joint contact pressures and optical trackers attached to the patella, tibia and femur (Chapter 4) were used to examine outcomes from the present study. Patellar stability was again examined by applying a 10N lateral displacing force to the patella via the polyethylene socket situated at the centre of the patella (Chapter 6).

7.5 Testing Procedure

Checks undertaken during testing outlined in Chapter 6 had ensured correct operation of all equipment and that any hysteresis of the specimens was minimised.

Pilot study testing found that soft tissue tightness could restrict the planned changes in TT positioning (Section 7.10.2). Therefore two incisions were made in the superficial fascial layer 5 mm outside both the medial and lateral edges of the TT. These were extended 2cm in a proximal direction and performed on all knees prior to commencement of testing. This step was used to represent the tissue releases, which would be necessary to perform to mobilise the patellar tendon during surgery, in a standardised and reproducible manner. Pressure, kinematic, and stability measurements were then recorded through a range of knee flexion at 10° increments from 0°-30° inclusive, then 60° and 90° flexion.

An extended TT osteotomy was performed by the one experienced orthopaedic surgeon, who ensured that the bone was cut in a coronal plane, parallel to the femoral condylar axis, on all eight knees tested. To standardise progressive changes of TT position eight identical metal T-plates with top plates were custom made to fix to the anterior aspect of each tibia (Figure 7.2) (Section 7.10.2).

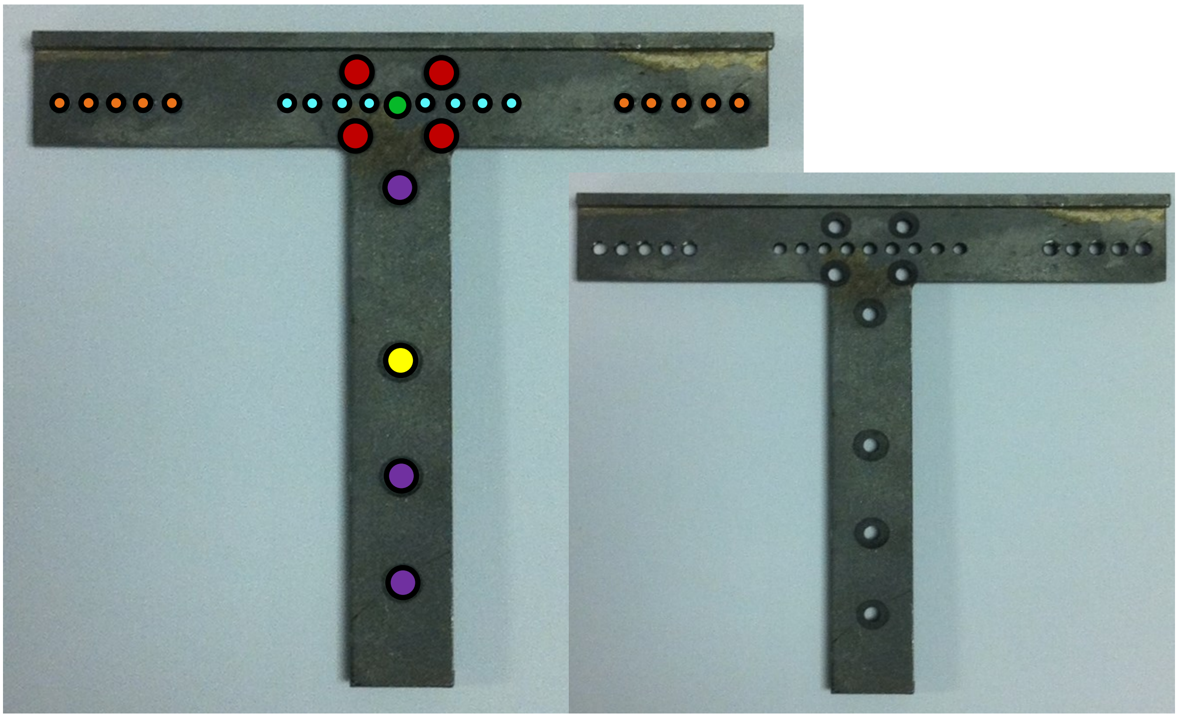

Figure 7.2 Photograph of the final T-plate design following pilot testing and refinement described in section 7.9.1. Colouring on the left image indicates the purpose of each of the holes:

Green hole: for positioning of the locking rod to align the location of the anatomical tibial tuberosity position.

Blue holes: for positioning of the locking rod to permit progressive 5mm medialisation and lateralisation of the tibial tuberosity.

Orange holes: for fixation of a top plate to clamp the tuberosity into position.

Red holes: for bicortical fixation of 4 bone screws to secure the plate to the anterior tibia.

Purple holes: for fixation of the plate with 8mm screws to the anterior tibia.

Yellow hole: a pivot point 50mm distal to the anatomical hole; the TT was rotated about this point.

During measurement and cutting, the knee was held in full extension in the test rig and the coronal plane of the tibia was assumed to be parallel to the base of the test rig. To ensure accurate re-positioning of the tuberosity following detachment from the tibia, a broken line was drawn across the proximal border of the TT. The most anterior central point of the tuberosity was then identified (Figure 7.3) and a 2mm hole drilled the full depth of the tibia from anterior to posterior, through the cortical bone, exiting the tibia posteriorly approximately 100-150mm deep, depending on the knee size. A further hole was also drilled 50mm distal to this as a reference point to 'pivot' the osteotomy from following fixation. This was to replicate the technique performed clinically as closely as possible (Koëter et al., 2007).

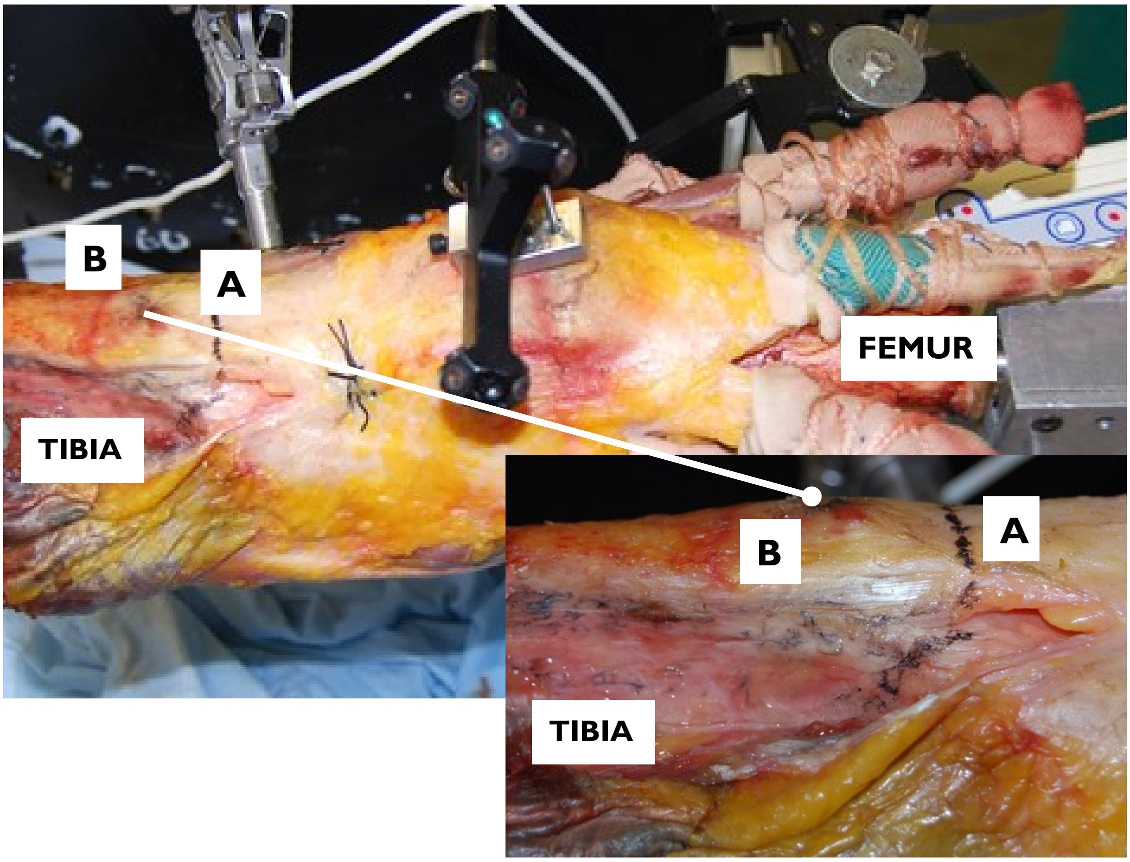

Figure 7.3 Left image of the knee positioned in the rig, with a black dashed line drawn along the proximal border of the tibial tuberosity (A). A mark has then been made at the most anterior point of the TT to drill the reference point prior to plate fixation (B). This is shown in greater detail in the right image with a close up and side on view of the tibial tuberosity.

Two further lines were drawn along the medial and lateral edges of the tibia and tuberosity, 10mm posterior to the described TT point. They were extended along the length of the tibia parallel to the defined axis. Any soft tissue covering these lines was removed with a No. 10 scalpel. The scalpel blade was also then pressed into the lines drawn along the medial and lateral tibial borders to make a groove (Figure 7.4). This made the osteotomy more accurate for the surgeon to perform.

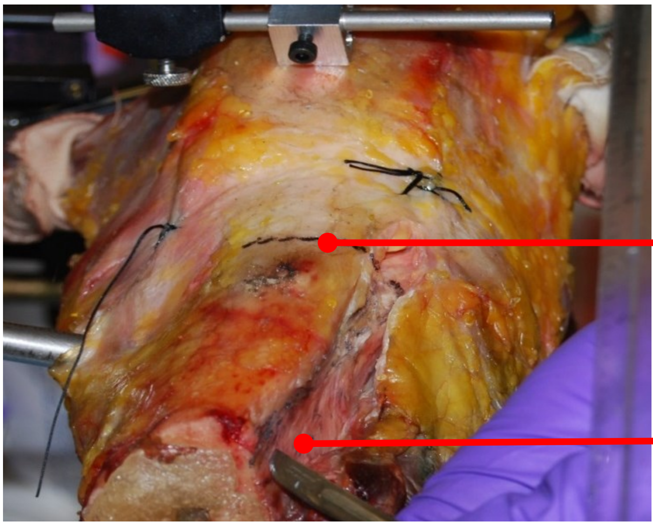

Figure 7.4 A distal to proximal view of one of the specimens positioned in the test rig. The soft tissues along the medial line of the tibial border have been cut away to expose the bone locally and a line drawn 10mm inferior to the previously defined TT point.

A cut was made, using an oscillating saw, at the lateral border of the tubercle and extended 120mm along to the distal tibia. A 4mm osteotome and No.10 scalpel blade were used to free the anterior tibial osteotomy away from any surrounding tissues allowing it to be elevated from the tibia. A second cut was made precisely 4mm beneath the first cut and the distal ends freed. A 4mm thick block of bone was then removed from the anterior tibia (Figure 7.5). This was the same thickness as the T-plate, meaning it could be inserted without influencing patellar kinematics or contact mechanics. Care was taken to ensure that the bone cuts were parallel to the base of the test rig, and thus in the coronal plane of the extended tibia, so that the tubercle wold not move anteriorly or posteriorly when it was displaced medially-laterally (Figure 7.6C).

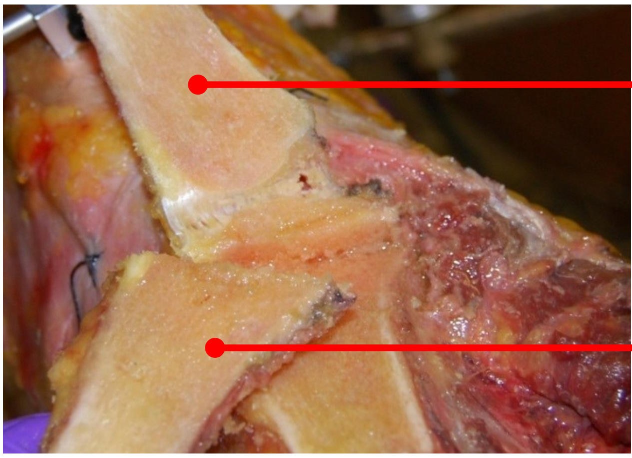

Figure 7.5 Removal of a 4mm thick bone block from the anterior tibia to enable the T-plate to be fitted into position.

The T-plate was then positioned with the green hole (Figure 7.2) aligned with the pre drilled hole at the anatomical centre of the TT. A transverse trough 5mm deep was then cut using an oscillating saw and a rongeur which enabled the T-plate to be fitted into position at a 6° varus angle relative to the shaft of the femur. This was measured using a goniometer and allowed the T-plate to be aligned along the mechanical axis of the femur (Yoshioka et al., 1987). A second 5mm trough was then made 120mm distal and parallel to the first. These troughs located the flanges on the T-plate and enabled the T-plate to be securely fixed to the anterior tibia, minimising any risk of it getting displaced during testing (Figure 7.6).

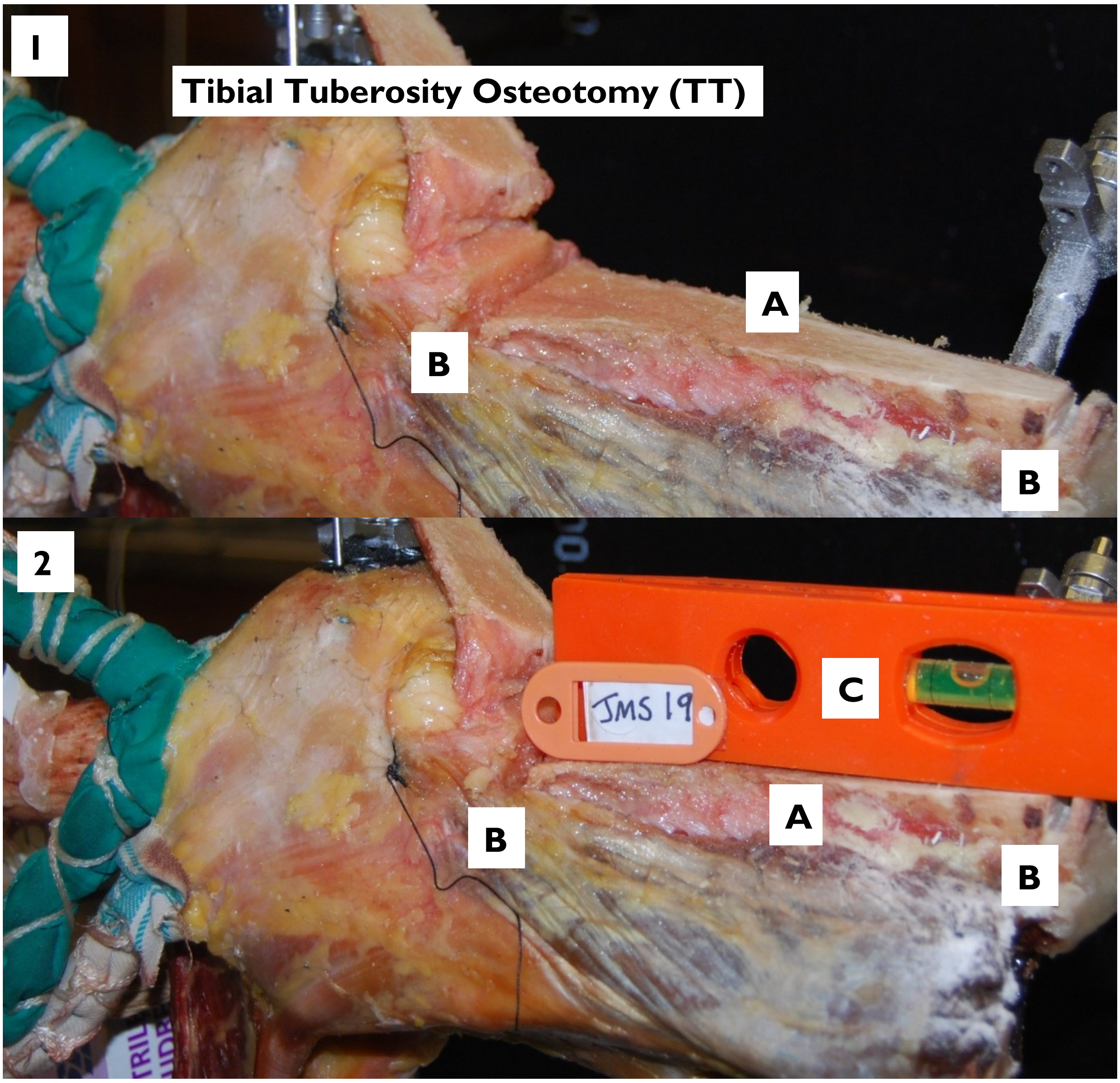

Figure 7.6: 1. Anterior tibia following osteotomy cuts and removal of bone block (A). Tibial osteotomy deflected from the anterior tibia as shown. Sunken transverse bone cuts, made at the proximal and distal tibia at a 6° valgus to the femur are shown (B). 2. The spirit level ensured standardisation of the bone cut position as discussed (C).

Following preparation, the T-plate was fixed to the anterior tibia. As described the central, proximal hole on the T-plate (green hole, Figure 7.2) was aligned with the previously drilled hole through the proximal anterior tibia, ensuring reproducible positioning of the plate over the anatomical tibial tuberosity site. The proximal and distal plate rims were inserted into the bone troughs. In addition four bicortical bone screws at the centre of the T-Plate (red holes, Figure 7.2) and three further 7mm screws (purple holes, Figure 7.2) were fixed into position down the body of the plate to ensure secure fixation of the plate throughout testing (Figure 7.7).

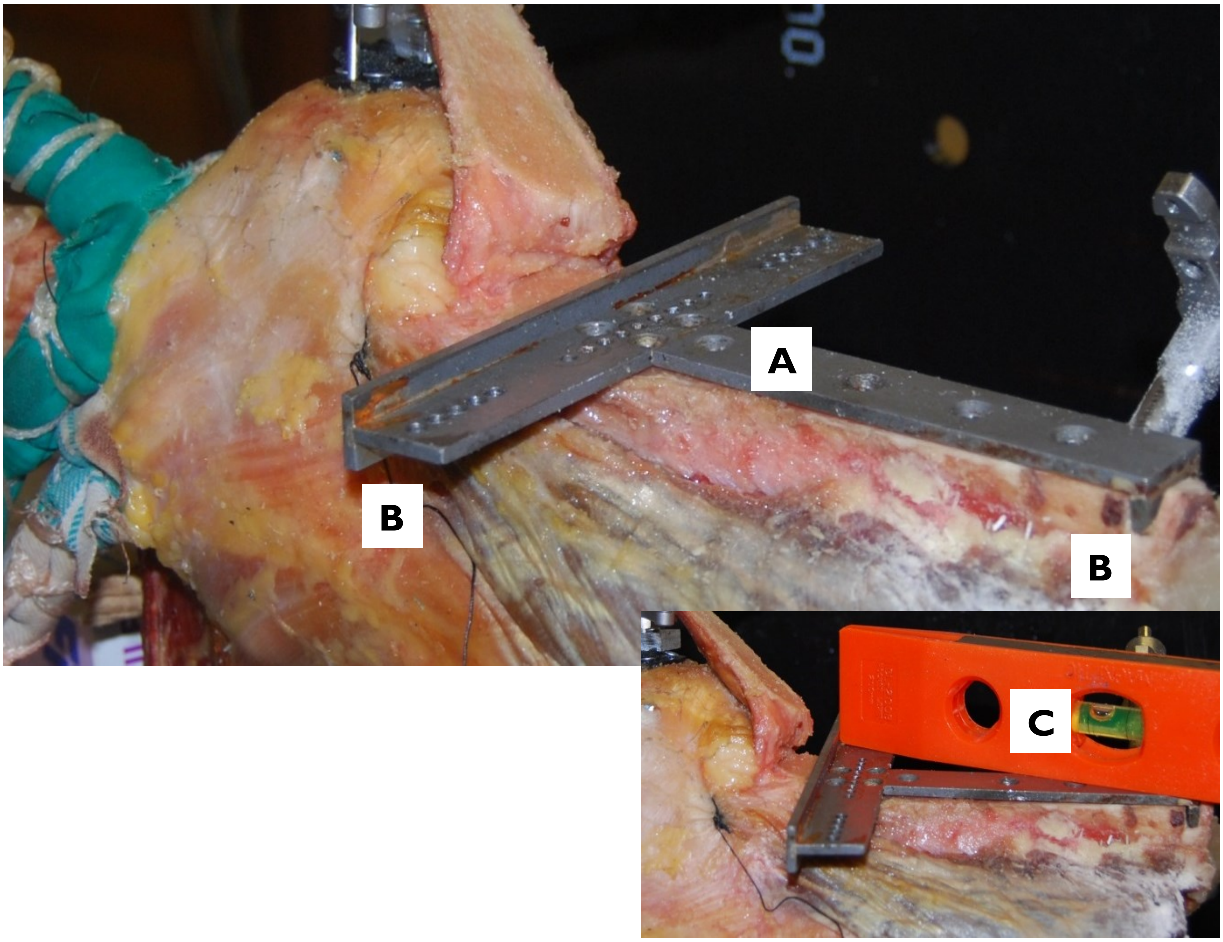

Figure 7.7 The T-plate is shown securely fixed to the anterior tibia, at a 6° valgus alignment to the line of the femur (A). The flanges at the end of the body of the T-plate are shown to securely fit into the pre-cut troughs at the proximal and distal ends of the tibia (B). The plate was secured into position with counter-sunk screws down the body of the plate and four bicortical screws shown in Figure 7.2. A spirit level confirmed the plate was parallel to the base of the test rig (C).

A rod was pushed through the 2mm hole drilled in the TT to locate its anatomical point. This was used to guide the TT progressively through 5mm increments medially and laterally, by positioning it in pre-drilled guide holes. To secure the osteotomy position, a plate was locked on top of the T-plate with one screw fixed 40mm medially and one 40mm laterally (Figure 7.8). A second rod was positioned through the previously drilled distal hole, 50 mm below the anatomical position of the TT, and through a hole on the T-Plate (yellow star, Figure 7.2). This provided a pivot to swing the osteotomy from to replicate the intraoperative procedure as described, and was maintained throughout all testing.

Figure 7.8 The locking plate in position, compressing the TT between it and the T-plate. Left image with the knee flexed to 90°. The guide rod for the anatomical position of the TT (green star) and the swinging point of the osteotomy (yellow star) are both highlighted.

7.6 Experimental Protocol

Measurements were taken on the intact knee with no plate in-situ, and then the knee with the plate in-situ and the TT in its 7 different positions. It was initially attempted to medialise and lateralise the TT by 20mm, as can be seen by the number of holes pre drilled along the T bar of the T-plate. However this was impossible due to soft tissue restraints and is discussed further in section 7.10.2. Therefore testing investigated the TT in its anatomical, 5mm, 10mm and 15 mm medialised and 5mm, 10mm and 15 mm lateralised positions. The order of testing for each specimen was randomised to avoid bias. Measurements were taken both with and without an external 10N displacing force applied. At the conclusion of testing, a re-test measurement was recorded with the TT back in its anatomical position to ensure there was no adverse effect on patellofemoral joint mechanics or kinematics as a consequence of the extreme TT medialisation or lateralisation.

7.7 Analysis

Custom written MATLAB scripts calculated mean and peak contact pressures and patellar motion from the raw data recorded by the Iscan and Toolviewer software (Appendix A+B). The co-ordinate system of each bone was again defined so that patellar lateral translation and tilt were taken to be positive. A power calculation determined a sample size of 8 necessary to determine a significant change with 80% power and 95% confidence (Kuroda et al., 2001). Dependent variables were: patellar translation and tilt, and medial and lateral mean and peak articular contact pressures. Data was analysed in SPSS, a Shapiro-Wilk test confirmed that the data was normally distributed.

A number of different analyses were performed on the data sets from each of the six different dependent variables measured as follows;

1. A two-way RM ANOVA was performed comparing the intact knee with no T-plate to the knee with the T-plate attached to the tibia and the TT in its anatomical position, across all flexion angles (0°, 10°, 20°, 30°, 60° and 90°).
2. A two-way RM ANOVA was performed on all dependent variables comparing the knee with the T-plate in position and the TT in its anatomical position with readings taken before and

following the progressive medialisation and lateralisation from 5mm-15mm at each angle of flexion (0°, 10°, 20°, 30°, 60° and 90°).

3. TT position (15mm, 10mm and 5mm medial and lateral and anatomical position) was compared with flexion angle (0˚, 10˚, 20˚, 30˚, 60˚ and 90˚) using a two-way RM ANOVA. Where TT position significantly influenced output, post-hoc paired t-tests with Bonferroni correction were applied comparing the anatomically positioned TT with each of the medialised and lateralised TT positions at individual flexion angles.

4. Pearson correlation analysis was performed on dependent variables at individual flexion angles where post-hoc tests identified a significant effect of TT position.

The final two analyses conducted both investigated the effect of the 10N lateral displacing force on the knee with the TT in different positions. These measurements are summarised in Table 7.1 below. Both were examined using a two way RM ANOVA comparing TT position and flexion angle. Post-hoc t-tests with Bonferroni correction were applied where significant differences were detected.

Table 7.1 Summary of Analysis Undertaken.

Section	Comparison	Measurement
7.7.5	**The combined effect of TT position and lateral displacing force**	Measurements taken at each flexion angle with the TT in the seven different positions with the 10N lateral displacing load applied (n condition +10N lateral load) were subtracted from the knee with the TT in the anatomical position with no 10N lateral load applied (i.e. (n condition+10N lateral load applied) – (plate in-situ with TT in its anatomical position + no external load applied)) e.g. (TT 5mm lateralised + 10N lateral load) – (plate in-situ with TT in its anatomical position and no external load applied)).
7.7.6	**The lone effect of the 10N lateral load application**	Measurements taken at each flexion angle at the anatomical and 3 lateralised TT positions (n condition) were subtracted from the corresponding measurement taken with 10N lateral load applied to the patella i.e. ((n condition+10N lateral load) – (n condition)) e.g. ((TT 5mm lateralised + 10N lateral load) – (TT 5mm lateralised)).

7.8 Results

7.8.1 The effect of plate insertion

No significant difference was found to be caused by the attachment of the T-plate to the anterior tibia for any of the variables examined: patellar translation (*P*=0.273) (Figure 7.9), patellar tilt (*P*=0.790) (Figure 7.10), mean medial (*P*=0.882) or lateral (*P*=0.624) and peak medial (*P*=0.355) or lateral (*P*=0.107) articular patellofemoral joint contact pressures.

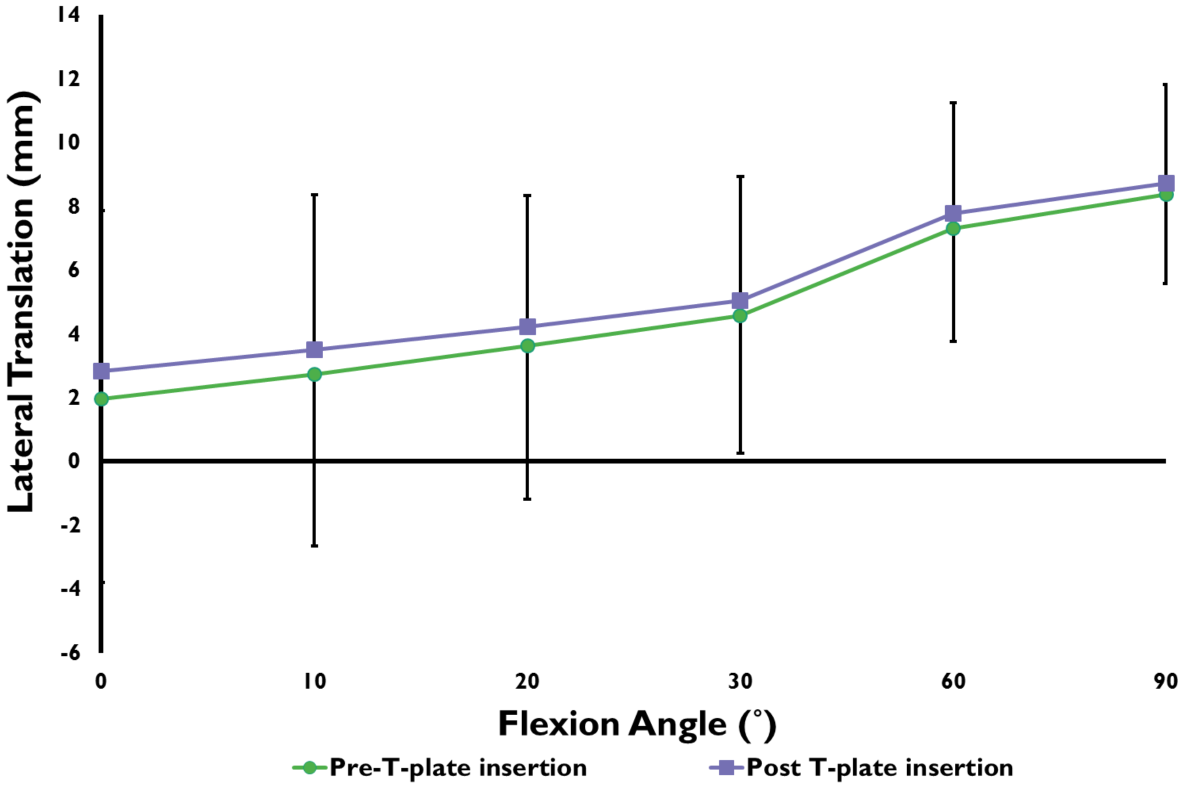

Figure 7.9 Patellar medial-lateral translation (mm; mean ± SD, n = 8) from 0˚ to 90˚ knee flexion shown before and following insertion of the T-plate to the anterior tibia.

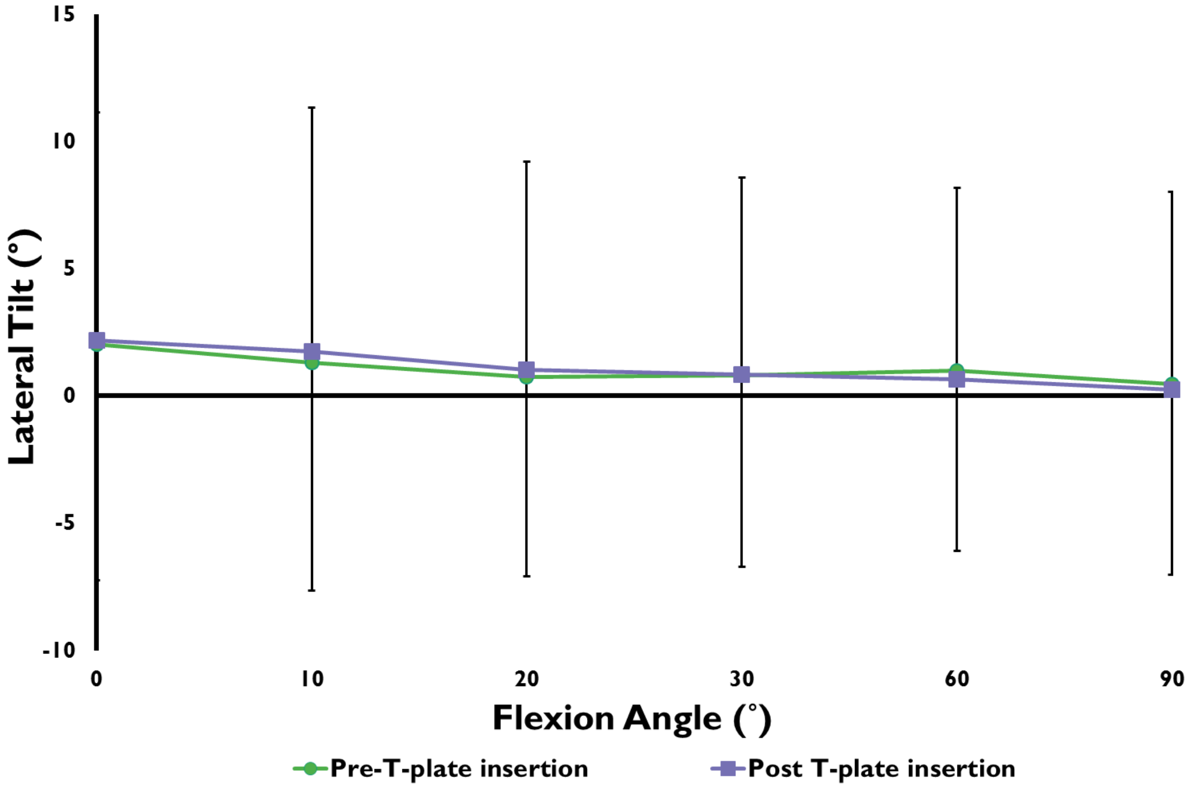

Figure 7.10 Patellar lateral tilt (°; mean ± SD, n = 8) from 0° to 90° knee flexion shown before and following insertion of the T-plate to the anterior tibia.

