

The Patellofemoral Joint: Form and Function



Dissertation submitted to the University of Oxford
in partial fulfilment
of the requirements for the degree of

Doctor of Philosophy

Andrew Paul Monk

St. John's College
Oxford

Michaelmas Term 2011

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Chapter 1

Thesis Overview

1.1 Thesis Aim

The overall aim of this thesis is to develop a fuller understanding of the patellofemoral joint (PFJ), with the longer term objective of being able to guide new designs for total knee replacement.

1.2 Structure of the Thesis

The premise of the investigations in this thesis is that anterior knee pain, a feature commonly associated with knee replacement will, in part, be due to abnormalities in PFJ kinematics. It is very likely that the shape of the joint surfaces and their interactions are important for normal pain-free function of the knee.

Following a literature review outlining the current literature and scope of the problem, the following subsections outline the separate studies that were undertaken to help advance our understanding of the PFJ.

1.2.1 Development of the PFJ

The basic structure of the knee has been conserved for over 300 million years [1]. Previous authors have postulated that inter-species variation in tibiofemoral biomechanics have been caused by anatomical changes associated with adoption of an upright stance for homonin gait [2, 3]. However, there have been no studies assessing the development of the patellofemoral joint throughout the human lineage. Such studies would provide insight into anatomical changes in the PFJ that have taken place to accommodate homonin bipedal gait and PFJ stability.

1.2.2 Morphology of the PFJ

Whilst numerous studies have described the normal anatomy of the knee in terms of simple measurements from bone, it has been shown that the cartilage covered articulating surface can be of a very different shape from that of bone [4, 5]. To date, there has been no work to describe whether the orientation of the trochlea groove's articulating surface has a consistent relationship with the orientation of the underlying bone, and the trochlea orientation in the replaced knee. A section of this thesis is dedicated to providing this description.

1.2.3 Development of Technique to Measure 3D Knee Kinematics

Previous kinematic analyses of the PFJ have been restricted mostly to two dimensional (2D) descriptions in the sagittal plane and it has not been possible to reliably measure coronal plane kinematics following knee replacement [6]. This thesis develops and validates a new method, MAUS (combining Motion Analysis with UltraSound), to address the coronal problem.

1.2.4 Measurement of PFJ Kinematics in Normal and Replaced Knees

It is likely that abnormal PFJ kinematics in both the normal and replaced knee might lead to pain and poor function. Using the developed MAUS technique, experiments were designed to enable a comparison of the kinematic profiles of normal and replaced knees, providing valuable insights into which part of the replaced knee creates abnormal movements.

1.2.5 Anatomical factors affecting PFJ kinematics

The most common kinematic abnormality is lateral subluxation of the patella [7]. This is associated with pain and dislocation [8]. Many anatomical abnormalities are said to predispose to subluxation [9, 10]. To date, no study has addressed the relative importance of these abnormalities. Investigations were performed to assess the relative importance of a range of anatomical variables to patella subluxation. Insight into this will help with the management of PFJ problems in the native knee, guide future operative technique for knee arthroplasty and help improve arthroplasty design.

1.2.6 Discussion

A discussion is provided which brings together the individual studies to comment further on the relationship between the form and function of the PFJ, summarising the important results, discusses the relevance of the work, and conclusions drawn from the studies.

Chapter 2

Introduction

This chapter summarises a literature review and highlights areas requiring investigation which are addressed in the later studies.

2.1 Outlining the Problem

Throughout the history of orthopaedic surgery, developments have been made to correct deformity, alleviate pain and restore function. One of the most successful examples is knee joint replacement surgery for degenerative arthropathies, with recent survivorship studies quoting near 100% success at 10 years [11]. However, despite recent advances in component design and implantation techniques, a significant proportion of patients experience problems relating to the patellofemoral joint (PFJ). Indeed, around 30-50% of referrals to clinics are for anterior knee pain and problems relating to the PFJ, both following knee joint replacement and in the native knee [12]. For the clinician, the origins of anterior knee pain are notoriously difficult to elucidate, and amongst others may include bony dysplasias, soft tissue abnormalities and even referred pain from hip pathology. The difficulty in establishing a diagnosis

is often further compounded by shortcomings with current imaging techniques, especially in obtaining satisfactory data regarding coronal plane movements of the patella in the replaced knee. This is borne-out in the literature with PFJ kinematic descriptions being largely restricted to the sagittal plane.

Much of the information available about the PFJ is surprising and contradictory; despite anterior knee pain and the PFJ being the leading source of complications following total knee replacement (TKR), at the time of writing there is no evidence to suggest whether or not to replace the patella articulating surface at operation [13, 14, 15, 16, 17]. A multi-centre, randomised control trial to assess the outcome of two groups; patella resurfacing versus non-resurfacing, has been underway since 1999 [18]. To date there has been no significant difference shown between the two groups. Similarly, during unicompartmental knee replacement (UKR) recent literature has suggested that the level of PFJ degeneration, as found at operation, has no bearing on the outcome of the procedure and therefore can largely be ignored, despite the patellar cartilage articulating with the metal femoral component following implantation [19]. Perhaps the most surprising example of all is that there are over one hundred described techniques for the surgical correction of patellofemoral dysplasia [20, 21].

In short, one of the great unsolved mysteries of orthopaedic surgery is the patellofemoral joint. Treatment strategies for patellofemoral joint problems must be based on a fundamental understanding of the PFJ which currently does not exist.

2.2 Anatomy of The Knee and Patellofemoral Joint

2.2.1 Terminology

Throughout this thesis the following terminology will be used;

Reference Planes

Axial (transverse): A horizontal plane splitting the body into upper and lower parts.

Coronal: A vertical plane splitting the body into anterior (front) and posterior (back) parts.

Sagittal (median): A vertical plane passing through the body from front to back dividing it into symmetrical left and right sides. Any plane parallel to the sagittal plane is termed para-sagittal (para-median) plane.

Alignment

Genu Varum: A coronal plane displacement of the tibia towards the midline.

Genu Valgum: A coronal plane displacement of the tibia away from the midline.

Physiological Valgus is a term given to the habitual posture of the normal knee with a valgus position of 5 degrees in men and 7 degrees in women.

Movements

The knee is described as a modified hinge joint with movements including flexion, extension, medial and lateral rotation. The flexion and extension noted at the knee

differ from true hinging in that there is rotational element due to the geometry of the articular surfaces of the distal femur and proximal tibia.

Flexion: Bending of adjacent body sections so that their anterior and posterior surfaces are brought together, eg. thigh and calf in flexion of the knee joint.

Extension: Moving apart of two adjacent body segments, eg. Straightening of the lower leg during extension about the knee.

Medial/Lateral Rotation: Movement about the longitudinal axis of a bone so that the anterior face of the bone moves towards (medial) or away from (lateral) the midline.

Abduction: Movement away from the midline.

Adduction: Movement of a body part towards the midline.

2.3 Anatomy

The knee joint comprises the medial and lateral tibio-femoral joints and a patellofemoral joint.

2.3.1 Embryology

Mrida-Velasco et al [22] reported the development of the knee from 50 embryonic and fetal lower limbs. The cartilage of the femoral and tibial epiphysis is seen around day 40, and ossification is seen at week 14. The knee joint cavity is first seen at the patellofemoral joint around day 52, with distinct separate tibiofibular joints visible from the 13th week onwards [22].

2.3.2 Trochlea

The distal end of the femur contains a trochlear groove, which is essentially concave and receives the convex posterior aspect of the patella which, during flexion at the knee joint, moves from a superior to an inferior position. The trochlea groove has traditionally been described as being roughly vertical, with the majority of femoral sulci lying between 0 - 5 mm lateral to the midline of the knee [23]. Trochlear dysplasia is characterised by a decreased height of the medial femoral condyle, a lateral trochlear facet, which is sometimes flat or even elevated, a flatter trochlear groove with associated decreased trochlear depth [24].

The extensor mechanism of the knee consists of a complex arrangement including the quadriceps muscle of muscles (vastus medialis, lateralis and intermedius), its common tendon, the patellar tendon, the patella itself and the retinaculæ. The patella is connected to the tibial tuberosity via the patella tendon/ligament [25].

The mechanical functions of the patella are to direct the force from the four parts of the quadriceps mechanism towards the tibial tuberosity and improve the efficiency of the quadriceps mechanism by increasing its lever arm. Throughout the flexion/extension movement at the tibio-femoral joint, the patella allows anterior displacement of the patella tendon moment arm.

2.3.3 Patella

As the largest sesamoid bone in the body, the patella is located within the extensor mechanism anterior to the knee joint. Tissue attachments to the patella include quadriceps tendon, patella ligament and medial and lateral retinaculæ. The patella is triangular in shape such that the apex points distally, providing attachment to the patella tendon, the upper part being the site of insertion of the tendon of quadriceps

femoris. The anterior surface is convex and separated from the skin by the prepatella bursa. The posterior surface is a smooth oval shape and provides the patella component to the patellofemoral joint complex. Crossed by a vertical ridge, the articulating surface is separated into medial and lateral facets [1].

Classification of Patella Shape

Wiberg, in 1941 [26] described a classification system incorporating the variation in relative proportions of the medial and lateral facets. Type 1 (10% of population) has roughly equally sized medial and lateral facets, Type 2 (65% population) having slightly larger lateral facets and Type 3 (25% population) having much larger lateral facets (Figure 2.1) [4, 26].

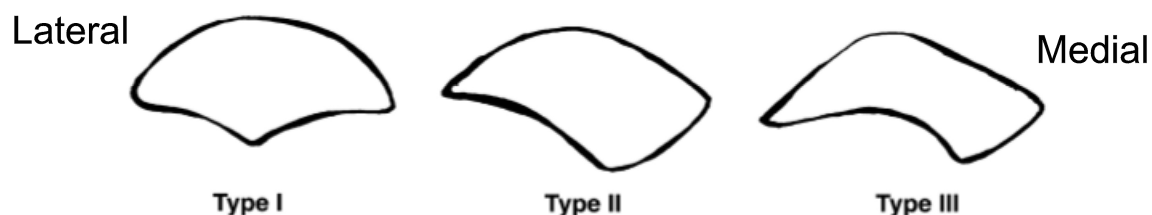


Figure 2.1: Showing Wiberg Classification of the Patella. Type I - equal sized facets, Type II - slightly larger lateral facet, Type III - much larger lateral facet. From Tecklenburg et al 2006.

Baumgartl later extended the classification of patellae to include the ‘Jaegerhut’ shape, without central ridge or medial facet [3]. A further facet, named the odd facet is found on the medial border of the patella, which only articulates with the femoral condyle in high flexion [27]. Grelsamer [4] described a classification of different patellae from measurements on sagittal plane radiographs, the ratio of maximal patella length to articular surface length;

Type 1 patellae have a morphology ratio between 1.2 and 1.5;

Type 2, over 1.5 (Cyrano appearance with a long distal pole);

Type 3, below 1.2 (short distal pole).

The Maldague classification describes the relative position of two radiographic lines (central ridge and lateral edge) as shown in Figure 2.2 [28].

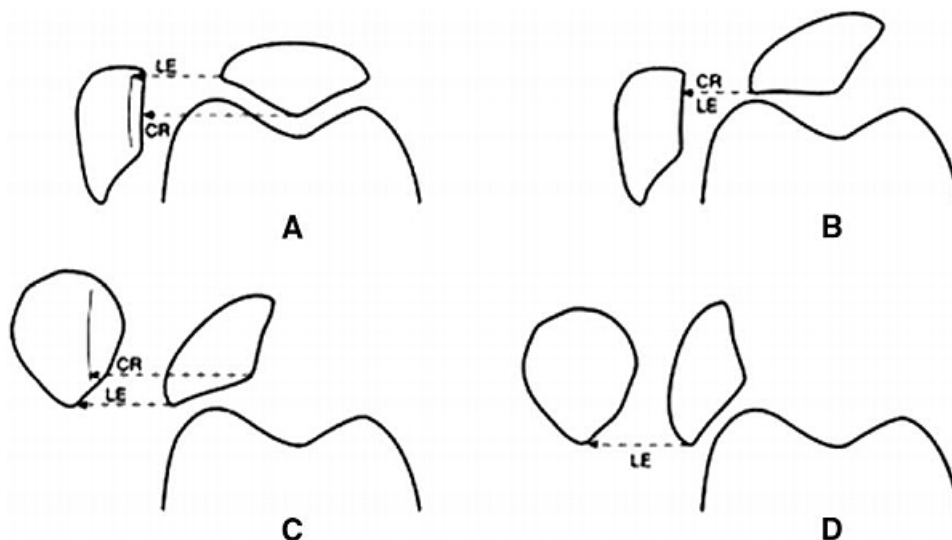


Figure 2.2: Schematic representation of patellar position on lateral views, according to the degree of external rotation (tilt). The modified Maldague-Malghems classification comprises 4 stages; (A) Stage 1 (normal position) both lines (central ridge, CR; lateral edge, LE) are concave posteriorly and separated by 5 to 10 mm. The anterior line is the lateral edge, the posterior line the central ridge. (B) Stage 2 (false lateral profile as described by Maldague and Malghem, or minor lateral patellar subluxation): both lines are superimposed. Only one straight line is visible. (C) Stage 3 (overwhelming lateral profile, or pronounced lateral subluxation): the anterior line is the central ridge, the posterior line is the lateral aspect of the patella and is convex. (D) Stage 4. the central ridge is no longer visible. The patella often covers the anterior cortex of the femoral metaphysis and appears ovoid in shape. From Christian et al 2006.

2.3.4 The patellofemoral joint

The patellofemoral joint is inherently unstable because the tibial tubercle lies lateral to the long axis of the femur. This gives rise to a tendency towards lateral subluxation of the patella during flexion at the knee joint. On entering the groove,

there are two anatomical features to compensate for the inevitable lateral migration of the patella. Brattsrom et al (1964) [29], describe the lateral aspect of the trochlea is angled such that the patella is directed towards the apex of the groove. The second stabilising factor is the tension provided by the medial soft tissue structures.

Patellofemoral contact area

Cadaveric studies have addressed the patellofemoral contact area during flexion and extension of the knee [30]. In extension, the contact area on the patella is a horizontal band across the patella from the medial to the lateral facet, and moves proximally throughout flexion, being located in the inferior, middle and superior thirds of the patella at 30°, 60° and 90° of flexion respectively. The size of the contact area increases with progressive flexion.

Goodfellow et al used a dye exclusion method [27] to characterise the patterns of patterns of patellofemoral joint contact (Section 2.5). Similar studies have measured the contact area with pressure sensitive film. The area of contact was measured to be 2.6 cm squared at 20° and 4.1 cm squared at 90°. After 90° of flexion the quadriceps tendon abuts the superior aspect of the trochlea. This acts to relieve some of the pressure from the patellofemoral joint. In conjunction with this, the patellofemoral contact area changes to from a single band to two distinct areas on the lateral and extreme medial, or ‘odd’ facets [31, 32].

Patellofemoral joint reaction force and pressure

The patellofemoral joint reaction force is a result of tension developing in the quadriceps and patellar tendons due to contraction of the quadriceps muscles. It is represented by the resultant vector of the quadriceps tendon force (QF) and the patella tendon force (PTF) across the knee joint at a known flexion angle (Figure 2.6) [33]. This relationship assumes that the movement of the patella is frictionless and that the two forces, QF and PTF are equal, whereas in reality QF has been shown to be closer to $1.5 \times \text{PTF}$ [34].

The articular cartilage layer on the posterior aspect of the patella is the thickest found in the human body, with a depth of up to 5 mm (Fulkerson, 2004) [33]. This reflects the forces that are transmitted through the joint. Melegari et al, 2008 [35] report a force of $0.5 \times \text{body weight (BW)}$ at 9° of flexion, $3.3 \times \text{BW}$ at 60° and up to $7.8 \times \text{BW}$ at high flexion, approximately 5000 N. Patellofemoral contact pressures during flexion increase up to 90° with the highest increase seen between 30° and 60° . At flexion beyond 90° the magnitude of pressure was seen to decrease until 120° , a feature attributed to the effect of unloading of the patellofemoral joint secondary to the abutment of the quadriceps tendon on the trochlea.

2.4 The Evolution of the Knee Joint

Understanding the biomechanics of such a complex system as the human knee joint might be facilitated by observing its origins. Comparative anatomical studies of our evolutionary ancestors allows an understanding into the origins of knee morphology, affording the surgeon valuable insights into the role of different features of the human

knee joint.

Literature regarding the development of the *Homo sapiens* knee joint is fraught with controversy and contradictory theories. To an extent the quadrapedal-bipedal shift remains shrouded in mystery, how is the ape to hominid transformation explained? At present, two mainstream paradigms exist to explain the origins of bipedalism; *knuckle walking* and *vertical climbing*. The plantigrade ‘heel-strike/toe-off’ gait pattern illustrated by humans for terrestrial walking can be seen in the hindlimbs of lower primates (eg. chimpanzee) and, with coincidental adaptations in the upper limb, is considered by some to be the key ‘terrestrial’ phase of the emergent bipedal gait [36].

The more popular, vertical climbing paradigm suggests that human hindlimb features might have been developed through tree climbing, highlighting common behaviours illustrated by climbing apes with developed hindlimbs [37]. However, the African ground apes such as gorilla, are powered largely by highly developed forelimbs.

The modern bi-pedal, human knee is far removed from our quadrapedal ancestors. Ichthyostega is the earliest known knee joint found in the fossil record of the Devonian period (365 million years ago), shown in Figure 2.3 [1]. There are distinct femoral, tibia and fibula bones. The distal femur demonstrates a distinct bicondylar articulating surface, a feature preserved until the modern day. Of note is the femoral articulation with both the tibia and fibula [38].