7.8.2 The effect of repeated testing

There was no significant effect found on any of the dependent variables when re-testing the TT in its anatomical position at the start of testing and following the progressive 15mm medialisation and lateralisation of the TT: patellar translation (P=0.231), patellar tilt (P=0.855), mean medial (P=0.322) and lateral (P=0.555) or peak medial (P=0.827) and lateral (P=0.139) articular patellofemoral joint contact pressures.

7.8.3 The effect of progressive TT medialisation / lateralisation

With the TT lateralised 15mm, lateral patellar translation and tilt increased by up to 4.5mm and 5.3°, whilst 15mm TT medialisation reduced lateral patellar translation and tilt by up to 5mm and 0.9° respectively. Tuberosity position had a significant effect on patellar translation (*P*=0.015) (Figure 7.11) and this was significantly influenced by knee flexion (*P*=0.011), with the patella moving more laterally in deeper flexion. Altering the TT position was also found to have a significant effect on patellar tilt (*P*=0.048) (Figure 7.12). A significant interaction between flexion angle and tuberosity position was identified for patellar tilt (*P*=0.042), with tuberosity transfer having a greater effect on patellar tilt in early flexion, before the patella engaged with the trochlear groove. Post hoc tests found 15mm lateralisation to have a significant effect at 0° on patellar tilt (*P*<0.05). Post hoc testing did not identify any significant effect on patellar translation at individual flexion angles.

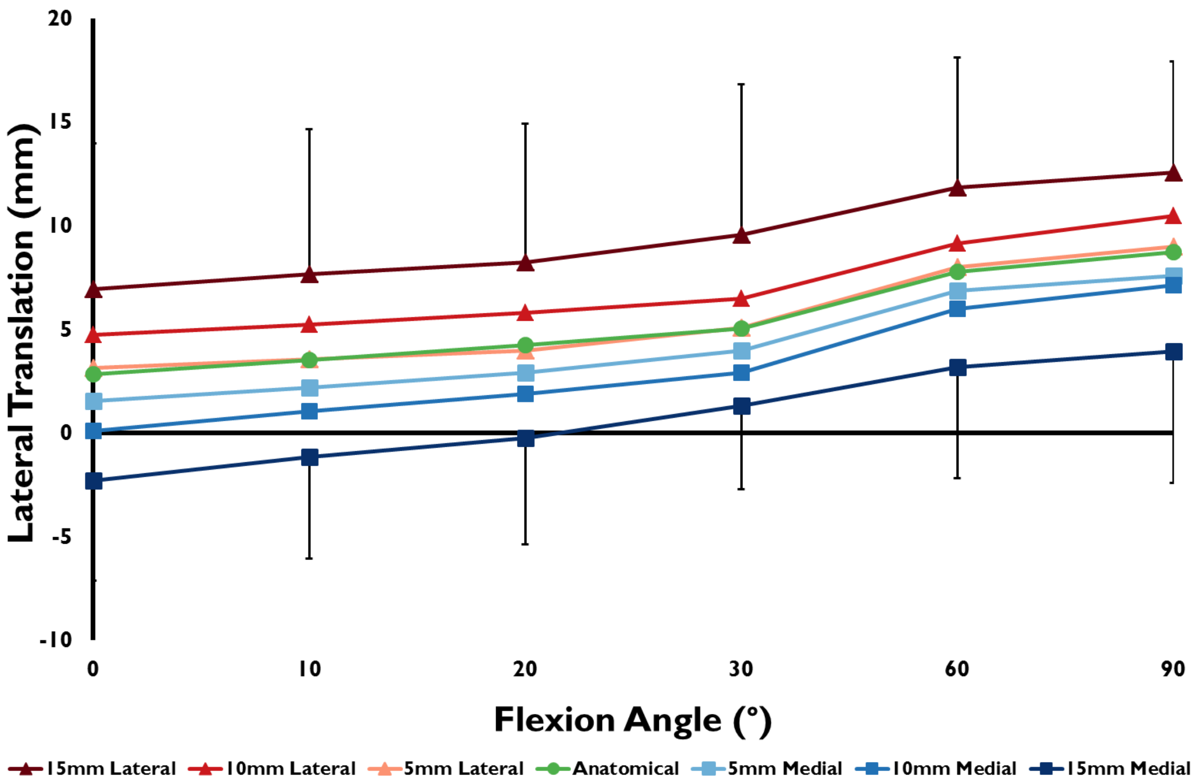

Figure 7.11 Patellar medial-lateral translation (mm; mean ± SD, n = 8) from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised.

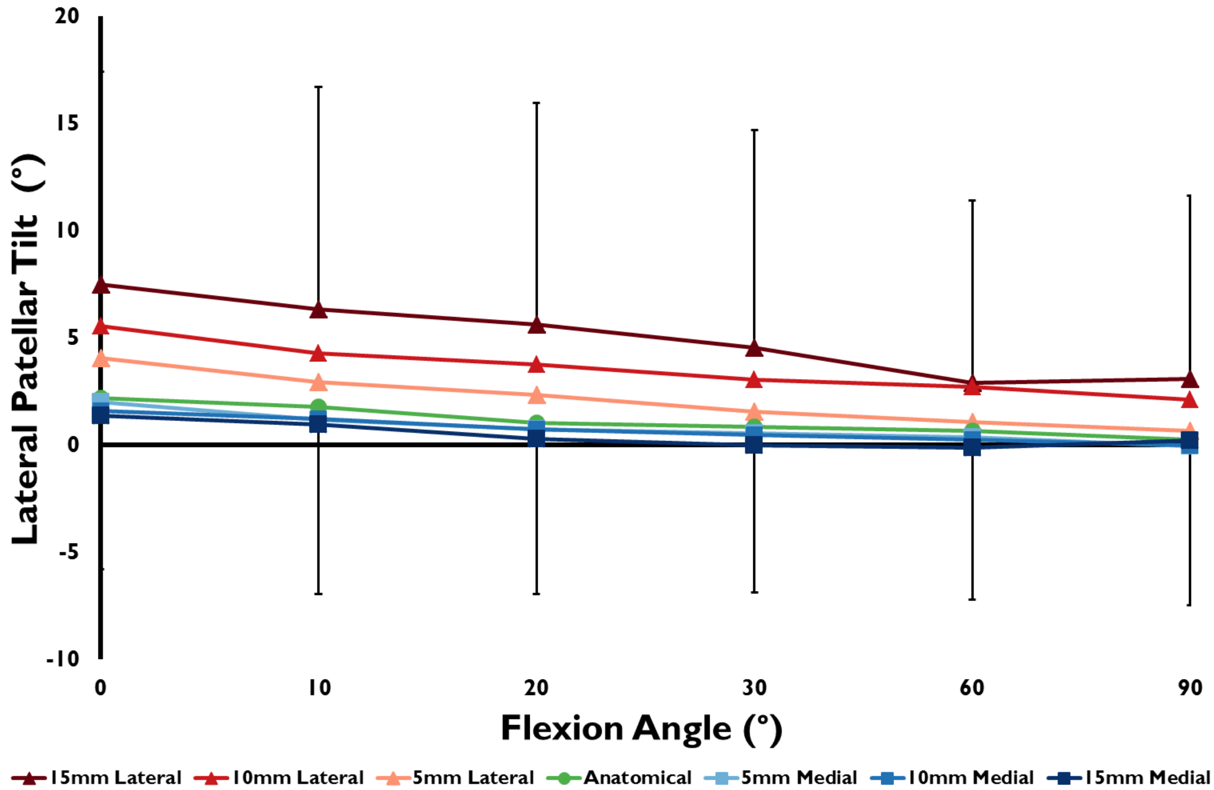

Figure 7.12 Patellar lateral tilt (˚; mean ± SD, n = 8) from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised.

Mean lateral pressure increased by 0.3MPa at 10° knee flexion following 15mm tuberosity lateralisation, and mean medial contact pressure increased by 0.1MPa at 20° following 15mm medialisation. Changing tuberosity position resulted in significant changes in mean lateral (P=0.001) and medial (P=0.015) patellofemoral contact pressure. Flexion angle was not found to have a significant effect on either measure: mean lateral (P=0.128) or medial (P=0.071) pressure. Significant post-hoc findings for mean pressure are highlighted in Figure 7.13.

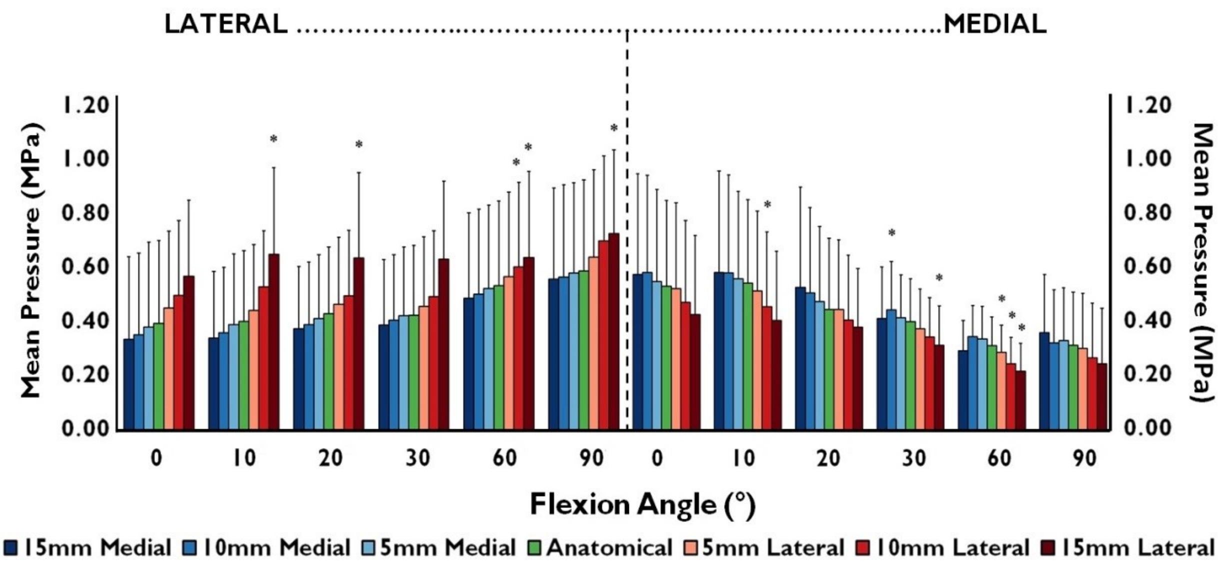

Figure 7.13 Mean lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean + SD, n=8) from 0˚-90˚ flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **P*<0.05 (compared to anatomical pressure).

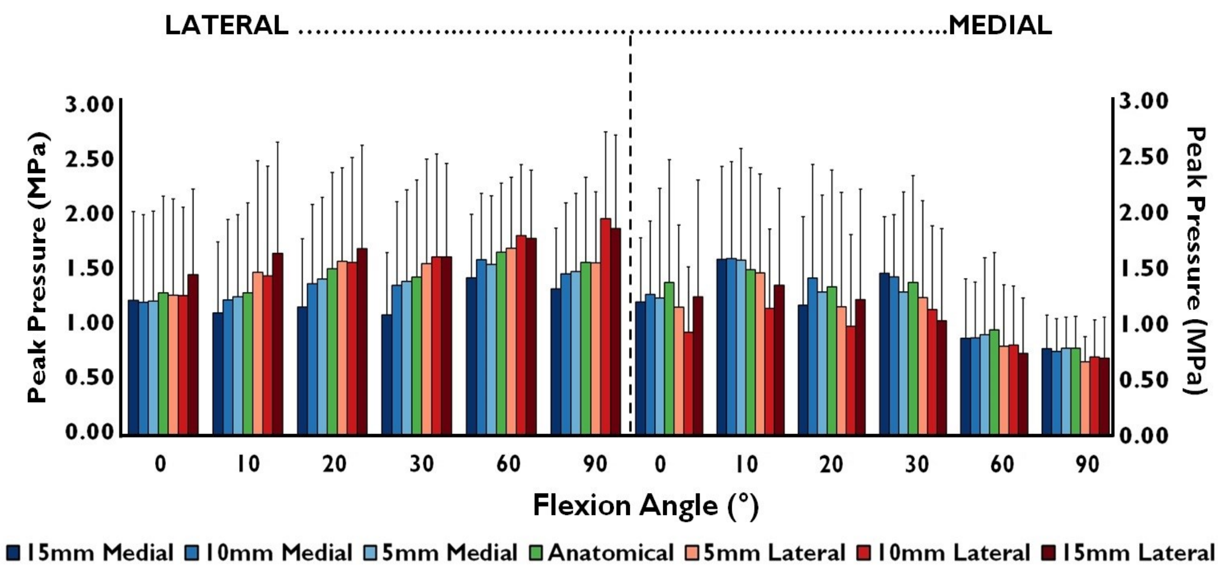

Figure 7.14 Peak lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean + SD, n=8) from 0˚-90˚ flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised.

Lateralisation of the TT by 15mm caused peak lateral pressures to rise by 0.4MPa at 10°. Altering TT position had a significant effect on peak lateral contact pressures (P=0.008), independent of flexion angle (P=0.349) (Figure 7.14), but TT position was not found to have a significant effect at any individual flexion angles during post-hoc testing following Bonferroni correction. TT position was not found to have any effect on peak medial patellofemoral joint pressures (P=0.108) (Figure 7.14).

A general trend of elevated lateral patellofemoral joint contact pressures as the TT was progressively lateralised was identified. However when the TT was medialised there was not the same sharp rise in contact pressures evident in the medial compartment. This is highlighted in Figure 7.15, where the pressures when the TT is medialised can be seen to be more closely related to the anatomical pressures, than the lateral facet pressure spikes evident when the TT was lateralised (peak pressure and mean pressure bars).

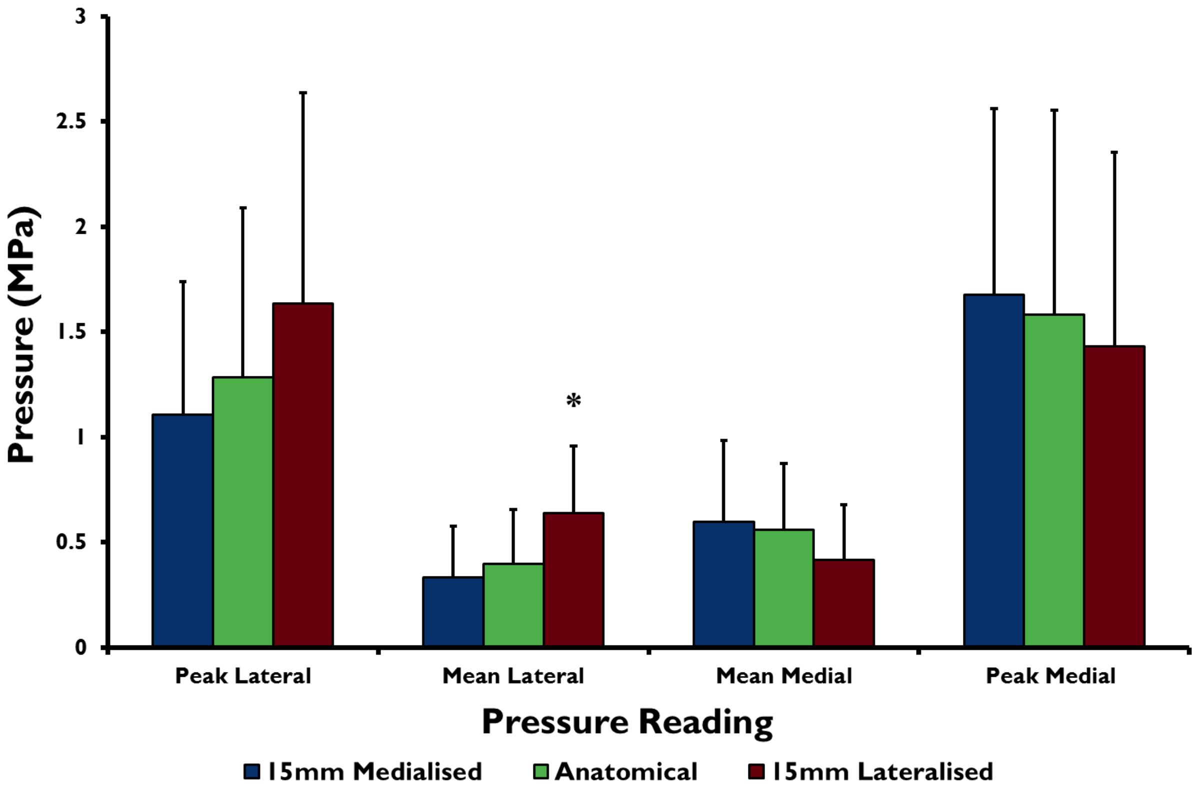

Figure 7.15 Patellofemoral joint contact pressures: peak lateral, mean lateral, mean medial and peak medial (MPa; mean + SD, n=8) measured at 10° knee flexion with the TT positioned: 15mm medialised, anatomical and 15mm lateralised. *P<0.05

7.8.4 Relationship between TT position and pressure change

A significant correlation was identified between patellar tilt and TT position at 0° (r=0.610, P<0.001): when the TT was lateralised, the patella tilted laterally. Significant correlations were also identified between TT position and mean lateral contact pressures at 10° (r=0.810, P<0.001), 20° (r=0.651, P<0.001), 60° (r=0.826, P<0.001) and 90° (r=0.792, P<0.001) and mean medial patellofemoral joint pressures at 10° (r=-0.717, P<0.001), 30° (r=-0.687, P<0.001) and 60° (r=-0.661, P<0.001). A linear regression established that TT position could significantly predict mean lateral pressure through knee flexion (0°-90°) (R^2=0.350-0.676, *P*<0.0005) (Figure 7.16).

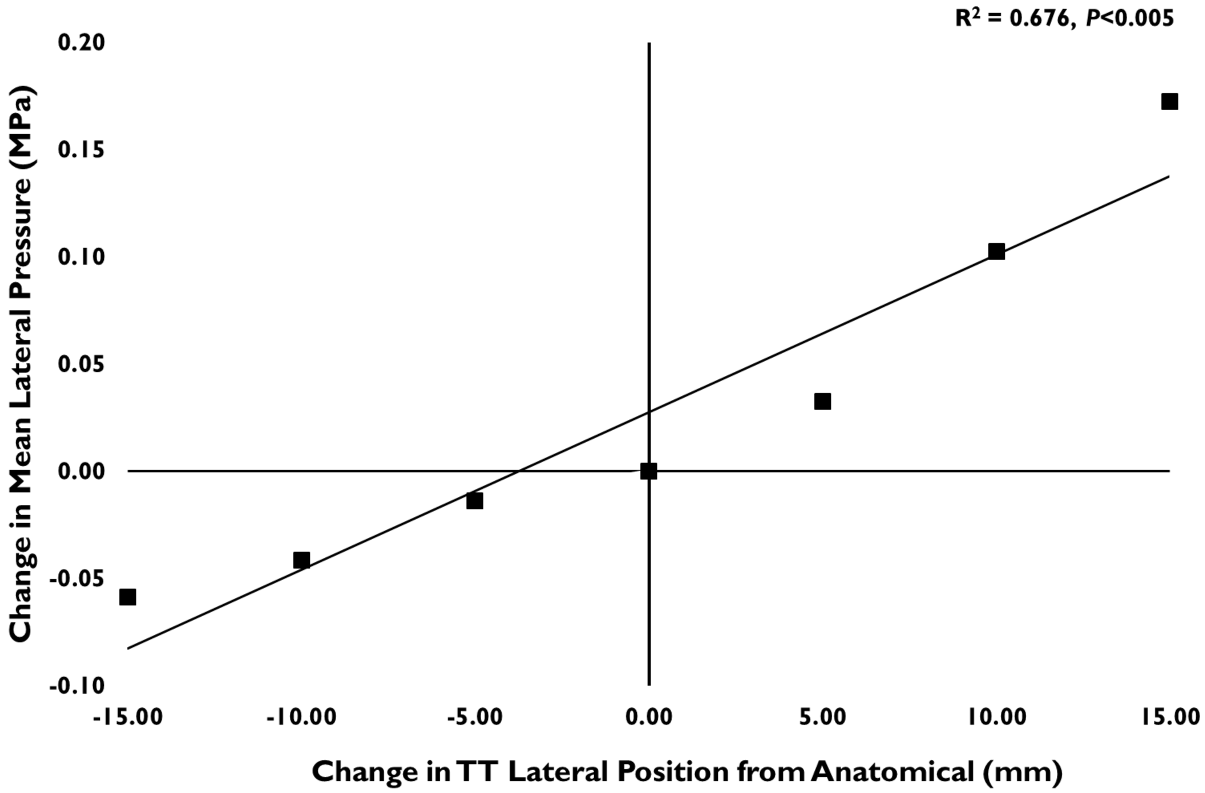

Figure 7.16 Change in mean lateral pressure (MPa) at 60° knee flexion plotted versus TT position (mm), with a regression value shown.

7.8.5 The combined effect of TT position and the 10N lateral displacing force on patellar stability

The effect of the 10N laterally directed force increased significantly as the TT was lateralised further from its anatomical position. With the TT in its anatomical position the patella was on average 2.8mm lateralised in full extension. The 10N lateral force increased lateral patellar translation by a further 0.3mm to 3.1mm lateral. Positioning the TT 15mm lateral resulted in the patella lateralising a further 4.1 mm, which was increased to 9mm lateral following the application of a 10N lateral load. Therefore the combined effect of lateralising the TT by 15mm and the application of the 10N lateral load equalled an extra 6.2mm subluxation (9mm – 2.8mm).

The combined effect of TT position and the application of 10N lateral load had a significant effect on patellar translation (*P*=0.014) (Figure 7.17). This effect was not found to be dependent on flexion angle, instead the effect was constant throughout knee flexion range (*P*=0.735). Post hoc testing identified differences between the anatomical TT and that positioned 15mm medial at 0° and 10° knee flexion, and the TT positioned 15mm lateral and the anatomical TT at 20°, 30° and 60° flexion.

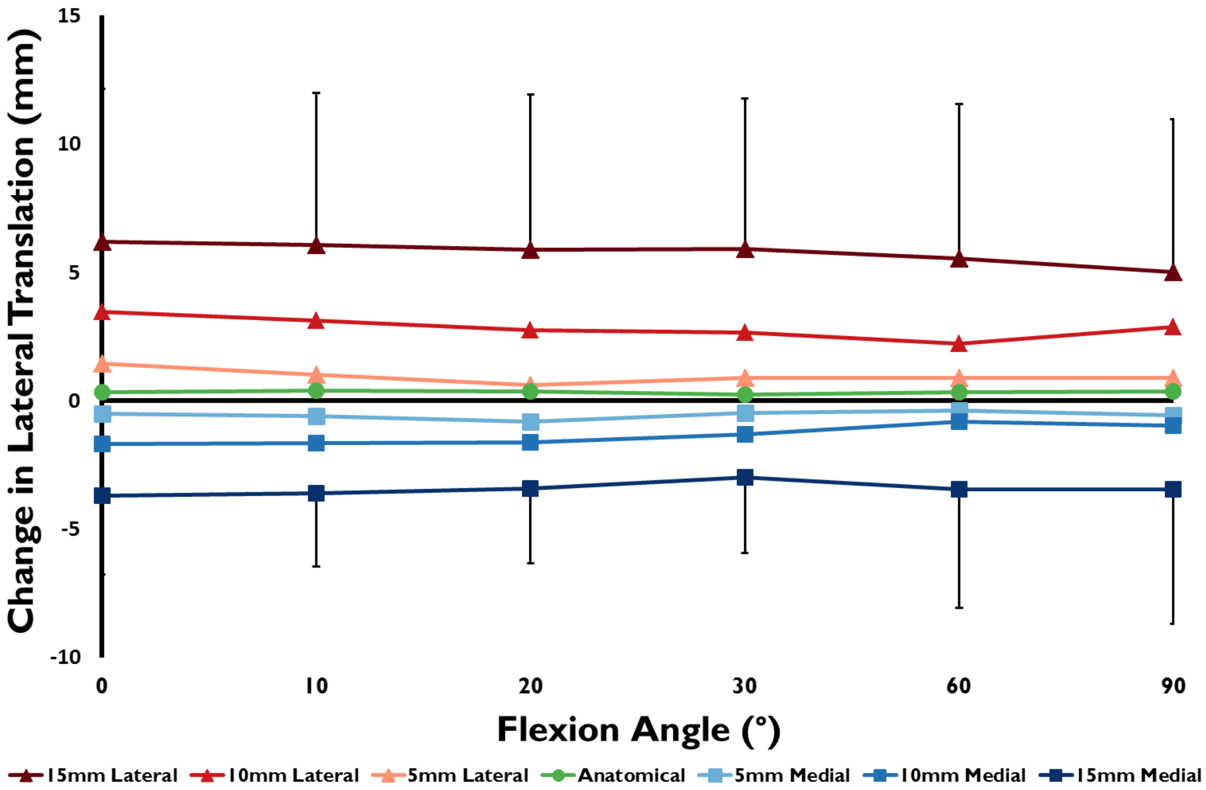

Figure 7.17 Change in patellar medial-lateral translation (mm; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0° to 90° knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised.

The combined effect of TT position and 10N lateral displacing force also had a significant effect on patellar tilt (*P*=0.024) (Figure 7.18). Flexion angle had a significant effect on the change in lateral tilt, with the effect of TT position and 10N lateral load application generally reducing in deeper flexion when the patella had engaged with the trochlea (*P*=0.006). Post hoc testing identified significant differences between the anatomically positioned TT and that positioned 15mm lateral at 0° and 10° knee flexion (both *P*<0.05).

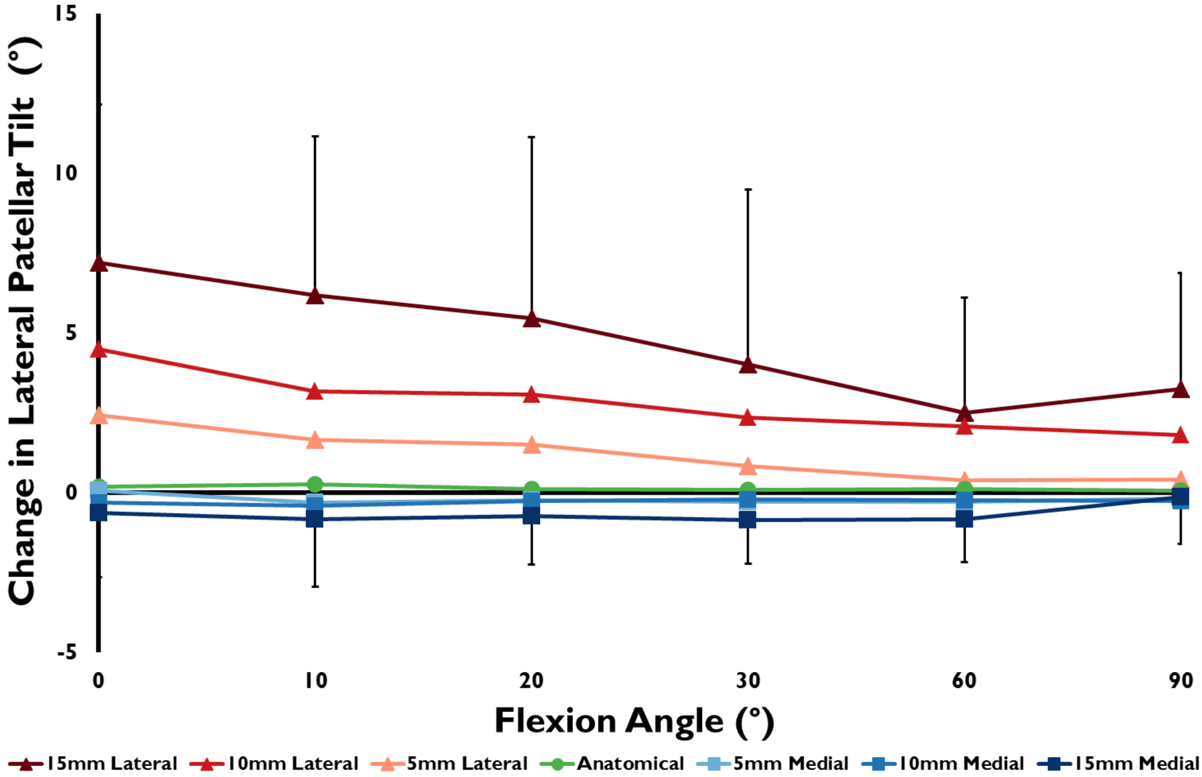

Figure 7.18 Change in patellar lateral tilt (˚; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT-TG positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised.

The combination of altered TT position and 10N lateral load also had a significant effect on mean medial (*P*=0.008) and mean lateral (*P*=0.002) patellofemoral joint contact pressures (Figure 7.19). These effects were reasonably constant throughout knee flexion and not found to be significantly influenced by knee flexion (mean medial: *P*=0.153 and mean lateral: *P*=0.297). This is shown in Figure 7.19, with significant post hoc tests highlighted.

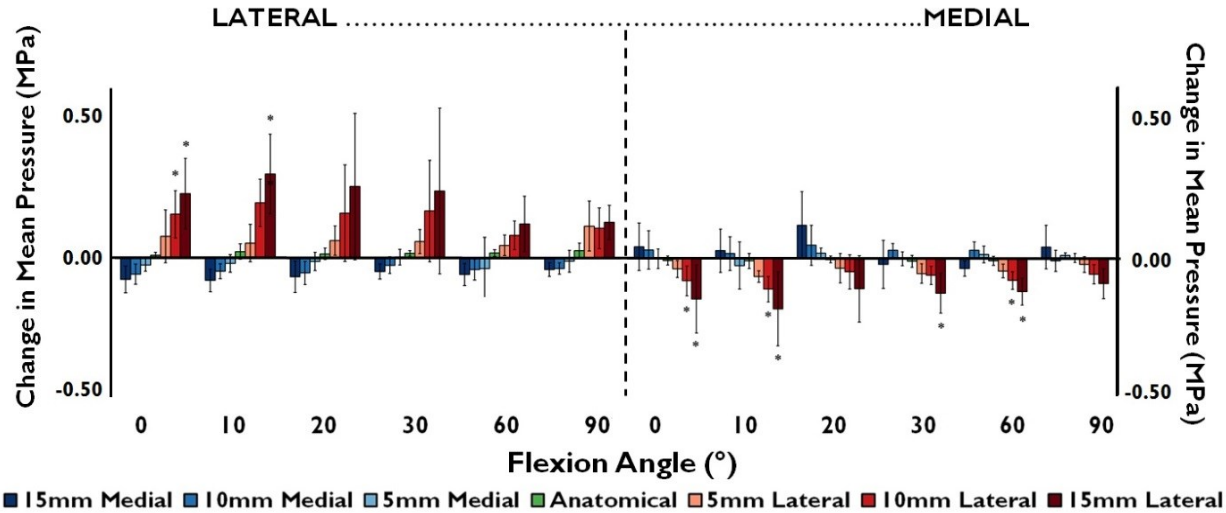

Figure 7.19 Change in mean lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean ± SD, n=8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. *P<0.05.