Any study regarding the evolutionary development of the knee requires an established lineage of specimens, representing the individual stages in the development of the

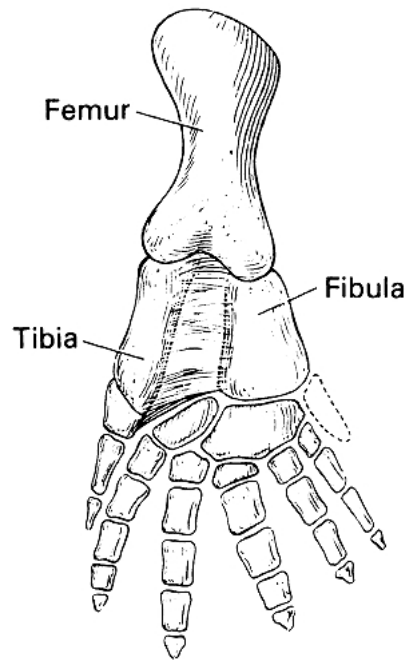


Figure 2.3: Left knee joint of *Ichthyostega*, the earliest known example of a knee joint from the fossil record. Of note are the bicondylar distal femur and both tibiofemoral and fibulofemoral articulations. From Dye, 1987.

modern form. To date, all studies regarding the ‘evolution’ of the knee, have included specimens far removed from the human lineage [1, 38, 39], including black bears, pigeons, and dogs. Indeed most studies have provided a description of the knee as a whole, with no study to date providing a description of the PFJ.

2.5 Kinematics and Kinetics

Kinematics

Kinematics is the branch of mechanics concerned with the motion of objects without reference to the forces which cause the motion. *Oxford Dictionary, Oxford University Press* [40].

Descriptions of knee movements are classically referenced to a co-ordinate system. The most widely used convention is Cole's modification of Grood and Suntay's Joint Co-ordinate System, which was introduced to standardise the description of knee joint motion [41, 42]. Cole's description provides movements relative to a body fixed coordinate system, described as three translations and three rotations around axes fixed to the long axis of tibia and femur, or six degrees of freedom.

The potential for error when measuring patellofemoral orientation has been demonstrated recently in a study comparing measurements of the trochlear groove, relative to four different medio-lateral planes of reference. Iranpour et al [43] used a common centre of the femoral head, and; (A) mechanical (centre of knee), (B) anatomic (inter epicondylar), (C) transcondylar (between spheres fitted to the posterior condyles), and (D) trochlear (medial and lateral apices) axes as shown in Figure 2.4. The study showed that slight changes in the orientation of the femur changed the measured differences in the trochlear position.

Tibiofemoral joint rotations include flexion/extension, axial rotation, adduc-

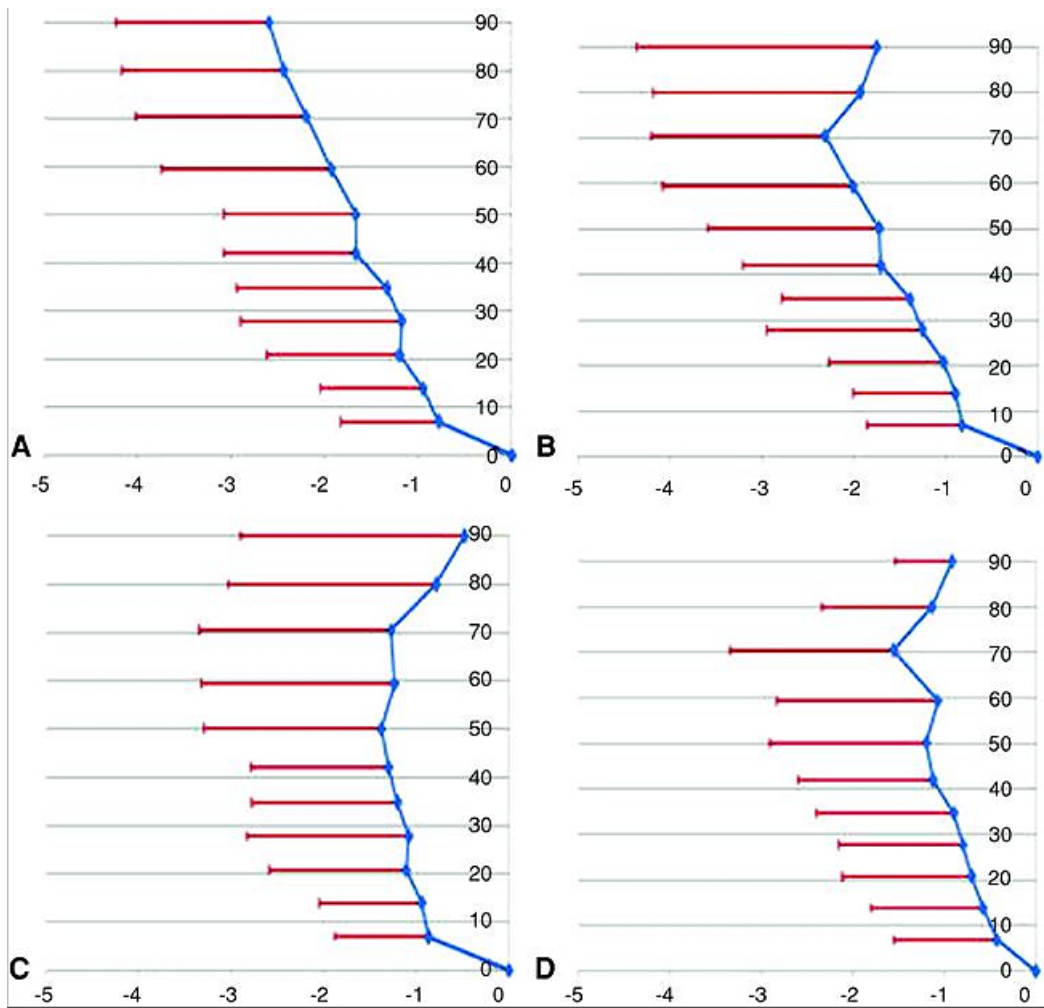


Figure 2.4: The effect of orientation on the apparent mediolateral translation of the trochlear groove is shown. The groove started at the (0,0) origin of each graph when the knee was in extension. As the knee flexed toward 90 degrees (along the vertical axis of each graph), the groove appeared to deviate laterally in relation to the starting point, shown along the X axis in millimeters after the femur was aligned to (A) mechanical, (B) anatomic, (C) transcondylar, and (D) trochlear axes (Iranpour, 2009).

tion and abduction. The translations are anterior/posterior, medial/lateral and proximal/distal. For the patellofemoral joint, rotations include flexion/extension, medial/lateral tilt and axial spin. The translations are anterior/posterior, medial/lateral and proximal/distal. Multiple cadaveric and *in vivo* studies have been performed to track these movements and are summarised in Section 2.6.2. Assessment of the

individual movements in isolation does not adequately describe the overall movement of the knee. To help describe the overall knee mechanism, alternative markers of knee movements have been studied, including contact points and the patellar tendon angle.

The early work of Goodfellow et al [27] shows a method of measuring patella migration during flexion of the knee using contact point movement. During flexion and extension the femur makes contact with the tibia in both medial and lateral compartments. The position of the contact points change with flexion angle of the knee and are dependent on the rotation and translation of the femur relative to the tibia. Describing the pattern of contact point movement provides a readily understandable summary of knee movements [44, 45].

Perhaps the most common measurement tool employed to describe knee kinematics is the patellar tendon angle (PTA) which is influenced by both tibiofemoral and patellofemoral movements. The PTA is the angle subtended by the patella tendon and the long axis of the tibia in the sagittal plane (Figure 2.5). The magnitude of the PTA is dictated by the relative positions of the distal femur and proximal tibia. Due to the cam shape of the distal femur, the PTA is larger in knee extension than in flexion. During flexion the distal femur moves posteriorly on the tibia and the anteroposterior width of decreases, the patella adopts a more posterior position, thereby reducing the value of the PTA. The PTA can be measured on sagittal plane fluoroscopy, and so can be used as a surrogate measure of sagittal plane kinematics [46, 47, 48, 49, 50, 51].

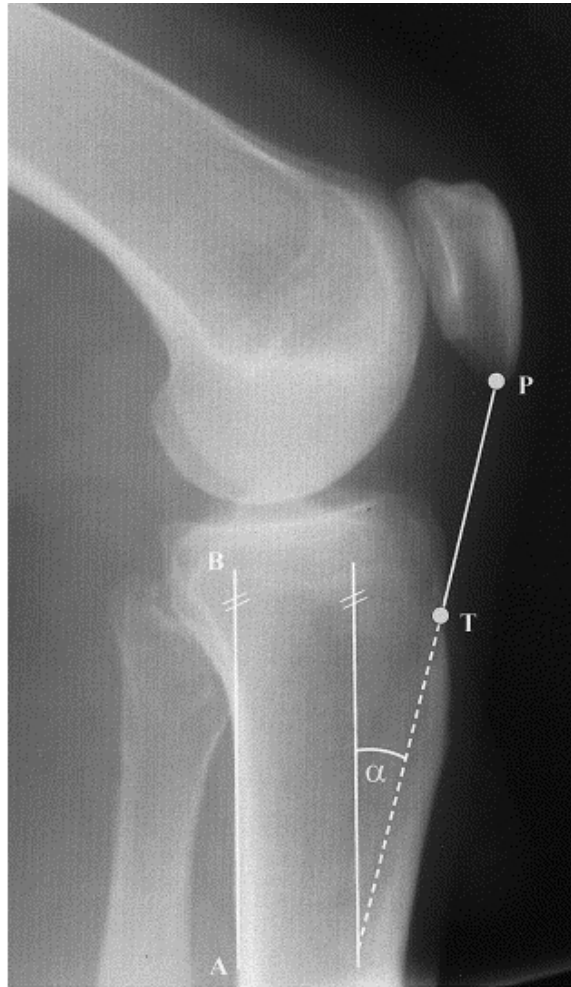


Figure 2.5: Showing the patella tendon angle (PTA). The long axis of the tibia (AB) is taken as the line of the posterior cortex of bone, measured from its proximal third. The line of the patella tendon (PT) is taken from its distal pole to the proximal margin of the tibial tuberosity. Reprinted with permission from Price (2004).

Patellofemoral kinematics

During knee flexion the patella flexes and externally rotates. The patella also tilts in the coronal plane during flexion. There is a constant medial tilt of the patella during high flexion as the ridge between the medial and odd facets engages the femoral condyle.

Kinetics

Kinetics is the branch of mechanics concerned with the motion of bodies under the action of forces. *Oxford Dictionary, Oxford University Press* [40].

The forces acting on the patella are the quadriceps force (QF), the patellar tendon force (PTF) and the patellofemoral contact force, which is referred to as the (PFF) as shown in Figure 2.6. The ratio of forces PFF:QF is dependent on the direction of pull through the patellar and quadriceps tendons ($\alpha + \beta$). The magnitude of the forces can be calculated using basic trigonometry.

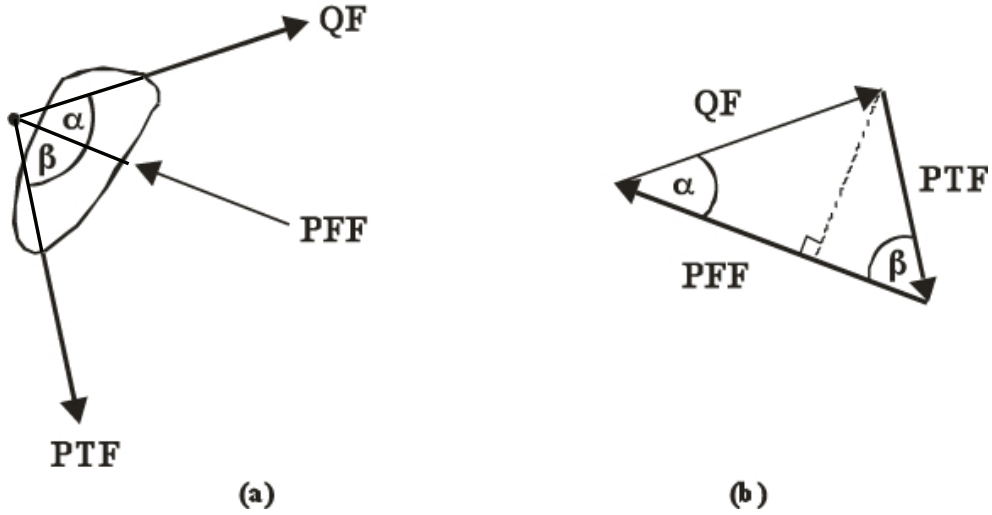


Figure 2.6: (a) Free body diagram of the patellofemoral joint forces and (b) the triangle of forces formed by the patellofemoral force (PFF), patellar tendon force (PTF) and quadriceps force (QF).

2.6 Methods of measuring PFJ kinematics and kinetics

2.6.1 Modelling of PFJ

Mathematical analysis of the knee involves theoretical representation of multiple interacting factors. Modelling is invaluable in the knee where subtle interplay of tendons, ligaments and joint surfaces create complex movement patterns. The kinematic characteristics can be modelled mathematically allowing for the application of modelling restraints and trialling of potential clinical scenarios (eg. anterior cruciate ligament compromise). The four-bar linkage mechanism used to describe the knee joint is an example of modelling the ligamentous constraints to the human tibiofemoral joint. More recent techniques include *finite element analysis* and *statistical shape modelling*.

Finite element analysis (FEA) allows for the inclusion of physical properties into the mathematical model. The earliest FEA models in orthopaedics were designed to represent the mechanical behaviour of skeletal parts [52]. Since then the method has been frequently applied to perform stress analyses of bones and bone/prosthesis interfaces and fracture fixation devices. Within the FEA models of the knee, the knee is divided into its component parts (bony and soft tissue), to which mathematical representations of the physical properties (eg. bone density, Young's modulus for ligaments) are ascribed. Computational methods are then used to analyse the changes in the elements (parts) created during movement. Many difficulties have been encountered when attempting to assign material properties to structures, with the complexity of biological structures (such as the trabecular architecture of the femur) often being underestimated [53].

Statistical shape modelling can be used in combination with advanced FEA to characterise the relationships between PFJ shape and kinematics/contact mechanics. The statistical shape model captures inter-subject geometric variability based on a ‘training set’ of functionally normal volunteer patients who undergo scanning (CT or MRI) to obtain a 3 dimensional geometry [54, 55].

Whilst computer modelling provides an attractive methodological option, allowing analysis of multiple different variables, current techniques carry the disadvantage of requiring robust validation using experimental data.

2.6.2 Imaging of PFJ kinematics

The use of radiographs, computed tomography (CT), magnetic resonance imaging (MRI) and ultrasound is well established in imaging the native patellofemoral joint [6, 44, 45, 56, 57, 58, 59, 60, 61, 62, 63]. This section outlines the principles of each imaging modality followed by a discussion of the recent literature including the methodological drawbacks.

Radiography

Radiography is an established method of obtaining excellent images of bone whereas soft tissues are poorly visualised. There are three established techniques for obtaining a radiograph: conventional analogue technique, computed radiography and digital radiography. The established analogue method employs a phosphor image plate cassette. Electrons within the phosphor crystals of the cassette become trapped at higher energy levels after being excited by X-ray photons. The image plate is then

read by a laser beam at which point the electrons return to a lower energy level, the difference in energy being proportional to the amount of radiation absorbed and manifest as light on a digital image [64]. Digital radiography utilises a flat panel detector. The detector converts the x-ray image to a digital signal that can then be transferred to digital imaging software such as PACS (picture archiving and communication system). This can be done directly and indirectly. The direct method converts x-ray photons into electrical charge via an amorphous selenium layer and a thin transistor array, with each electrode representing a pixel. The indirect method converts the x-ray photons into light, via a caesium iodide scintillator. The light is detected by a silicon photodetector array and converted into a digital signal. Fluoroscopy involves the real-time acquisition of digital radiographic data. Using the flat panel detector multiple sequential images are arranged to form a video loop that can be exported in a variety of video formats. Patellofemoral kinematics can be assessed in this manner by keeping the knee joint in the field of view whilst simultaneously flexing and extending the knee. In this manner, previous studies have assessed sagittal plane kinematics of the knee joint with reference to the patella tendon angle and the patella flexion angle. When considering kinematic analysis of the knee, a drawback of radiography is that it can only provide image data in one plane. The advent of computed radiography allowed for sophisticated data processing with a dynamic range of greater than one hundred times that of analogue methods.

Computed Tomography

The origins of computed axial tomography date back to 1968 and the experiments of an engineer named Godfrey Hounsfield [65]. Through his experimental work on reconstructing the contents of a box from projections in random directions,

Hounsfield found that treating the box as a series of slices, rather than a volume simplified the problem dramatically. He tested the same theorem with a series of matrices and after addition of simulated absorption values found the predictions to be highly accurate. CT images are formed from a series of back projections of x-ray transmissions taken in multiple directions through a target tissue. The acquisition of CT data can be in multi-slice or helical format, can be fused with other data formats such as real-time CT fluoroscopy, and is commonly viewed as a Cartesian-coordinate map, as slices or 3D reconstructions [66]. The initial scale used for describing the attenuation coefficients within body tissues, was related to that of water with each Hounsfield unit being 0.1% of the attenuation of water.

Hounsfield Units for Materials used in Knee Surgery	
Material	Hounsfield Units
Adipose Tissue	-80
Water	0
Collagen	250
Cortical Bone	1,600
Aluminium	2,300
Iron	34,000
Iodine	141,300
Lead	205,000

Table 2.1: Showing the range of Hounsfield Units (HU) for materials found in the knee.

The values in Table 2.1 are representative of materials when scanned with a beam energy of 65 keV (voltage equivalent of 120 kV). Interestingly, at lower operating energies e.g. 55 keV, the HU value for cortical bone is greater than 2000, whereas soft tissue values change very little in comparison. In this way, as with conventional radiographs, CT provides an excellent way of imaging bone, but not soft tissue. Other high atomic number materials such as those found in metal joint prostheses also show large degrees of variation of HU with different beam energies [66]. This

provides potential for CT hardware setups that might allow imaging of the metalware in post-operative knee joints. The use of CT in assessing the patellofemoral joint is well established, with kinematic assessment being possible using sequential static positions of flexion [10, 61, 62, 63, 67], or true dynamic assessment using as described by Muhle in 1999 using ultrafast or helical scanning [66, 68].

Magnetic Resonance Imaging

Magnetic Resonance Imaging (MRI) is a non-ionising method of providing digital representation of entire tissue planes and has no demonstrable adverse biological effects. Tissue representation in an MR image is directly related to the constituent fat and water composition. At the atomic level these tissues are all compounds of carbon, hydrogen, oxygen and phosphorus. The theory of MR image production is based on the behaviour of these compounds within a magnetic field [69]. Within the atom the proton spins about its axis producing a localised magnetic field. Within an entire tissue the axes of these atoms are randomly aligned, however when a magnetic field is passed across the tissue the atoms align with the axis of the magnetic field. The MR image is formed from the additional magnetic moments that are created from protons within this field. It follows therefore that the strength of the magnetic field measured in Tesla (T) dictates the relative response of the magnetic moments within the tissue.