A significant effect on peak lateral patellofemoral contact pressures was also identified (P=0.001), with peak pressures seen to progressively rise as the TT was lateralised. This effect was not found to be dependent of flexion angle (P=0.485). There was no significant effect of the 10N lateral load application and TT position identified on peak medial joint contact pressures (P=0.096). Post hoc tests did not identify significant changes (Figure 7.20).

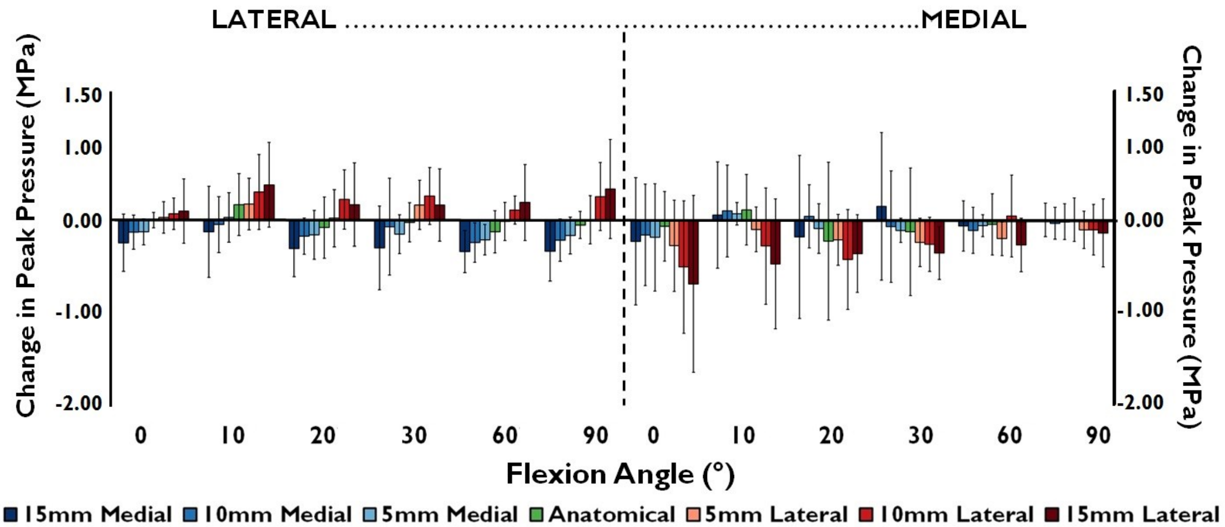

Figure 7.20 Change in peak lateral (left side of the graph) and medial (right side of the graph) patellofemoral joint contact pressures (MPa; mean ± SD, n=8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised.

7.8.6 The lone effect of the 10N lateral load applied to the TT in its anatomic and lateralised positions

When the TT was positioned 15mm laterally, the patella was positioned 6.9mm laterally. Following the application of a 10N lateral displacing force, this was found to increase to 9mm laterally, indicating an increase of 2.1mm (9mm - 6.9mm). In comparison, in the anatomically positioned TT, the patella began 2.8mm lateral, increasing to 3.1mm following the application of a 10N lateral displacing load. This resulted in only a 0.3mm change (3.1mm-2.8mm). Therefore the further lateral the TT was positioned, the greater the tendency for its displacement to increase following the application of a 10N lateral load (Figure 7.21). The change imposed by the 10N lateral load was found to have a significant effect on patellar translation (P=0.006), with the effect generally greater in early knee flexion before the patella engaged with the trochlear groove (P=0.042). Post hoc testing identified a significant effect of the 10N lateral load application at 0° and 10° when the TT was 10mm and 15mm lateralised compared to the TT in its anatomical position (P<0.05).

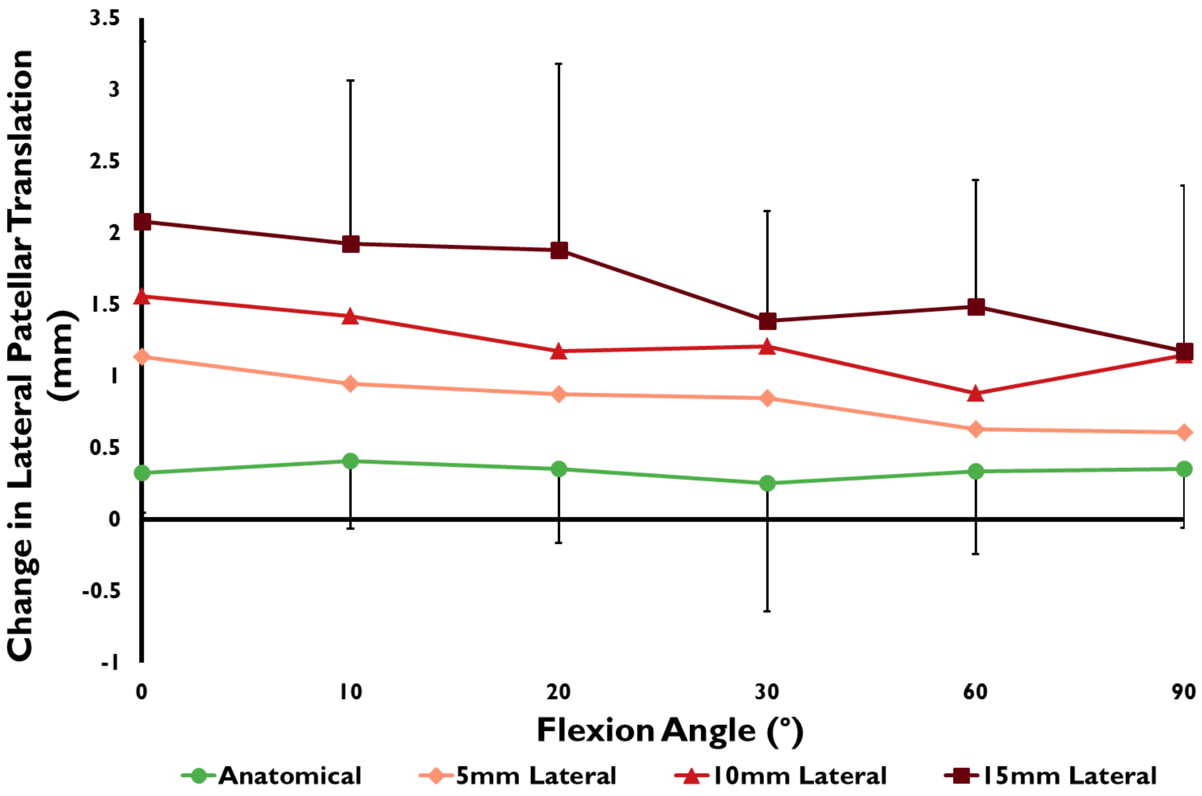

Figure 7.21 Change in patellar medial-lateral translation (mm; mean ± SD, n = 8) caused by the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned anatomically and 5mm, 10mm and 15mm lateralised.

Patellar tilt was also significantly affected by the 10N load application ($P<0.001$), a factor which was found to have a greater effect in early flexion ($P<0.001$) (Figure 7.21). Post hoc testing identified a significant destabilising effect of the 10N load at 0° and 20° when the TT was 15mm lateralised compared to in its anatomical position ($P<0.05$).

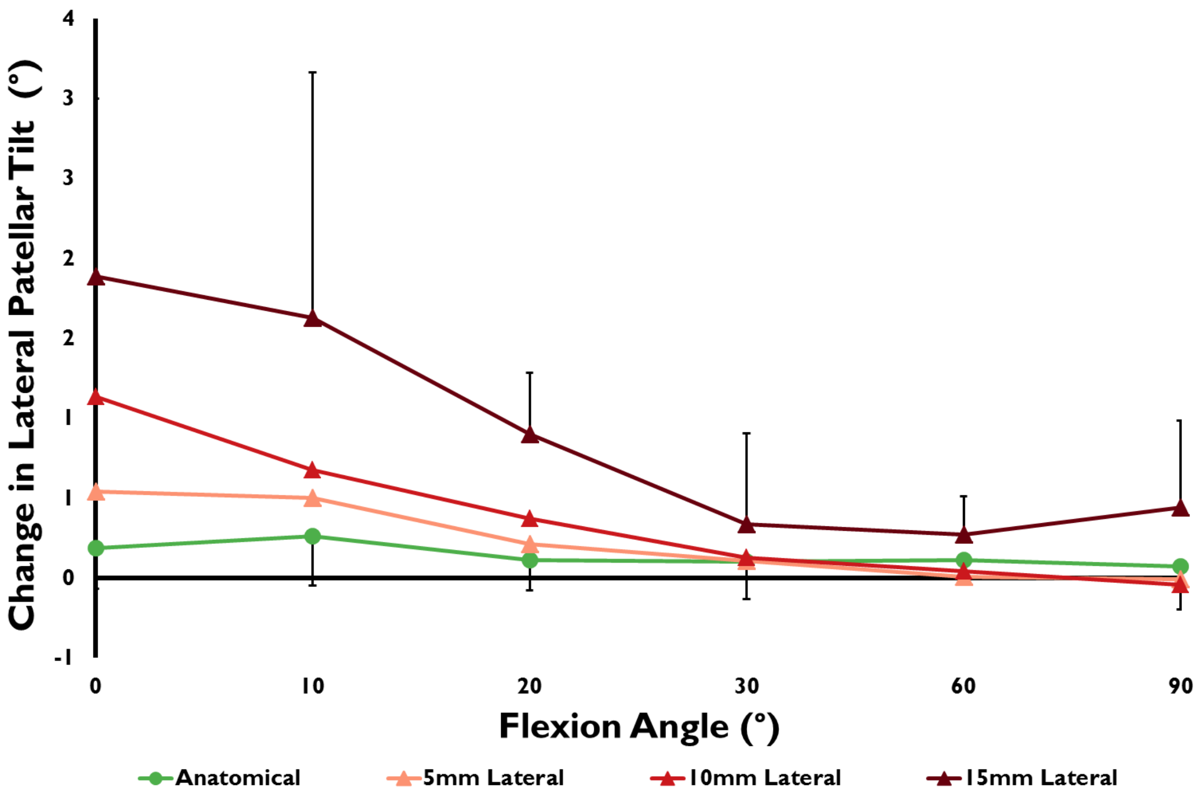

Figure 7.22 Change in patellar lateral tilt (°; mean ± SD, n = 8) caused by the application of 10N lateral displacing force to the patella from 0° to 90° knee flexion with the TT positioned anatomically and 5mm, 10mm and 15mm lateralised.

The 10N lateral load application to the patella was not found to have a significant effect on any of the patellofemoral joint articular contact pressures measured: mean medial ($P=0.132$) and lateral ($P=0.123$) or peak medial ($P=0.362$) and lateral ($P=0.398$) contact pressures. The pressure results all lacked a clear trend in response to the 10N lateral load application and are typified in Figure 7.23 and Figure 7.24 which shows the effect of 10N lateral load application on mean lateral and peak medial patellofemoral joint pressures.

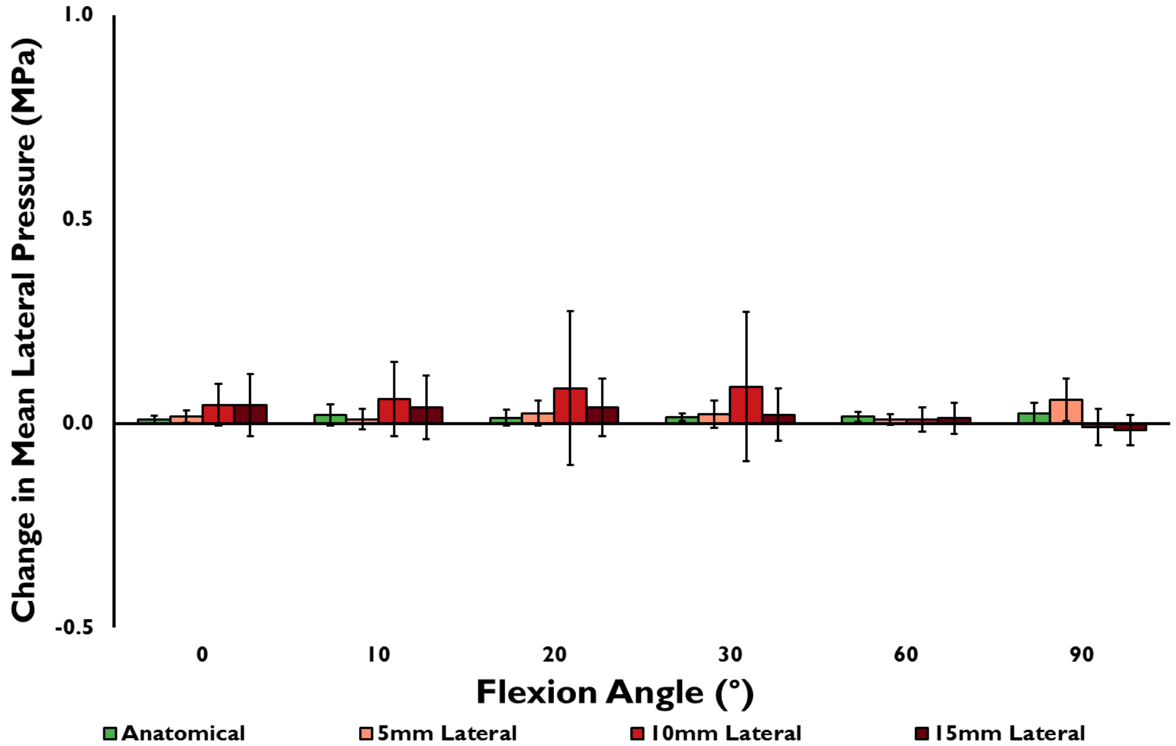

Figure 7.23 Change in mean lateral patellofemoral joint contact pressures (MPa; mean ± SD, n=8) caused by the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned anatomically and 5mm, 10mm and 15mm lateralised.

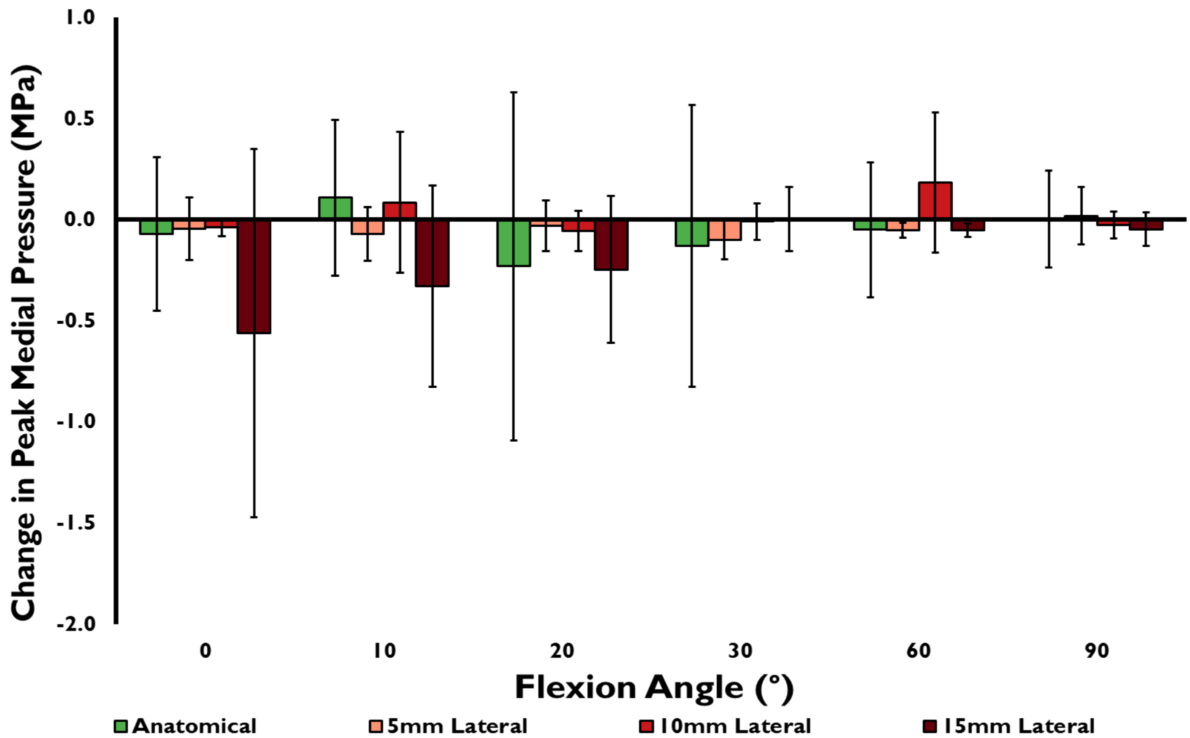

Figure 7.24 Change in peak medial patellofemoral joint contact pressures (MPa; mean ± SD, n=8) caused by the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned anatomically and 5mm, 10mm and 15mm lateralised.

7.9 Discussion

The current study identified that progressive lateralisation of the TT caused increased lateral patellar tracking, elevated lateral patellofemoral facet contact pressures and reduced patellar stability. These findings correlate well with clinical reports of increased incidence of patellar dislocation and pain in patient populations with elevated TT-TG distances (Dejour et al., 1994; Pennock et al., 2013). A significant correlation between mean lateral contact pressure and TT position was identified. This has the potential to be used in the pre-surgical planning of tibial tubercle transfer operations. Importantly, in comparison to the lateralisation effects, medialisation of the TT did not cause corresponding excessive increases in medial contact pressures or patellar motion. Rather, medial pressures tended to reach a plateau during progressive medialisation, unlike lateral pressures during lateralisation which continued to rise. The greater effect of TT lateralisation compared to medialisation is hypothesised to occur as a consequence of the high stiffness of the ITB and lateral retinaculum as opposed to the medial retinaculum and MPFL. The present findings provide a rationale for tibial tuberosity transfer procedures in appropriate populations of patients.

This study had limitations inherent to *in vitro* testing which are discussed further in Chapter 9. In addition, a number of factors which may have impacted on the results require to be highlighted. Firstly alteration of the TT position was performed along a linear plate in order to standardise testing. This prevented any alteration to A-P position; however this may not be representative of what occurs intra-operatively. Secondly the lateral load applied to the patella was relatively small in comparison to that which would be expected to occur *in vivo*. This may have underestimated the effect of TT positioning on stability. However the load was decided upon following extensive pilot testing and review of the literature, as discussed in Chapter 6. Furthermore, although the load applied was low, it was comparable to the muscle and ITB loads applied in the experiment and therefore avoided inflicting any damage to the local soft tissues.

Progressive TT lateralisation resulted in corresponding rises in lateral patellar tilt and translation and elevated lateral joint contact pressures. Contact pressure findings and lateral patellar tracking are consistent with prior cadaveric work in this area (Koëter et al., 2007; Ramappa et al., 2006). Findings highlight that patients with elevated TT-TG distances can suffer increased lateral patellofemoral joint pressures as a result. Clinically this is concerning since osteoarthritis is reported to most commonly affect the lateral compartment of the patellofemoral joint (Casscells and Ward, 1978; Iwano et al., 1990). Referring to Figures 7.11-7.14 it is evident that 5 mm lateralisation had only a small effect on patellar kinematics and contact mechanics. However, adverse effects are seen to significantly rise when the TT is lateralised 10mm or 15mm. Given the mean TT-TG distance in this study was

10.4mm, five millimetre increases in TT-TG would equal 15.4mm suggesting that TT-TG surgery would mainly be indicated in patients with TT-TG distances greater than this. This finding is supported by clinical reports of pathological TT-TG distances (Balcarek et al., 2011; Dejour et al., 1994) and provides useful information to aid in clinical decision making.

Interestingly a moderate relationship was identified between TT position and mean lateral contact pressures (Figure 7.16). This demonstrated a consistent trend for mean lateral patellofemoral joint contact pressures to progressively elevate as the TT was lateralised and reduce as the TT was medialised. This is clinically relevant given the previously discussed prevalence of degenerative changes in the lateral patellofemoral joint compartment and the recognised lateral patellar facet syndrome often present in patients with patellofemoral joint pathology (Iwano et al., 1990; Johnson, 1989). Traditionally these problems have been treated with lateral release (Kolowich et al., 1990), however recent evidence has suggested the adverse effects of such techniques on patellar stability (Merican et al., 2009). The present study findings provide support for tibial transfer surgery as an option in these cases to provide alleviation of the pressures in the lateral patellofemoral joint compartment.

Tibial tuberosity medialisation presents the challenge to obliterate instability and reduce lateral patellofemoral joint contact pressures whilst avoiding over correction, which may cause increased medial patellar tracking or joint pressures. Prior cadaveric and computer modelling work has found that TT-TG reduction can cause significantly elevated medial patellofemoral joint contact pressures (Benvenuti et al., 1997; Koëter et al., 2007; Kuroda et al., 2001). Indeed Huberti and Hayes (1984) found patellofemoral joint pressures from reducing the Q-angle were greater than those resulting from increasing the Q-angle. A further study investigated 10mm TT medialisation, finding it to have no significant effect on patellar tracking, but a negative influence on peak joint contact pressures (Ramappa et al., 2006). This is likely a consequence of the dissection of the medial and lateral retinacula undertaken by the authors to enable them to insert the pressure sensor into the patellofemoral joint. In contrast, medialising the TT in the current study did not have a significant effect on peak patellar pressures and although mean medial contact pressures were elevated, the effect was very small (approximately: 0.1 MPa). Medial patellar translation was seen to rise, however this did not correspond with large changes in the contact pressures measured, and patellar tilt was largely unaffected by medialisation. Furthermore the tendency following TT medialisation was for the medial contact pressures to reach a plateau following an initial increase after the 5mm transfer (Figure 7.13 and Figure 7.14).

The study highlights a contrast in results between progressive TT medialisation with that of lateralisation which, resulted in a trend for continuing elevated lateral joint contact pressures as the TT was further lateralised. The difference in findings between the present study and prior work probably relates to methodological variations, with the current study loading the ITB, whereas prior work had not (Benvenuti et al., 1997; Kuroda et al., 2001). The position and orientation of the thick distal transverse ITB fibres support its role in providing strong medial restraint to the patella (Merican and Amis, 2008), whereas the MPFL is known to be a thinner fascial band (Baldwin, 2009). These findings provide a rationale for TT medialisation in patients with patellar instability and elevated TT-TG distances. However in the context of sub-optimal clinical outcomes and the common reporting of medial facet osteochondral defects reported in patellar dislocation populations (Nomura and Inoue, 2005; Nomura et al., 2003); pre-operative planning and confirmation of patellar tracking intra-operatively with or without nerve stimulation (Pritsch et al., 2007) appear advisable to ensure successful surgical outcome.

This study is thought to be one of the first to directly examine patellar stability in knees with increased TT lateralisation. The results highlight that as the TT was moved to a more lateralised position, the effect of the 10N lateral load applied to the patella significantly increased lateral patellar tilt and translation. This highlights the destabilising effect of TT lateralisation on the patella and correlates with the increased incidence of patellar dislocation in populations of patients with high TT-TG distances (Dejour et al., 1994). The increase in instability caused by lateralising the TT was significantly reduced following TT medialisation. These findings provide a rationale for TT medialisation in the treatment of patients with elevated TT-TG distances and the clinical presentation of patellar instability. Contact pressure did not significantly change as a consequence of the 10 N lateral load. This may in part be due to the failure of the 10N lateral displacing force to simulate *in vivo* loads as previously discussed. In addition it could owe to the lack of sensitivity of the Tekscan pressure sensors identified during pilot testing in Chapter 4. However, as discussed, this was the best available instrument for the examination of patellofemoral joint contact pressures.

Transfer of the TT is a widely performed procedure on patellofemoral patients, however as highlighted post-operative outcomes reported have been varied and inconsistent, with some implying it can cause accelerated osteoarthritis (Aglietti et al., 1994; Arnbjornsson et al., 1992). The present study provides a rationale for surgical intervention in the form of tibial tubercle transfer surgery in patients with elevated lateral compartment patellofemoral joint pressures, patellofemoral instability and elevated TT-TG distances. Medialisation of the TT was not found to result in significant increases in medial joint pressures, however careful pre-operative planning and confirmation of intra-operative tracking clinically are recommended to ensure optimal post-operative outcome and avoidance of complications resulting from over medialisation.

7.10 Conclusion / Key Findings

Progressive TT lateralisation resulted in:

1. Elevated lateral contact pressures.
2. Increased lateral patellar tracking.
3. Reduced patellar stability.

Meanwhile medialisation of the TT:

1. Restored patellar stability.
2. Increased medial patellar tracking.
3. Did not result in excessive changes in pressure akin to those identified when the TT was lateralised.

These findings provide some rationale for TT transfer surgery in patients with elevated TT-TG distances and patellofemoral pathology to help alleviate symptoms.

7.11 Addendum

The aim of this chapter of the thesis was to determine the effect of progressively medialising and lateralising the TT. In order to do this effectively a number of factors required consideration. These are summarised in the following section, with the rationale for the final methodology outlined.

7.11.1 Plate design

Initially when planning the method for the TT medialisation and lateralisation a number of practices were performed on the knee specimens, which had been tested during the first 3 experimental chapters of this thesis. Prior work investigating cadaveric tibial osteotomies had reported successfully fixing the tuberosity directly to the tibia, as occurs intraoperatively (Ramappa et al., 2006), hence, this was the plan initially for the present study. However it was quickly established that following osteotomy of the TT, it would not be possible to fix the TT at progressive medial and lateral locations on the tibia for a number of reasons:

- ✗ The bone was not good quality in some knees, and tended to crumble, making accurate positioning and re-positioning of the TT impossible.
- ✗ In some of the smaller knees it was not possible to move the TT 15-20mm medially or laterally as the anterior aspect of the tibia was not wide enough to accommodate this.
- ✗ It was found that the position and orientation of the tubercle could not be controlled accurately.

It was realised therefore that some form of fixation device required to be developed to enable standardised, progressive TT medialisation and lateralisation. Following pilot testing in the laboratory it was determined that the device would be required to enable:

- ✓ Secure fixation to the tibia, to prevent it from being dislodged or moving during the testing, once the quadriceps muscles were loaded.
- ✓ It should be capable of being attached to the anterior tibia without inducing any changes to patellar contact mechanics or kinematics which could influence results.

A similar prior study had devised a sliding plate for the TT to be moved progressively medially and laterally (Kuroda et al., 2001). However this method still possessed some inaccuracy in the precise placement of the TT and lacked the ability for it to be consistently re-tested at 5mm intervals along the plate. It was therefore decided that a 'locking plate' would be the most appropriate device to fix to the anterior tibia. This would permit secure fixation to the tibia and enable standardisation of medialisaiton and lateralisation of the TT since this could be undertaken by fixing the TT to each of the progressive positions. With this in mind, two different prototypes were made up to pilot test in the laboratory. The first is shown in Figure 7.25.

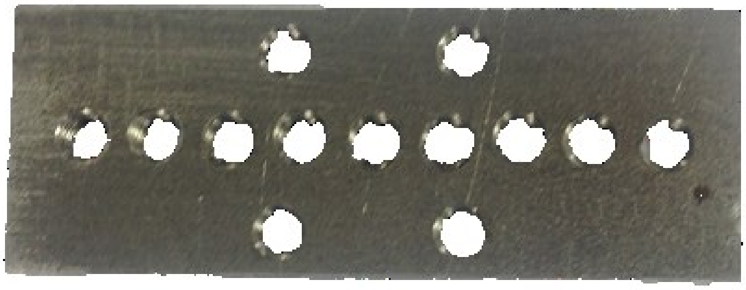

Figure 7.25 The first plate developed for fixation to the anterior tibia to enable progressive TT medialisation and lateralisation. The central row of holes shown are for the TT to be positioned in its anatomical location (central hole), and then at progressive 5mm intervals up to 20mm medially and laterally. The top and bottom row of holes were for screw fixation of the plate to the anterior tibia.

A second prototype was also developed for the initial round of pilot testing (Figure 7.26). It was designed on a slight curve. This was to address concerns that when the TT was moved sideways along a straight line that it would cause a rise in tension in the patellar tendon to develop which would not be accounted for. In order to address this, the curved plate was developed to account for this as the TT was moved and positioned progressively medially and laterally.

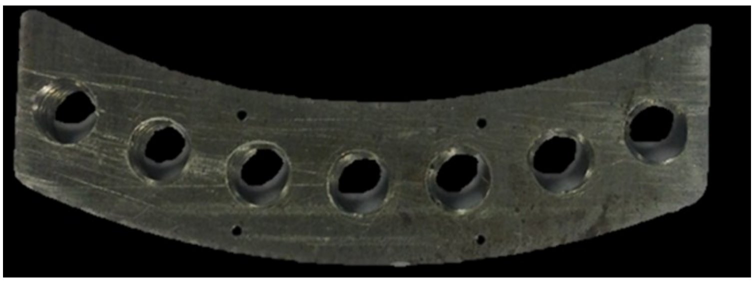

Figure 7.26 The second plate developed for fixation to the anterior tibia. The central hole is to correspond to the anatomical TT position and the holes to the right and left sides to permit progressive 5mm medialisation and lateralisation up to 15mm. The small holes in the top and bottom rows of the plate were used to fix K-wires through as an alternative to the screws planned for use in the first image.

Following a further round of testing it was realised that neither plate was suitable for testing. This was primarily due to the fact that it was impossible to securely fix either to the anterior tibia. Both the K-wire and screw fixations failed to work adequately to prevent motion of the plate in relation to the bone. It was therefore attempted to cement the plate in position at the anterior tibia. Unfortunately this was not successful since, even following careful application, the cement tended to occlude the pre drilled holes for TT fixation. Furthermore the cement took up space beneath the plate, which was difficult to standardise, and as such would be a potential variable in altering the A-P position of the plate. A third plate was therefore developed for further pilot testing (Figure 7.27).

The curved versus straight bar design raised a number of questions. Ultimately it proved very challenging to standardise the curve design of the plate since the specimen sizes were very varied. There was a range in the distances between the distal patellar tendon and the TT of the specimens to be tested, making it impossible to standardise the position and angle of the holes along the curved plate. Observation of the patellar tendon when the TT was medialised and lateralised along the straight bar identified no evidence of excessive patellar tendon stretching or loading. Therefore it was decided to use a straight plate design throughout testing. This ensured standardisation of plate position and of the testing conducted across all the specimens. Limitations were clearly outlined and the method ensured further variables, in particular secondary changes to patellar height, were not introduced to the experiment.