Precession

As protons spin about their individual axes within the main magnetic field the attraction of the field is balanced by the centrifugal force and the proton begins to

precess. The relationship between the precessional frequency w , and the magnetic field strength Bo is given by the Larmor equation

$$w = x Bo$$

where x is a physical constant for the proton [69].

Resonance

The net magnetism of precessing magnetic moments is small and cannot be measured directly. Instead it is measured indirectly using a second perturbing magnetic force which is applied perpendicular to the magnetic force B and when tuned to the Larmor frequency, will cause resonance of the magnetic signal, which can in turn be measured as a function of signal strength.

2.6.3 Ultrasound

Medical ultrasound utilises sound waves with frequencies within the 1 - 12 MHz range (audible range is less than 20 kHz). Ultrasound waves are generated from piezoelectric crystals that vibrate at specific frequencies when an alternating current is passed across them, the same crystals acting as continuous receivers. The speed of sound is relatively constant in different body tissues and so the time taken to receive an emitted signal is directly related to the distance (depth) from the emitting probe. The ratio of amplitude of the emitted and received pulses is called the transmission coefficient and is dependent of the impedance caused by the tissue. [70]

The received energy to the crystals can be displayed as A -, B - or M - mode, which correspond to different display methods to characterise the signal.

- A - mode imaging. This shows the **amplitude** of the as a measure of depth and reflected energy.
- B - mode imaging. Strength of the reflected signal is characterised as **brightness**.
- M - mode imaging. The strength and depth of the received signal is characterised by the **motion** of a curve of varying amplitude and curvature.

2.6.4 *In Vivo* experiments

Gait analysis provides a non-invasive and non-ionising measurement tool for *in vivo* knee kinematic analysis. Its origins in analysis of diseased knee date back to the 1970s [71], with more recent refinements allowing in depth simultaneous analysis of knee kinematics and kinetics [72, 73, 74, 75]. The major drawback of this technique is skin movement artefact. Skin markers placed over bony landmarks introduce inaccuracies related to the reconstruction and orientation of the underlying limb during execution of locomotion. Recent techniques have utilised mathematical solutions to refine the inaccuracy in establishing joint centre positions, however to date this form of analysis remains too blunt a tool for detailed analysis (Section 5.5.4) [76, 77, 78].

Open MRI scanning has allows for dynamic measurement of PFJ kinematics [79, 80],

but is limited to the native knee. Attempts to use MRI to measure knee kinematics following knee arthroplasty have been unsuccessful due to distortion of the images from metal artefact.

The use of individual radiographs to study knee movements was popularised in the 1980s and involved obtaining multiple sequential radiographs during movement but has subsequently lost favour due to ethically unacceptable radiation doses [81, 82].

Other plain radiographic techniques include Roentgen stereophotogrammetric analysis (RSA) which requires the insertion of <1 mm radio-opaque tantalum balls surgically inserted into the objects (eg. prosthesis and bone). Using two simultaneous radiographic projections the positional relationship of the two objects can be described (eg. bony movements or prosthesis migration) with a high degree of accuracy [83, 84]. The use of single and bi-planar fluoroscopic imaging to quantify PFJ kinetics has become increasingly popular recently, due to lesser radiation doses than those required for conventional radiographs and/or RSA. When comparing the native with replaced knee, descriptions have been restricted to the sagittal plane, mostly analysing tibiofemoral contact points and/or comparing changes in PTA in the native and replaced knee [46, 85, 86, 87, 88].

Summary

For a comprehensive assessment of patellofemoral kinematics following knee arthroplasty, movements in six degrees of freedom must be measured. This requires visualisation of the patella from full extension through to complete flexion of the

knee. Descriptions of patellar movements following implantation of prostheses have previously been restricted to sagittal plane kinematic descriptions, using fluoroscopy. To date no study has reliably measured coronal plane movements *in vivo*, of the patella following implantation of metal prostheses.

Whilst each imaging modality described here has been used in some way to measure patellofemoral kinematics, from an experimental design standpoint only some can be applied *in vivo* to assess movements of the patella, especially following knee arthroplasty. The position of the patella in the trochlear groove is dependent on the interaction of the bony and soft tissue restraints. The degree of constraint provided by bone or soft tissue alters as a function of height within the trochlear groove. In the flexed knee, when the patellofemoral joint reaction forces are maximal, the bony factors will likely influence the patellar position in the groove to a greater degree than the soft tissue stabilisers. However, in extension the position of the patella has been shown to be entirely dependent on which quadriceps muscles are contracting [89]. Furthermore, muscle balance during patellar movements is an important factor when comparing passive and active patellofemoral kinematics studies. To date, no reliable technique has been developed that can be applied to all patients (in the native and replaced knee) to allow analysis of coronal plane patellar movements during activities representative of normal daily life.

2.7 Clinical Problems

Anterior knee pain represents 30 - 50% of all orthopaedic out-patient referrals and is the most frequent cause of problems after knee replacement [12, 90]. Whilst the investigation of pain pathways and mechanisms is beyond the scope of this

thesis, effort is made in this thesis to investigate potential mechanical reasons for pain.

2.7.1 OA of the Knee including PFJ

The incidence of isolated symptomatic patellofemoral osteoarthritis has been estimated at 8% in women and 2% in men [9]. Radiographic studies have noted three patterns of wear, isolated patellofemoral, medial tibiofemoral, and medial tibiofemoral/patellofemoral. The patterns of wear in the patellofemoral joint medial, lateral and global.

Grading of Chondromalacia Patella

Chondromalacia of the patella is a general term applied to describe damage to the undersurface of the patella. It can be classified using the modied Shahriaree grading system [91].

- Grade 1.

Arthroscopic findings: Softening of articular cartilage; T1-weighted MRI findings: Partial width focal decreased signal areas of patellar cartilage on T1-weighted sequences, not extending to cartilage surface.

- Grade 2.

Arthroscopic findings: Blistering of articular cartilage with surface abnormality; T1-weighted MRI findings: Focal area of sharply margined decreased signal extending to the articular surface.

- Grade 3.

Arthroscopic findings: Cartilage fibrillation; T1-weighted MRI findings: Indistinct focal areas of decreased signal extending to the articular surface.

- Grade 4.

Arthroscopic findings: Full thickness cartilage ulceration; T1 weighted MRI findings: Full-thickness decreased signal abnormalities with associated subchondral bone changes.

Patellofemoral arthroplasty (PFA) is a salvage procedure for the treatment of isolated patellofemoral osteoarthritis. To date its use remains controversial. Whilst follow-up studies have reported excellent survivorship between 64% and 95.8% at 5 years [10, 11], failures predominantly arise due to progression of osteoarthritis to the tibiofemoral compartments [12]. Progression of osteoarthritis is not recorded when addressing disease in tibiofemoral compartments using primary medial unicompartmental arthroplasty [13].

Total Knee Replacement (TKR) has recognised complications which can adversely effect outcome [14]. Murray and Frost [92] reveal that despite the reported survivorship of 97% using revision as an end-point, 30% of patients report pain at some stage by seven years following operation. The source of pain in these patients is predominantly from the PFJ which represents the origin for the majority of complications following TKR [93, 94, 95].

The process of knee arthroplasty replaces the native joints with an artificial surfaces which in turn alters the kinematics of the knee [75].

2.7.2 Knee arthroplasty

Total knee replacement (TKR) and Patellofemoral Joint Replacement (PFJR) are standard treatments for end-stage osteoarthritis when conservative measures have failed [96]. Data from joint registries demonstrate increasing demand for these procedures. In 2004, Dixon et al reviewed the Department of Health statistics between 1991 and 2000 for primary and revision Total Knee Replacement and Total Hip Replacement (THR) [97]. The incidence of primary TKR doubled, with revision TKR increasing by 300%; primary THR increased by 18% whilst revision THR doubled. Based on these trends, the authors predicted a 63% rise in TKR to 54,000 primary TKR procedures and 47,000 primary THR by 2010. Figure 2.7 shows the 2010 National Joint Registry (NJR) data for England and Wales [98], comparing the results for TKR with THR.

There has been a progressive increase in the numbers of primary TKR operations performed between 2005 and 2010, increasing from 62,217 cases to 79,263 cases, showing earlier predictions of 54,000 procedures to be conservative [97]. It is known that compliance is not 100% for the NJR, and was lowest at the start of the register. The Swedish Knee Arthroplasty Register (SKAR) initiated in 1975 has 100% compliance by all Swedish institutions performing arthroplasty, and shows a similar trend to the UK data. Most notable are the statistics regarding younger patients [99]. In Sweden, knee surgery for osteoarthritis (i.e. high tibial osteotomy and arthroplasty) in patients less than 55 years of age has doubled during the past ten years, with the use of TKR increasing five-fold [100]. Similarly, Australian registry data supports this trend, showing a 40% increase in the number of knee

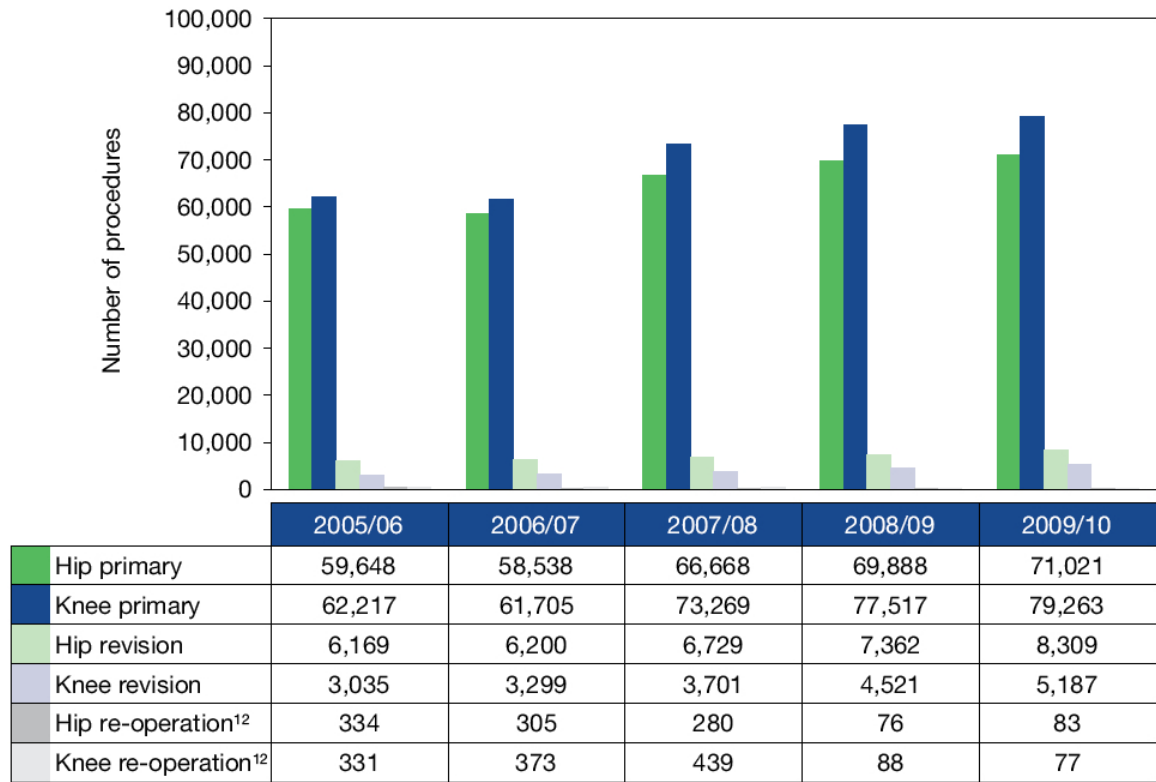


Figure 2.7: Showing National Joint Registry data on hip and knee operations from 2005 - 2010. From NJR 2010.

reconstructions over the last 5 years [101]. This trend is predicted to continue to increase in the future, with recent studies predicting the demand for primary TKR to increase by more than 670% by 2030 in the USA alone, with the revision rate predicted to increase by 600% [102]. In comparison, the projected demand for primary total hip replacement (THR) is less acute, reflecting more closely the predicted change in demographics over time. Primary hip replacement is predicted to rise 174% by 2030, with the revision rate expected at 137%, in line with ageing trends.

TKR is considered to be a successful operation with quoted survival rates of up to 98% at 20 years post-implantation, using revision as an end-point [103, 104, 105]. However, the use of revision as a measure of success has been questioned. The

circumstances influencing the decision to revise are often multi-factorial and the use of revision as a measure of success or failure of an implant has been widely shown to be a poor reflection of the patient experience [106, 107, 108]. From a patients perspective, the binary end-point of *revision* (failure) versus *no revision* (success) overlooks the spectrum of outcome experienced by individuals, and does not allow for identification of the early signals of later failure.

It is important to distinguish between survivorship and satisfaction. Previously the evaluation of a procedure was largely from the surgical perspective, concerning (a) the technical aspects of component positioning, and (b) the durability of the component fixation in terms of loosening and/or other complications. More recently studies have shown that the concerns and priorities of surgeons and their patients differ when considering the outcome of a surgical procedure [109]. The use of patient reported outcome measures (PROMS) is now widespread, their gravitas reflected in the recent implementation of the UK Department of Health policy ¹. In an outcome study comparing survival with satisfaction using the Oxford Knee Score (OKS) [111], around 30% of knee replacement patients were found to not be satisfied with the outcome of the procedure, and complained of moderate to severe pain, despite a quoted 97.5% survival rate (using revision as an end-point) at 7 years [92]. Similarly, two recent studies have compared the differences in patient satisfaction following TKR and THR, using simple satisfaction scores (eg. 1= '*very happy*', 2= '*happy*', 3= '*OK but not perfect*', and 4= '*never happy*') [112, 113]. These authors noted that whilst both TKR and THR had comparable success using revision as an end-point, the percentage of patients that were '*very happy*' was much higher in the THR group compared with the TKR group, 54% vs 4%, whilst the percentage of

¹From April 2009 the audit of all hip and knee joint replacements has been performed using PROMS such as the Oxford Hip and Knee Scores (OHS and OKS) [110].

‘*never happy*’ patients was larger for the TKR group than the THR group, 7% vs 1%. The potential reasons for these discrepancies are diverse and include, amongst others: patient perceptions and expectations, pain management and measurement bias. Investigation into potential causes of such profound differences between the groups is beyond the scope of this thesis. However, it is clear that despite recent advances in prosthetic design and implantation techniques, a sizeable proportion of post-operative TKR patients remain unsatisfied with the outcome. The reasons for this are not clear. Whilst PROMS provide valuable insight into whether there might be issues with certain prosthetic designs, they only serve to highlight problems within large groups. They are a relatively blunt tool for analysing function. The assessment of function of knee replacement designs can be performed using kinematic analyses, which have been shown to be effective at highlighting differences between groups with small numbers.

Contemporary prosthetic PFJR and TKR designs aim to restore normal kinematics, however, abnormal patellofemoral joint (PFJ) kinematics after joint replacement remain a recognised factor leading to post-operative complications [114].

2.8 Patellofemoral instability

Subluxation and dislocation typically occurs during 0 - 30° flexion and infrequently in the terminal phases of extension. The causes for this can be broadly classified into two groups [9];

Poor patellar engagement into the trochlear groove. When the knee is flexed, the patella is seen to sublux laterally in the early stages, and then either dislocates completely or returns to the apex of the groove more, in greater degrees of flexion.

This situation can occur with patellar dysplasia, with a high patella (patella alta) and with trochlear dysplasia. The consequence of patella alta, with or without trochlear dysplasia, is that the patella is not located in the proximal part of the groove.

Failure of Patella to stay in the trochlear groove. Despite being aligned properly in the early stages of flexion, the patella is seen to migrate from the apex of the groove. The major trend observed in the dysplastic patellofemoral joint is for the patella to move laterally. There are a small group of patients that dislocate their patellae medially, but they all tend to have undergone a procedure involving medialisation of the tibial tubercle and lateral release. The proposed factors that cause the patella to migrate laterally in flexion include; large valgus angle at the knee, trochlear dysplasia in the form of a shallow groove or deficient lateral buttress (lateral condylar hypoplasia) [115], and imbalance of the soft tissues about the patella. The medial soft tissues may be deficient, seen in the vastus medialis muscle following recurrent dislocations or wasting, and the lateral soft tissues may be too tight from fibrosis [9].

Surgical options for correction

There is no single procedure that can be applied to all cases of patellofemoral instability. The available operations can be grouped into bony and soft tissue reconstructions and/or proximal and distal re-alignments.

Bony reconstruction

Reconstruction of the trochlear groove can be attempted in a variety of ways [116]. Albee in 1919 described the method of trochlear osteotomy to create a lateral buttress

[117]. Alternatively, the trochlear groove can be deepened, by excavating bone from beneath the groove and building up the surface by the use of osteochondral flaps [118, 119, 10].

Distal realignment of the tibial tubercle is performed when there is poor engagement of the patella and can be applied to patients with poor engagement due to a dysplastic proximal trochlea and those patients with patella alta. Medialisation of the tibial tubercle has been described by Elmslie [120], which is most often employed to correct lateral subluxation of the patella caused by the tibial tubercle lying too far laterally.

Soft tissue reconstruction

The soft tissue tension around the patella can be adjusted by increasing the tension in medial structures and by performing lateral releases. The role of lateral release in the treatment of patellofemoral instability remains controversial. Dandy and Desai in 1994, reported the procedure to be effective provided there is no evidence of patella alta and there is no ligamentous laxity, with the procedure being more effective than tibial tubercle transfer or medial soft tissue tightening procedures [121]. Clifton et al (2010) performed a systematic review of literature regarding lateral release for patellofemoral instability and reported satisfaction between 30 - 100%. [122].

The medial soft tissues can be plicated or tightened by the use of fascia, tendon, imbrication or advancement of the vastus medialis tendon. This procedure can be employed on the rare occasion that there is only laxity of medial structures, but is more commonly performed as an adjunct to other procedures.