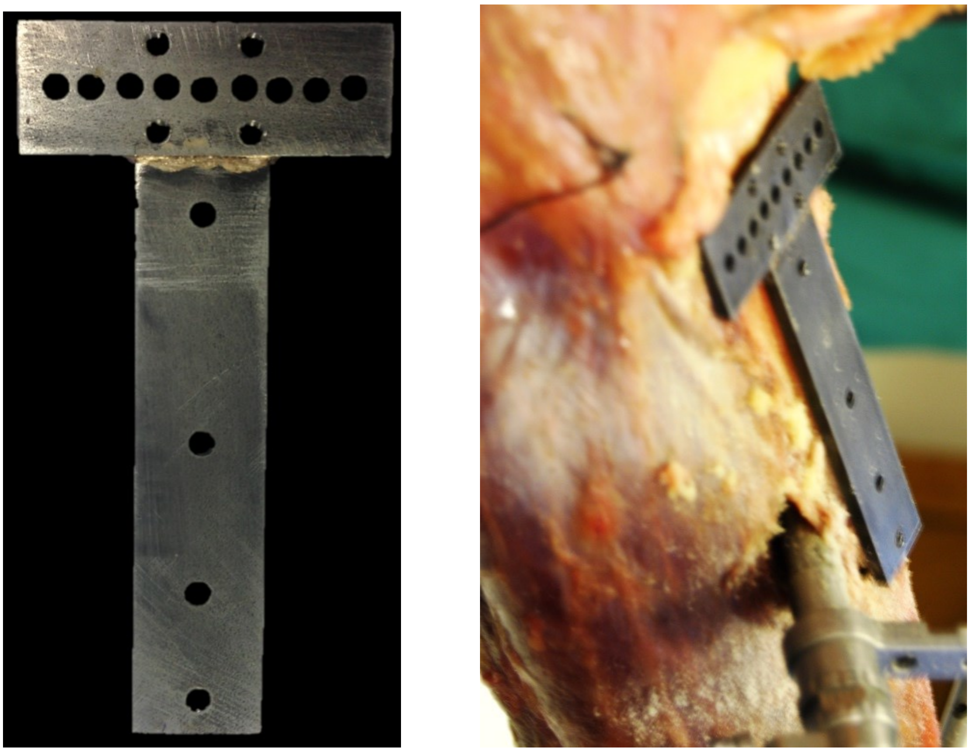

Figure 7.27 The left image shows the third prototype to be developed with a bar to be secured down the tibia for added fixation. The right image shows the plate attached to a specimen during pilot testing.

The third round of pilot testing with the plate highlighted further problems with the design. The plate was more secure than previously with the bar along the body of the tibia as shown in Figure 7.27. However because it was secured into position at the outset, testing was able to proceed further than previously. This identified a tendency for the plate to loosen after a number of rounds of testing. Furthermore, the TT was fixed into the plate with a screw. Unfortunately following multiple tests, the TT began to degrade and the bone weakened. This raised concerns that the bone may fracture during testing, which would invalidate any further results from the specimens.

A final design was therefore developed to address both of these problems. Firstly a cross bar was added along the top and bottom of the T-plate (Figure 7.28). This was to help the plate withstand the loading from the muscle tensions as the knee was flexed through motion and to also help with fixation. Secondly the top of the plate was extended out, so it was significantly wider than previously.

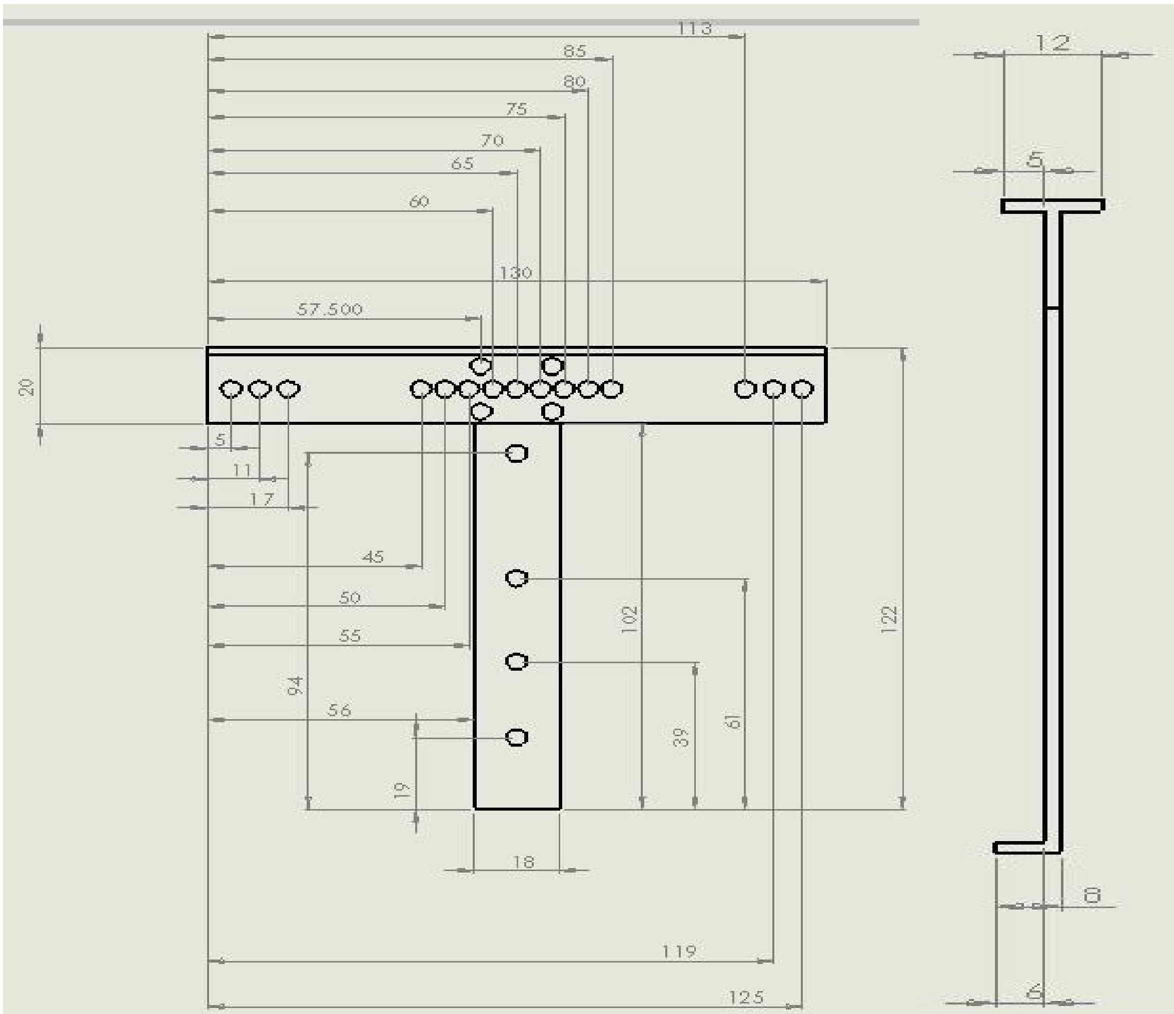

Figure 7.28 Drawings for the final plate design following pilot testing and refinement. Measurements are all shown in (mm) and made in Solidworks (2012).

Lastly a second plate was developed to sandwich on top of the base T-plate (Figure 7.29). This aligned identically with the top plate of the T-bar plate. The TT was positioned between the two plates and they were then clamped to one another using the wide fixations of the base plate and top plate. This was pilot tested and worked very well to ensure the TT osteotomy was securely fixed, but not vulnerable to fracturing.

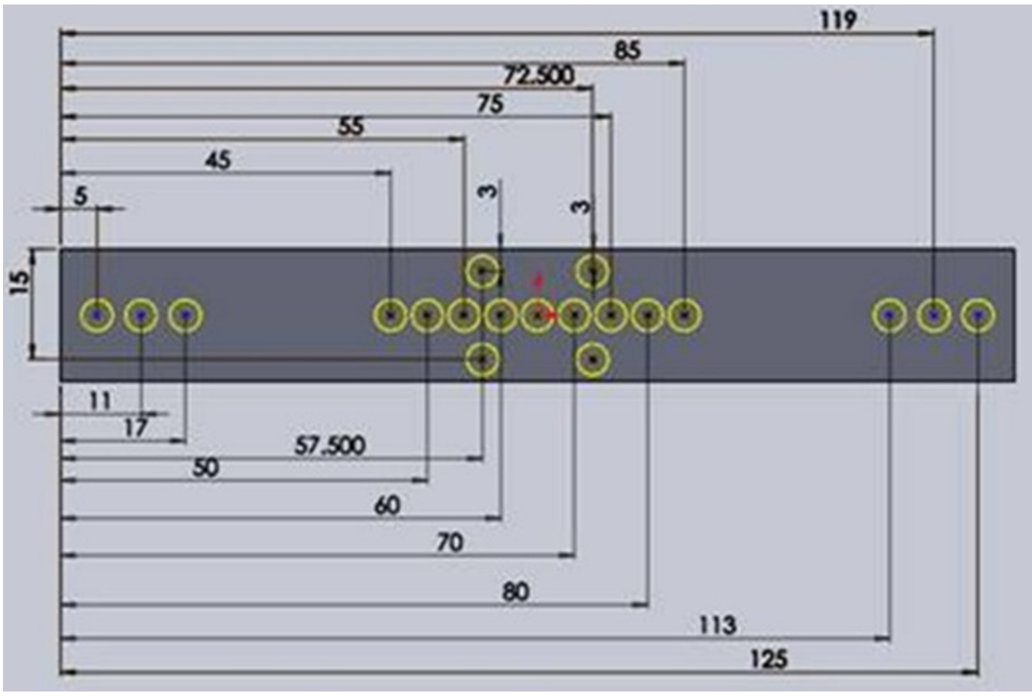

Figure 7.29 Diagram showing the top plate design to align with the T-plate in Figure 7.28 (made in Solidworks (2012))

The finished T-plate and refined method for T-plate attachment to the tibia is oultined in section 7.4, and Figure 7.2 and Figure 7.8.

Pilot testing identified no significant effect of plate insertion on any of the outcomes examined in the study: patellar tilt or translation, medial or lateral, peak or mean patellofemoral joint articular contact pressures. This was however also checked for all specimens tested, and the findings are reported in section 7.7.1.

7.11.2 Tibial tuberosity progressive medialisation and lateralisation

Following refinement of the T-plate, a full specimen was pilot tested. It had been planned to perform medialisation and lateralisation of the TT at 5mm, 10mm, 15mm and 20mm in turn. However during testing of the first specimen it became evident it was very challenging to displace the TT beyond 10mm due to restraint from the local soft tissues. This was discussed with the surgeons advising on the project and they reported this was a problem often encountered intraoperatively. It was advised that in this circumstance, small soft tissue releases would be performed medially and laterally, just proximal to the TT. These were performed by the surgeon to the superficial fascial layer: a standardised distance of 5mm wider than the medial and lateral edges of the TT, and were then extended 2cm in a proximal direction. Following these incisions the TT could be comfortably medialised and lateralised to 15mm; however it could still not easily be displaced to 20mm. In order to avoid inflicting any damage on the soft tissue structures, it was decided to limit testing to having the TT positioned: 15mm, 10mm and 5mm medial, in its anatomical position and 5mm, 10mm and 15mm lateral.

7.11.3 Conclusion

Following extensive pilot study testing based on prior literature and trial and error, a plate design was finalised as described for use in the present study. The final design ensured no movement between the plate and bone was permitted following fixation. It did not cause disruption to patellar kinematics or patellofemoral joint mechanics.

CHAPTER 8

TT-TG and MPFL

8.1 Background

The patellofemoral joint articulation has been highlighted as one of the most biomechanically complex in the human body with a combination of ligamentous, muscular and bony constraints maintaining patellar stability (Senavongse and Amis, 2005) (Chapter 2). Prior work has recognised that dysfunction, of varying severity, in one or more of these structures is commonly associated with patellar dislocation (Baldwin, 2009; Dejour et al., 1994). A wide range of surgical procedures are proposed for the management of these patients, meaning the treating clinician often has the challenge of reasoning and determining the optimal intervention for use in individual cases. The previous chapters determined the adverse effects of MPFL transection and TT lateralisation, both of which resulted in excessive patellar tilt and lateral joint contact pressures. This concluding experiment in the series of cadaveric studies therefore set out to investigate whether there is a threshold at which the effects of TT malalignment can be restored with MPFL reconstruction alone, or whether surgery in patients with extensor mechanism malalignment should always constitute, at least in part, a TT transfer procedure.

In the operative management of patellar dislocation patients, two main approaches are considered. The soft tissue approach addresses damage to the medial retinaculum and specifically the MPFL. Reconstruction of this ligament has gained increasing popularity in the last 5-10 years. Its biomechanical role in maintaining patellar stability is highlighted in Chapter 4. The minor trauma, early mobilisation and successful clinical outcomes in patients undergoing MPFL reconstruction support its implementation (Christiansen et al., 2008). However this procedure fails to address bony malalignment such as femoral or tibial rotation which are frequently present in patients with recurrent patellar dislocation (Dejour et al., 1994). Alternative surgical approaches such as TT transfer procedures, which involve reducing the lateral vector applied to the patella through medialisation of the TT and extensor mechanism, are typically more aggressive; necessitating longer recovery periods, restricted post-operative mobilisation and an increased complication risk (Aglietti et al., 1994; Kohn et al., 2004).

Current literature supports the conservative primary management of patients following primary patellar dislocation in the absence of an osteochondral fragment (Stefancin and Parker, 2007). In the patient with TT alignment within normal limits, MPFL reconstruction alone appears an appropriate treatment to restore patellofemoral joint contact mechanics and kinematics (Schöttle et al., 2010; Servien et al., 2011) (Chapter 4). However in the patient with a combination of MPFL injury and abnormally high TT-TG distance, there is limited objective evidence to guide surgical decision making in the use of one intervention over another, or indeed whether there is a threshold up to which MPFL reconstruction can satisfactorily restore patellar mechanics. Prior research has suggested interventions aimed at the passive stabilisers of the knee, including the MPFL, carry less risk than those which involve more aggressive realignment of the active structures such as the extensor apparatus during TT transfer surgery (Ostermeier et al., 2007a). It has previously been reported that MPFL reconstruction alone provided excellent clinical outcomes at 5 years follow up in patients with trochlear dysplasia which was not treated (Steiner et al., 2006). These findings suggest that addressing the underlying bony malalignment is not always necessary to ensure surgical success. The present study therefore set out to investigate the effect of progressive TT medialisation and lateralisation following MPFL transection and reconstruction on patellar tracking, stability and patellofemoral joint contact mechanics.

8.2 Aims

Based on findings summarised in Chapters 4 and 7, it was hypothesised that:

1. MPFL transection in combination with progressive TT lateralisation would:
    a. Result in elevated lateral patellofemoral joint contact pressures and increased lateral patellar motion.
    b. Result in a reduction of patellar stability as the TT was lateralised.
2. MPFL reconstruction would:
    a. Restore patellar kinematics and patellofemoral joint mechanics when the TT was positioned in its anatomical or 5mm lateralised positions, but not when in 10mm or 15mm lateralised positions.
    b. Restore patellar stability when the TT was 5mm lateralised but not when positioned 10mm or 15mm lateralised.

8.3 Clinical Relevance

This chapter presents findings which typically relate to populations of patients suffering patellar dislocation in the presence of abnormal anatomy. The aim is to determine whether these patients can be treated in a similar manner to those populations considered in Chapters 5 and 6, or whether additional intervention is necessary to optimise clinical outcome.

8.4 Materials and Methods

8.4.1 Specimen Preparation

The local research ethics committee granted approval for this study. In order to conserve cadaveric specimens, the same specimens described in Chapter 7 were used for testing in this study. The knees were already positioned in the test rig, with the Tekscan sensor and optical trackers in place. The testing followed directly on from Chapter 7, thus the T-plate was also in position. Knees were not moved from the test rig between the experiments. This ensured data sets could be compared directly without the addition of any external variables which could have influenced results.

8.5 Testing Procedure

With the TT in its anatomical position, the MPFL was transected and measurements taken with the TT in each of the seven different positions (15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised), at six different angles of knee flexion (0°, 10°, 20°, 30°, 60° and 90°) both with and without the 10N lateral load applied to the patella. The MPFL was then reconstructed by one experienced orthopaedic surgeon and the measurements repeated for all combinations of the variables.

The first fascial layer in the area of the medial epicondyle and adductor tubercle was carefully exposed, with care taken to cause no disruption to the medial retinaculum. The femoral attachment of the MPFL was identified and marked with a pin to ensure anatomical MPFL reconstruction could be performed. The MPFL was then transected with a No 10 scalpel blade near the femoral attachment to minimise trauma to the retinaculum and adjacent structures. Measurements were recorded as described above.

MPFL reconstruction was performed using a double strand gracilis tendon. The tendon was embedded in the superomedial border of the patella. Fixation of the graft to the patella was made via two through bone sutures which were tied together at the superolateral patellar border. A femoral tunnel was drilled at the pre-marked anatomical femoral attachment of the MPFL. The free graft ends were sutured together and passed between the second and third layers of the medial retinaculum and into the femoral tunnel, with the sutures fed from medially to laterally. The graft was fixed with a clamp laterally at the femur with 2N of tension applied. This technique is described in full in section 5.4 of the thesis. The graft was always fixed with the TT positioned in its anatomical position with the knee flexed to 30°. These parameters were determined based on results in Chapter 5 defining the optimal reconstruction of the MPFL.

At each stage of the experiment, both when the MPFL was transected and when it was reconstructed, testing order of TT position, flexion angle and 10N lateral load application was randomised to minimise any order bias.

8.6 Analysis

Raw data was processed using custom written MATLAB scripts (Appendix A+B) and analysed in SPSS. Tekscan files were also manually checked to confirm the processed data. A Shapiro-Wilk test was performed and determined that data sets were normally distributed.

The data for the specimens with the MPFL intact and the TT in its seven positions reported in chapter seven was included in the analysis to enable the effect of MPFL state (intact, transected and reconstructed) to be examined.

Dependent variables were patellar tilt and translation, mean and peak medial and lateral facet articular contact pressures. The primary factors investigated were TT position (15mm, 10mm, 5mm medialised, anatomical, 5mm, 10mm and 15mm lateralised), MPFL state (intact, transected, reconstructed) and flexion angle (0˚, 10˚, 20˚, 30˚, 60˚, 90˚). Two sets of data were analysed:

1. Six RM ANOVA (one on each of the dependent variables) were performed on the data sets obtained when no 10N external load was applied to the patella. This examined the effects of TT position, MPFL state and knee flexion angle.
2. A further RM ANOVA was performed on each of the dependent variables. The purpose of this was to investigate patellar stability. As in the previous sections, measurements taken from the knee with the TT in its anatomical position, MPFL intact and no 10N lateral load applied to the patella were subtracted from the corresponding flexion angle reading of each combination of TT position and MPFL state when the 10N lateral displacing load was applied

(i.e. (n condition+10N lateral load applied) – (plate in-situ with TT in its anatomical position (MPFL intact and no external load applied)).

Since the objective of this chapter was to investigate the relationship of TT position and MPFL status, a third RM ANOVA (as performed in Chapters 6 and 7) to directly compare each individual recording before and after the application of the 10N lateral displacing load was not undertaken.

8.7 Results

8.7.1 The combined effect of MPFL state and TT position

8.7.1.1 Patellar Translation

MPFL state influenced patellar tracking. With the MPFL intact and the knee in full extension, the patella was positioned 2.8mm and 4.7mm lateral when the TT was positioned in its anatomical and 10mm lateralised positions respectively. Lateral patellar translation increased to 3.3mm and 5.8mm (anatomical and 10mm lateral positions) following MPFL transection and was restored to 2.5mm and 4.0mm lateral respectively following MPFL reconstruction. These trends are highlighted in Figure 8.1 and Table 8.1.

MPFL state was found to have a significant effect on patellar translation (P<0.001), with MPFL transection resulting in increased lateral patellar tracking, which was typically fully or at least partially restored with MPFL reconstruction. The state of the MPFL had a significant interaction with flexion angle (P=0.001); MPFL transection and reconstruction typically had a greater effect on elevating and restoring patellar translation during early knee flexion (Figure 8.1). No significant interaction was found between TT position and MPFL state (P=0.096), with the effect of changing MPFL status tending to be reasonably constant throughout the differing TT positions (Figure 8.1).

No significant post hoc tests were identified for patellar translation when the TT was displaced 5mm (Figure 8.2A). Figure 8.2B+C highlight that with the TT displaced 10mm laterally and the MPFL transected, lateral patellar translation significantly increased at 0° and 10°, and at 0°, 10° and 20° following 15mm TT lateralisation. These effects were restored following MPFL reconstruction with no significant effect in any of the TT lateralised positions identified (all P>0.05). However there was a significant effect at 10°, 20° and 30° on patellar translation when the TT was positioned 15mm medially and the MPFL reconstructed (P<0.05).

Table 8.1 Patellar medial-lateral translation (mm) for the MPFL intact, transected and reconstructed knee with the TT in its anatomical, 5mm lateral, 10mm lateral and 15mm lateral positions. Results shown through knee flexion range (mean and standard deviation (SD), n=8).

Flexion Angle (°)	TT Anatomical (mm)		TT 5mm Lateral (mm)		TT 10mm Lateral (mm)		TT 15mm Lateral (mm)	
	1. MPFL INTACT							
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	2.8	5.0	3.2	5.2	4.7	5.6	6.9	7.0
10	3.5	4.9	3.6	4.6	5.2	5.4	7.7	7.0
20	4.2	4.1	4.0	4.1	5.8	5.2	8.2	6.7
30	5.1	3.9	5.1	3.8	6.5	4.8	9.6	7.3
60	7.8	3.5	8.0	3.6	9.1	4.0	11.8	6.3
90	8.7	3.1	9.0	2.4	10.5	3.5	12.6	5.4
	2. MPFL TRANSECTED							
Flexion Angle (°)	**TT Anatomical (mm)**		**TT 5mm Lateral (mm)**		**TT 10mm Lateral (mm)**		**TT 15mm Lateral (mm)**	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	3.3	4.8	3.9	4.9	5.8	5.1	9.0	6.2
10	4.1	4.4	4.1	4.3	6.4	5.0	9.0	6.6
20	4.8	4.0	4.7	4.2	9.0	4.9	9.9	6.6
30	5.7	3.9	5.9	4.1	7.6	4.7	10.0	6.6
60	8.2	3.4	8.2	3.5	9.6	3.7	11.4	6.2
90	9.1	2.9	9.2	1.8	10.9	3.4	11.8	6.1
	3. MPFL RECONSTRCUTED							
Flexion Angle (°)	**TT Anatomical (mm)**		**TT 5mm Lateral (mm)**		**TT 10mm Lateral (mm)**		**TT 15mm Lateral (mm)**	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	2.5	4.3	2.7	4.5	4.0	5.0	7.0	6.1
10	3.2	4.1	3.0	4.1	4.9	4.6	7.4	6.2
20	4.0	4.0	3.5	3.7	5.2	4.5	8.0	6.6
30	5.0	3.5	5.3	4.2	6.4	4.5	9.1	6.5
60	7.7	3.5	8.3	3.7	9.3	3.9	11.7	6.2
90	8.8	3.0	9.2	2.2	10.7	3.5	12.2	5.0

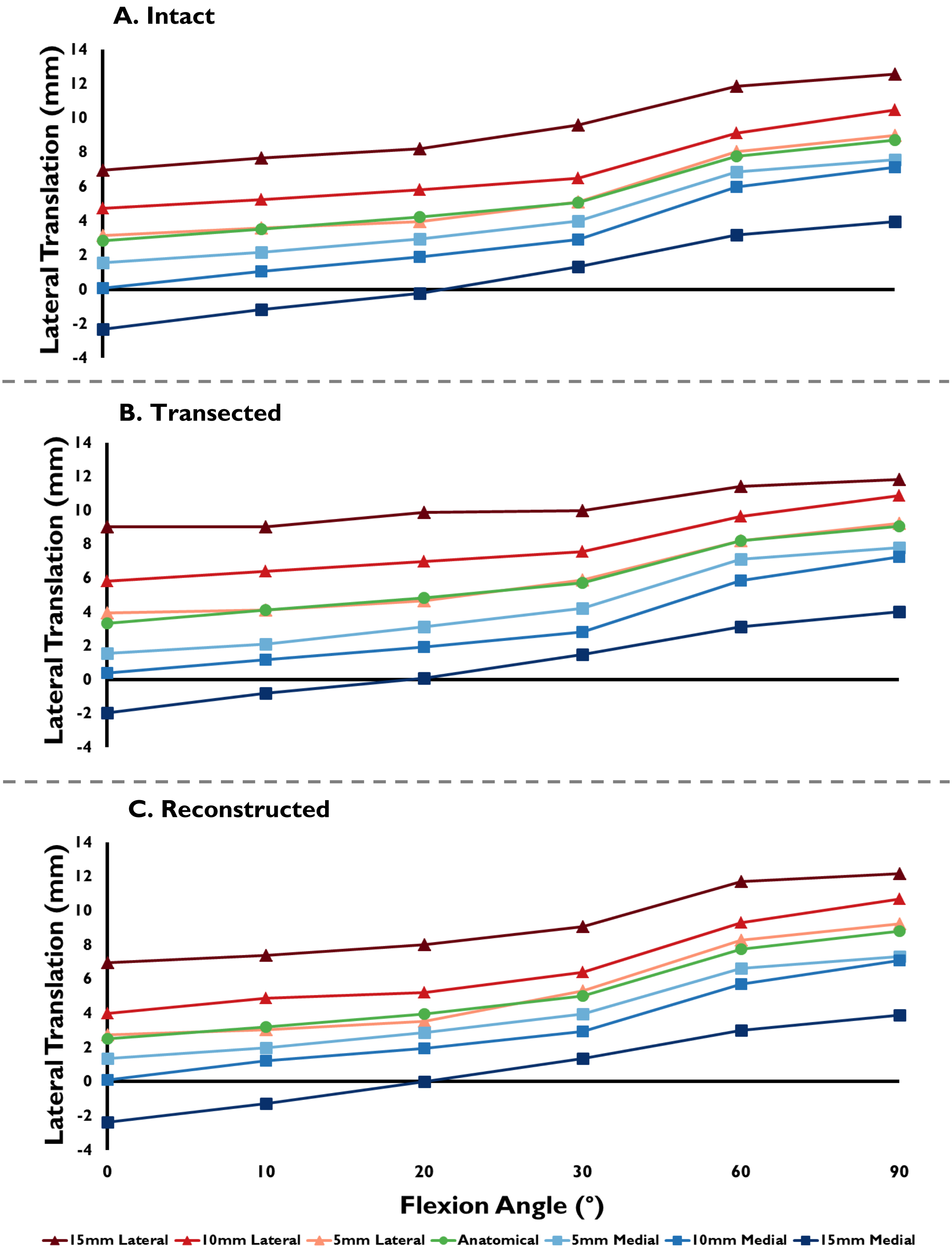

Figure 8.1 Patellar medial-lateral translation (mm; mean, n = 8) from 0° to 90° knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. results for the intact knee, **B**. results for the knee with the MPFL transected and **C**. results for the knee with the MPFL reconstructed.

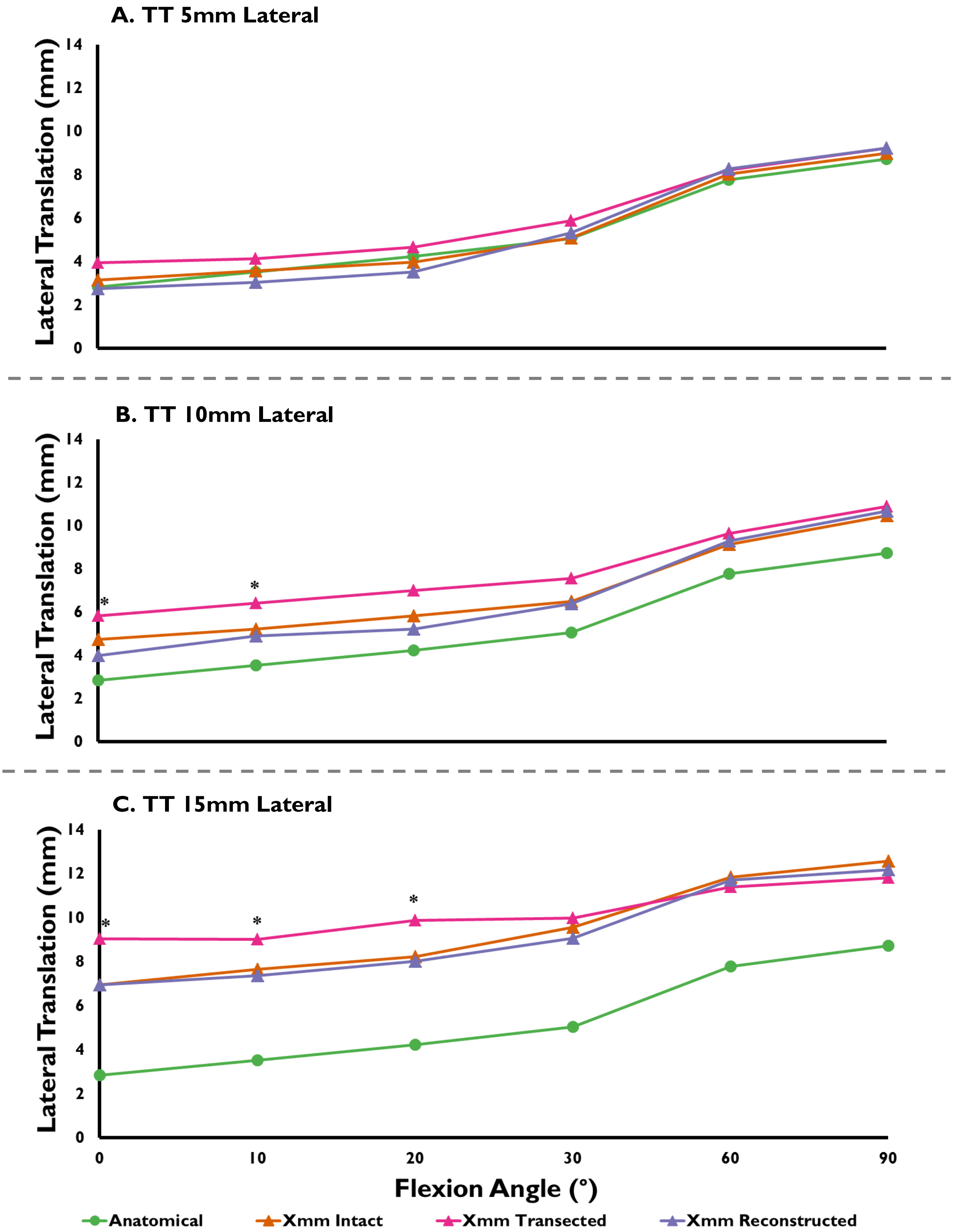

Figure 8.2 Patellar medial-lateral translation (mm; mean, n = 8) from 0˚ to 90˚ knee flexion. Green line is identical throughout, representing the TT positioned in its anatomical position (with MPFL intact). Data then shown in each graph corresponding to the TT: **A**: 5mm lateralised, **B**: 10mm lateralised and **C**: 15mm lateralised. Orange line = MPFL intact, pink line = MPFL transected, purple line = MPFL reconstructed. *P<0.05.

8.7.1.2 Patellar Tilt

Patellar tilt was more sensitive to TT lateralisation than medialisation as highlighted in Chapter 7 and Figure 8.3. In the intact knee at full extension, the patella rested at 2.8° and 7.5° lateral with the TT in its anatomical and 15mm lateralised positions respectively. Lateral patellar tilt increased to 3.6° and 10° (anatomical and 15mm lateral) following MPFL transection and reduced to 1.7° and 6.8° respectively following MPFL reconstruction. These trends are highlighted in Figure 8.3 and Table 8.2.

MPFL state was found to have a significant effect on patellar tilt, with lateral patellar tilt typically increasing following MPFL transection and reducing following reconstruction (P<0.001). A significant interaction was also identified between the state of the MPFL and flexion angle (P<0.001). Generally the effect of the MPFL state on tilt reduced with increasing knee flexion as the patella engaged and was constrained by the trochlear groove. A significant interaction was identified between TT position and MPFL state (P=0.048). Increasing TT lateralisation tended to increase the effect of MPFL transection and reduce the ability of MPFL reconstruction to restore patellar tilt. MPFL reconstruction restored patellar tilt to close to the intact state when the TT was in its anatomical and 5mm lateralised positions, but not with the TT in 10mm or 15mm lateralised positions (Figure 8.4).

Figure 8.4A shows that with the TT displaced 5mm, patellar tilt at 0° was significantly increased following MPFL transection. However this was restored after MPFL reconstruction, with no significant differences between the anatomical TT and 5mm lateralised TT evident (P>0.05). Figure 8.4B shows that with the TT lateralised 10mm, patellar tilt was significantly increased at 0°, this was increased further following MPFL transection and was not restored following MPFL reconstruction (P<0.05). Figure 8.4C highlights the significant effect of 15mm TT lateralisation on patellar tilt at 0° and 20°. Increases in patellar lateral tilt caused by TT lateralisation were not restored by MPFL reconstruction at either 0º or 20º (P < 0.05).

No significant patellar tilt post-hoc tests were identified for any of the measurements taken following medialisation of the TT (all P>0.05).

Table 8.2 Patellar medial-lateral tilt (°) for the MPFL intact, transected and reconstructed knee with the TT in its anatomical, 5mm lateral, 10mm lateral and 15mm lateral positions. Results shown through knee flexion range (mean and SD, n=8).

Flexion Angle (°)	MPFL INTACT							
	TT Anatomical (°)		TT 5mm Lateral (°)		TT 10mm Lateral (°)		TT 15mm Lateral (°)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	2.2	9.0	4.1	8.9	5.5	9.1	7.5	9.9
10	1.8	9.6	2.9	8.5	4.3	9.3	6.3	10.4
20	1.0	8.2	2.3	7.5	3.8	8.3	5.6	10.3
30	0.8	7.8	1.6	7.6	3.1	8.4	4.5	10.1
60	0.7	7.5	1.1	7.4	2.7	8.3	2.9	8.5
90	0.3	7.8	0.7	7.9	2.1	8.2	3.1	8.5
Flexion Angle (°)	**MPFL TRANSECTED**							
	TT Anatomical (°)		TT 5mm Lateral (°)		TT 10mm Lateral (°)		TT 15mm Lateral (°)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	3.7	8.9	5.2	8.9	6.4	9.1	9.9	9.6
10	2.8	8.5	4.0	8.8	5.6	8.8	8.3	9.7
20	2.3	7.4	3.4	7.6	4.9	7.9	7.2	9.6
30	1.0	7.5	2.2	7.5	3.8	8.3	5.4	9.6
60	0.7	7.5	1.1	7.2	3.1	8.1	3.3	8.6
90	0.3	7.7	0.8	7.8	2.4	8.0	3.8	8.8
Flexion Angle (°)	**MPFL RECONSTRCUTED**							
	TT Anatomical (°)		TT 5mm Lateral (°)		TT 10mm Lateral (°)		TT 15mm Lateral (°)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	1.7	7.5	3.3	8.6	5.0	9.1	6.8	9.7
10	0.9	7.7	2.4	8.2	3.8	9.0	5.7	10.1
20	0.2	7.3	1.9	7.3	3.4	7.8	5.2	10.1
30	-0.1	7.3	1.1	7.5	2.4	7.6	4.0	9.7
60	0.1	7.2	1.0	7.5	1.4	7.4	3.0	7.6
90	-0.3	7.9	0.4	8.0	1.1	7.7	2.6	7.7

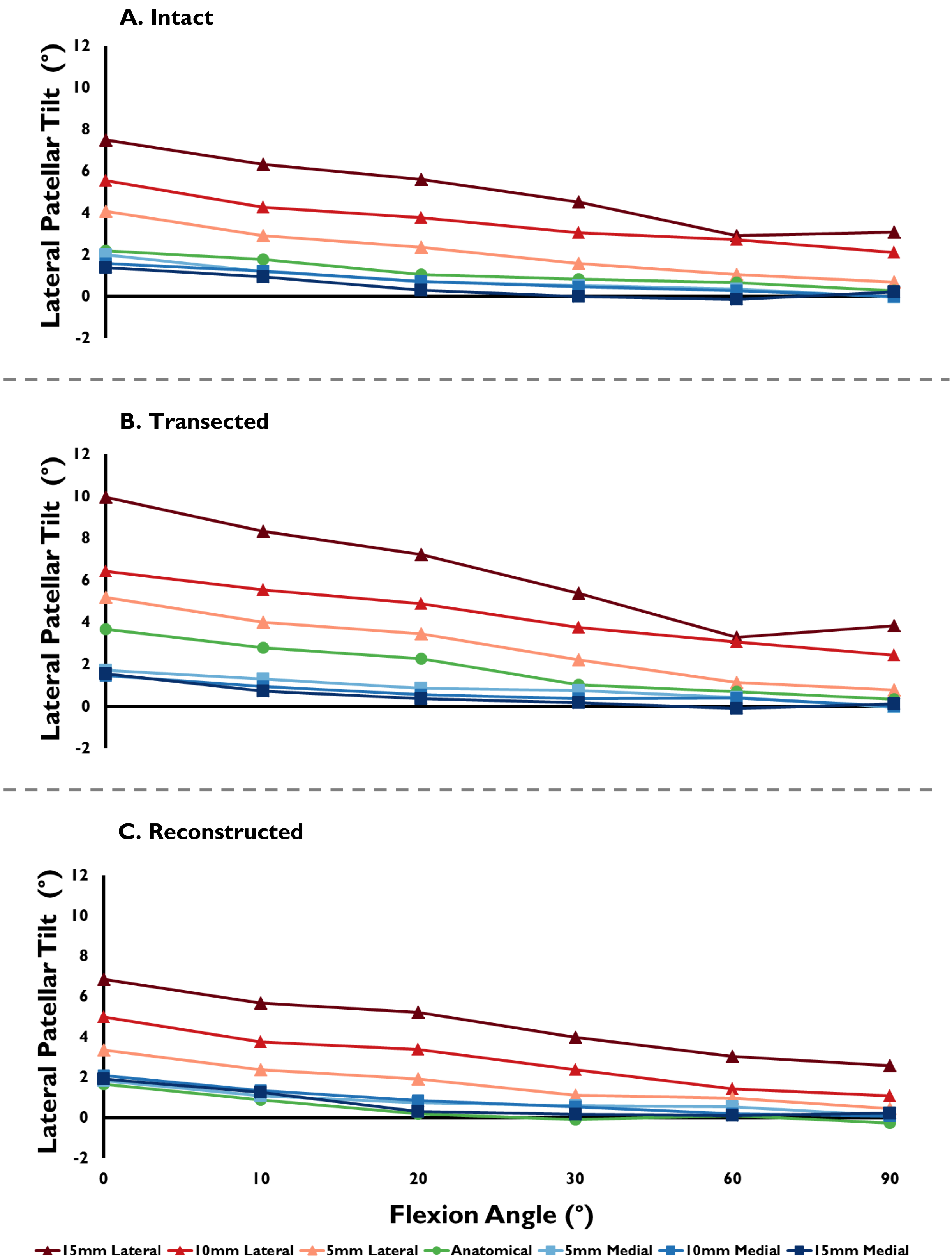

Figure 8.3 Patellar medial-lateral tilt (°; mean, n = 8) from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. results for the intact knee, **B**. results for the knee with MPFL transected and **C**. results for the knee with the MPFL reconstructed.

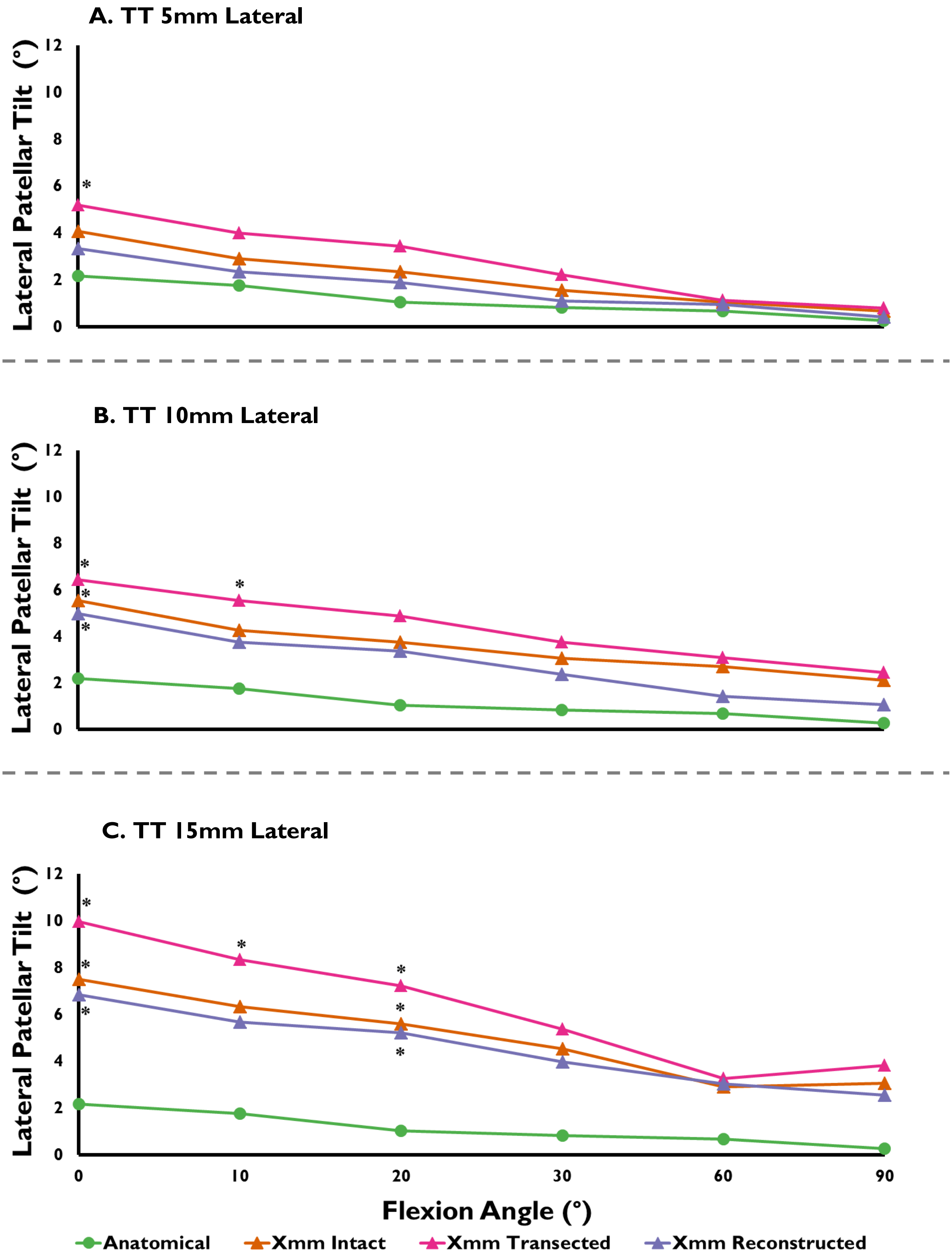

Figure 8.4 Patellar medial-lateral tilt (°; mean, n = 8) from 0° to 90° knee flexion. Green line is identical throughout, representing the TT positioned in its anatomical position (with MPFL intact). Data then shown in each graph corresponding to the TT; **A**: 5mm lateralised, **B**: 10mm lateralised and **C**: 15mm lateralised. Orange line = MPFL intact, pink line = MPFL transected, purple line = MPFL reconstructed. *P<0.05.

8.7.1.3 Mean Lateral Contact Pressures

There was a trend of increasing lateral contact pressures following TT lateralisation and MPFL transection (Figure 8.5). For some TT positions, pressures were restored after MPFL reconstruction, but this was not consistent. With the TT in its anatomical position in full extension, mean contact pressures were; 0.39MPa, 0.46MPa and 0.37MPa respectively for intact, transected and reconstructed MPFL states. Meanwhile TT lateralisation to 15mm gave pressures of 0.49MPa, 0.58MPa and 0.46MPa respectively for corresponding states at 0º.

A significant effect of MPFL status was identified, with mean lateral pressures typically rising following MPFL transection and reducing with MPFL reconstruction ($P<0.0001$). TT position and MPFL state demonstrated a significant interaction ($P<0.001$), with MPFL transection and reconstruction generally having a greater effect when the TT was more lateralised. Flexion angle was not found to have a significant effect on mean contact pressures ($P=0.171$); instead pressures remained reasonably constant throughout full knee flexion range (Figure 8.5).

Post hoc testing identified significant effects of 15mm TT lateralisation and medialisation at 0° with the MPFL intact. Following MPFL transection there was a significant effect on pressures with the TT 5mm lateralised at 0°, with the TT 10mm lateralised at 0°, 10°, 20° and 60° and the TT 15mm lateralised at 0°, 10°, 20° and 30°. MPFL reconstruction resulted in significant differences in the 15mm lateralised TT across all angles of flexion, and with the TT medialised 15mm at 0° and 60°.

8.7.1.4 Mean Medial Contact Pressures

A significant effect of MPFL state was identified on mean medial contact pressures ($P<0.0001$). Flexion angle was not found to have a significant effect ($P=0.103$), with Figure 8.6 highlighting mean pressures typically remained reasonably constant across all angles of knee flexion. A significant interaction between TT position and MPFL state was identified ($P=0.001$). This was reflected with an enhanced effect of MPFL transection and reconstruction when the TT was progressively lateralised.

Significant post hoc tests are highlighted in Figure 8.6 ($P<0.05$). Significant differences in pressures remained following MPFL reconstruction with the TT lateralised 15mm at 0° and 10° knee flexion and with the TT 10mm lateral at 10° flexion (all: $P<0.05$). There was also a significant effect of 15mm TT medialisation on the contact pressures at 60° flexion ($P<0.05$).

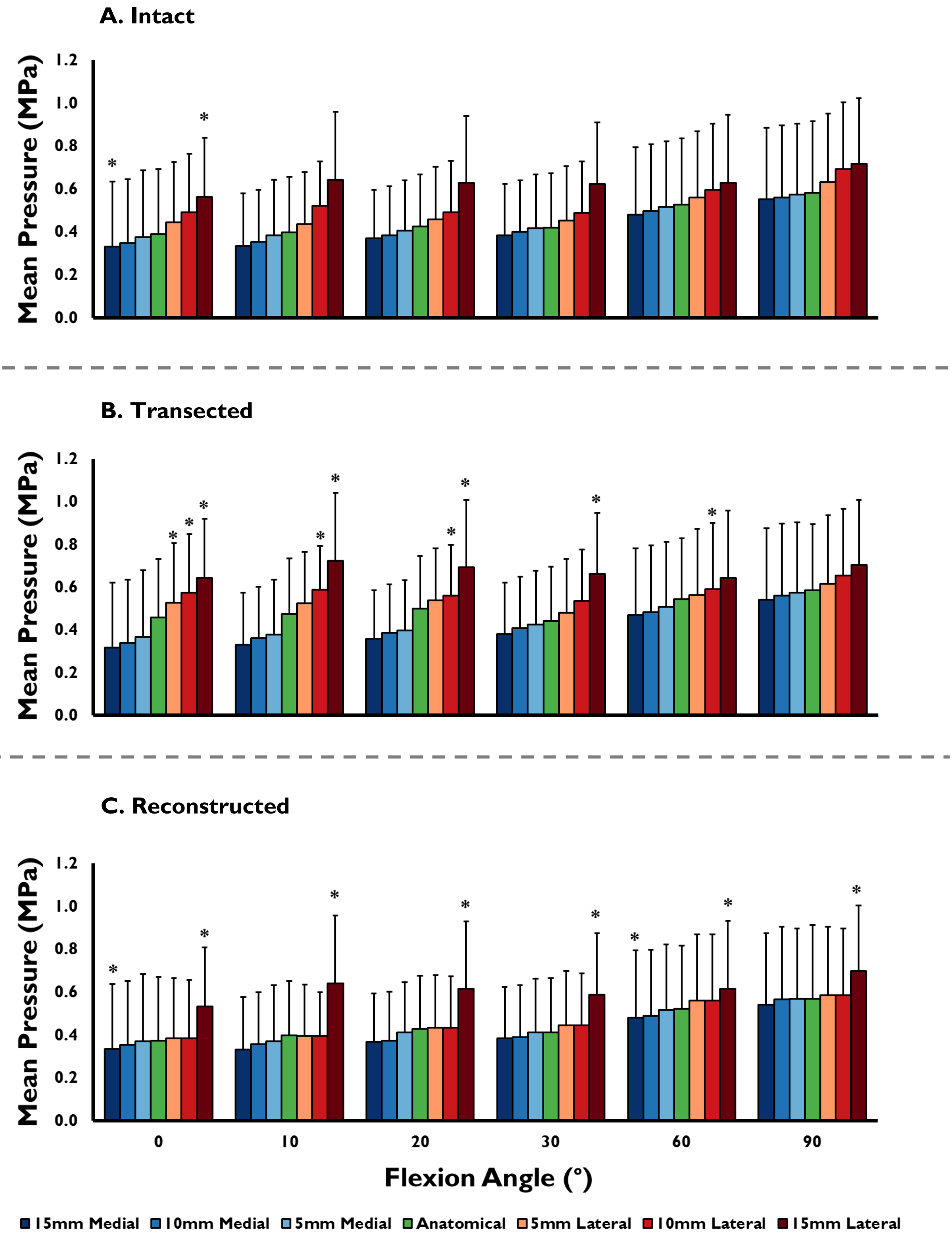

Figure 8.5 Mean lateral patellofemoral joint contact pressures (MPa; mean + SD, n = 8) from 0° to 90° knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. results for the intact knee, **B**. results for the knee with the MPFL transected and **C**. results for the knee with the MPFL reconstructed. *P<0.05.

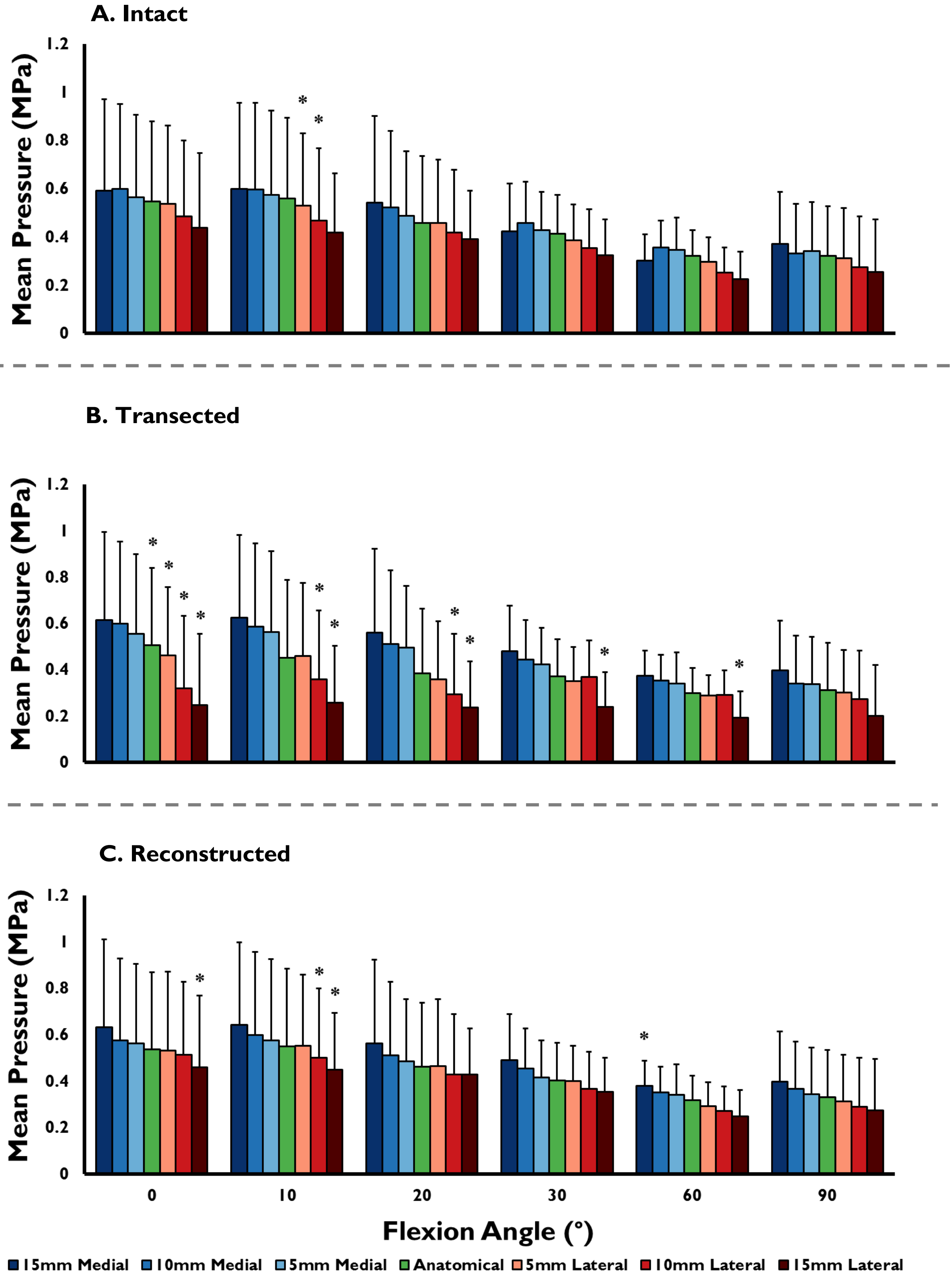

Figure 8.6 Mean medial patellofemoral joint contact pressures (MPa; mean + SD, n = 8) from 0° to 90° knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. results for the intact knee, **B**. results for the knee with the MPFL transected and **C**. results for the knee with the MPFL reconstructed. *P<0.05.

8.7.1.5 Peak Lateral Contact Pressures

MPFL state was not identified to have a significant effect on peak lateral pressure as highlighted in Figure 8.7 (*P*=0.746). Flexion angle was not found to have a significant effect on peak lateral patellofemoral joint pressures, with the pressures instead seen to remain reasonably constant through full flexion range (*P*=0.327). No significant interaction was identified between TT position and MPFL status (*P*=0.814), with altering MPFL status not found to have any consistent effect on peak lateral pressure outcomes.

8.7.1.6 Peak Medial Contact Pressures

Peak medial pressure results also highlighted a lack of a consistent trend in response to altering MPFL state from intact to transected and reconstructed. Analysis did not identify a significant effect of MPFL state on peak medial pressures (*P*=0.135) (Figure 8.8). There was a significant effect of flexion angle identified (*P*=0.044), with a tendency for medial contact pressures to reduce in deeper knee flexion as the patella translated laterally for all TT and MPFL states. No significant interaction between MPFL state and TT position was identified (*P*=0.069).

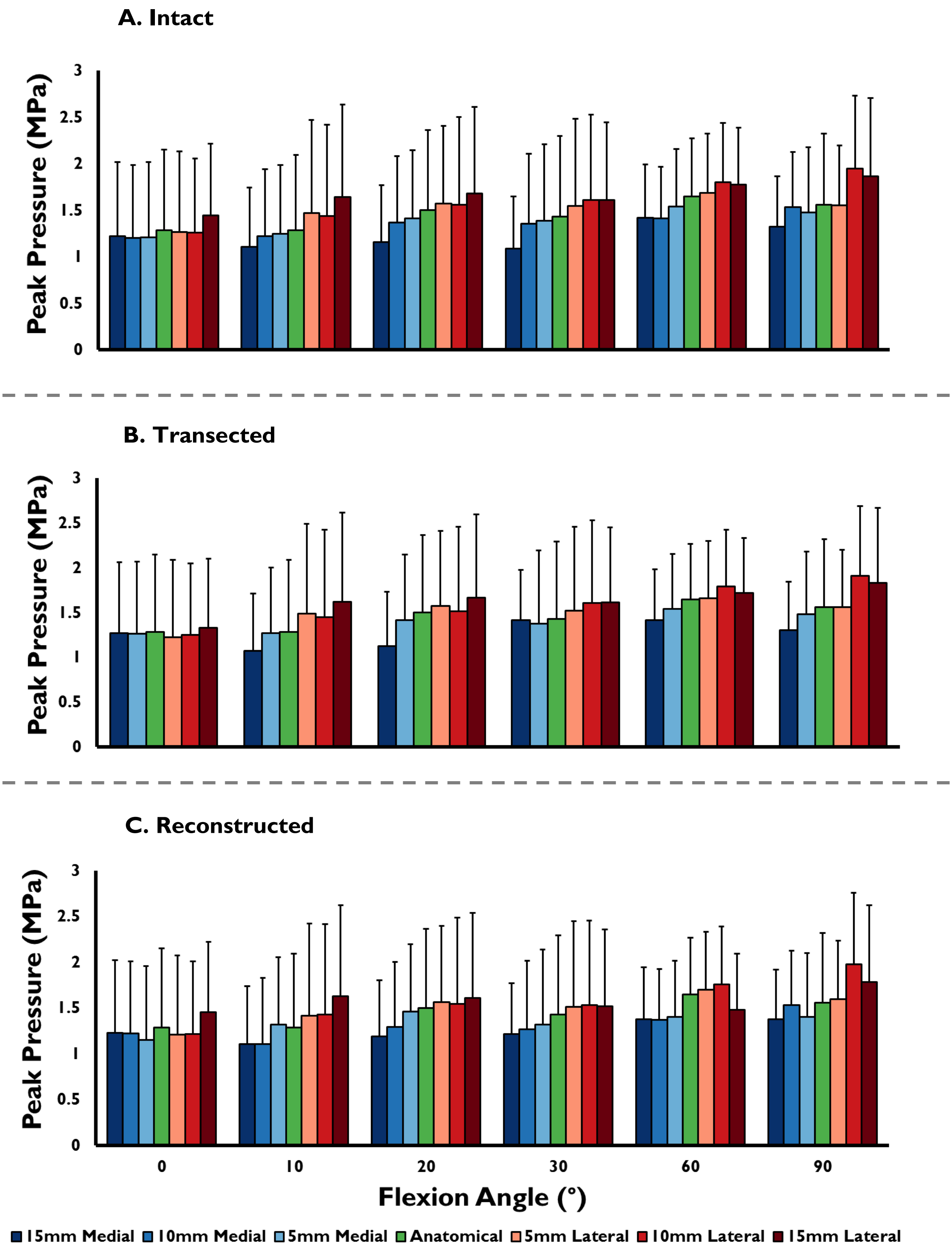

Figure 8.7 Peak lateral patellofemoral joint contact pressures (MPa; mean + SD, n = 8) from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. results for the intact knee, **B**. results for the knee with the MPFL transected and **C**. results for the knee with the MPFL reconstructed.

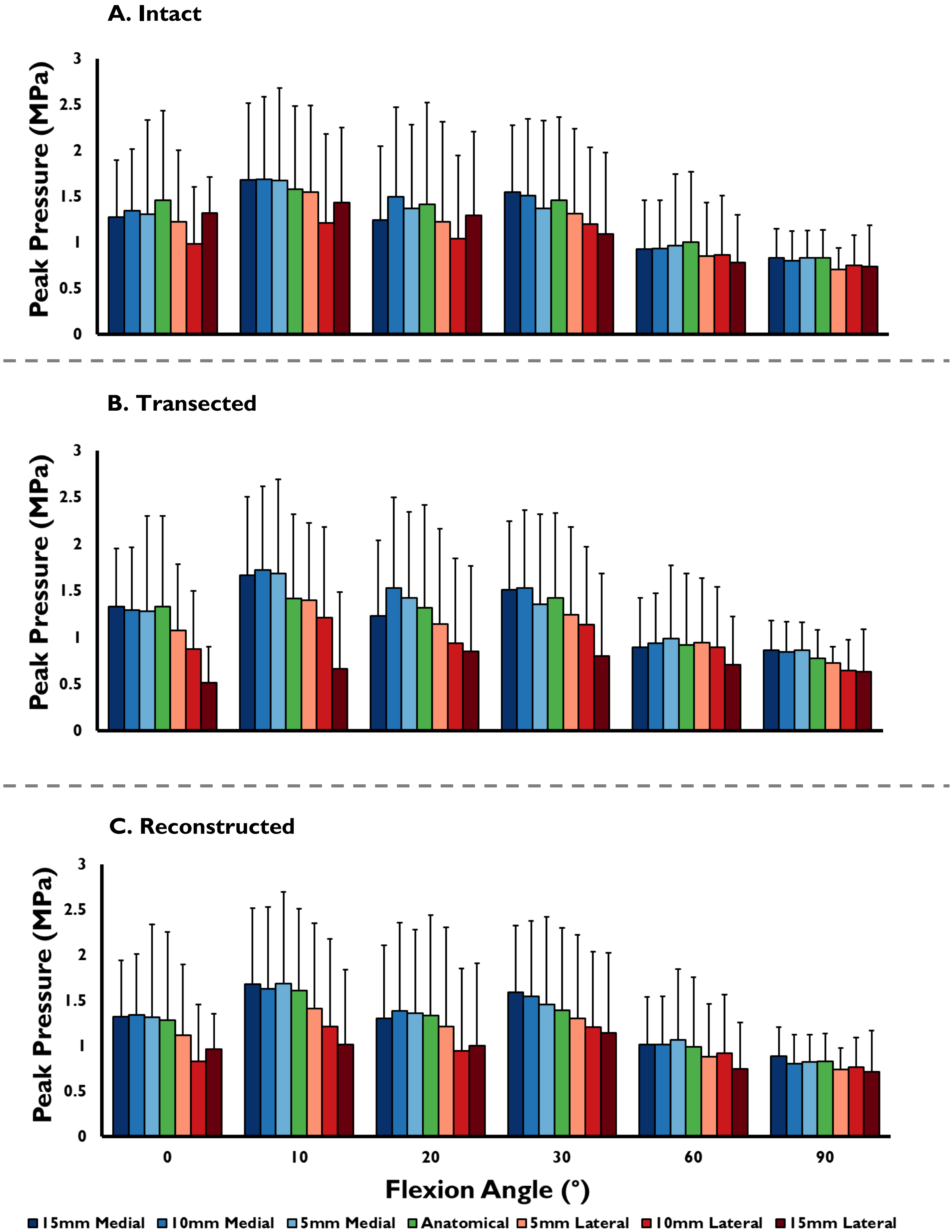

Figure 8.8 Peak medial patellofemoral joint contact pressures (MPa; mean + SD, n = 8) from 0° to 90° knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. results for the intact knee, **B**. results for the knee with the MPFL transected and **C**. results for the knee with the MPFL reconstructed.