The Elmslie-Trillat procedure adjusts both bony and soft tissue structures, combining the lateral release procedure with medial transfer of the tibial tubercle and medial soft tissue plication [120]. Other popular operative alternatives include the Roux-Goldthwaite procedure, a distal patella tendon realignment procedure involving a vertical split of the patella tendon. The lateral half is then passed beneath the medial, and attached onto the tibia, pulling the patella medially.

Whilst there are a range of operations available as outlined above, opinion differs as to the most appropriate surgical principle that should be followed when correcting patellofemoral instability; those that address bony, and those that address soft tissue abnormalities. Certain authors advocate that the fundamental abnormalities requiring correction are bony in origin; that patella alta and trochlear dysplasia should be addressed prior to soft tissue balancing [123, 124, 125]. Contrary to this, other authors focus entirely on adjusting soft tissues with a recent attention being focussed on reconstruction of the medial patellofemoral ligament [126, 127, 128].

Whilst there are over 100 described procedures for the treatment of patellofemoral dysplasia and instability [20], there is no consensus as to the ideal method(s) of treatment. Previous studies have documented the relative strengths of soft tissue restraints around the patella, [129] however it remains unclear as to which anatomical structures or features or combination of structures are more important in preventing pathological subluxation of the PFJ. It is likely that certain anatomical features, such as a high riding patella, or lax soft tissues affect normal patellofemoral kinematics. It is possible that a hierarchy of anatomical features exists without which normal movements of the patella cannot occur. As yet this hierarchy has not been established.

2.9 Specific Objectives

1. To determine how the PFJ has developed throughout human evolution. This will give insights into PFJ anatomy and function.
2. To provide a detailed description of the 3D anatomy of the PFJ, with particular reference to the orientation of the femoral trochlear groove.
3. To develop a method of assessing 3D PFJ kinematics *in vivo* both in replaced and native knees.
4. To study the coronal plane kinematics of normal and replaced knees, as abnormalities in kinematics are likely to be source of symptoms.
5. To study the anatomical abnormalities that lead to abnormal coronal plane knee kinematics manifesting in subluxation.

Chapter 3

Phylogeny of the PFJ - An Evolutionary Perspective

3.1 Introduction

The basic structure of the knee has been conserved for over 300 million years. The knee of all terrestrial tetrapods and bipeds is derived from the analogous structure of the sarcopterian lobe-finned fish from the early Devonian period, alive 370 million years ago (Mya) [1]. The earliest fossil evidence of a load-bearing knee is that of *Ichthyostega* (360 Mya), in which the femur, tibia and fibula are all present, with the bicondylar shape of the distal femur already evident [38]. This feature is common to all knees in the human lineage, but is qualitatively different between species with different locomotive strategies.

In humans, normal knee kinematics is the result of a complex interaction between joint surface morphology and soft tissue constraints. Darwinian theory suggests that, via the process of natural selection, modern day knee morphology has evolved to best

suit its environment. However, this appears counter-intuitive when one considers the precarious, dislocation-prone arrangement of the PFJ; with valgus orientation, shallow trochlear groove and poor lateral trochlear buttress. During evolution, the adoption of an upright stance greatly altered both the internal loading environment and the kinematics of the knee. Hominin bipedal gait requires, in addition to loading in extension and flexion, loading during axial rotation of the knee, features not present throughout our non-bipedal ancestry. To date, there has been no specific report to chronicle the changes that brought about the precarious arrangement of the modern human knee joint.

Before describing the details of the human knee joint anatomy and kinematics it is helpful to gain a fuller understanding of the origins and development of the unique arrangement of this joint. It is possible that developmental trends might exist, illustrating important features of hominin anatomy that might be useful when considering TKR design.

Aim

The aim of this study was to describe the evolution of the PFJ in terms of its morphology (trochlear depth and sulcus angulation) and orientation (medio-lateral and antero-posterior) relative to the tibiofemoral knee compartments by analysing knee joints throughout human lineage.

3.2 Materials and Methods

3.2.1 Specimens

A collaboration with the Natural History Museums of London and Oxford, and the Department of Zoology, University of Oxford provided access to distal femora from the Devonian, Triassic, Jurassic, Cretaceous, Miocene, Paleolithic, Pleistocene periods to the modern day. Specimens included amphibious reptiles (eg. Hellbender), dinosaurs, shrews, tupaiae, lemurs, African ground apes, Lucy (*A. afarensis*), Turkana Boy (*H. erectus*), *H. neanderthalis* and modern .

Gait Type	Specimen	Million Years Ago	Period	N
Sprawling	Cryptobranchus	340	Devonian	2
	Iguana tuberculata	310	Permian	2
	Varanus flavascens,			1
	Caiman crocodilus,			2
	Crocodilus porosus,			4
	Tupaia minor	70	Cretaceous	1
Climbing	Perodicticus potto	63 - 70		1
	Pachylemur insignis,			1
	Palaeopropithecus ingens			1
	Callithrix jacchus	40		1
	Pongo pygmaeus	14		1
Erect	Gorilla	7		2
	Pan troglodytes	6	Miocene	2
	Australopithecus aferensis	3.5		1
Bi-pedal	Australopithecus africanus	2	Paleolithic	1
	Homo ergaster	0.5		1
	Homo erectus	0.5		1
	Homo neanderthalis	0.3	Pleistocene	2
	Homo sapiens	0		10
			Total	37

Table 3.1: List of specimens organised chronologically.

Gait Types

The specimens can be broadly classified into four gait categories, namely sprawling, climbing, erect and bi-pedal and are detailed in Table 3.1, organised chronologically. The earliest locomotion methods shown by the reptiles involved movement primarily with both the hip and knee flexed (*sprawling* gait) [130]. Normal locomotion is slow demonstrating very small excursions at the hip and knee joint which remain essentially in a fixed flexion position, and dependent on lateral flexion of the spine. A notable exception is in the rare event of reptilian running, which demonstrates larger flexion and extension movements at the hip and knee.

The subsequent gait pattern of our arboreal ancestors (*climbing* gait) demonstrates a large range of motion at the knee joint utilised for activities such as climbing and jumping. When stationary the knee is predominantly held in flexion, allowing for ground foraging and rest, whilst at other times the hind legs are used for jumping and swinging, in combination with a well developed tail. To accommodate these movements, the range of movement at the knee joint is from full extension to approximately 130 degrees of flexion.

During the period of the African ground apes and the running monkeys, the knee is seen to be held comfortably in extension during static stance. Although these species are predominantly quadrupedal, a considerable amount of time is spent ‘*erect*’ with the legs in full extension, due to their ability to walk upright for short periods of time and stand erect when reaching for food.

By the time of the Australopithecines and subsequent homonins, the majority of the time is spent upright on both feet, and the ‘*bi-pedal*’ gait strategy is well established. At this stage, the knee functions as a weight bearing, modified hinge joint.

3.2.2 Geometry Measurement

Computed Tomography (CT) data of each specimen was obtained using a 64 slice CT scanner (GE Light speed CT-1, GE Healthcare, Buckinghamshire, United Kingdom). Data sets were obtained using a helical scan with an inter-slice distance of 0.625 mm. Raw CT DICOM images were imported into MIMICS software (MIMICS Version 12.0.1, Materialise, Leuven, Belgium) for three dimensional reconstruction. Automated CT segmentation was performed to generate the three-dimensional (3D) geometries of distal femora for all specimens, examples of which are given in Figure 3.1.

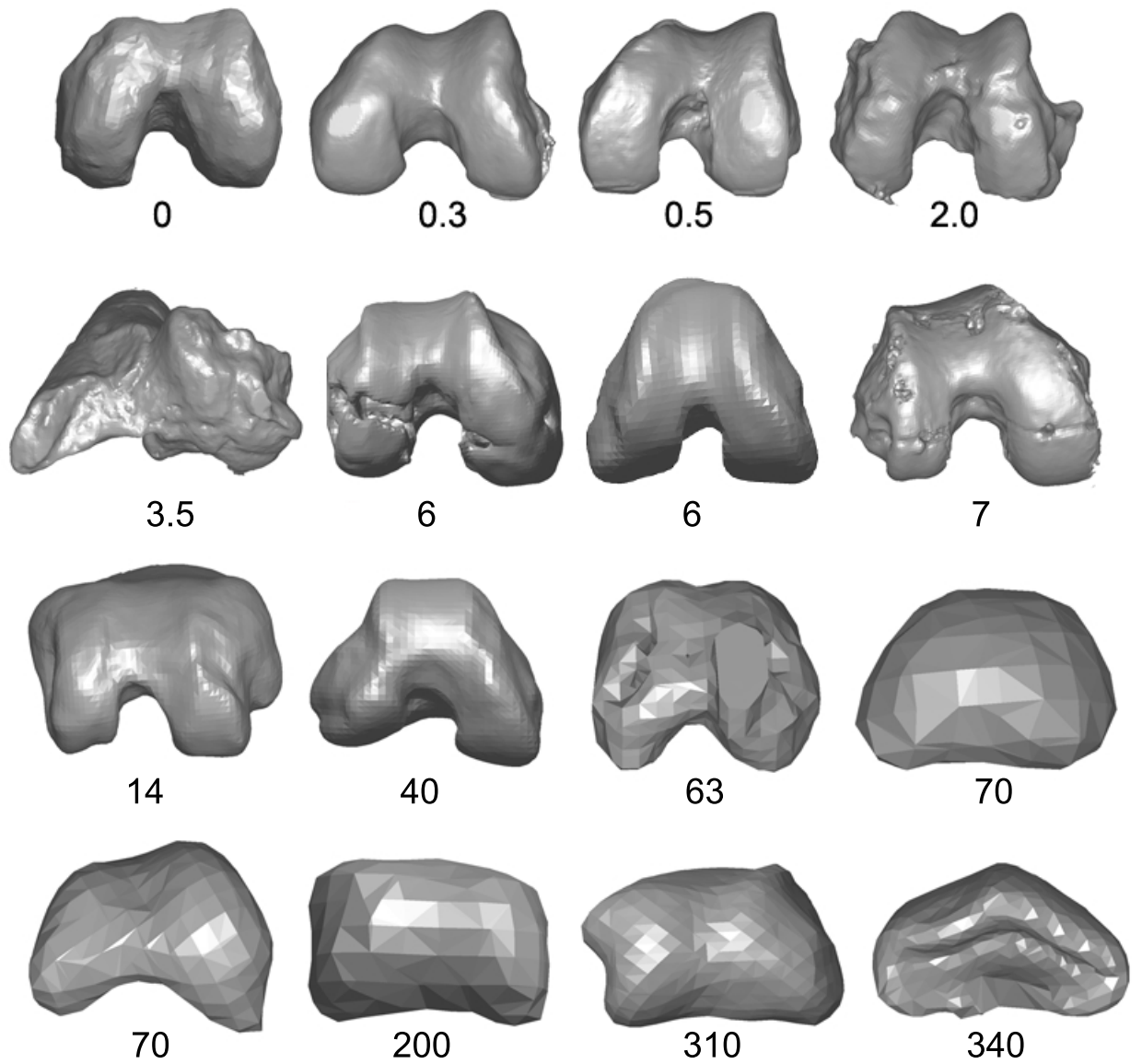


Figure 3.1: Showing distal femora ordered in Millions of years of age.

Alignment

The distal femoral geometry for each specimen was aligned using 3D manipulation software, Geomagic (Geomagic Studio 11, Geomagic Inc., North Carolina, USA). All measurements were taken relative to a coronal plane defined using the medial and lateral femoral condyles, and the shaft of the femur as shown in Figure 3.2a. Three

points were defined using the centres of spheres fitted to the posterior articulating surface of the *medial and lateral femoral condyles* and as certain older specimens no longer had a complete femoral head but intact shaft, the *midpoint of the femoral shaft* was used (centre of the femoral canal at an estimated height halfway between the proximal surface of the greater trochanter and the most distal part of the articular surface of the femoral condyles). A *closest-fit* algorithm available in Geomagic software was used to fit each sphere, the position and size of which is dictated by a surface boundary which is manually selected. The boundary of the posterior femoral condyle subtended a 90° arc from halfway between the anterior and posterior edges of the femur, to the most posterior extent of the posterior condyle. The femoral shaft sphere was dictated by a 2 mm cuff fitted to the mid-shaft of the femur.

A line drawn between the centres of the posterior condyles represented the posterior intercondylar line, PICL (Figure 3.2b).

The centre of the trochlea (C) was represented as the centroid of all points on the 3D reconstructed mesh within the boundary of the femoral trochlea. The trochlear area was selected manually (red area shown in Figure 3.3b) from which an automated centroid was calculated using Geomagic software (Figure 3.3b).

3.2.3 Measurements

The evolution of patellofemoral joint was measured in terms of;

- 1) Evolving depth of the trochlea; measured as the ratio of depth of the trochlea to depth of the distal femur in the axial plane. Following alignment of the specimen,

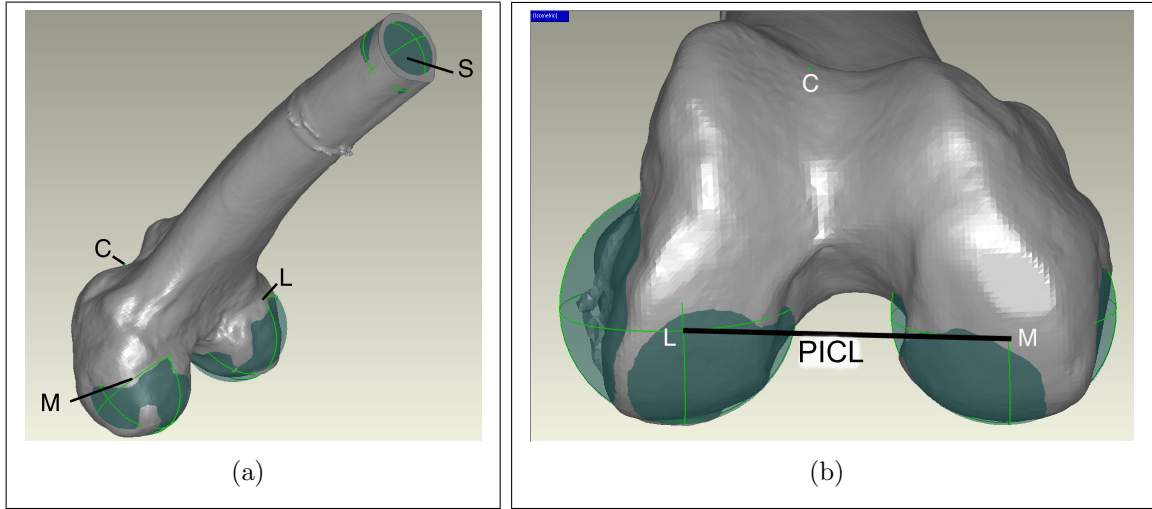


Figure 3.2: Oblique (a) and Distal (b) views of a Homo Neanderthal right femur. Showing spheres fitted to postero-medial (M); postero-lateral (L) condyles; midpoint of the femoral shaft (S); centroid of the femoral trochlea (C); and the posterior intercondylar line (PICL).

the depth of the trochlea was measured as the maximum AP distance from the most anterior point of the distal femur to the deepest part of the trochlear groove (A) and the depth of the distal femur was defined as the maximum AP distance occupied by the distal femur (B), shown in Figure 3.4.

2) Trochlear sulcus angles (S); the angle subtended by three points; the apex of the medial and lateral walls of the trochlea and the deepest part of the trochlear groove. This was measured at a height midway between the proximal trochlea and the roof of the intercondylar notch (Figure 3.5).

3) Displacements of the trochlea position throughout evolution; in the antero-posterior direction, (3a) and medio-lateral direction, (3b) (Figure 3.6).

3a) The antero-posterior position of the trochlea was defined as the aspect ratio of the perpendicular A-P distance from the PCL to the centroid of the trochlea (C) /

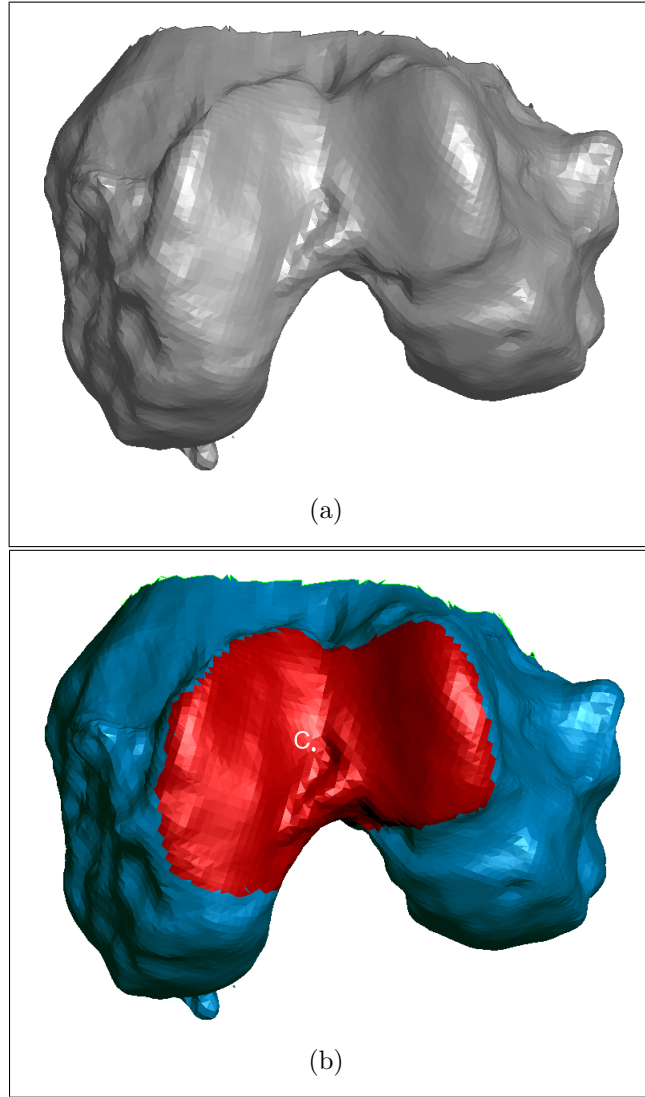


Figure 3.3: (a) Right distal femur of *H. ergaster* (b) showing the manually selected trochlear boundary (red) and automated centroid of the femoral trochlea (C).

the length of the PCL .