8.7.2 The combined effect of MPFL state, TT position and 10N lateral load

8.7.2.1 Patellar Translation

In the intact knee flexed to 20° the patella was positioned 4.2mm lateral. This increased to 4.6mm lateral following the application of a 10N lateral load to the patella, resulting in an overall increase of 0.4mm lateral translation (4.6-4.2). Following MPFL transection the resting position of the patella at 20° increased to 4.8mm, increasing further to 6.1mm following lateral load application, making an overall change of 1.9mm (6.1-4.2). With the MPFL reconstructed, patellar translation reduced to 4mm, increasing to 4.5 mm following lateral load application. This meant the change in position from the original position was reduced to 0.3mm (4.5-4.2). In contrast to this, in the knee with the TT 15mm lateralised, flexed to 20°, the patella began positioned 8.2mm laterally, this increased to 10.1mm laterally following 10N lateral load application. This resulted in a change of 5.9mm from the start position of 4.2mm (10.1-4.2). Following MPFL transection the patella moved more laterally to 9.9mm, increasing to 11mm following lateral load application, a change of 6.8mm from the start position (11-4.2). Following reconstruction the patella was positioned 8mm lateral, increasing to 9.2mm following lateral load application, indicating an overall lateralisation of 5mm from the start position (9.2-4.2). It is therefore evident that following MPFL transection the trend was for the patella to translate more laterally, indicating a destabilising effect. This was restored to some extent following MPFL reconstruction (Figure 8.10).

MPFL state was found to have a significant effect on patellar translation following the application of a 10N lateral load, with translation significantly increasing following MPFL transection and reducing following reconstruction ($P<0.001$). A significant interaction was identified between the ligament state and TT position ($P=0.006$). Results were reasonably constant throughout knee flexion (Table 8.3 and Figure 8.9), therefore flexion angle was not found to have a significant effect on translation ($P=0.745$).

Post hoc testing failed to identify any significant effect of TT position and MPFL state on individual readings at specific angles of knee flexion (all $P>0.002$).

Table 8.3 Change in patellar medial-lateral translation (mm) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion. Shown for the TT in its anatomical, 5mm, 10mm and 15mm lateralised positions. Results shown through knee flexion range (mean and SD, n=8).

Flexion Angle (°)	MPFL INTACT							
	TT Anatomical (mm)		TT 5mm Lateral (mm)		TT 10mm Lateral (mm)		TT 15mm Lateral (mm)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.33	0.28	1.46	1.58	3.46	3.09	6.19	5.95
10	0.41	0.47	1.01	1.62	3.12	3.67	6.07	5.91
20	0.35	0.52	0.61	1.57	2.77	3.44	5.87	6.05
30	0.26	0.89	0.88	1.62	2.66	3.57	5.91	5.86
60	0.34	0.58	0.88	1.88	2.24	3.35	5.55	6.01
90	0.36	0.42	0.89	1.75	2.89	4.23	5.03	5.94
Flexion Angle (°)	**MPFL TRANSECTED**							
	TT Anatomical (mm)		TT 5mm Lateral (mm)		TT 10mm Lateral (mm)		TT 15mm Lateral (mm)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	1.64	0.89	2.64	1.69	4.78	3.12	7.83	4.87
10	1.88	1.23	2.11	2.03	4.59	3.94	7.50	5.78
20	1.86	1.46	1.75	2.13	4.32	4.19	6.72	5.51
30	1.98	1.10	1.88	1.67	4.13	3.42	6.60	5.74
60	1.31	0.39	1.00	1.63	3.13	2.94	5.81	5.47
90	1.23	0.76	1.36	1.66	3.54	3.44	5.16	5.79
Flexion Angle (°)	**MPFL RECONSTRUCTED**							
	TT Anatomical (mm)		TT 5mm Lateral (mm)		TT 10mm Lateral (mm)		TT 15mm Lateral (mm)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.29	0.43	0.49	0.59	1.82	2.01	4.80	4.92
10	0.41	0.50	-0.03	1.37	2.06	2.58	4.62	5.52
20	0.31	0.73	0.00	1.65	2.24	3.50	4.98	5.77
30	0.53	0.77	0.90	1.77	2.23	3.40	5.20	5.97
60	0.71	0.65	0.77	1.76	2.27	3.11	5.21	6.19
90	0.45	0.89	0.88	1.94	2.89	4.05	4.55	6.27

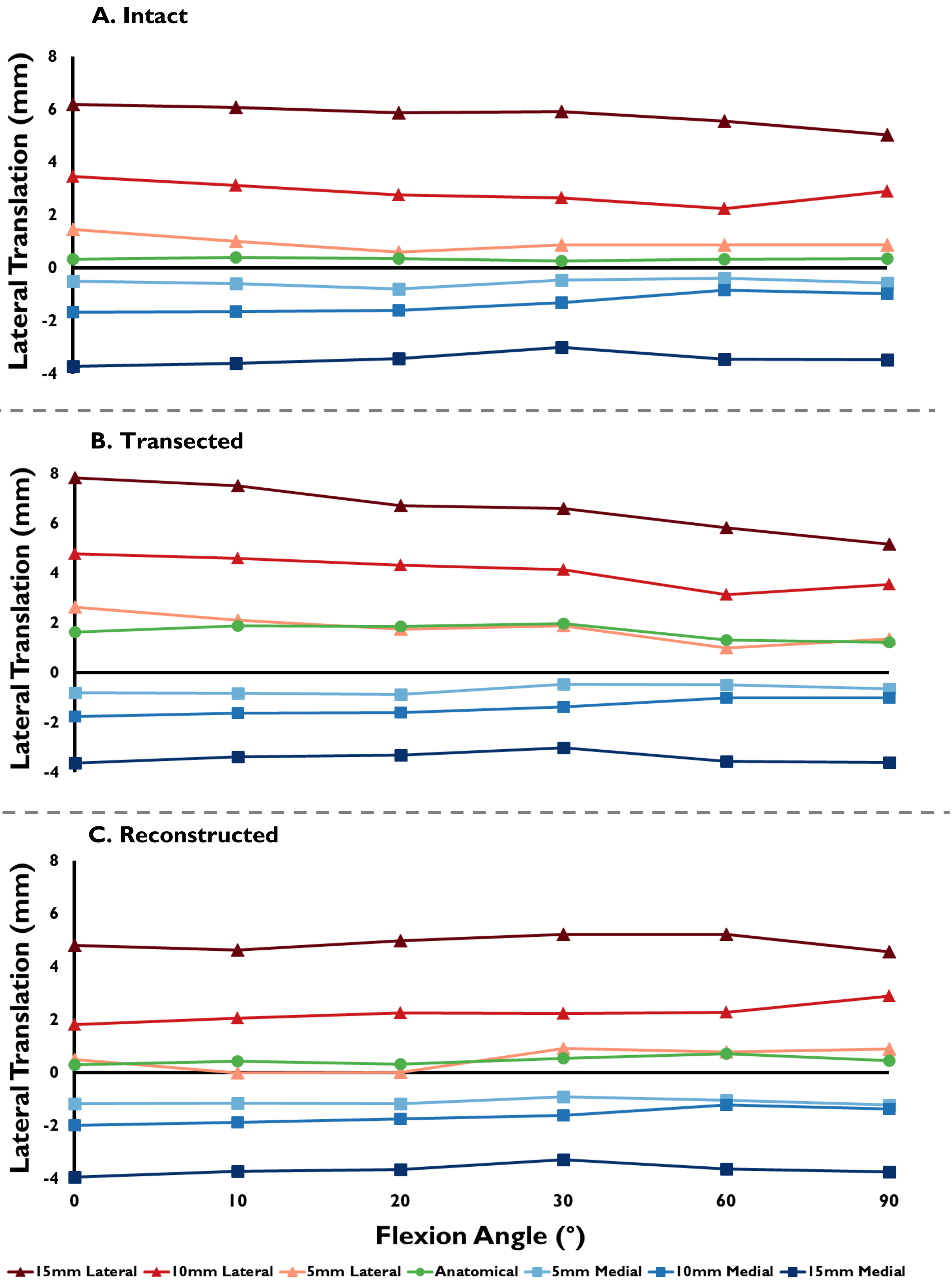

Figure 8.9 Change in patellar medial-lateral translation (mm; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. Shown with the MPFL: **A**. intact, **B**. transected and **C**. reconstructed.

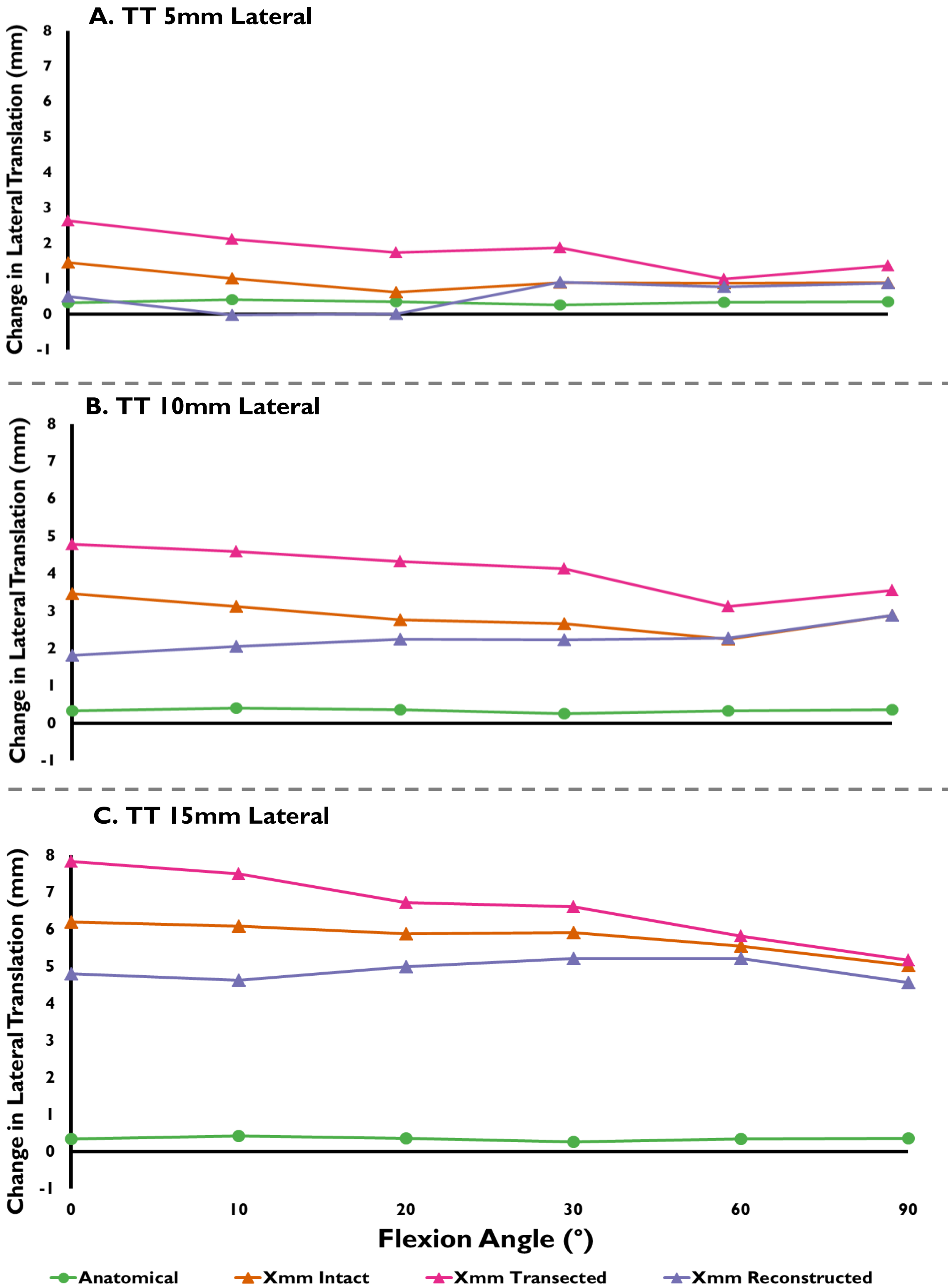

Figure 8.10 Change in patellar medial-lateral translation (mm; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion. Green line is identical throughout, representing the TT positioned in its anatomical position (with MPFL intact). Data then shown in each graph corresponding to the TT: **A**: 5mm lateralised, **B**: 10mm lateralised and **C**: 15mm lateralised. Orange line = MPFL intact, pink line = MPFL transected, purple line = MPFL reconstructed.

8.7.2.2 Patellar Tilt

In the fully extended knee, the patella on average was tilted 2.2° lateral. This increased to 2.4° following the application of a 10N lateral load, making a change of 0.2° (2.4-2.2). When the MPFL was transected, the patella was tilted 3.7° lateral, increasing to 4.8º with lateral load application. This made a change of 2.6º from the start position (4.8-2.2). Finally MPFL reconstruction reduced the lateral patellar tilt to 1.7º, and permitted only 2.7º tilt following 10N lateral load application. This meant a change of only 0.5º from the start position (2.7-2.2). When the knee was extended and the TT 15mm lateralised, the patella was positioned at 7.5º lateral, increasing to 9.4º following the application of the lateral displacing force. This equated to a 7.2º change from the start position (9.4-2.2). Following MPFL transection patellar tilt increased to 10º, which increased further to 11.2º following the lateral load application, giving a change of 9º from its starting position with the TT in its anatomical position (11.2-2.2). MPFL reconstruction reduced patellar tilt to 6.8º and only permitted 6.9º tilt following the lateral load application. This meant the patella lateralised 4.7º from its initial position, less than in the intact knee (6.9-2.2). Overall the results indicate an increase in lateral patellar tilt following TT lateralisation, which was not restored following MPFL reconstruction ($P<0.05$).

MPFL state had a significant effect on patellar tilt following the application of the 10N lateral force to the patella ($P<0.001$). The effect of the load tended to increase following MPFL transection and then reduce following MPFL reconstruction (Figure 8.12 and Table 8.4). A significant interaction between MPFL state and TT position was also present, with TT lateralisation and MPFL transection resulting in a marked increase in lateral patellar tilt. Knee flexion angle also had a significant effect on patellar tilt, with the effect of both TT positioning and MPFL state tending to reduce in deeper flexion as the patella engaged with the trochlear groove ($P=0.007$) (Figure 8.11).

Post hoc testing identified significant effects of MPFL transection on patellar tilt when the knee was in full extension and the TT positioned 5mm, 10mm and 15mm laterally, as highlighted in Figure 8.12. No significant post hoc effects were identified as a result of TT medialisation for any of the MPFL states.

Table 8.4 Change in patellar lateral tilt (°) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion. Shown for the TT positioned in its anatomical, 5mm, 10mm and 15mm lateralised positions. Results shown through knee flexion range (mean and SD, n=8).

Flexion Angle (°)	MPFL INTACT							
	TT Anatomical (°)		TT 5mm Lateral (°)		TT 10mm Lateral (°)		TT 15mm Lateral (°)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.18	0.25	2.44	1.67	4.49	3.67	7.20	4.95
10	0.26	0.31	1.66	2.46	3.18	4.17	6.20	4.96
20	0.11	0.19	1.51	1.89	3.09	3.27	5.46	5.67
30	0.10	0.23	0.84	1.09	2.35	3.24	4.03	5.45
60	0.11	0.11	0.39	0.51	2.08	3.30	2.52	3.58
90	0.07	0.27	0.41	1.03	1.80	2.97	3.25	3.62
Flexion Angle (°)	**MPFL TRANSECTED**							
	TT Anatomical (°)		TT 5mm Lateral (°)		TT 10mm Lateral (°)		TT 15mm Lateral (°)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	2.58	0.98	3.81	1.60	5.88	2.77	9.1	4.4
10	1.63	1.02	2.92	2.27	5.08	3.60	8.0	4.9
20	1.83	1.11	3.65	1.40	4.76	3.22	7.1	5.5
30	0.64	0.75	2.06	1.80	3.48	3.08	5.2	5.4
60	0.39	0.51	1.14	1.69	2.88	3.29	3.5	3.6
90	0.62	0.60	0.82	1.36	2.32	3.78	4.5	3.4
Flexion Angle (°)	**MPFL RECONSTRUCTED**							
	TT Anatomical (°)		TT 5mm Lateral (°)		TT 10mm Lateral (°)		TT 15mm Lateral (°)	
	MEAN	SD	MEAN	SD	MEAN	SD	MEAN	SD
0	0.49	0.82	1.51	1.34	2.97	2.24	4.71	4.93
10	0.12	0.59	1.01	1.76	2.25	2.39	4.05	4.74
20	0.09	0.63	1.24	1.52	2.69	2.35	4.20	5.18
30	-0.04	0.46	0.47	0.86	1.68	1.96	3.39	5.18
60	0.05	0.33	0.37	0.54	1.75	3.07	2.59	3.29
90	0.12	0.49	0.35	0.82	1.62	2.86	2.40	2.91

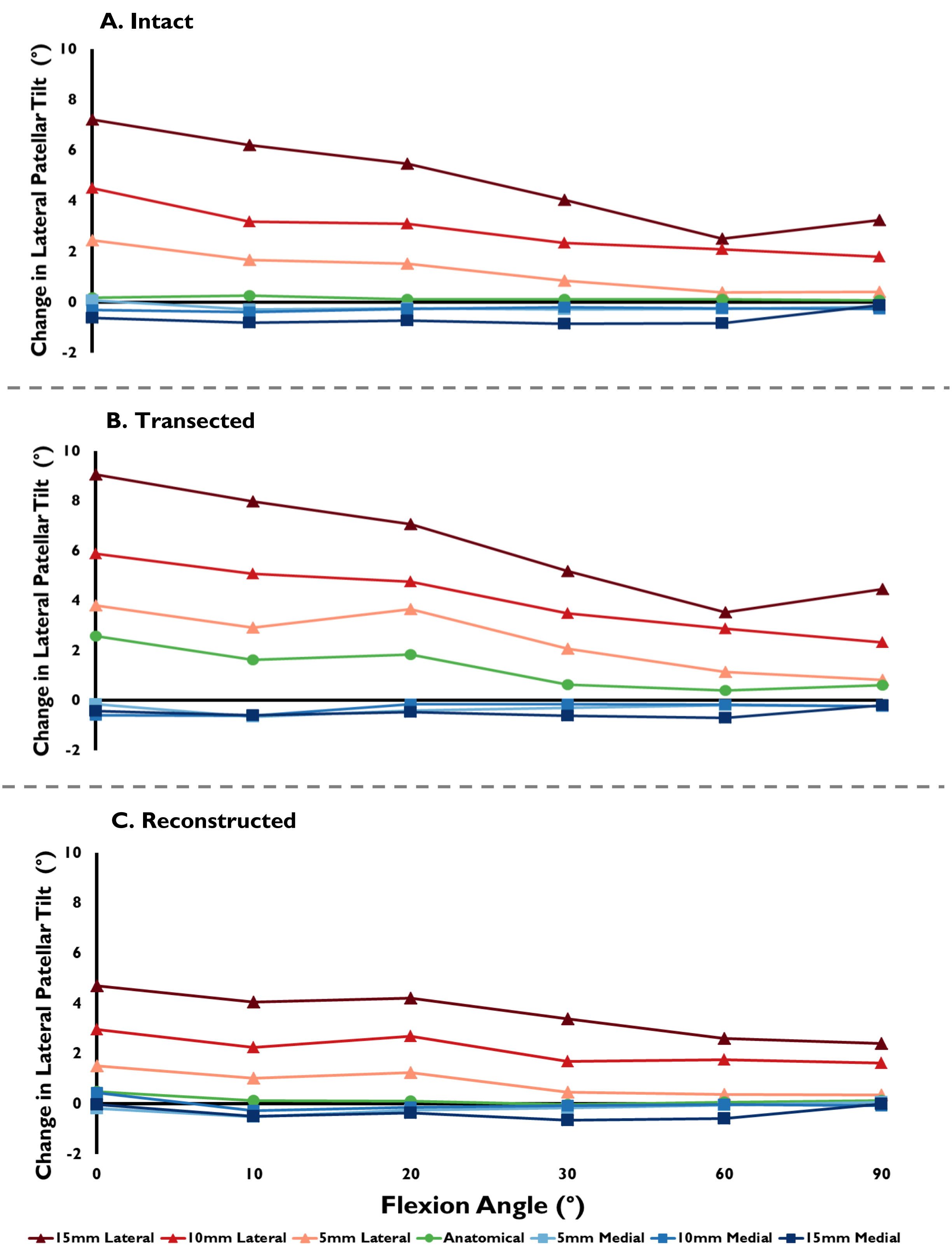

Figure 8.11 Change in patellar lateral tilt (°; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. Shown with the MPFL: **A**. intact, **B**. transected and **C**. reconstructed.

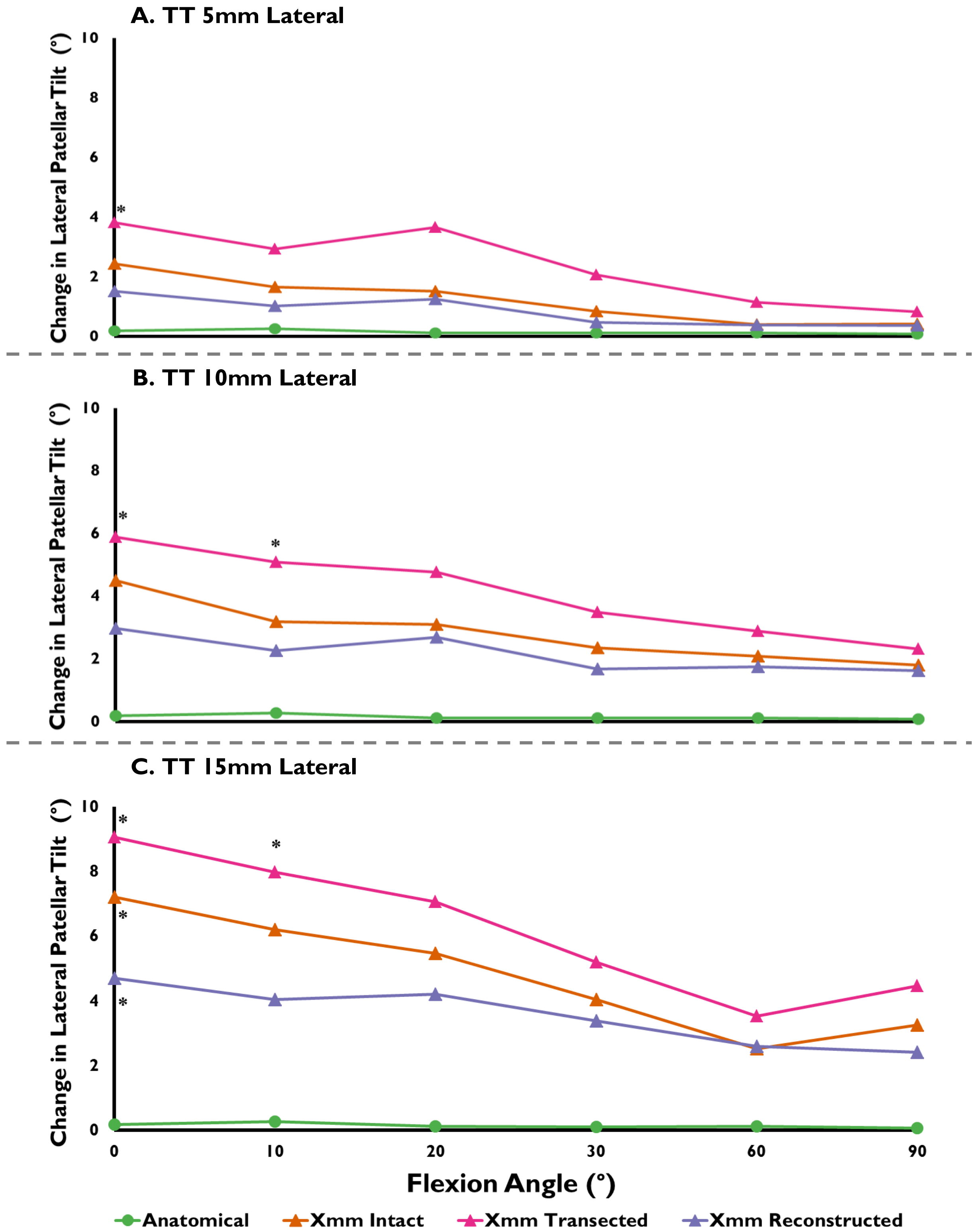

Figure 8.12 Change in patellar lateral tilt (°; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion. . Green line is identical throughout, representing the TT positioned in its anatomical position (with MPFL intact). Data then shown in each graph corresponding to the TT: **A**: 5mm lateralised, **B**: 10mm lateralised and **C**: 15mm lateralised. Orange line = MPFL intact, pink line = MPFL transected, purple line = MPFL reconstructed.

8.7.2.3 Mean Medial Contact Pressures

A significant effect of MPFL status was identified on mean medial contact pressures following the application of a 10N lateral load to the patella (P<0.001). With the TT in its anatomical position, the knee in 10º flexion and the 10N lateral load applied to the patella, the mean medial patellofemoral joint contact pressure was 0.55MPa. This reduced to 0.4MPa following MPFL transection and was restored to 0.52MPa following MPFL reconstruction. Similarly with the TT 5mm lateralised and 10N load applied to the patella at 10º knee flexion, the mean mean medial contact pressure was 0.37MPa. This reduced to 0.22MPa following transection of the MPFL before increasing to 0.41MPa after reconstruction. This trend was identified consistently throughout testing (Figure 8.13). The pressure change was significantly influenced by knee flexion, with the effect of the 10N lateral load application typically reducing as the knee was flexed further (P<0.001). There was a significant interaction of MPFL state and TT position, with the change in mean pressure generally increasing as the TT was medialised or lateralised further (P<0.001).

Significant post hoc tests are highlighted in Figure 8.13. MPFL transection resulted in significant reductions in mean medial contact pressures throughout early knee flexion, following MPFL transection (all; P<0.05). Significant differences in pressures remained following MPFL reconstruction with the TT lateralised 15mm at 60°, but were otherwise not found to be significantly different to the corresponding TT in its anatomical position with the MPFL intact at the corresponding flexion angle.

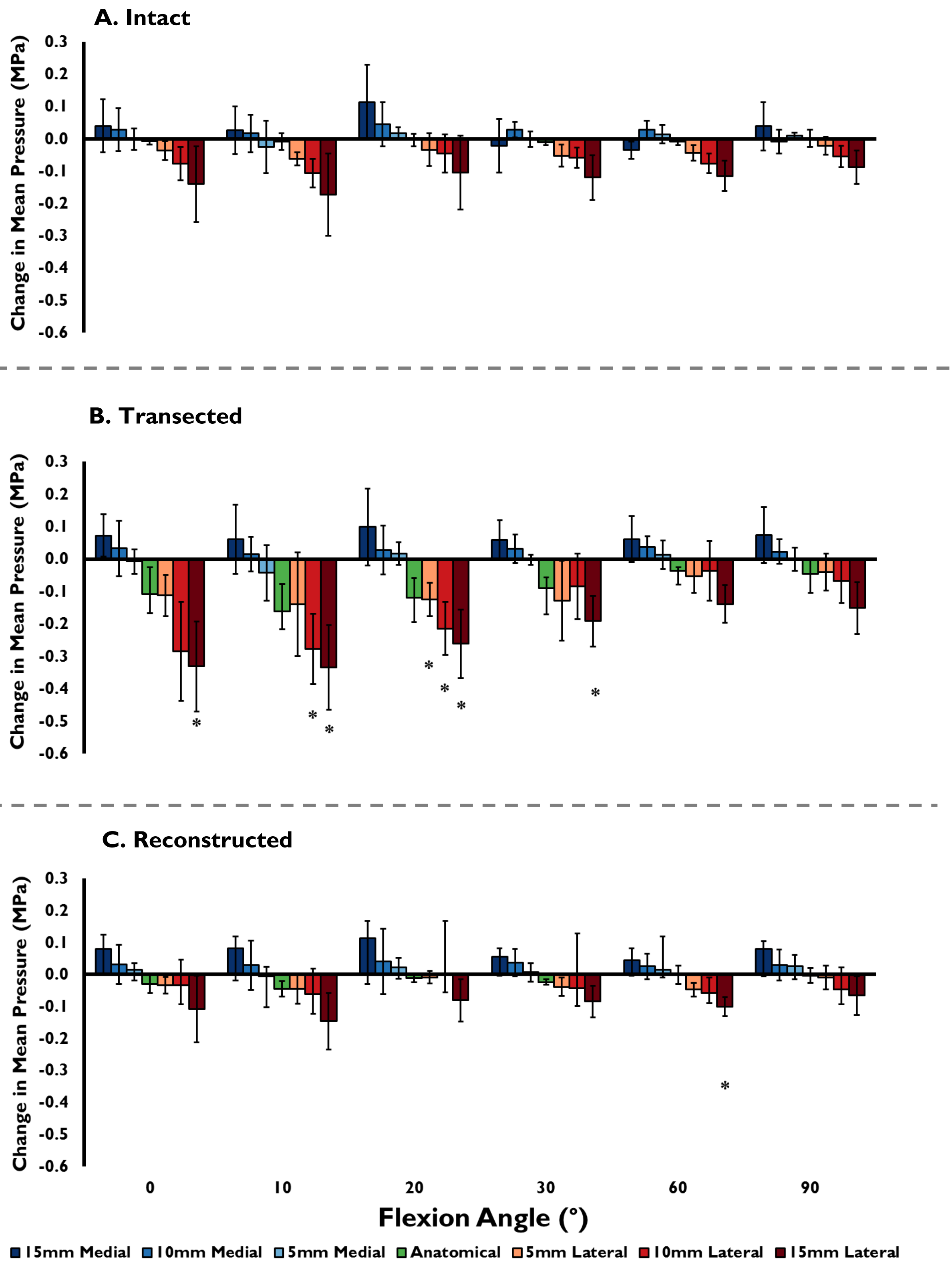

Figure 8.13 Change in mean medial patellofemoral joint contact pressures (MPa; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0° to 90° knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. Showing results for the intact knee, **B**. showing results for the knee with MPFL transected and **C**. showing results for the knee with the MPFL reconstructed. **P*<0.05.

8.7.2.4 Mean Lateral Contact Pressures

Mean lateral pressure data did not demonstrate clear trends of patellofemoral joint behaviour following the application of the 10N load to the patella (Figure 8.14). There was no significant effect found of MPFL status on mean lateral contact pressure changes from the intact state with the TT in its anatomical position following the application of a 10N lateral load to the patella (P=0.149). Furthermore no significant interaction of TT position and MPFL results was identified (P=0.128). However flexion angle did have a significant effect on the mean lateral pressures, with these tending to be much greater in early flexion than deeper flexion (P=0.009).