3b) The medio-lateral position of the trochlea was established as the horizontal displacement of the trochlear centroid. Using the trochlear centroid (C) and the two condylar sphere centres (M + L), a triangle was formed. The length of the lines CM and CL were calculated as the scalar Euclidean distances between the two points. The medio-lateral position of the trochlea was represented by the ratio of CM/CL, a

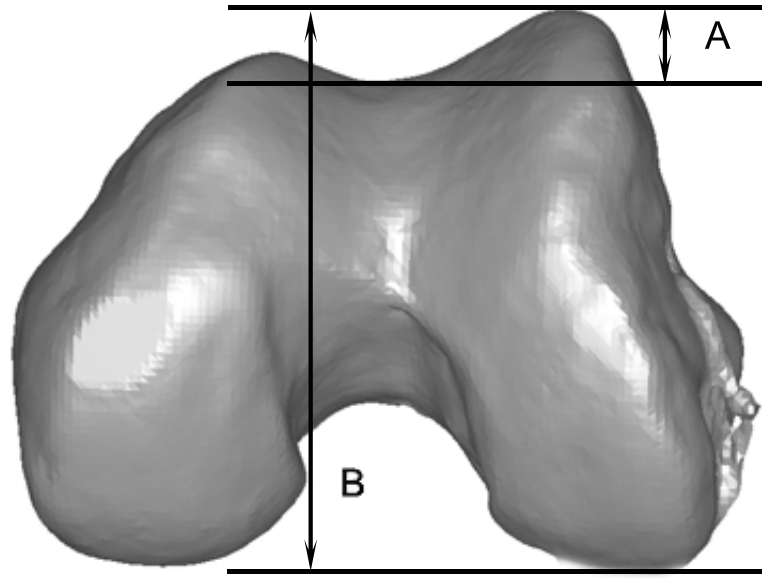


Figure 3.4: Showing measurements used for calculating the ratio of depth of trochlear groove (A) and depth of the distal femur (B). Ratio = A/B .

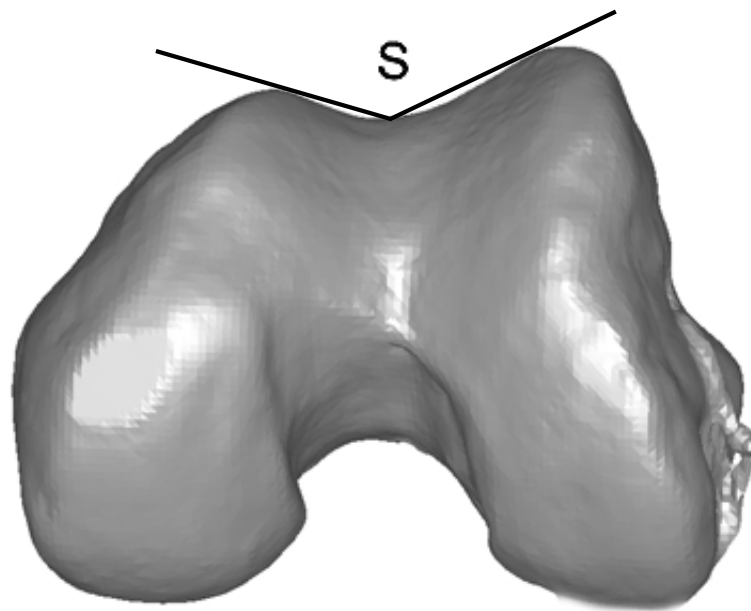


Figure 3.5: Showing measurement of sulcus angle, S.

value of greater than one representing a more laterally placed trochlea and a value of less than one, more medial.

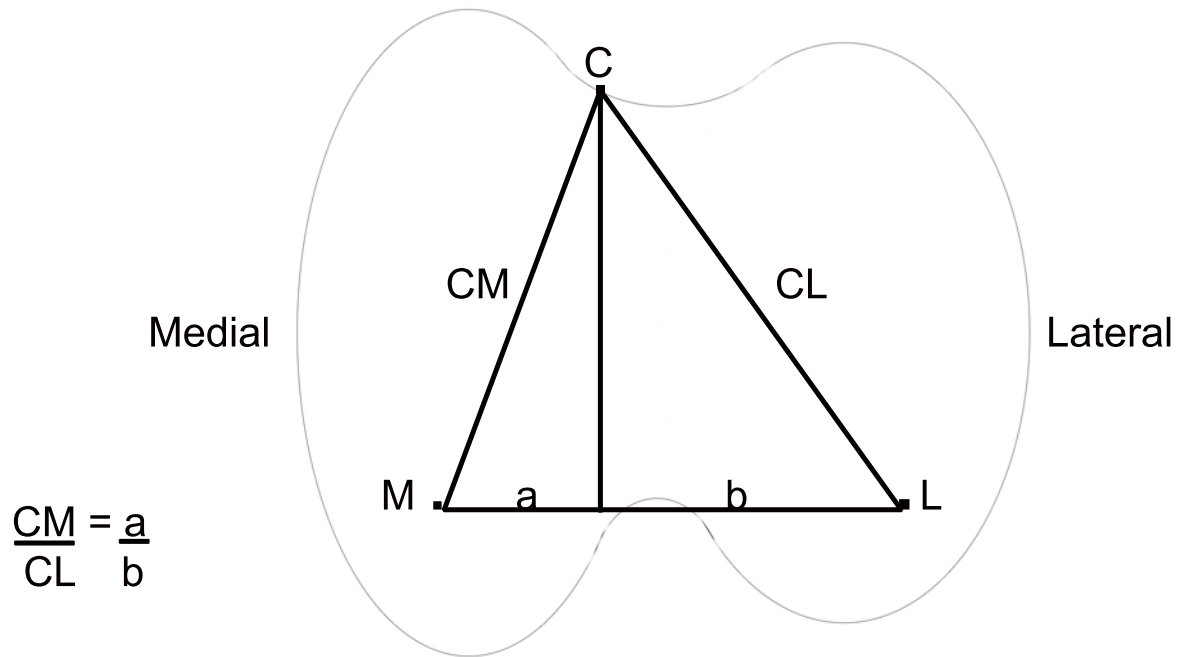


Figure 3.6: Showing measurements required to calculate displacements of trochlear groove. C = trochlear centroid, Posterior condylar line is represented by the distance between the medial (M) and lateral (L) posterior condyles.

Measurement of sulcus angles for all specimens were repeated by two independent observers at two separate time points (two months apart). Inter-observer error was assessed using correlation coefficients.

3.3 Results

3.3.1 1) Evolving depth of the trochlea.

Figure 3.7 shows the change in depth of the trochlea relative to the depth of the distal femur through evolution. From the amphibious reptiles in the Devonian period (340-410 Mya) to the tree dwelling lemurs of the Eocene epoch (38-23 Mya), the trochlea occupied a depth of less than one twentieth of the overall depth of the distal femur. A change in trend is most distinct from the period of the African Ground Apes (14 Mya) through to the Miocene period of running monkeys (6 Mya), when the trochlea occupied around a tenth the depth of the femur. By the hominin era of the Pliocene period there was a peak in the depth, with the trochlea around one third of the distal femur, followed by a decrease in depth to the present day depth of one tenth of the distal femur, roughly similar to the chimpanzee.

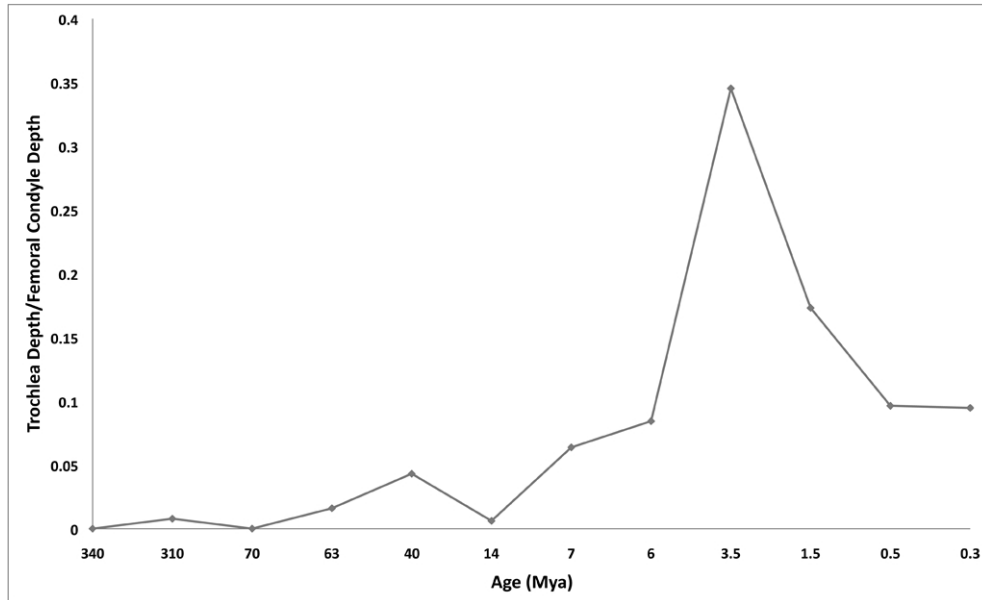


Figure 3.7: Showing the change in depth of the femoral trochlea relative to the maximum depth of the femoral condyle throughout evolution.

3.3.2 2) Sulcus angles.

From the quadrupedal reptiles of the Devonian period (360 Mya) to the early lemurs of the Miocene period, the sulcus angle remained relatively flat (150 - 175°). A dramatic change can be seen during the time of human arboreal ancestors with orangutan, *Pongo Pygmaeus* (140°), followed by the gorilla (108°). This acute sulcus angle was maintained during the early homonins (3.5 Mya) of *A. Aferensis* (108°), *H. ergaster* (109°) and *H. erectus* (120°) before a flattening out to the current angulation in *H. sapiens* of 138° (Figure 3.8).

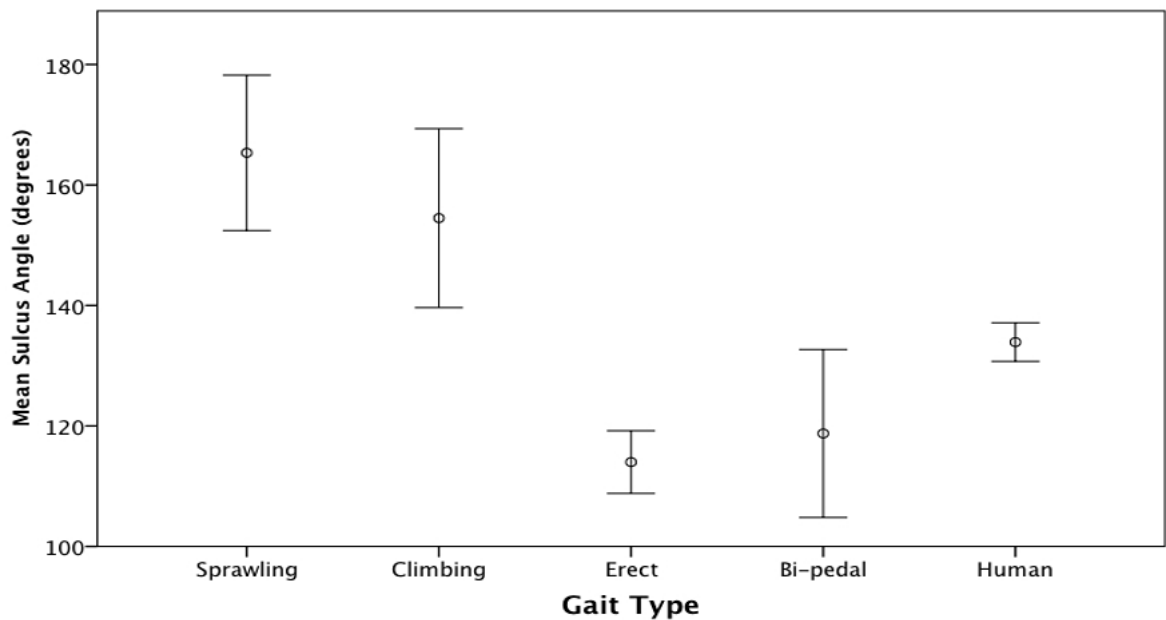


Figure 3.8: Showing the change in sulcus angle throughout evolution.

3.3.3 3) 3D migration of the trochlea.

Antero-Posterior position of trochlea

Aspect ratios (Antero-Posterior depth / Width) for the distal femur are demonstrated in Figure 3.9. The trochlear centroid remained anteriorly placed (AP distance greater than Medio-Lateral) until the hominin era (4 Mya) when the trochlear centroid moved more posteriorly, as the knee shape changed from being elongated in the AP plane to a more rounded shape with an aspect ratio roughly equal to 1 (Figures 3.7, 3.9).

Medio-lateral orientation of trochlea

Despite starting with a laterally orientated trochlea in the Devonian period (360 Mya), the majority of human evolution was spent with either a neutral or medially placed trochlea (310 - 4 Mya). From the time of the earliest hominins (*A.afarensis*, 3.5 Mya) the trochlea moves to a more lateral position, peaking in the Pleistocene period before moving to the less lateral, present day position. Notable exceptions to the general trend include the lemurs (63 Mya), the running monkeys (14-6 Mya) and the iguana, *V.flavascens* (13 Mya).

Inter-observer variability measured using correlation coefficients was 0.987. A value between 0.61 to 0.80 represents 'good' agreement and more than 0.80, 'very good' agreement.

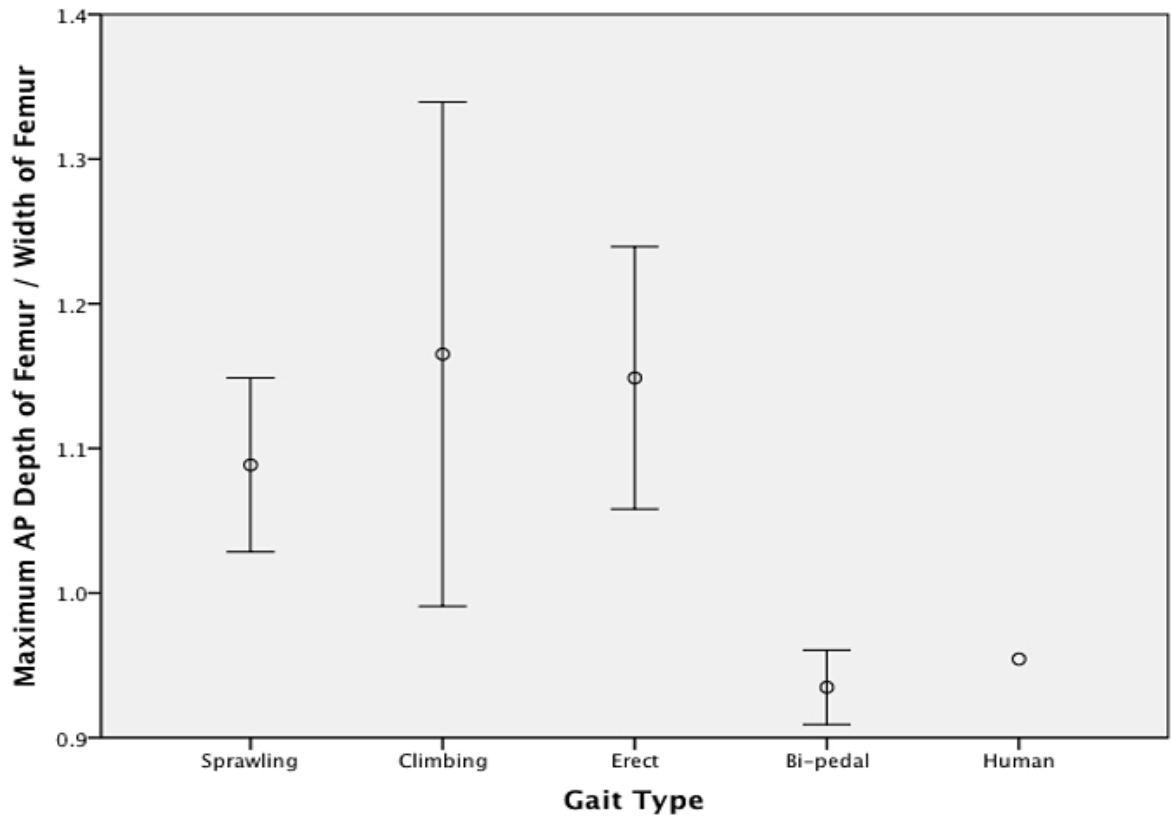


Figure 3.9: Showing the change in Antero-Posterior position of the trochlea throughout human evolution. A value greater than 1 represents an AP distance greater than the width of the distal femur.

3.4 Discussion

Previous studies have highlighted the evolutionary importance of where, when and why, man stood up [131]. Whilst other studies have compared the overall knee structure between different species, to date no study has described the evolution of the PFJ [1]. This study has chronicled changes in the shape and orientation of the patellofemoral joint throughout human ancestry. Included here are various different species spanning 400 million years, with their varying gait types including: sprawling, climbing, erect and bi-pedal.

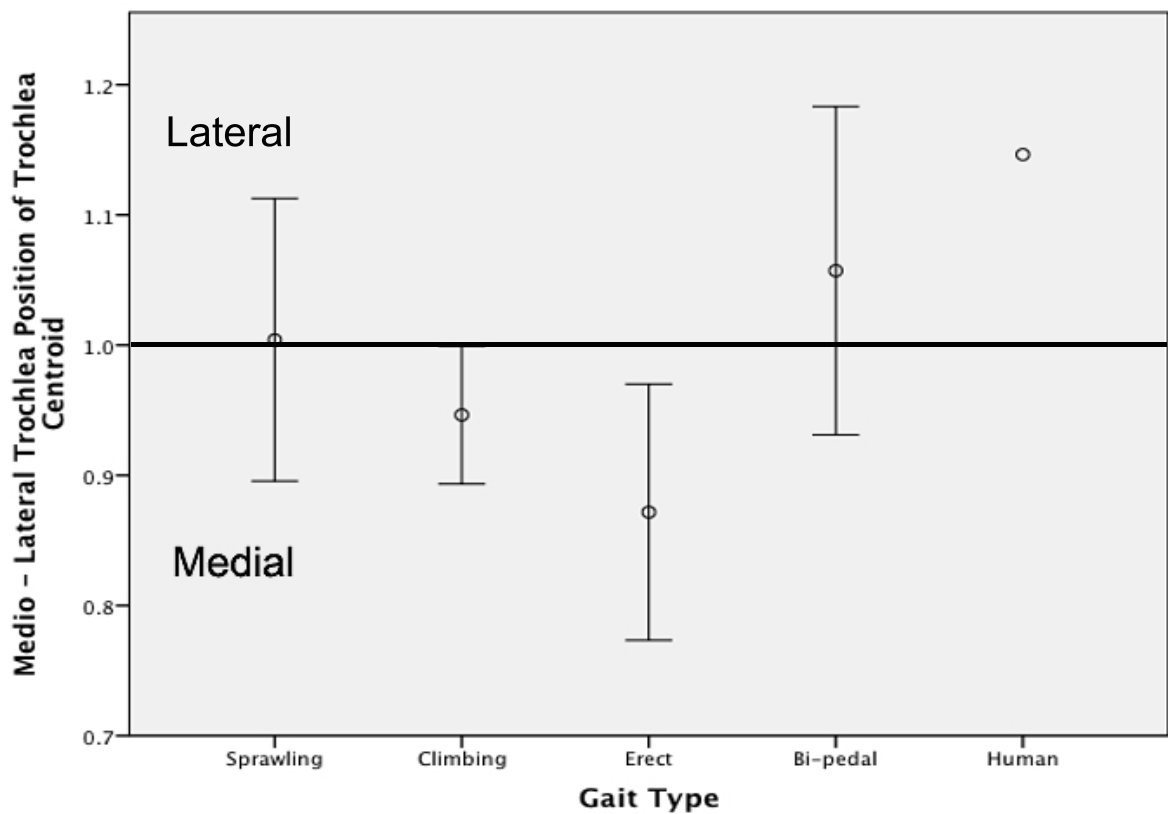


Figure 3.10: Showing the change in Medio-Lateral position of the trochlear groove throughout human evolution. A value greater than 1 represents a more lateral position; less than one, medial.

The data presented demonstrates that throughout evolution, at certain key points dramatic changes are noted in the developmental trends of the PFJ shape and morphology, which coincided with different loading environments within individual species. From a functional standpoint, the most poignant moment in our evolution is the adoption of upright stance and associated bipedal gait. With little or no weight bearing being performed by the arms, the entire body weight is carried through the hind limbs, through a range of 130° flexion, and as a consequence the loading environment in the knee joint changed dramatically.

Development of the trochlea

The development of the trochlea has followed a close association with gait type. During the time of our earliest ancestors (eg. Hellbender, 340 Mya) the trochlea was flush with the anterior femur. The sprawling gait of these amphibious reptiles rarely required extension beyond around 100° and these species demonstrate ‘semi-fixed flexion’ gait. By 13 Mya and the arrival of the African ground apes and running monkeys (gorilla, orangutan and chimpanzee), the trochlea is well formed and is accompanied by a bony patella. Interestingly, there is a similar arrangement to that of the apes shown 50 million years earlier demonstrated by certain species of lemur (with a cartilagenous patella) which is then not seen in intermediate species such as the marmoset and pongo.

This data reveals that the key shift in trend towards the current shape and orientation was with the African ground apes and especially, the gorilla. The gorilla is notoriously poor at balancing on two limbs, being top heavy with cumbersome arms (Figure 3.11). However, it is unique during this time in that a) it utilises a full range of movement at the knee from high flexion when seated to full extension during normal gait and b) is one of the earliest examples of periodic bipedal gait. It is at this point that this study has shown the deepest trochlear groove, with a sulcus angle of 109° , providing maximal stability whilst accommodating loads being transmitted to the femur via the bony patella. The relative flattening of the trochlea seen in hominins since the great apes is likely due to the recent change in loading conditions. When considering the associated change in aspect ratio of the distal femur recently to a more rounded shape, it is likely that axial tibio-femoral rotations

(dictated by the cruciate ligaments) played a significant role. This, in combination with the altered loading environment of the PFJ associated with bi-pedal gait, has lead to a more shallow trochlear groove.



Figure 3.11: Silverback gorilla with legs in extension.

Of particular interest in the data is the unique knee arrangement of the iguana, who has a PFJ which is more lateral, posterior and shallow compared to similar species with a sprawling gait, whose PFJ is characteristically medial and anterior. The explanation for this may lie in their unique gait strategies, which are in some ways akin to the hominin form [132]. The desert iguanas for example are traditionally considered to exhibit a sprawling gait pattern, but in reality are very able to sprint with full extension at the knee (Figure 3.12), a feature traditionally associated with bipedal gait. Similarly, the same trend seen in association with the lemurs (63 Mya)

coincides with the appearance of a patella which most likely evolved to increase the lever arm to offset the enormous compressive stresses generated via the quadriceps tendon during the hyperflexed phase of leaping [133].

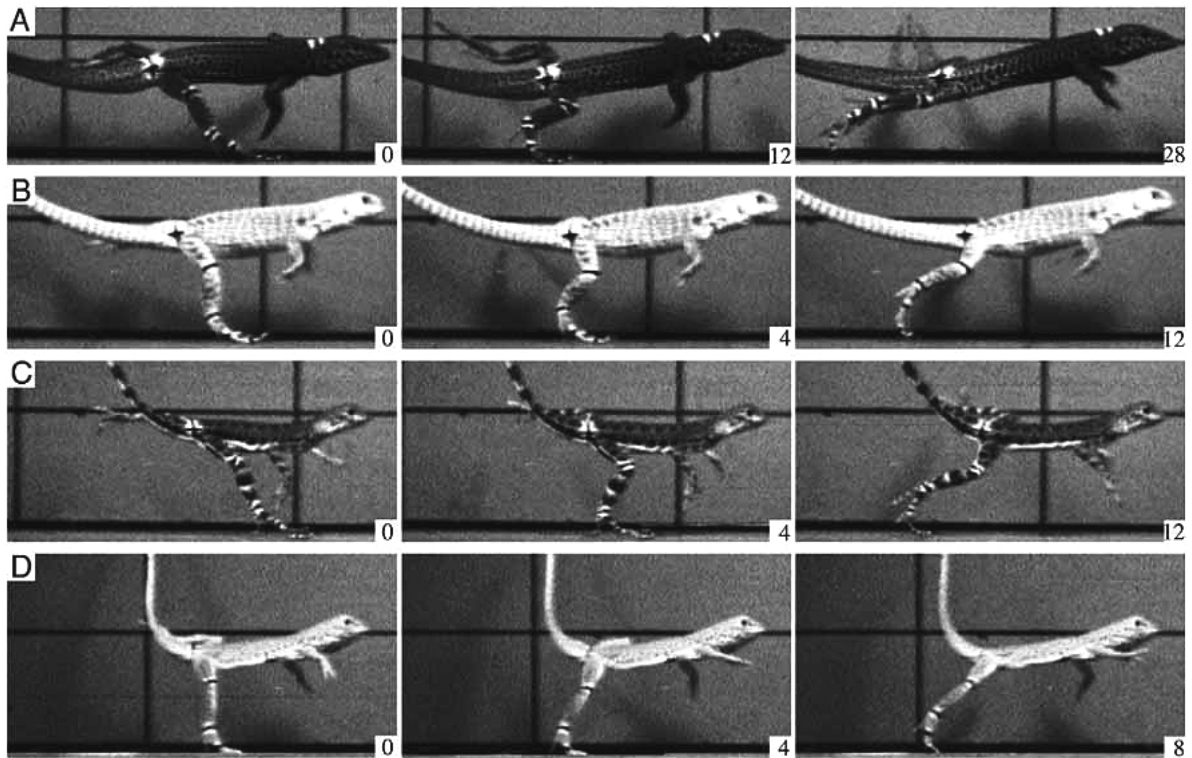


Figure 3.12: Reptilian examples of bipedal sprinting in full knee extension (from top to bottom: whiptail lizard, desert iguana, fringe-toe lizard, horned lizard). Irschick and Jayne, 1999.

Development of a more circular shape of the knee

Aspect ratios of the distal femur revealed that through evolution the shape of the distal femur (and proximal tibia) changed from being triangular shaped to a more circular knee joint. Whilst there is no specific data here regarding the development of the cruciate ligaments, it is likely that their influence on axial rotation altered the

forces through the knee joint throughout the range of flexion. The cruciate ligaments would have caused a radial re-distribution of the line of action of the forces and over time, via altered biomechanics, created a more rounded knee.

3.5 Limitations

The main drawback of this study is the lack of information regarding the individual specimens. Certain species including the hominins contain only one sample. It is unknown whether the individual represents a healthy specimen or perhaps suffered from a developmental abnormality, resulting in morphological dysplasia. Similarly, there is no indication of whether the specimen represents a young or old member of the species.

Whilst the data tracking the centroid of the trochlea points towards a recent ‘posteriorisation’ of the trochlea during the hominin era, it is important to remember that the actual position of the patellar tendon and quadriceps lever arm will be on the anterior aspect of the patella, not included in this study. Without patella specimens there can be no estimate of the extra distance from the bony trochlea to the patella tendon which would include; trochlear cartilage; patellar cartilage and patella.

3.6 Summary

Human gait requires flexion, full extension and rotation necessitating a compromise of stability. It has required anteriorisation and lateralisation of the patellofemoral joint, flattening of the trochlear groove and development of the cruciate ligaments to accommodate rotation. This study has highlighted the evolved hominin patellofemoral position and key geometries of the human knee, which might be useful for further

work.

Chapter 4

Bony and cartilaginous Anatomy of the PFJ

Chapter 3 described the evolutionary developments that have occurred throughout human ancestry leading to the modern day human position and orientation of the PFJ. This chapter investigates the shape and geometrical arrangement of the trochlea groove of the normal human patellofemoral joint. The aim is to understand further the path that the patella takes when articulating with the distal femur during knee flexion. The trajectory of the deepest part of the trochlear groove is described in normal human knees and compared with that of the prosthetic designs used in later clinical studies.

4.1 Introduction

Normal patellofemoral motion is dictated by the complex interaction of morphological constraints (bone and overlying cartilage shape) and surrounding soft tissue stabilisers (ligaments and tendons). Bony constraints have been shown to assert

a stronger influence on normal PFJ kinematics compared to soft tissue stabilisers acting on the patella after 20° flexion [134, 135]. Indeed, it is widely recognised that patella tracking is influenced by femoral trochlear geometry [136] with significant changes in PFJ kinematics noted following trochleoplasty [124]. Whilst traditional radiological assessments of trochlear morphology (defined as form and structure) and dysplasia have been made from bone [123, 137], the cartilage articular surface of native knee joints has been shown to have a different morphology from the underlying bone [62, 138]. In a study of patients with trochlear dysplasia and normal knees, van Huyssteen et al showed that the both the sulcus angle and contour of the cartilage trochlea was different from the bone in all cases [5].

Whilst there is a wealth of kinematic data available from cadaver and *in vivo* studies comparing normal and post-arthroplasty patients (Section 2.5) and the importance of the trochlear morphology in knee arthroplasty is well recognised [139], few studies have investigated the differences in the trochlear morphology of normal and replaced knees. Previous authors have attempted to describe the trochlea orientation by applying mapping the circular path of the patella through the trochlear groove during flexion [140], and by applying cartilage maps to the joint surface [141], however, few studies have addressed the path of the apex throughout the flexion arc [43].

The intention of this study was to compare directly the trajectory of the patella during flexion by comparing the nadir (lowest point) of the native bony *and* cartilaginous trochlea with those of prosthetic TKR and PFJR designs. In the context of this thesis, knowledge of the trajectory of the trochlea groove will prove useful when later considering movements of the patella during flexion. It is likely that the shape of the trochlea influences patella position during flexion, and so is an important

consideration when attempting to replicate the normal anatomy in prosthetic design (Section 8.2).

4.2 Materials and Methods

4.2.1 3D Trochlea Geometries

Native Knees

Retrospective analysis of MRI scans of normal subjects (n=20, 10M:10F, Mean age 29 yrs (range 15-46) taken with the leg in extension was performed. Subjects with normal MRI scans of the knee were selected at random from the database at the host institution. Exclusion criteria included any previous surgery, developmental abnormality, known dysplasia or reported anatomical abnormality. Sagittal, coronal and axial MRI scans were acquired using either 1.5 T or 3.0 T MRI scanners (Siemens Magnetom, Phillips, Guildford, UK) at the host institution. Minimum slice thickness for sagittal scans was 1.5 mm. Semi-automated segmentation was performed to acquire cartilage and bone datasets for each knee using the LiveWire algorithm available in MIMICS software (Version 10, Materialise Interactive Medical Image Control System, Mimics, Leuven, Belgium, 2009). Following manual selection of areas of cartilage, the LiveWire algorithm highlights localised areas of equivalent image intensity to directly map cartilage boundaries. Although still requiring manual user input, the technique represents a significant advance in comparison to previous manual segmentation techniques. 3D reconstructions were created from segmented data sets for cartilage and bone on the distal femur and patella.

Prosthetic Geometries

Bilateral 3D geometries of the three types of femoral components used in the later clinical study (Chapter 6) were acquired from the component manufacturers. The TKR designs included the AGC and Vanguard prostheses (Biomet Inc, Warsaw, Indiana, USA). The prosthesis used for the isolated patellofemoral joint replacement (PFJR) group was the FPV (Wright Medical Technology Inc, Arlington, TN, USA). Antero-posterior views of the prostheses can be seen in Figure 4.1.

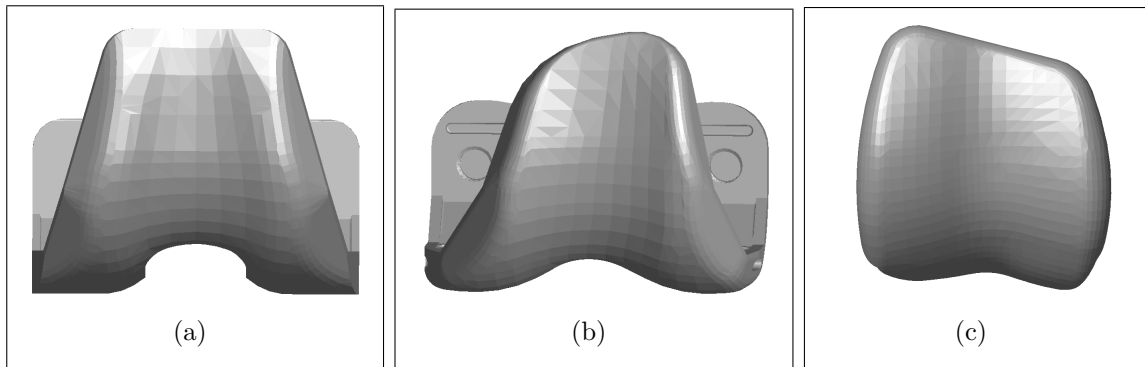


Figure 4.1: Antero-posterior projections of 3D femoral component geometries obtained using Geomagic software; (a) neutral AGC TKR (b) left Vanguard TKR and (c) right FPV PFJR.

4.2.2 Measurement of Patella Trajectory

A trajectory is defined as the path that a moving object follows through space as a function of time. The position of the trochlear groove nadir from superior to inferior was considered a guide to the position of the apex of the patella in a normal patellofemoral joint and therefore as a surrogate marker of the trajectory of the patella during flexion. Whilst this does not account for any subluxation due to soft tissue constraints, it does reflect how the patella might position itself anatomically during flexion within the trochlear groove.

The medio-lateral location of the deepest part (apex) of the trochlea groove varies as a function of height within the trochlea, therefore measurements of the position were taken at five reproducible heights. These heights correspond to the patella contact position on the femur at flexion positions of 20° (engagement), 30°, 45°, 60° and 90° [27] (Figure 4.2), used for the later described kinematic studies, Section 6.2.5.

1. 0.5 cm distal to top of the trochlea cartilage.
2. 0.5 cm proximal to the roof of the notch.
3. Midway between lines 1 and 2.
4. Midway between lines 1 and 3.
5. Midway between lines 2 and 3.

The 3D geometry for each distal femur was aligned in Geomagic software (Geomagic Studio 11, Geomagic Inc., North Carolina, USA) using the technique described in Section 3.2.2. Prosthetic geometries were aligned to manufacturer’s intended specifications. In the coronal plane, the horizontal distance between a vertical line intersecting the most lateral tip of the lateral condyle to the apex of the trochlea groove was measured for each knee at each of the five heights outlined above. The measurement technique is further detailed in Figure 4.3. For the prostheses, instead of using the lateral condyle as a reference point, following alignment, a vertical line intersecting the most lateral edge of the prosthesis was used. This measurement method could be used on any knee, regardless of side.