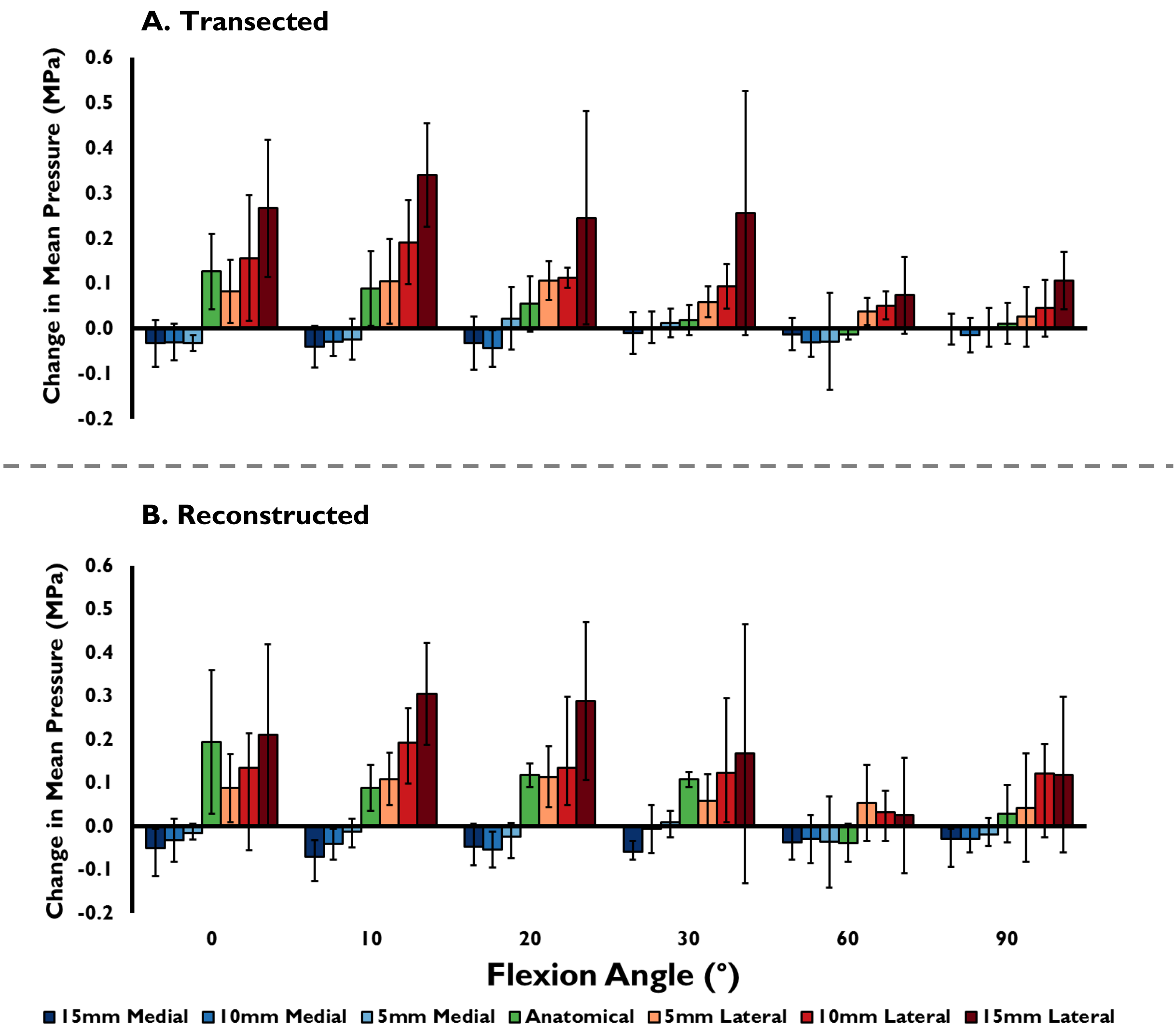

Figure 8.14 Change in mean lateral patellofemoral joint contact pressures (MPa; mean ± SD, n = 8) caused by the combination of altering TT position and the application of a 10N lateral displacing force to the patella from 0˚ to 90˚ knee flexion with the TT positioned: 15mm, 10mm, 5mm medialised, anatomical and 5mm, 10mm and 15mm lateralised. **A**. Showing results for the knee with the MPFL transected and **B**. showing results for the knee with the MPFL reconstructed.

8.7.2.5 Peak Patellofemoral Contact Pressures

No significant overall effect was identified on the peak lateral patellofemoral joint contact pressures as a result of MPFL status (P=0.139) following the application of the 10N lateral load to the patella. There was also no interaction identified between MPFL status and TT position (P=0.228).

The application of a 10N load to the patella did not cause any significant effect on peak medial contact pressures of MPFL condition; intact, transected or reconstructed (P=0.394). There was also no interaction identified between the TT position and MPFL status (P=0.443).

8.8 Discussion

MPFL transection resulted in significantly elevated lateral patellar tilt, translation, mean lateral contact pressures and reduced mean medial contact pressures during early knee flexion (all: P<0.05). These effects were significantly increased with progressive TT lateralisation (all: P<0.05). MPFL reconstruction restored patellar translation and mean lateral contact pressures to the intact state when the TT was in its anatomical, 5mm and 10mm lateralised positions. However these were not restored when the TT was 15mm lateralised. Patellar tilt and mean medial contact pressures were restored following 5mm lateralisation (P>0.05) but not after 10mm or 15mm TT lateralisation (P<0.05). No significant effect was found on peak medial or lateral contact pressures as a result of MPFL reconstruction or transection in combination with TT medialisation or lateralisation (all: P>0.05). There was a reduction in patellar stability, identified as increased lateral patellar tilt and translation and reduced medial contact pressures, as a combined result of MPFL transection and TT lateralisation (P<0.05). Mean medial pressures and patellar tilt were restored when the MPFL was reconstructed and the TT was in its anatomical, 5mm or 10mm lateralised positions (all P > 0.05), but not when in its 15mm lateralised position (P<0.05). Patellar translation was restored following MPFL reconstruction with the TT in all positions (P>0.05).

There are a number of limitations relating to this specific study which it is important to acknowledge. Firstly the tensioning protocol for the present study meant the knees were positioned at 30º flexion with 2N tension applied to the gracilis graft. This was based on prior findings in chapter 5 identifying the optimal MPFL reconstruction method (Stephen et al., 2013b). Two Newton was identified as the tension necessary to hold the MPFL graft taut, but not stretched, during pilot testing. However clinically if the TT was excessively lateralised, the surgeon may be inclined to apply greater tension to the MPFL graft to help to medialise the patella and centre its tracking in the trochlear groove. This experiment has not investigated the effect this would have had on outcome measures. It could be hypothesised based on findings in Chapter 5 that this would result in increased medial patellar tilt and medial patellofemoral joint contact pressures. However a further study would be required to establish if this is the case in the presence of a lateralised TT. The use of a strain transducer was considered in the present investigation to enable this to be investigated (Fleming et al., 1993). However following cadaveric behaviour observed in the prior studies, concerns were raised that inflicting higher tensions to the graft *in vitro* could result in failure of the bony attachment or of the graft itself. Furthermore the study protocol already necessitated a large time spent data gathering and the addition of further variables would have significantly increased the testing time which could invalidate results due to soft tissue failure. Graft tensioning in combination with TT position is therefore suggested as a factor which merits further examination and evaluation in future studies.

The challenge and lack of consensus in determining the optimal treatment pathway for patients following patellar dislocation is well documented, and is recognised to be more complex in cases of abnormal lower limb alignment such as elevated TT-TG or trochlear dysplasia (Ostermeier et al., 2007a; Stefancin and Parker, 2007; Steiner et al., 2006). Commonly clinical decision making in these cases can come down to surgical experience / preference alone with a lack of rigorous evidence available to influence the clinical reasoning process (Ostermeier et al., 2007a). Chapter 5 summarised published successful clinical outcomes and presented *in vitro* results which found MPFL reconstruction to restore patellar kinematics and contact mechanics following MPFL transection. However little prior research has examined the threshold at which MPFL reconstruction can restore patellar kinematics and patellar contact mechanics in patients with abnormal anatomy. A recent clinical case series summarised that MPFL reconstruction alone provided good outcomes at 5 year follow up in a population of patients with trochlear dysplasia, suggesting soft tissue reconstructive techniques have the potential to compensate for bony pathology. Given the mean TT-TG distance in the knees used in the present study was 10.4mm, findings suggest that MPFL reconstruction alone would be sufficient to restore mean patellofemoral joint contact pressures and patellar tilt and translation in cases of TT-TG distances less than 15mm.

Study findings support the guidelines reported in the widely cited paper by Dejour et al (1994), which suggests TT transfer surgery as appropriate for patients with TT-TG distances of 20mm or greater. Furthermore a clinical trial reported successful post-operative outcomes in a population of patients following patellar dislocation, with MPFL reconstruction undertaken on patients with TT-TG distances under 15mm, whilst those with TT-TG distances over 15mm were treated with TT transfer surgery in combination with MPFL reconstruction (Schöttle et al., 2005). Ellera Gomes et al (2004) reported optimal outcomes following MPFL reconstruction were obtained from those patients with normal Q-angle measurements, suggesting a role for TT transfer in a select population of patellar dislocation patients. Prior work has highlighted the controversy surrounding the use of TT transfer surgery in the treatment of patellar instability (Chapter 7). Given that most of the predisposing factors for patellar dislocation are static factors (injury to the MPFL or bony geometry), it is recognised that the MPFL plays a vital role regardless of the presence of malalignment. Present study findings certainly question the suggestion that TT transfer should be conducted in patients with TT-TG distances above 10mm as previously reported (Tecklenburg et al., 2010). However further clinical studies are recommended to confirm this.

This is thought to be the first study to examine progressive TT lateralisation and the effect of MPFL transection and reconstruction. It is clear on review of clinical literature that there remains a lack of consensus relating to the threshold of TT-TG distance up to which MPFL alone sufficiently restores patellar stability, kinematics and contact mechanics following patellar dislocation. Transection of the MPFL resulted in significant increases in lateral patellar tilt, translation and mean lateral patellofemoral joint contact pressures and a significant reduction in mean medial joint contact pressures. These effects were significantly increased as a consequence of progressive TT lateralisation. Reconstruction of the MPFL restored patellar kinematics, stability and patellofemoral joint mechanics in the intact knee and that with the TT lateralised by 5mm (to approximately 15mm). However, reconstruction failed to restore joint mechanics and kinematics when the TT was lateralised by 10mm or 15mm (TT-TG over 20mm). These findings provide evidence to suggest that in patients suffering patellar dislocation, with TT-TG distances of up to 15mm, that MPFL reconstruction alone should restore patellar stability, joint contact mechanics and patellar kinematics, and thus alleviate clinical symptoms. Meanwhile those patients with TT-TG distances in excess of 15mm may benefit from an additional TT procedure to ensure post-operative restoration of joint contact mechanics and patellar kinematics. Clinical studies may now be used to examine how these results translate into functional outcomes of patients.

8.9 Conclusion / Key Findings

Combined MPFL transection and TT lateralisation resulted in:

1. Elevated lateral patellofemoral joint contact pressures.
2. Reduced medial patellofemoral joint contact pressures.
3. Increased lateral patellar tracking and tilt.
4. Reduced patellar stability.

Reconstruction of the MPFL restored joint kinematics, contact mechanics and patellar stability when the TT was 5mm lateralised (TT-TG distance of approximately 15mm), but failed to do so when the TT was 10mm or 15mm lateralised (approximate TT-TG distance of 20-25mm). Over-medialisation of the TT to 15mm in combination with MPFL reconstruction resulted in significant increases in medial patellar translation and mean medial contact pressures.

CHAPTER 9

CONCLUSIONS and FUTURE WORK

9.1 Summary

Incompetence of the MPFL has been recognised as an integral factor in patellofemoral joint instability. This thesis has provided an in depth examination of its structure, considering its anatomy, biomechanical behaviour and reconstruction procedures. Key findings from each section are highlighted.

9.1.1 MPFL femoral attachment and ligament length change patterns

Prior literature showed that the natural behaviour of the MPFL was poorly understood and crucially the correct landmark for anatomical reconstruction was poorly defined. Chapter three set out to determine the length change pattern of the native MPFL and the effect of non-anatomical femoral and differing patellar attachments. This was undertaken using a suture attached to an LVDT. A secondary aim of this section of work was to recommend a reproducible femoral attachment site for undertaking anatomical MPFL reconstruction.

The native MPFL was found to be close to isometric with a mean maximal length change pattern of 2.1mm from 0°-110° knee flexion. Non-anatomical points were found to cause a significant loss of isometry. The proximal femoral attachment resulted in up to a 6.4mm mean lengthening and the distal attachment up to a 9.4mm mean shortening through 0°-110° of knee flexion, resulting in significantly non-isometric grafts ($P<0.05$). Malpositioning the femoral tunnel in the proximal-distal axis had much larger effects than moving it in the A-P axis, while moving the patellar attachment had a smaller effect. The femoral attachment point for MPFL reconstruction, taking the A-P femoral condyle diameter to be 100%, was identified 40% from the posterior, 50% from the distal, and 60% from the anterior border of the medial femoral condyle (Figure 9.1). The key findings of this chapter allow the surgeon to radiographically preoperatively validate, through the use of reliable anatomical landmarks and via the 40-50-60% rule (if the condyle is of normal shape), the most isometric tunnel placement for MPFL reconstruction.

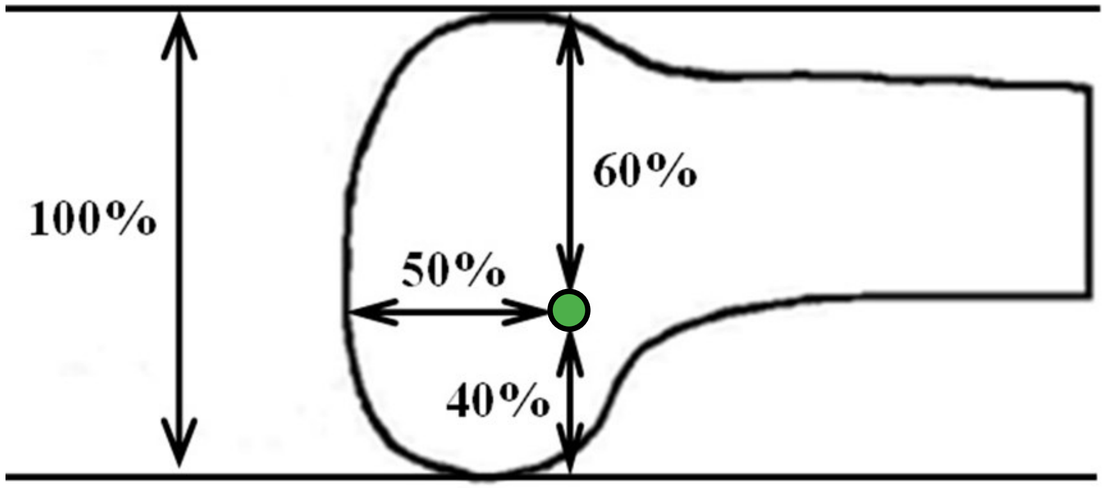

Figure 9.1 The MPFL attachment is defined in relation to the size of the medial femoral condyle: if A-P size was 100%, then the MPFL attachment is: 40% from the posterior, 50% from the distal and 60% from the anterior outline.

9.1.2 MPFL Transection

Rupture or injury of the MPFL is reported to occur during every patellar dislocation (Sillanpää et al., 2009b; Weber-Spickschen et al., 2011). Therefore it was important to determine the effect of MPFL rupture on patellofemoral joint contact mechanics and kinematics. Chapter four was conducted to answer this research question. The LVDT and suture, pressure sensitive film between the patella and trochlea and a set of optical trackers attached to the patella, femur and tibia were used to examine patellar kinematics and contact mechanics during the experiment.

Measurements were recorded from 0°-90° knee flexion with the MPFL intact and were repeated following MPFL transection. A significant increase in the distance between the patellar and femoral MPFL attachment points was noted following transection as measured by the LVDT ($P<0.05$). MPFL transection resulted in significantly increased lateral translation and lateral tilt of the patella in early knee flexion ($P<0.05$). Peak and mean medial patellofemoral joint contact pressures were significantly reduced and peak lateral contact pressures significantly elevated in early knee flexion following MPFL transection ($P<0.05$). MPFL transection resulted in significant alterations to patellofemoral joint tracking and contact pressures, which may cause alterations to articular cartilage health in the long term. The findings add to previous literature providing a rationale for MPFL reconstructive surgery.

9.1.3 MPFL Reconstruction

Based on findings from the transection study it was logical that the next step would be to determine if there was an optimal method of undertaking MPFL reconstruction. Cases of poor clinical outcome following surgery were evident on review of the literature. Commonly these were as a consequence of incorrect femoral tunnel position and inappropriate graft tensioning during surgery. This study therefore set out to investigate three variables: femoral tunnel position, graft tension and fixation angle for graft tension to be applied. Data on patellar kinematics, measured by optical tracking, and patellofemoral contact pressures, measured by pressure-sensitive films, were used as outcome measures.

The MPFL was transected and then reconstructed using a double strand gracilis tendon graft. For a graft tensioned to 2N, anatomically placed MPFL reconstruction restored intact medial and lateral joint contact pressures and patellar tracking (P>0.05), but femoral tunnels positioned proximal or distal to the anatomical origin resulted in significant increases in peak and mean medial pressures and medial patellar tilt during knee flexion or extension, respectively (P<0.05). Grafts tensioned with 10N or 30N also caused significant increases in medial pressure and tilt (P<0.05). Graft fixation with 2N tension at 30° or 60° knee flexion restored all measures to intact values (P>0.05), but fixation at 0° caused significant increases (P <0.05) in medial joint contact pressures compared to intact knees.

Graft over-tensioning, or femoral tunnels positioned too proximal or distal, caused significantly elevated medial joint contact pressures and increased medial patellar tilting. The importance of correct femoral tunnel position and graft tensioning in restoring normal patellofemoral joint kinematics and articular cartilage contact stresses was clearly evidenced. The chapter presents a protocol for graft tensioning which restored normal patellar tracking and contact conditions.

9.1.4 The Effect of Quadriceps Weakness

An alternative treatment used in the primary management of patients following patellar dislocation is physiotherapy (Sillanpää and Mäenpää, 2012). This commonly focuses on strengthening exercises for the quadriceps musculature. Specifically, weakness in the Vastus Medialis Muscles (VMM) has been associated with the development of patellofemoral joint pain and instability. This section of the thesis therefore investigated the effect of a reduction in VMO and VML muscle tension on patellofemoral joint kinematics, contact mechanics and stability.

Measurements were repeated for three conditions: 1. With knees physiologically loaded; 2. With the VMO muscle unloaded and 3. With the VMO and VML unloaded. Measurements were also repeated for the three conditions with a 10N lateral displacing force applied to the patella. Reduction of VMM tension resulted in significant increases in lateral patellar tilt, translation, elevated lateral and reduced medial joint contact pressures and reduced patellar stability (all: $P<0.05$). The results identified comparable effects of VMM tension reduction and MPFL sectioning (Chapter 4), on altering patellar tracking and contact mechanics. These findings add support to the role of quadriceps strengthening programmes for first time patellar dislocators and patellofemoral pain patients with apparently normal joint anatomy.

9.1.5 Tibial Tuberosity-Trochlear Groove Distance

The final section of the thesis set out to explore the effect of MPFL reconstruction in the presence of abnormal anatomy. Patellar dislocation commonly occurs in association with elevated TT-TG distances, and tibial tuberosity transfer is a successful surgical procedure used in the treatment of these patients. Limited literature exists examining the effect of tibial tuberosity lateralisation or progressive medialisation on patellar stability, kinematics and patellofemoral joint contact mechanics, therefore this was explored as the penultimate part of this thesis. Outcome measures were again patellofemoral joint contact pressures and patellar kinematics.

Lateralisation of the TT significantly elevated lateral joint contact pressures and lateral patellar tracking, and reduced patellar stability ($P<0.05$). There was a significant correlation between mean lateral pressure and TT-TG position ($r=0.810$, $P<0.001$) at 10° knee flexion. Tibial tubercle medialisation resulted in increased medial patellar tracking ($P<0.05$), however it also reduced lateral facet contact pressures ($P<0.05$), and did not cause elevated peak medial contact pressures or patellar tilting (Both: $P>0.05$). This data provides a rationale for undertaking tibial tuberosity transfer surgery in appropriate carefully selected clinical populations with elevated TT-TG distances and patellar instability.

9.1.6 MPFL and TT-TG

The last section of this thesis examined the effect and interaction of MPFL status (intact, transected or reconstructed) and TT position (15mm, 10mm and 5mm medialised, anatomical and 5mm, 10mm and

15mm lateralised) on patellar kinematics, joint mechanics and stability. The aim was to determine if there was a threshold beyond which MPFL reconstruction would no longer restore joint mechanics and kinematics and more aggressive surgery such as tibial tuberosity transfer surgery would be indicated.

MPFL transection resulted in significantly elevated lateral patellar tilt, translation, mean lateral contact pressures and reduced mean medial contact pressures during early knee flexion (all: $P<0.05$). These effects were significantly increased with progressive TT lateralisation (all: $P<0.05$). MPFL reconstruction restored patellar translation and mean lateral contact pressures to the intact state when the TT was in its anatomical, 5mm and 10mm lateralised positions. However these were not restored when the TT was 15mm lateralised. Patellar tilt and mean medial contact pressures were restored following 5mm lateralisation ($P>0.05$) but not after 10mm or 15mm TT lateralisation ($P<0.05$). No significant effect was found on peak medial or lateral contact pressures as a result of MPFL reconstruction or transection in combination with TT medialisation or lateralisation (all: $P>0.05$).

Considering the mean TT-TG in this study (10.2mm), findings suggest that in patients with TT-TG distances of up to 15mm patellofemoral joint kinematics and contact mechanics will be satisfactorily restored with MPFL reconstruction alone. However in those patients with TT-TG distances greater than 15mm, more aggressive procedures such as tibial tuberosity transfer may be indicated to ensure restoration of joint mechanics post-surgery. The question of whether additional procedures are also necessary in order to ensure a successful clinical outcome is beyond the scope of this work *in vitro*.

9.2 Limitations

There are inevitable limitations to this series of reported cadaveric experiments, many of which are inherent to all *in vitro* testing. Consideration and reasonable attempts were made to address and minimise the majority of these (as far as possible). Many of the limitations are discussed within their relevant chapters and specifically in the addendum sections at the conclusion of some chapters. More general limitations pertaining to the cadaveric experiments and outcome measures are discussed in further detail in the following sections.

9.2.1 Specimens

The knees used for these studies were elderly. For the first set of experiments (Chapters 3-5) the mean age of the specimens was 73.5 years, with the second section of the thesis using knees with a mean age of 64 years (Chapters 6-8). Typically it has been highlighted that patellofemoral joint problems will manifest in younger populations (Chapter 2). Additionally, the specimens were all confirmed with imaging to have normal patellofemoral joint anatomy present. Unfortunately it is not possible to know directly how the results from tests on normal, elderly knees extrapolate to those younger patients with abnormal patellofemoral joint anatomy, such as trochlear dysplasia or patellar alta, who are commonly reported to suffer patellar dislocation (Fithian et al., 2004). Some studies have created pathology in normal knees to investigate the effects of these pathologies (Amis et al., 2008). However in the context of the current work it was determined this would be challenging to control and therefore testing was performed on 'normal' specimens where the limitations of the work could be clearly identified and defined.

9.2.2 Test Rig and Muscle Loading

A common practice during *in vitro* testing is determining muscle loading contributions based on the cross sectional area of the muscle in question. This makes the assumption that the ratio of these loads is constant throughout the cycle and does not take into account the different activation times of the muscles which occurs *in vivo* during functional movements (Ivanenko et al., 2004). The set up in the present study did not simulate weight-bearing activity, which has been suggested to enhance patellar stability, but was akin to an open chain knee extension movement against gravity. Care was taken to load the muscles in physiological directions previously reported from cross sectional area studies (Farahmand et al., 1998a) but the tensions were constant and lower than those estimated to occur *in vivo* (Cohen et al., 2001). In the absence of rigorous techniques to measure muscle activation *in vivo* this was determined an appropriate compromise. Although the forces in this experiment would likely be exceeded *in vivo*, the analysis performed in each of the chapters involved the comparison of one condition to a number of others. Therefore the nature of the changes found was unlikely to alter; although it is acknowledged that they may be larger *in vivo* (Müller et al., 2009).

9.2.3 Measurement Methods

The method used for detecting soft tissue length changes, the LVDT and attached suture, has been validated previously (Ghosh et al., 2009) and therefore allowed accurate measurements during motion of the loaded knee. It was was confirmed by micrometer to be accurate to ± 0.01mm prior to testing commencement. Test re-test measurements taken during pilot testing highlighted its very good reliability (ICC>0.89) (Chapter 3). Therefore the results can be presented with assurance. The optical tracking system was found to have very good reliability during pilot testing (Chapter 4) and was reported in the literature to have an overall volume root mean square (RMS) distance error of 0.35mm for a single marker (Wiles et al., 2004), with the traxtal tracker used for measurement of patellar tracking having a reported accuracy of 0.04mm with a precision of 0.03mm (Merican and Amis, 2009). This meant the kinematic data could also be presented with confidence.

The Tekscan pressure sensors have numerous issues, which are discussed in detail in section 4.9.1. The system, however, was the best available to measure patellofemoral contact pressures and offered distinct advantages over other systems. The main frustration with the use of Tekscan was the lack of a rigorous, reasoned method by which to condition, equilibrate and calibrate the sensors. Extensive pilot testing was undertaken during the experimental stage of Chapter 4 to investigate these variables prior to commencing testing. However it was disappointing that none of the Tekscan advice manual, the Tekscan representative or the sensor manufacturers based in the USA were able to provide standardised, detailed information on the optimal method for sensor preparation.

The Tekscan has additional problems of not conforming to the shape of the patellofemoral joint, meaning that creasing can occur as the knee is flexed. Again pilot testing meant these effects could be markedly minimised with practice and increased skill at positioning the sensor in the joint. Temperature and liquid exposure are both also known to affect the Tekscan sensors (Bachus et al., 2006; Jansson et al., 2013). The laboratory temperature was controlled with a thermometer and fan as rigorously as possible, however it did of course at times change during the day and from day to day. Liquid exposure of the sensors was only to the minimal joint fluid present in the patellofemoral joint capsule, which was fairly constant across all knees. However these factors could have impacted on study findings.

The Tekscan sensors were inserted using a method intended to minimise disruption to the retinacula but this required opening the proximal patellofemoral joint capsule and lifting / dissection of VI from the femur. A proximal approach to access the patellofemoral joint was the least invasive method to permit insertion of the sensor to the patellofemoral joint whilst leaving the local soft tissues intact. It is unknown how this affected patellar contact pressures, however this was constant throughout testing, meaning that the data sets were comparable. The test-retest repeatability of the Tekscan system meant that small pressure changes could not be identified (Chapter 4). However, the significant effects of interventions in the series of studies reported generally provided pressure changes larger than this and so these could be identified reliably.

Lastly, stability testing on the patella, undertaken in the final three experimental chapters, could have been performed using a number of different methods as highlighted in the addendum of Chapter 6. A higher load may have given more significant and representative results; however the use of the 10N load was based on prior literature and concerns over minimising any changes caused to the local soft tissues during protocols that required many tests.

9.3 Future Work

9.3.1 Cadaveric Studies

The nature of cadaveric studies provides a key advantage over *in vivo* testing: the ability to use direct, invasive testing methods. Measurement measures can therefore often be more rigorous, valid and reliable than those recorded directly on live participants. Rightly, there are also less ethical issues with the use of specimens than those encountered *in vivo*. These factors make cadaveric testing an excellent tool for the researcher and clinician. As discussed throughout the thesis, a number of future studies would be beneficial to investigate some further areas not addressed by the present study series. These come under two main headings and are discussed below.

9.3.1.1 MPFL Reconstruction

Section 5.10.1 addresses a number of factors pertaining to MPFL reconstruction which would benefit from further investigation through cadaveric study. These include, but are not limited to:

- The type of graft used during reconstruction.
- The technique for MPFL reconstruction: specifically comparing the quadriceps swing over method with the more traditional direct patellar and femoral attachment methods.

- The effect of anteriorly or posteriorly placed femoral tunnels. It is hypothesised that these would have less effect than proximal-distal malpositioning but the effect of this on patellofemoral contact mechanics or patellar kinematics has not been investigated.
- Graft tensioning applied at the patellar attachment versus that applied at the femoral attachment. Is there any difference between the two methods?
- The creation of patella alta and/or trochlear dysplasia in sets of specimens and the combined examination of this anatomy with MPFL reconstruction. This would be the same as the TT-TG portion of this thesis in Chapters 7 and 8. The effect of these abnormal bony geometries on the 40-50-60% rule also merits investigation.
- Future work could examine the effect of TT anteriorisation, which could be predicted to reduce patellofemoral contact force across the range of flexion / extension. It is not certain that that would reduce contact pressure because of upsetting joint congruence.
- Further, the effect of TT anteromedialisation could be investigated.
- The same test set up could be used to examine tightness and release work around the lateral retinaculum. This has not been investigated with the use of joint contact pressure as an outcome.

9.3.1.2 Test Rig

This study replicated an open-chain knee extension movement. A number of different rigs were appropriate to use in this series and these are examined in Chapter 3. Future work should look at devising a test rig to investigate full lower limb motion, including the hip and ankle joints, since these are known to influence patellofemoral biomechanics (Chapter 2). This is likely to represent an extensive body of research to determine the cross sectional area and line of action of the hip and ankle muscles and as such was unfortunately beyond the scope of the present study.

9.3.2 Computer Modelling

Undoubtedly the future of orthopaedic surgery is moving towards greater pre-operative planning with the use of computer simulation models likely to play a key role in this in the future. Hopefully the 40-

50-60% rule determined in this work will be possible to apply clinically via the use of such methods. Computer models have the potential to enable rigorous pre-surgical planning for subject specific cases and also have a role in determining those patients for whom surgery will be successful and those patients that may be better managed conservatively. There is a trend in the literature of computer modelling studies predicting surgical outcomes and these undoubtedly have a critical role in driving forward clinical success.

9.3.3 Clinical Trials

This thesis is composed of *in vitro* work. In order to translate the results of cadaveric studies to the clinical setting, *in vivo* studies are vital. Many of the findings reported in this thesis need to be investigated in the clinical setting. The examination of clinical outcomes in patients following MPFL reconstruction over longer time frames is certainly indicated. This will also help to establish whether or not there are any adverse effects as a consequence of femoral tunnel malpositioning or graft overtensioning. Certainly some smaller series of cases present in the literature already support these findings (Bollier et al., 2011; Thaunat and Erasmus, 2009). Ideally however future studies should also include objective outcome measures such as kinematic data or force plate analysis alongside the subjective outcomes currently reported on.

The role of MPFL reconstruction in joint preservation is also relevant to investigate. The lateral compartment of the patellofemoral joint is known to suffer the most with degenerative joint changes so intervention in the form of MPFL reconstruction following patellar dislocation may be appropriate to preserve the joint in the longer term and prevent osteoarthritis in later life (Chapter 4). A large scale, prospective randomised controlled trial into the management of patients post patellar dislocation to compare surgical and conservative physiotherapy management of these patients is appropriate. Long term follow up of patients is critical to ascertain any adverse effects of failing to adequately address the injured medial structures at the time of dislocation.

There is also a need for long term studies to be conducted on populations of patients who have undergone TT transfer procedures. These could investigate patient reported outcome scores in addition to radiographic outcomes to look at joint degeneration, and the role of correction of TT position. In addition it would be of interest to perform gait laboratory assessments on these patients both directly post-operatively and up to several years post-operatively to compare lower limb kinematic before and following the procedure.

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APPENDICIES

Appendix A: Tekscan Analysis

A.1 Background

A MATLAB script was developed and written to analyse the raw Tekscan data and obtain peak and mean pressures on the lateral and medial facets of the patellofemoral joint. This was undertaken by a current PhD student and a summer student (Punyawan Lumpaopong and Robin Fox) from the biomechanics group. It was manually checked for accuracy several times prior to use.

Tekscan raw data was saved as movie files (.fxs) at the time of data recording. This was then processed following the steps below and analysed using the MATLAB script. The script was designed to analyse data sets obtained from the experiment, in which the knee was flexed from 0º to 90°.