The trajectory of the trochlear groove was plotted as medio-lateral displacement versus flexion angle for the following five surfaces; (i) bone, (ii) cartilage, and the

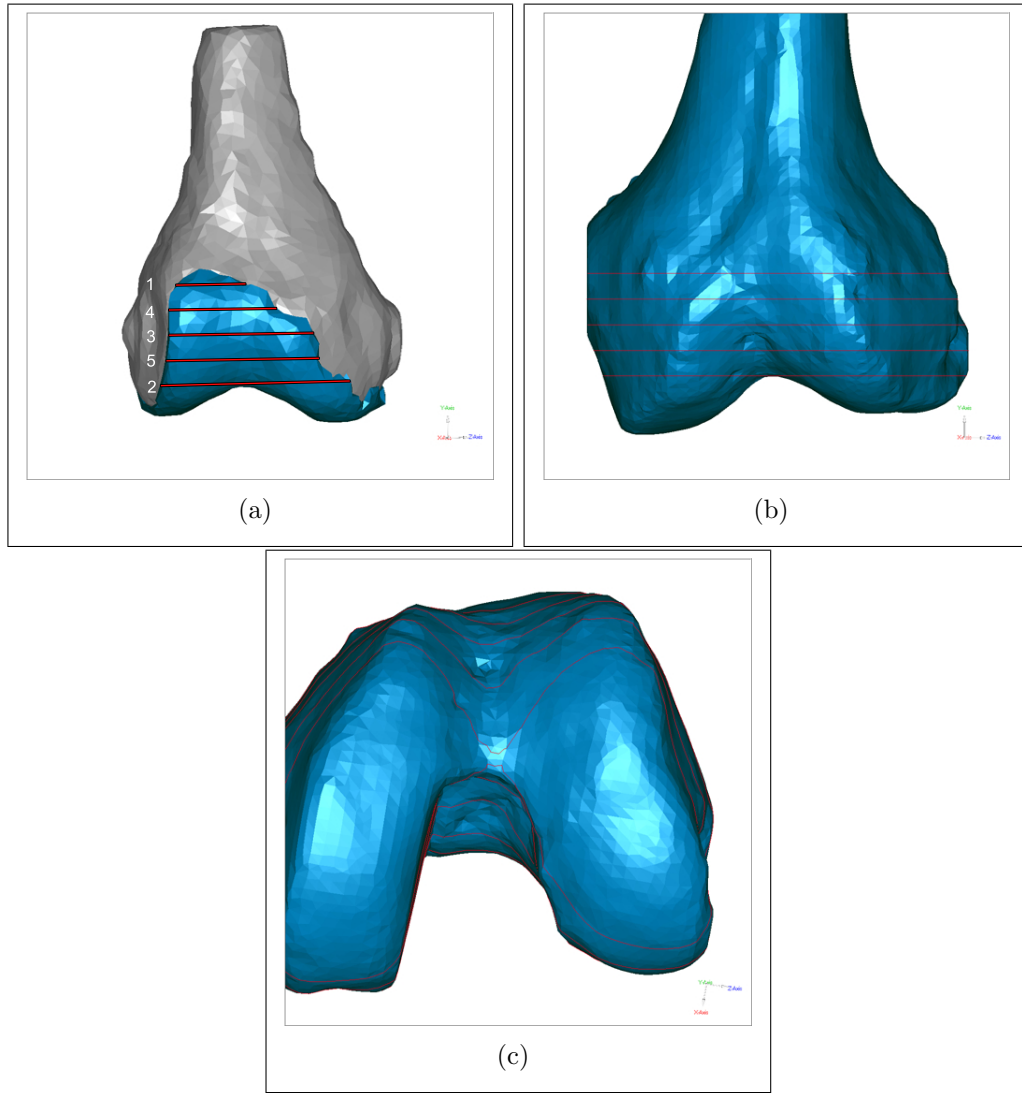


Figure 4.2: Showing the five levels at which apex positions were measured for (a) cartilage and (b) bone. (c) shows distal view of levels used.

three prostheses outer surfaces; (iii) AGC, (iv) Vanguard and (v) FPV to assess for similarities in the orientation of the trochlear groove.

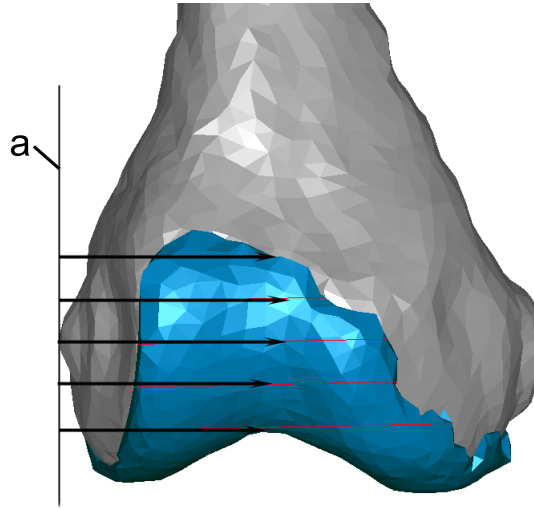


Figure 4.3: Showing theory of measurement technique used to plot displacement of trochlea groove from lateral condyle (a) at 5 different heights.

4.2.3 Data Analysis

Whilst qualitative analysis gives an indication of similarities/differences in the *direction* that the patella is directed during flexion, further exploration was required to assess whether there were any significant differences in the *magnitude* of the displacement of the position of the trochlear groove in the replaced knee compared with normal bone and cartilage.

Exploratory data analysis was performed to establish whether there were differences in the magnitude of the trochlea groove displacement between the five surface types (bone, cartilage, AGC, Vanguard, FPV) using the Kruskal-Wallis analysis of variance (ANOVA) test. The data were considered non parametric due to the limited number of prosthetic geometries. This revealed that there were significant differences between the mean movements between the groups ($p < 0.001$) across all ranges of flexion. Post-hoc non-parametric tests were then used to assess for significant

differences in the displacement of the apex position between knee types (cartilage vs bone, cartilage vs prostheses, bone vs prostheses) at each stage throughout the flexion range (0° to 20° , 20° to 45° , 45° to 60° and 60° to 90°) using the Mann-Whitney test. Statistical tests were performed using SPSS version 18.0 (SPSS Inc, Chicago, IL, USA).

4.3 Results

4.3.1 Direction of movement

Figure 4.4a shows the distribution of trajectories of the patella for bone whilst Figure 4.4b shows the range of distribution of trajectories for the cartilage layer. Of note is the lateral direction of both the bone and cartilage trajectories between 0° and 90° , followed by a medial movement between 60° and 90° .

The mean displacement of trajectory over the entire flexion range for the bony trochlea was 1.03 mm (SD 2.69 mm), and mean 0.62 mm (SD 2.60 mm) for cartilage.

Figure 4.5 compares the mean bone and cartilage trochlear trajectories with the trajectories of the three prosthetic implants used in the clinical study (Chapter 6). The lateral trajectory direction shown by the bone and cartilage is not replicated by the arthroplasty groups. The AGC prosthesis has a neutral trajectory down the trochlea without medial or lateral deviation whereas both the Vanguard TKR and the FPV PFJR exhibit a medially orientated trajectory. Whilst the FPV is seen to be orientated more medially than the Vanguard through to 60° , the FPV trochlear groove trajectory undergoes a change in direction medially between 60° and 90° of knee flexion.

4.3.2 Magnitude of movements

Table 4.1 shows a comparison between trochlea types at the different stages of flexion.

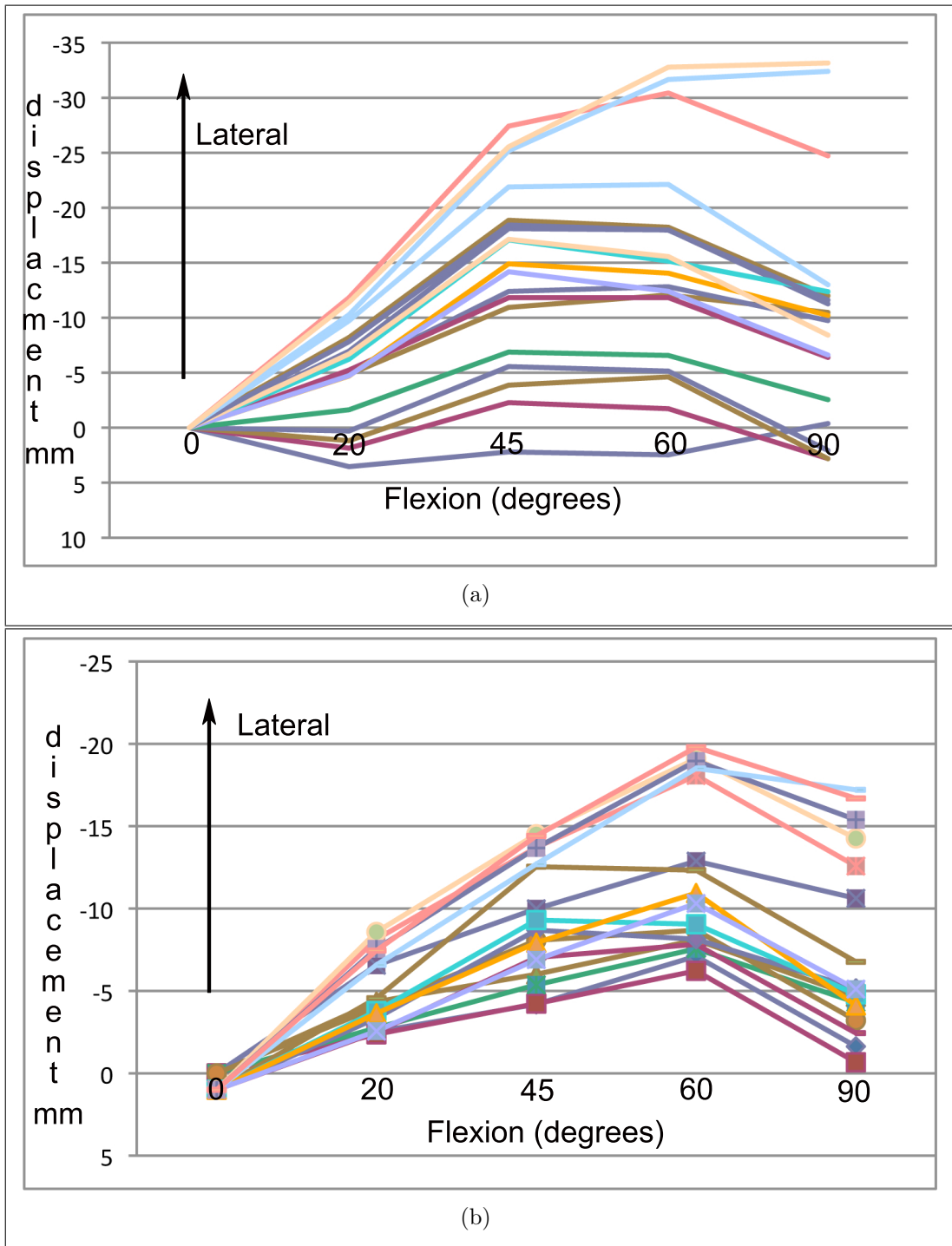


Figure 4.4: Showing all subjects' change in position of the apex of the trochlea groove versus height for (a) bone and (b) cartilage (1 = top, 5 = bottom).

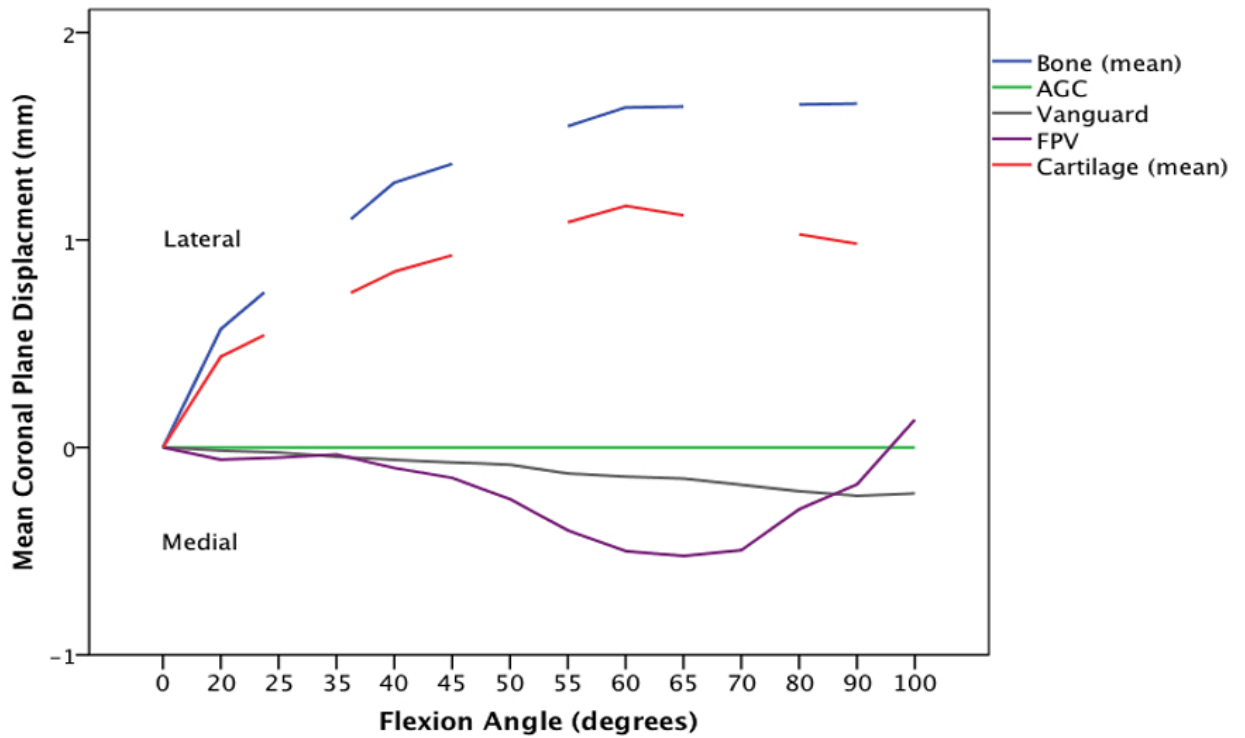


Figure 4.5: Showing the trajectories of the apex of the trochlea groove in normal bone, cartilage, PFJR and TKRs.

When comparing the magnitude of the displacements for bone compared with cartilage, there was no significant difference in the sizes of the movements across the entire flexion range, from extension to 90° ($p = 0.73 - 0.98$). This similarity in the magnitude of movements was only replicated when comparing the cartilage movements with those of the AGC prosthesis ($p = 0.14 - 0.74$). In all other comparisons of bone versus Vanguard, bone versus FPV, bone versus AGC, cartilage versus Vanguard and cartilage versus FPV, there were significant differences seen at points across the flexion range ($p < 0.05$).

Differences in trochlea groove position					
	Angle range	0° to 20°	20° to 45°	45° to 60°	60° to 90 °
Comparison of types					
Cart	Bone	p=0.754	p=0.731	p=0.754	p=0.975
Bone vs	AGC	p=0.011	p=0.166	p=0.05	p=0.411
	Vanguard	p=0.011	p=0.166	p=0.05	p=0.365
	FPV	p=0.007	p=0.131	p=0.011	p=0.166
Cart vs	AGC	p=0.74	p=0.14	p=0.373	p=0.74
	Vanguard	p=0.775	p=0.005	p=0.133	p=0.775
	FPV	p=0.775	p=0.003	p=0.033	p=0.319

Table 4.1: Summary table comparing magnitude of movements throughout the flexion range for the 5 trochlea types: Bone, Cartilage, AGC, Vanguard and FPV. Similar movements (not significantly different) are shown in bold.

4.4 Discussion

It is likely that the path of the patella during knee flexion is influenced by the orientation of the trochlea groove, and as such knowledge of the shape and orientation of the trochlea groove is critical when considering potential causes of anterior knee pain, especially in the post-operative knee replacement patient. It is possible that abnormal movements of the patella might cause pain via abnormal loading of the soft tissues.

The aim of this study was to describe the orientation of the trochlear groove in both normal and replaced knees. The bone and cartilage trajectories of the trochlea groove of 20 normal patients was compared with that of two commonly used TKRs, the AGC and Vanguard femoral components, and that of the FPV PFJR.

In normal knees, for most of the arc of flexion the trajectory of the trochlear groove apex was orientated progressively laterally between 0° and 60° for both cartilage and

bone (Figure 4.5). Between 60° and 90° whilst the mean bony trajectory followed a more neutral, central pathway, the cartilage trajectory was seen to move medially. Neither of these trends were reproduced by any of the knee prostheses. Indeed far from being a predominantly lateral directed trochlea groove, the Vanguard and FPV have a *medially* orientated trochlea, whilst the AGC TKR showed a neutral straight path (Figure 4.5). This observation would imply that if the patella were to follow the orientation of the underlying trochlea groove, then the direction of displacement in the replaced knee is opposite to that of the native knee.

Paired analysis of the change in trochlea apex position between the five positions of flexion, in all five types of knee revealed that only the AGC prosthesis with the neutrally orientated trochlea groove was not statistically different from the native cartilage trochlea. In all other combinations of bone versus prosthesis and cartilage versus prosthesis, the trochleae showed statistically different arcs. This further emphasises the inability of the replaced knee to mimic the normal anatomy.

The human knee joint is an example of a well balanced system of interacting bony and soft tissue components. Slight alterations in some part of the system will potentially unbalance the system as a whole. Previous authors have debated for over 30 years the multiple simultaneous manoeuvres that occur in the tibiofemoral and patellofemoral joints during knee flexion [27, 142, 143]. The spiral mechanism of the knee joint incorporates axial rotation at the tibiofemoral articulation. As the soft tissues of PFJ complex bridge the tibiofemoral joint (from quadriceps of the femur to the tibial tubercle), the movements at the tibiofemoral joint will affect the loads in the patellofemoral joint and the two cannot be considered in isolation. Any deviation of a prosthesis from normal PFJ anatomy has the potential to alter the patellar

position, a situation which can only be compounded during repetitive movements of the knee. The normal strain in soft tissues surrounding the PFJ will be increased or decreased, altering the movements of the patella and/or causing pain.

The closest performance of a prosthesis in this study to that of the normal knee was that of the AGC TKR, with no significant differences noted when comparing the changes of the trochlea apex position throughout the flexion arc. This finding is in keeping with the clinical success of the prosthesis with a reported cumulative survival rate of the femoral component of near 100% at 20 years [144]. Whilst revision as an end-point does not give any idea of patient satisfaction or anterior knee pain, it is an accepted measure of success (Section 2.7.2). Of note was the closer results of the prostheses' measurements to the cartilage than to the bone. The data from this study would imply that future femoral components should be designed to represent the articular cartilage rather than the underlying bone.

Limitations

Measurements taken from the prostheses were performed here in accordance with manufacturers guidance. This entailed alignment of specimens in Geomagic, to the engineering specifications. It is likely that in the clinical setting the alignment of the prosthesis varies, dependent on uncontrolled variables including correct placement of femoral drill guides and surgeon's choice. Ideally, future study would include a direct comparison between pre-operative, native femoral geometry and post-operative prosthetic geometries. This study has only measured prosthetic geometries used in later studies of this thesis. Further work is needed to see whether similar results are obtained for alternative prostheses. All subjects and prostheses were considered

to be standard adult size. No attempt was made to scale movements according to dimensions of the individual subject's distal femur.

4.5 Summary

These data suggest that the shape of articular surfaces of the distal femur might be important in controlling PFJ kinematics. It is possible that abnormalities in trochlea shape may lead to abnormal patella kinematics. Further investigation is required to assess whether during knee flexion, the pathway of the patella follows the trajectory of the underlying trochlea groove in normal and replaced knees. Should the patella follow the trochlear groove orientation, then it follows that the trochlear groove orientation in replaced knees should mimic that of the normal knee which, in the current prostheses, is not the case.