Steps for analysing Tekscan data:

1. In the *I-Scan*™ Pressure Measurement System:
    a. Load a Tekscan movie file (.fxs).
    b. Load the calibration and equilibration files (Chapter 4).
    c. Save the file as an ASCII file (.asm).

2. In the MATLAB shell
    a. Move the ASCII file(s) of the data set into the MATLAB current folder.
    b. Run the MATLAB script.
    c. Enter the distance between the pointer and the patellar ridge in order to separate the medial and lateral facets. Enter a positive value for right knees and negative value for the left knees.
    d. Enter the Tekscan row and column numbers of the pointer.

A.2 Tekscan Script

```
%Punyawan Lumpaopong;Robin Fox
%23.11.11 Version 1.0
%Filename:PamTekscan231111.m
%-----------------------------------------------------------------------
%DESCRIPTION:
%  This script extracts results and data from Tekscan film records of  patellofemoral
%   joint tests, having the following brief testing procedure.
%    1) A knee is set on the rig. Tekscan film is inserted into the joint cavity.
%    2) A pointer is imprinted on the film at about the middle of the patella to locate patellar position on the
%        Tekscan film.
%    3) Quadriceps muscles are loaded. The knee is flexed from 0 to 90 degrees at 10 degree
%         intervals.
%   Ten separate ".asm" files are recorded at each flexion angle and stored in the same folder.
%
%INPUTS:
%    1) The distance between the pointer and the patellar ridge (measured).
%    2) Pointer row and column of each record (read from I-Scan(R)software).
%To exit, enter the number '000' to the Pointer row or column input.
%
%OUTPUTS:
%  ORDER OF DATA PRINTED IN .XLS FILE (25 items):
   %----------------------------------------------------------------------
   % 1) Angle
   % 2) Pointer row
   % 3) Pointer column
   % 4) Ridge row
```

```
  % 5) Ridge column
  % 6) Contact area
  % 7) Max stress
  % 8) Max stress row
  % 9) Max stress column
  % 10) Max medial stress
  % 11) Max medial stress row
  % 12) Max medial stress column
  % 13) Max lateral stress
  % 14) Max lateral stress row
  % 15) Max lateral stress column
  % 16) Medial contact area
  % 17) Lateral contact area
  % 18) Medial-lateral distance of max medial stress from pointer
  % 19) Medial-lateral distance of max medial stress from ridge
  % 10) Proximal-distal distance of max medial stress
  % 21) Medial-lateral distance of max lateral stress from pointer
  % 22) Medial-lateral distance of max lateral stress from pointer
  % 23) Proximal-distal distance of max lateral stress
  % 24) Mean medial stress
  % 25) Mean lateral stress
  %---------------------------------------------------------------------
%INITIALISE WORKSPACE
clear
clc
%PREALLOCATE MEMORY
results=zeros(10,25);rawdata=cell(10,1);data=cell(10,1);modata=cell(10,1);
%SET UP CONSTANTS
angles=transpose(0:10:90);
```

```
sensel_width=1.3; %mm

sensel_area=1.69;%mm^2 area of one sensel

%ACQUIRE DATA & INPUTS

distance_pr=input('Distance between the pointer and the ridge (in mm)=');

d=round(distance_pr/sensel_width);

asmfnames=dir('*.asm');%acquire all .asm filenames

for k=1:10;%0 to 90 degrees at 10 degrees increment

  %% 1)ANGLE

  %% 2)POINTER ROW

  %% 3)POINTER COLUMN

  %% 4)RIDGE ROW

  %% 5)RIDGE COLUMN

  rawdata{k,1} =importdata(asmfnames(k,1).name);

  data=rawdata;%to keep raw data for checking

  pointer_r=input(['Pointer row (',num2str(angles(k,1)),')=']);

  if pointer_r==000;

    return

  end

  if pointer_r>44 || pointer_r<1

    disp('Invalid, must be [1:44], re-enter: ');

    pointer_r=input(['Pointer row (',num2str(angles(k,1)),') = ']);

  end

  pointer_c=input(['Pointer column (',num2str(angles(k,1)),')=']);

  if pointer_c==000;

    return

  end

  if pointer_c>44 || pointer_c<1

    disp('Invalid, must be [1:44], re-enter: ');

    pointer_c=input(['Pointer column (',num2str(angles(k,1)),') = ']);
```

```
end
ridge_r=pointer_r;
ridge_c=pointer_c+d;
results(k,1)=angles(k,1);%store results 1-5
results(k,2)=pointer_r;
results(k,3)=pointer_c;
results(k,4)=ridge_r;
results(k,5)=ridge_c;
%% 6)CONTACT AREA
% modify data
% 6.1)Set a specified area, outside of which all values should be zero, if
% not, they are set to zero
% Approximate area is between columns 10 and 40, and rows 13 and 33:
% i====ROWS, j====COLS for data.data(i,j)
for j=(1:10) % columns 1 to 10
   data{k,1}.data(:,j)=0;
end
for j=(40:44) % columns 40 to END
   data{k,1}.data(:,j)=0;
end
for i=(1:13) % rows 1 to 13
   data{k,1}.data(i,:)=0;
end
for i=(33:44) % rows 33 to END
   data{k,1}.data(i,:)=0;
end
% 6.2)Clean up values which are surrounded by zeroes
for i=(2:44-1)
   for j=(2:44-1)
      if data{k,1}.data(i+1,j)==0 && data{k,1}.data(i-1,j)==0 && data{k,1}.data(i,j+1)==0 &&
```

```
data{k,1}.data(i,j-1)==0
            data{k,1}.data(i,j)=0;
         end
      end
   end
   % 6.3)Clean up values that have '0's on three out of four sides
   oldmean=mean(mean(data{k,1}.data)); %find mean of each data
   newmean=1; %assign an arbitrary value to newmean
   while newmean~=oldmean
      oldmean=newmean; %re-define oldmean to keep going
      for i=(2:44-1)
         for j=(2:44-1)
            if data{k,1}.data(i+1,j)==0 && data{k,1}.data(i-1,j)==0 && data{k,1}.data(i,j+1)==0
               data{k,1}.data(i,j)=0;
            elseif data{k,1}.data(i+1,j)==0 && data{k,1}.data(i,j-1)==0 && data{k,1}.data(i,j+1)==0
               data{k,1}.data(i,j)=0;
            elseif data{k,1}.data(i+1,j)==0 && data{k,1}.data(i-1,j)==0 && data{k,1}.data(i,j-1)==0
               data{k,1}.data(i,j)=0;
            elseif data{k,1}.data(i-1,j)==0 && data{k,1}.data(i,j-1)==0 && data{k,1}.data(i,j+1)==0
               data{k,1}.data(i,j)=0;
            end
         end
      end
      newmean=mean(mean(data{k,1}.data));
   end
   % 6.4)store modified data to a new variable
   modata{k,1}.data=data{k,1}.data;
   % extract results
   % nnz function finds number of non-zero cells in data
```

```
num_cells=nnz(data{k,1}.data); %equivalent to number of contact cells

contact_area=num_cells*sensel_area;

results(k,6)=contact_area; % store result 6

%% 7)MAX STRESS

%% 8)MAX STRESS ROW

%% 9)MAX STRESS COLUMN

[max_v max_ind]=max(data{k,1}.data(:));

[r c]=ind2sub(size(data{k,1}.data), max_ind);

    % Find out how much difference there is between max and surrounding cells...

    % Surrounding values must be less than 10% difference

min_val_req=0.9*max_v; % the minimum value required to satisfy our conditions

    % if any of the 8 surrounding cells have a value below threshold, do not

    % average!

if data{k,1}.data(r+1,c)<min_val_req || data{k,1}.data(r,c+1)<min_val_req ||
data{k,1}.data(r+1,c+1)<min_val_req || ...

        data{k,1}.data(r-1,c+1)<min_val_req || data{k,1}.data(r-1,c)<min_val_req || data{k,1}.data(r-
1,c-1)<min_val_req ||...

        data{k,1}.data(r,c-1)<min_val_req || data{k,1}.data(r+1,c-1)<min_val_req

    max_actual=max_v;

else
max_actual=(data{k,1}.data(r+1,c)+data{k,1}.data(r,c+1)+data{k,1}.data(r+1,c+1)+data{k,1}.data(r
-1,c+1)+data{k,1}.data(r-1,c)+data{k,1}.data(r-1,c-1)...

        +data{k,1}.data(r,c-1)+data{k,1}.data(r+1,c-1)+max_v)/9;

end

results(k,7)=max_actual; % store results 7-9

results(k,8)=r;

results(k,9)=c;

%% 10)MAX MEDIAL STRESS

%% 11)MAX MEDIAL STRESS ROW

%% 12)MAX MEDIAL STRESS COLUMN

%% 13)MAX LATERAL STRESS
```

```
%% 14)MAX LATERAL STRESS ROW
%% 15)MAX LATERAL STRESS COLUMN
% separate medial and lateral data using ridge column
med_data=data{k,1}.data(:,(ridge_c+1:44));
lat_data=data{k,1}.data(:,(1:ridge_c));
% find maximum medial stress
[maxmed_v maxmed_i]=max(med_data(:));
[r_med c_med]=ind2sub(size(med_data), maxmed_i);
% find acutal position in array 'data' since medial is the 'right' side
% of the array
rdata_med=r_med; %rows aren't shifted
cdata_med=c_med+ridge_c; %counted from the ridge
% find maximum lateral stress
[maxlat_v maxlat_i]=max(lat_data(:));
[r_lat c_lat]=ind2sub(size(lat_data), maxlat_i);
%actual position in array 'data' doesn't change since lateral is 'left'
%side of array
rdata_lat=r_lat;
cdata_lat=c_lat;
results(k,10)=maxmed_v; % store results 10-15
results(k,11)=rdata_med;
results(k,12)=cdata_med;
results(k,13)=maxlat_v;
results(k,14)=rdata_lat;
results(k,15)=cdata_lat;
%% 16)MEDIAL CONTACT AREA
%% 17)LATERAL CONTACT AREA
%% 18)MEDIAL-LATERAL DISTANCE OF MAX MEDIAL STRESS FROM POINTER
%% 19)MEDIAL-LATERAL DISTANCE OF MAX MEDIAL STRESS FROM RIDGE
```

```
%% 20)PROXIMAL-DISTAL DISTANCE OF MAX MEDIAL STRESS
%% 21)MEDIAL-LATERAL DISTANCE OF MAX LATERAL STRESS FROM POINTER
%% 22)MEDIAL-LATERAL DISTANCE OF MAX LATERAL STRESS FROM RIDGE
%% 23)PROXIMAL-DISTAL DISTANCE OF MAX MEDIAL STRESS
%find medial and lateral contact areas
num_cells_med=nnz(med_data); % number of medial contact cells
num_cells_lat=nnz(lat_data); % number of lateral contact cells
med_area=sensel_area*num_cells_med;
lat_area=sensel_area*num_cells_lat;
results(k,16)=med_area; %store results 16-17
results(k,17)=lat_area;
%find distance to max medial and lateral stresses
mldist_lat_p=(cdata_lat-pointer_c-1)*sensel_width;%from pointer
mldist_lat_r=(cdata_lat-ridge_c-1)*sensel_width;%from ridge
mldist_med_p=(cdata_med-pointer_c-1)*sensel_width;
mldist_med_r=(cdata_med-ridge_c-1)*sensel_width;
pddist_lat=(r_lat-ridge_r-1)*sensel_width*(-1);
pddist_med=(r_med-ridge_r-1)*sensel_width*(-1);
results(k,18)=mldist_med_p; % store results 18-23
results(k,19)=mldist_med_r;
results(k,20)=pddist_med;
results(k,21)=mldist_lat_p;
results(k,22)=mldist_lat_r;
results(k,23)=pddist_lat;
%% 24)MEAN MEDIAL STRESS
%% 25)MEAN LATERAL STRESS
% group all non-zero elements in arrays to find mean
latnz=nonzeros(lat_data);
mednz=nonzeros(med_data);
```

```
  % Find mean of previous arrays (latnz and mednz) and output
  lat_mean_v=mean(latnz);
  med_mean_v=mean(mednz);
  results(k,24)=med_mean_v; % store results 24-25
  results(k,25)=lat_mean_v;
end
header={'1Angle' '2PRow' '3PCol' '4RRow' '5RCol' '6ContactArea' '7Max' '8MaxRow' ...
  '9MaxCol' '10MaxMed' '11MaxMedR' '12MaxMedC' '13MaxLat' '14MaxLatR' '15MaxLatC' ...
  '16CAMed' '17CALat' '18MedML_P' '19MedML_R' '20MedPD' '21LatML_P' '22LatML_R' ...
  '23LatPD' '24MeanMed' '25MeanLat'};

xlswrite('Tekresults',header,1,'A1');
xlswrite('Tekresults',results,1,'A2');
```

Appendix B: Kinematic Analysis

B.1 Background

A MATLAB script was developed, by a current PhD student (Punyawan Lumpaopong) in the biomechanics group, to analyse the motions of the patella and tibia relative to the femur and to illustrate the motions from the raw tracking data obtained from the optical tracking system.

The algorithm and script were verified and validated using a simple rigid body motion experiment before the script was fully developed to analyse patellofemoral joint kinematics. Patellar and tibial motions were calculated in relation to the reference frame of the femur.

B.1.1Usage

The proper formats of the data and correct filenames were required in order to run the script. The script clearly explains the required data preparation. The outputs, 3D motions of the patella and tibia were returned as MATLAB variables which were saved for later analysis.

B.2 The MATLAB script

```
%Punyawan Lumpaopong
%18/08/11 v1.0
%Biomechanics Group, Department of Mechanical Engineering,
%Imperial College London
%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%%
%SCRIPT NAME: PTIntermotion1.m
%
%DESCRIPTION:
%   This script calculates patellofemoral joint motion from optical tracking data (3D position:X,Y,Z,
%and orientation data:q0,q1,q2,q3-quaternion format) obtained from cadaveric knee motion study.
%The data are measured using optical tracking system (The Polaris(R),Northern Digital Inc).
%   The script only applies to cadaveric knee motion study which:
%   1) uses 'RIGHT KNEE',
%   2) the femur is fixed, and
%   3) tracking data is collected intermittently from 0 to 90 degrees at 10
%   degrees increment (data are saved in 10 separate folders).
%
%OUTPUTS:
%   Three output varibles will be saved in current Matlab folder.
%   1) Presults.mat
%   2) Tresults.mat
%   3) colname.mat
%   Variable Presults.mat and Tresults.mat are 10-by-6 matrix (10 positions
%and 6DOF motions). The motions of the patella and tibia are RELATIVE TO THE FEMUR!
```

```
%   File colname.mat shows column headers coresponded to columns in
%Presults.mat and Tresults.mat.
%
%COLLECTING OPTICAL TRACKING DATA:
%   Three optical trackers are attached to the femur, patella and tibia.
%At initial postion, normally at full extension (0 degrees), three anatomical
%landmarks of each bone are digitised using digitising probe and tracking
%data at this position are collected.
%   Recommended anatomical landmarks:
%   1) Femur: 1)Lateral epicondyle
%             2)Medial epicondyle
%             3)Proximal end of femoral intermedullary rod
%   2) Patella: 1)Lateral border
%               2) Medial border
%               3) distal border
%   3) Tibia:   1)Lateral condyle
%               2)Medial condyle
%               3)Distal end of tibial intermedullary rod
%   When the tibia is flexed to next positions, only tracking data are collected.
%
%DATA PREPARATION:
%   To analyse kinematic data using this script, raw data from above
%requires rearrangement into the following files and folder:
%   1)Anatomical landmark files
%     Rearrange raw data into 9 text files (.txt) for 9 anatomical landmarks
%     and name them (case sensitive) as follow:
%      1)Femur: AF1Lateral.txt,AF2Medial.txt,AF3End.txt
%      2)Patella: AP1Lateral.txt,AP2Medial.txt,AP3End.txt
%      3)Tibia: AT1Lateral.txt,AT2Medial.txt,AT3End.txt
%     When copying data to each file, simply take all 9 columns of related data
%     and delete all headers.
%   2)Tracking data files
%     Duplicate the files and simply rename the new files as the following example:
%     Examples: F0.txt is the data of the femur at 0 degrees.
%               P40.txt is the data of the patella at 40 degrees.
%               T90.txt is the data of the tibia at 90 degrees.
%   3)Create a new folder (any name). Move all files created above to the
%     folder.
%   4)Save the patellar tracking data at 0 degrees flexion or now P0.txt
%     as "int_PTracker.txt". The file name refers to the initial tracker
%     position of the "intact" knee.
%   **When analysing different experiment of the same knee while no new anatomical landmark
%   positions are not measured, instead of doing 3),user must copy the file "int_PTracker.txt" of the
%   intact knee to the new created folder. This will allow recalculation of the new anatomical
%   landmark position.
%
%SCRIPT USAGE:
% 1)Point "Matlab Current Folder" to the folder created above.
% 2)Run the script!
%
%CUSTOM FUNTIONS REQUIRED:
% 1)invhtm() %inverse of homogeneous matrix
% 2)cangle() %extracting cardan angle from transformation matrix
%
%TO COMPARE RESULTS ACROSS KNEES or DIFFERENT EXPERIMENTS OF A KNEE:
%  Another m-file "PFJgraphs.m" has been developed to do the job.
%-----------------------------------------------------------------------
%Initialisation
```

```
clear
clc
%1111111111111111111111111111111111111111111111111111111111111111111111
%                      IMPORTING KINEMATIC DATA
%1111111111111111111111111111111111111111111111111111111111111111111111
%import all anatomical landmark files from current folder
%import patellar and tibial tracker files from current folder
lmfiles=dir('A*.txt');
ffiles=dir('F*.txt');
pfiles=dir('P*.txt');
tfiles=dir('T*.txt');
int_PTracker=importdata('int_PTracker.txt');
alandmarks=cell(1,9); %preallocate cell array
ftrack=cell(1,10);ptrack=cell(1,10);ttrack=cell(1,10);
for i=1:9,%landmark loop
   alandmarks{1,i}=importdata(lmfiles(i,1).name); %data stored in cell array
end
for i=1:10,%tracker position loop
   ftrack{1,i}=importdata(ffiles(i,1).name);
   ptrack{1,i}=importdata(pfiles(i,1).name);
   ttrack{1,i}=importdata(tfiles(i,1).name);
end
%calculate average values of anatomical points and store data in cell array
%'F','P' and 'T'
F=cell(1,3);P=cell(1,3);T=cell(1,3);
for j=1:9,%landmark loop
   if j<=3%for the femur
      F{1,j}=mean(alandmarks{1,j}(:,2:8));
   elseif j>3 && j<7%for the paltella
      P{1,j-3}=mean(alandmarks{1,j}(:,2:8));
   elseif j>=7%for the tibia
      T{1,j-6}=mean(alandmarks{1,j}(:,2:8));
   end
end
%calculating average values of tracking data
PT=cell(1,10);TT=cell(1,10);
for k=1:10,
   PT{1,k}=mean(ptrack{1,k}.data(:,2:8));
   TT{1,k}=mean(ttrack{1,k}.data(:,2:8));
end
%store the average values in matrices: 'FLandmarks','PLandmarks',
%'TLandmarks','PTracker'and 'TTracker.
FLandmarks=transpose([F{1,1}(1,5:7);F{1,2}(1,5:7);F{1,3}(1,5:7)]);
PLandmarks=transpose([P{1,1}(1,5:7);P{1,2}(1,5:7);P{1,3}(1,5:7)]);
TLandmarks=transpose([T{1,1}(1,5:7);T{1,2}(1,5:7);T{1,3}(1,5:7)]);
Flexion=[0:10:90];
PTracker=zeros(10:8);%preallocate matrix
TTracker=zeros(10:8);
for m=1:10,
   PTracker(m,:)=[Flexion(1,m) PT{1,m}(1,1:7)];
   TTracker(m,:)=[Flexion(1,m) TT{1,m}(1,1:7)];
end

%2222222222222222222222222222222222222222222222222222222222222222222222
%                 GENERATE LOCAL COORDINATE SYSTEM
%                 FOR THE FEMUR, PATELLA AND TIBIA
%2222222222222222222222222222222222222222222222222222222222222222222222
%   o-origin = midpoint between medial and lateral anatomical landmarks
```

```
%   x-axis   = midpoint-lateral
%   y-axis   = midpoint-posterior
%   z-axis   = midpoint-distal
%------------------------------------------------------------------------
%%%THE FEMUR%%%:
%femoral body fixed axes based on anatomical axes (non-orthogonal axes).
p4F=0.5*(FLandmarks(:,2)+FLandmarks(:,1));%calculate midpoint or o-origin.
rxF=FLandmarks(:,1)-p4F;%calculate x-axis vector
rzF=p4F-FLandmarks(:,3);%calculate z-axis vector
ryF=cross(rzF,rxF);%calculate y-axis vector
%calculate transformation matrix to global coordinate
TFaG_c=[rxF/norm(rxF) ryF/norm(ryF) rzF/norm(rzF) p4F;0 0 0 1];
invTFaG_c=invhtm(TFaG_c);
%TFaG is constant (_c) because the femur is fixed.
%------------------------------------------------------------------------
%%%THE TIBIA%%%:
%tibial body fixed axes based on anatomical axes (non-orthogonal axes).
p4T=0.5*(TLandmarks(:,2)+TLandmarks(:,1));
rxT=TLandmarks(:,1)-p4T;
rzT=TLandmarks(:,3)-p4T;
ryT=cross(rzT,rxT);
%calculate transformation matrix to global coordinate
TTaG_0=[rxT/norm(rxT) ryT/norm(ryT) rzT/norm(rzT) p4T;0 0 0 1];
invTTaG_0=invhtm(TTaG_0);
%TTaG is not constant but varies at different knee flexion angles
%calcualte global coordinates of 3 anatomical points
rTaG_in=[TLandmarks transpose(TTracker(1:1,6:8));1 1 1 1];
%calculate local coordinates of 3 anatomical points (constant)
rTaTa_c=invTTaG_0*rTaG_in;
%------------------------------------------------------------------------
%%%THE PATELLA%%%:
%patellar body fixed axes based on anatomical landmarks (orthogonal axes)
p4P=0.5*(PLandmarks(:,2)+PLandmarks(:,1));
rxP=PLandmarks(:,1)-p4P;
ryP=cross(PLandmarks(:,3)-p4P,PLandmarks(:,1)-p4P);
rzP=cross(rxP,ryP);
%Transformation matrix to global coordinate
TPaG_0=[rxP/norm(rxP) ryP/norm(ryP) rzP/norm(rzP) p4P;0 0 0 1];
invTPaG_0=invhtm(TPaG_0);
%TPaG is not constant but varies at different knee flexion angles
%global coordinates of 3 anatomical points
rPaG_in=[PLandmarks transpose(int_PTracker.data(1,6:8));1 1 1 1];
%local coordinates of 3 anatomical points (constant)
rPaPa_c=invTPaG_0*rPaG_in;
%determine new patellar anatomical landmarks from tracking data of current
%kinematic data set
TPG_0=[transpose(quat2dcm(int_PTracker.data(1:1,2:5))) transpose(int_PTracker.data(1:1,6:8));0 0
0 1];
invTPG_0=invhtm(TPG_0);
TPaP_c=invTPG_0*TPaG_0;
TPG_0n=[transpose(quat2dcm(PTracker(1:1,2:5))) transpose(PTracker(1:1,6:8));0 0 0 1];
TPaG_0n=TPG_0n*TPaP_c;
rPaG_0n=TPaG_0n*rPaPa_c;
rPaG_inn=rPaG_0n;

%3333333333333333333333333333333333333333333333333333333333333333333333
%           CALCULATE TIBIAL MOTIONS FROM TRACKING DATA
%3333333333333333333333333333333333333333333333333333333333333333333333
```

```
for i=1:1:10,
   angle=TTracker(i,1);
   %calculate transformation matrices from tibial tracker frame to global
   %frame
   TG_R=transpose(quat2dcm(TTracker(i:i,2:5)));%initial rotation matrix calculated from raw
quaternion
   TG_T=transpose(TTracker(i:i,6:8));%initial displacement from the origin of global coordinate
system to local coordinate system.
   TG=[TG_R TG_T;0 0 0 1];%homogeneous transformation matrix
   eval(['TTG_' num2str(angle) '=TG;']);%store homogeneous transformation matrix for each angle

   %plot to see how the coordinate system of the tracker moves
   Lenght = 50;
   figure(1);
   for j=1:1:3,
      xaxis=plot3([TG(1,4), TG(1,4)+Lenght*TG(1,j)],[TG(2,4), TG(2,4)+Lenght*TG(2,j)],
[TG(3,4), TG(3,4)+Lenght*TG(3,j)]); hold on
      yaxis=plot3([TG(1,4), TG(1,4)+Lenght*TG(1,j)],[TG(2,4), TG(2,4)+Lenght*TG(2,j)],
[TG(3,4), TG(3,4)+Lenght*TG(3,j)]); hold on
      zaxis=plot3([TG(1,4), TG(1,4)+Lenght*TG(1,j)],[TG(2,4), TG(2,4)+Lenght*TG(2,j)],
[TG(3,4), TG(3,4)+Lenght*TG(3,j)]); hold on
   end

   %calculate constant transformation from tibial anatomical axis (Ta) to tibia tracker (T)
   invTTG_0=invhtm(TTG_0);
   TTaT_c=invTTG_0*TTaG_0;

   %calculate TTaG_(angle)
   eval(['TTaG_' num2str(angle) '=TTG_' num2str(angle) '*TTaT_c;']);
   eval(['TTaFa_' num2str(angle) '=invTFaG_c*TTaG_' num2str(angle) ';']);

   %calculate relative motions the tibia to the femur
   eval(['Tmotions_' num2str(angle) '=[transpose(TTaFa_' num2str(angle) '(1:3,4)) ncangle(TTaFa_'
num2str(angle) ')];']);

   %to move bone: calculate position vectors of 3 points on the bone for
   %each flexion angle
   eval(['rTaG_' num2str(angle) '=TTaG_' num2str(angle) '*rTaTa_c;'])

end

%44444444444444444444444444444444444444444444444444444444444444444444444
%             CALCULATE PATELLAR MOTIONS FROM TRACKING DATA
%44444444444444444444444444444444444444444444444444444444444444444444444
for i=1:1:10,
   angle=TTracker(i,1);
   %calculate transformation matrix from patellar tracker frame to global
   %frame
   PG_R=transpose(quat2dcm(PTracker(i:i,2:5)));
   PG_T=transpose(PTracker(i:i,6:8));
   PG=[PG_R PG_T;0 0 0 1];
   eval(['TPG_' num2str(angle) '=PG;']);

   %plot to see how the coordinate system of the tracker moves
   Lenght = 50;
   figure(1);
   for j=1:1:3,
      xaxis=plot3([PG(1,4), PG(1,4)+Lenght*TG(1,j)],[PG(2,4), PG(2,4)+Lenght*PG(2,j)], [PG(3,4),
```

```
PG(3,4)+Lenght*PG(3,j)]); hold on
       yaxis=plot3([PG(1,4), PG(1,4)+Lenght*TG(1,j)],[PG(2,4), PG(2,4)+Lenght*PG(2,j)], [PG(3,4),
PG(3,4)+Lenght*PG(3,j)]); hold on
       zaxis=plot3([PG(1,4), PG(1,4)+Lenght*TG(1,j)],[PG(2,4), PG(2,4)+Lenght*PG(2,j)], [PG(3,4),
PG(3,4)+Lenght*PG(3,j)]); hold on
   end

   %calculate constant transformation from patellar anatomical axes (Pa) to patellar tracker (P)
   invTPG_0n=invhtm(TPG_0n);
   TPaP_c=invTPG_0n*TPaG_0n;

   %calculate TTaG_(angle)
   eval(['TPaG_' num2str(angle) '=TPG_' num2str(angle) '*TPaP_c;']);
   eval(['TPaFa_' num2str(angle) '=invTFaG_c*TPaG_' num2str(angle) ';']);

   %calculate joint angle from function cangle()
   eval(['Pmotions_' num2str(angle) '=[transpose(TPaFa_' num2str(angle) '(1:3,4)) ncangle(TPaFa_'
num2str(angle) ')];']);

   %to move bone: calculate position vectors of 3 points on the bone for
   %each flexion angle
   eval(['rPaG_' num2str(angle) '=TPaG_' num2str(angle) '*rPaPa_c;'])

end

%store results in matrices 'Presults' and 'Tresults'
Presults=[Pmotions_0;Pmotions_10;Pmotions_20;Pmotions_30;Pmotions_40;Pmotions_50;Pmotions
_60;Pmotions_70;Pmotions_80;Pmotions_90];
Tresults=[Tmotions_0;Tmotions_10;Tmotions_20;Tmotions_30;Tmotions_40;Tmotions_50;Tmotion
s_60;Tmotions_70;Tmotions_80;Tmotions_90];

%555555555555555555555555555555555555555555555555555555555555555555555
%                   PLOT PATELLAR AND TIBIAL MOTIONS
%                        IN 2D AND 3D GRAPHS
%555555555555555555555555555555555555555555555555555555555555555555555
%store parameters in cell array for plotting
rTaG={rTaG_0,rTaG_10,rTaG_20,rTaG_30,rTaG_40,rTaG_50,rTaG_60,rTaG_70,rTaG_80,rTaG_9
0,};
rPaG={rPaG_0,rPaG_10,rPaG_20,rPaG_30,rPaG_40,rPaG_50,rPaG_60,rPaG_70,rPaG_80,rPaG_90,
};

%how bones move: plot calculated anatomical landmarks of each flexion (add
%to figure(1))
figure(1)
for b=1:10,
   plot3(rTaG{1,b}(1,1:4),rTaG{1,b}(2,1:4),rTaG{1,b}(3,1:4),'-+m'); hold on
   plot3(rPaG{1,b}(1,1:4),rPaG{1,b}(2,1:4),rPaG{1,b}(3,1:4),'-+m'); hold on
end
title('Trackers and bones plot')
xlabel('x-axis')
ylabel('y-axis')
zlabel('z-axis')
grid on;axis equal;
%plotting bones motion in 12 figures (5 and 3 figures for tibia and patellar, respectively)
x=zeros(1,10);%vector to store angle values
tnames={'Tibial Medial-Lateral Shift','Tibial Anterior-Posterior Draw','Tibial Proximal-Distal
Translation',...
   'Tibial Flexion','Tibial Abduction/Adduction','Tibial Internal/External Rotation',...
```

```
    'Patellar Medial-Lateral Shift','Patellar Anterior-Posterior Translation','Patellar Proximal-Distal
Translation',...
    'Patellar Flexion','Patellar Lateral Rotation','Patellar Lateral Tilt'};
ynames={'Lateral shift (mm)','Posterior draw (mm)','Distal Translation (mm)',...
    'Extension (deg)','Abduction(deg)','External Rotation (deg)',...
    'Lateral Shift (mm)','Posterior translation (mm)','Distal Translation (mm)',...
    'Extension (deg)','Lateral Rotation (deg)','Lateral Tilt (deg)',};
xname='Knee flexion angle (deg)';

for f=1:12,%figure loop must start at Figure2
    y=zeros(1,10);
    for k=1:10,%data loop from 0 - 90 degrees
        x(k)=TTracker(k,1);
        figure(f+1);
        if f<=6
            y(k)=Tresults(k,f);
        else
            y(k)=Presults(k,f-6);
        end
        plot(x,y,'-+m')
    end
    title(tnames(f))
    xlabel(xname)
    ylabel(ynames(f))
end
%save Presult, Tresults, tname as MAT-files
save('Presults.mat','Presults');
save('Tresults.mat','Tresults');
save('tname.mat','tnames');
save('yname','ynames');
```