It is likely that anterior knee pain following knee arthroplasty is secondary to an unfamiliar loading pattern enforced through the host bone and soft tissues, dictated by the foreign shape of the prosthesis.

Chapter 5

PFJ Kinematics: MAUS Development and Validation

5.1 Introduction

The results of the previous chapter (Chapter 4) have demonstrated differences in shape and orientation of the PFJ between the native adult human knee and current knee prostheses. As function follows form, it is likely that movements of the patella will be influenced by the shape of the distal femur. As a consequence, the patella kinematics of the adult native knee may well be different from that of current prostheses.

Previous studies have extensively investigated tibiofemoral and patellofemoral kinematics in the normal knee and following arthroplasty, both *in vivo* and *in vitro*. The majority of reliable *in vivo* kinematic descriptions are restricted to the sagittal plane and have repeatedly shown marked differences between normal and replaced knees, and also between different types of knee replacement [46, 145, 146, 147]. These

differences, however, only partly explain the range of outcome observed with different knee replacements. It is likely that coronal or transverse plane patellofemoral kinematics are also important, and may be more significant when considering causes of post-operative pain. However, the radiological measurement of coronal plane PFJ kinematics after knee replacement presents numerous challenges, many of which are considered in this chapter. In addition, there is considerable variability in the post-operative position of joint replacement components which adds a further confounding factor to the interpretation of results. To obtain a fuller understanding of the range of outcome following knee arthroplasty, information on coronal plane kinematics is required.

This chapter details the development of a novel technique for the measurement of coronal plane patellofemoral kinematics. Importantly, this technique was designed to be applicable to both normal *and* replaced knees. A brief background of the experimental options is given and the justification for the selected imaging methodology is provided.

The first step in establishing a new method for coronal plane measurement was to review and consider the imaging and experimental techniques available.

5.2 Methodological Considerations

5.2.1 *In Vitro* versus *In Vivo* Studies

When considering intricate biomechanical movements (eg. subtle axial rotations of the tibia/femur during flexion) or situations where the investigation of interest is difficult to control or measure *in vivo* (eg. patella kinematics following knee replacement), the use of cadaveric or post-mortem specimens has become commonplace [148, 149, 150]. During such kinematic assessments of the knee joint, the experimental procedure usually aims to represent simplified physiological loading in order to simulate realistic positions of the femur, tibia and patella. This has traditionally been achieved via the application of load through the quadriceps tendons following intra-medullary fixation of the femur and the tibia to a rig [149, 151, 152]. During everyday weight-bearing activities, the kinematics of the knee joint are influenced by ankle and forefoot position and the subtle antagonistic action of individual quadriceps and hamstring muscles producing balanced rotations within the knee. The majority of cadaveric studies simulate knee extension by only applying forces through the quadriceps mechanism, whilst others provide a single constant force across all the hamstring muscles [153]. These studies have the methodological drawback of not addressing the antagonistic action of the individual hamstring muscles, an essential force when considering physiological rotations within the knee [154].

In vivo measurements of everyday activities such as squatting, normal gait and step activity provide relevant kinematic data [155]. Indeed the leading activities in which anterior knee pain is experienced following knee replacement arthroplasty include

walking, shopping and climbing stairs, dressing and getting out of a bed or a chair (in the squat position) [156]. During these activities the leg is predominantly in the stance phase of the gait cycle, implying that the range of motion at the ankle is considerably less than that of the hip and knee. It follows that, when designing a study to investigate pain, one should mimic these activities wherever possible.

Relevant kinematic study of the PFJ should be 1) an *in vivo* study with 2) the subject upright, weight-bearing through their feet, performing an activity of daily living.

5.3 Choice of Imaging Modality

Methods of measuring kinematics of the patella in the native knee are well established [9, 61, 145, 157]. Following implantation of a knee prosthesis however, certain movements, such as coronal plane patellar displacements become more difficult to measure.

The potential use of different imaging modalities available at the host institution for measuring coronal plane movements of the patella were evaluated using a cadaveric human knee. Investigations were performed as outlined below before and after implantation of a size 7 Scorpio TKR, without the patella re-surfaced (Stryker, Mahwah, New Jersey, USA).

5.3.1 Radiographs and Fluoroscopy

Three dimensional patterns of patellar motion can be reconstructed from uni-planar and bi-planar image matching techniques, using registration algorithms relating 2D images (plane radiograph or fluoroscopy) to a 3D model derived from CT or MRI data. Whilst this has been shown to be effective in the native knee, there has been limited success *in vivo* following knee replacement [158].

Methods

Standard antero-posterior (AP) and lateral (Lat) radiographs were obtained. A fluoroscopic approach was then investigated to determine whether a ‘C-arm’ fluoroscopic gantry (Ziehm 8000, Ziehm Imaging, Nuremberg, Germany) commonly used in the operating theatre setting could be orientated such that coronal plane displacements of the patella could be measured. To acquire coronal plane images of the patella with the femoral component *in situ*, the C-arm was orientated proximo-distally (PD), such that the knee flexed into the x-ray beam between the source and camera of the C-arm, providing a supero-inferior image (similar to the traditional ‘skyline’ view) of the patella. Sequential images were acquired at 10° intervals of knee flexion from 0 - 120°, determined using a goniometer whilst a lunge movement was performed (Figure 5.1). This was repeated twice. The images were later analysed to assess whether both the medial and lateral edges of the patella could be visualised at each 10° interval, from which the position of the midpoint of the patella could be calculated. Table 5.1 shows a summary of the positions at which the patella could be identified across all repeats, in both the native knee and following arthroplasty.

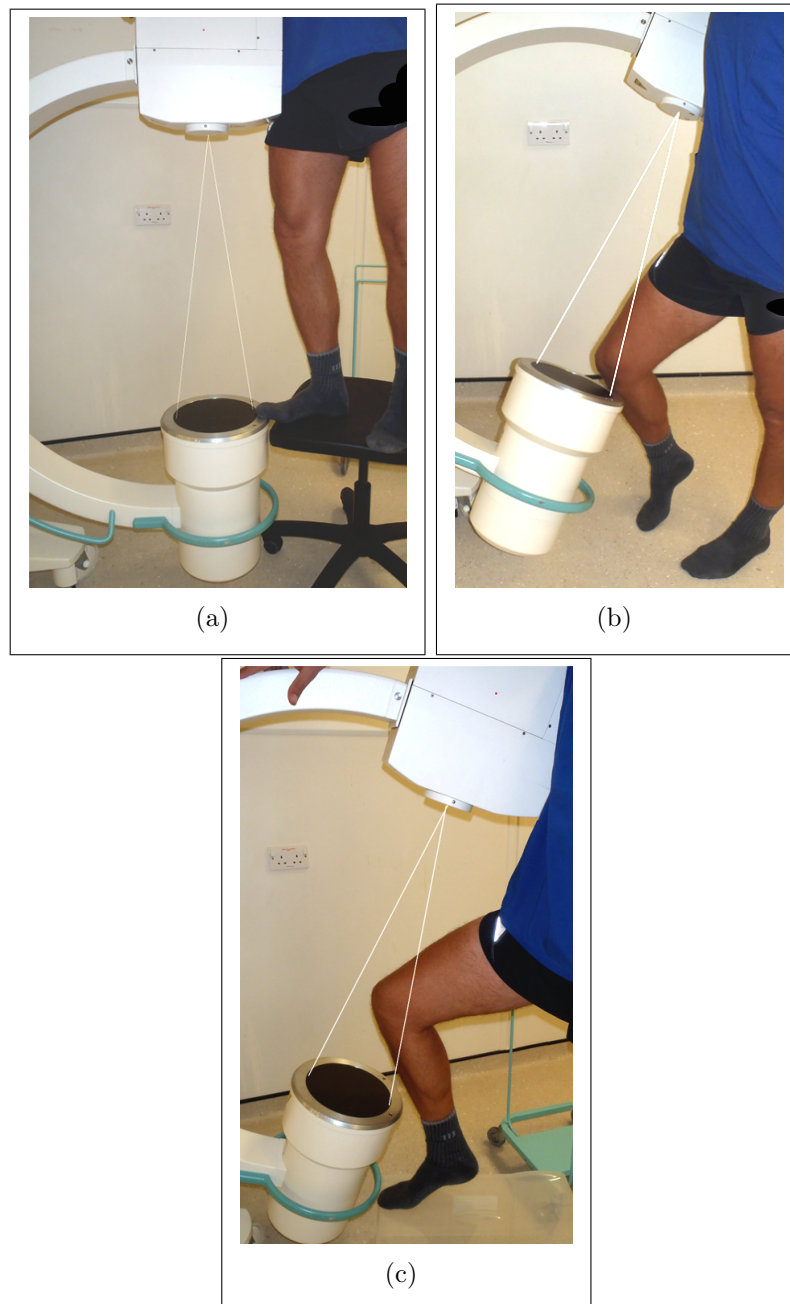


Figure 5.1: Showing spatial difficulties (a) obtaining the skyline patellofemoral view below 50 degrees of flexion (b) within a fluoroscopic C-arm field of view (white lines).

Results

Using the implanted cadaveric knee, the antero-posterior view of the patella was obscured by the femoral component (Figure 5.2a) whilst an adequate view of the

knee was seen from the lateral plane radiographs (5.2b).

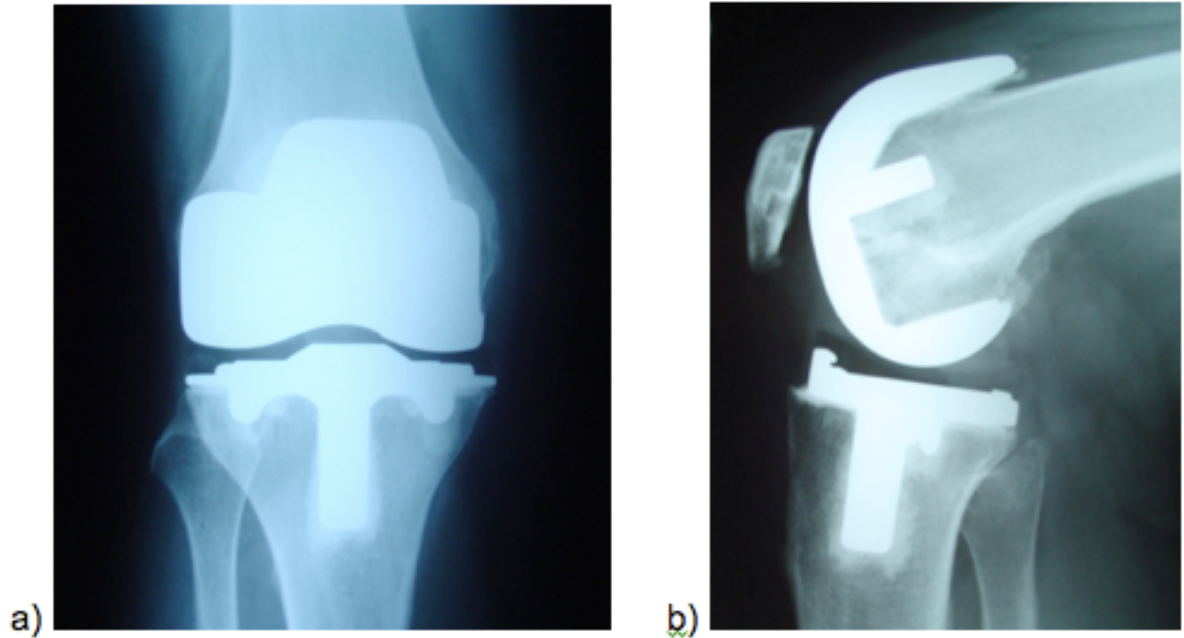


Figure 5.2: a) Antero-Posterior and b) Lateral radiographs of a right knee following implantation of TKR.

The orientation of the leg and the fluoroscopy C-arm were such that no images of the patella could be obtained with the leg in extension or during the early stages of flexion (Table 5.1). In the native knee, the patella could be visualised from 50° to 120° of knee flexion, whereas following implantation the patella was obscured from view by the prosthesis until 70° of knee flexion. In addition, the view of the patella was again obscured at flexion angles higher than 90°.

Discussion

Whilst the proximo-distal C-arm arrangement proved difficult, the lateral view has been shown to be excellent and so a technique was trialled using two partial

Presence of Patella in Fluoroscopy Image		
Flexion Angle (°)	Native	Post Arthroplasty
0	X	X
10	X	X
20	X	X
30	X	X
40	X	X
50	✓	X
60	✓	X
70	✓	✓
80	✓	✓
90	✓	✓
100	✓	X
110	✓	X
120	✓	X
Patella visualised	58%	25%

Table 5.1: Showing presence or absence of patella using fluoroscopy throughout flexion.

(oblique) views that could then be co-registered to an index view of the patella obtained using either another imaging modality (such as CT or MRI) or imaging taken prior to knee replacement surgery being performed. It was not logistically possible, with the available equipment, to orientate the gantries so that the subject could perform a lunge or squat whilst obtaining satisfactory views of the patella throughout the flexion range. Of particular note was the difficulty in obtaining clear views in extension and early flexion. Using plane radiography, images could be taken of the patella in early flexion by positioning the x-ray source anterior to the knee and the image plate posterior to the knee (or vice versa) in an oblique arrangement. To obtain fluoroscopic images, however, the subject was required to flex the knee. This exercise, with the available fluoroscopic devices, did not allow for equipment to be placed between the heel and leg, especially in the latter stages of flexion. For this reason the C-arm had to be placed anterior to the leg, and the subject ‘lunged’ forwards. As a consequence, there were no suitable images from the

early stages of flexion. By 70° of flexion, the knee joint could be manoeuvred to a position between the C-arm where adequate views of the patella could be obtained in both the native and replaced knee. In the latter stages of flexion, when the patella was in the intercondylar notch, the view was again obscured by the prosthesis. In short, the only images that could be obtained of the patella in the replaced knee were between 70° to 90° of knee flexion, accounting for 25% of the flexion range compared with 58% in the native knee, between 50° to 120°. Fulkerson et al demonstrated that in the stance phase of gait, from 0° to 30° of flexion is when the patella is most unstable and so forms an essential range for assessment for any PFJ kinematic study [9, 159]. This range was not visible in either native or replaced knees.

Conclusions

The use of radiographs and fluoroscopy for measuring PFJ kinematics was deemed inappropriate for this study for two reasons primarily; the inability to image the patella in the replaced knee throughout the flexion range and the lack of availability of fluoroscopy in this institution. Secondly, the host institution is a regional tertiary referral centre for orthopaedics that has three C-arm fluoroscopy machines, all of which are in use daily and none of which were available for research purposes.

5.3.2 Preliminary investigations into the feasibility of using CT and MRI at host institution

The use of CT and MRI has been shown to be successful in measuring PJJ kinematics in the native knee [140, 160]. Previous studies have highlighted issues with image artefact following implantation of components, the majority of which are made from

metal alloys [161]. Alloys that contain ferro-magnetic metals cause extensive metal artefact during CT and MRI scans [162, 163]. Previous studies have quantified the CT measurement errors using phantoms constructed from a range of different metal alloys. Jakel and Reiss (2007) report a percentage change in the Hounsfield Units (HU) range observed on the introduction of certain metals, as shown in Table 5.2 [164].

Errors associated with metal artefact with CT	
Ion/Alloy	Alteration in range of HU
Titanium	3.0%
Tungsten	3.5%
Steel	8.0%
Titanium/Steel (TKR)	5.0 - 18.0%
Total Range	3.0 - 18.0%

Table 5.2: Metal artefact errors associated with CT (HU = Hounsfield Units), Jakel and Reiss (2007).

Similar quantification measurements have been made using MRI. Venook et al [165] quantified metal artefact by analysing the maximum magnitude of shift of pixels directly adjacent to a TKR during a 1.5 T MRI scan, citing errors of up to 12 mm depending on the metal composition of the prosthesis (Table 5.3).

Errors associated with metal artefact with MRI	
Medium	Maximum Magnitude of Shift Artefact of Pixels
Air	0.4 mm
Titanium	7.0 mm
Chromium	12.0 mm
Total Range	0.4 - 12.0 mm

Table 5.3: Metal artefact errors associated with MRI, Venook et al, (2006).

Recent advances in Metal Artefact Reduction Sequences (MARS) and improved image acquisition techniques such as altered spin-echo sequences, increasing the

amplitude of the frequency encoding (FE) gradient and view-angle tilting techniques have lead to improvements in the quality of images [166, 167]. However, not all institutions have this technology available and there are reports of associated blurring of the images following manipulation [166].

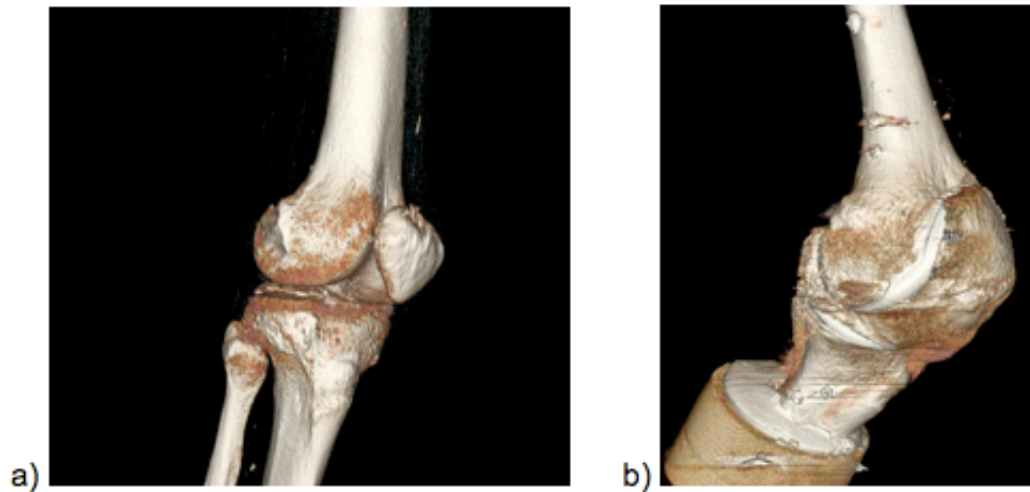


Figure 5.3: 3D CT reconstructions a) before and b) following implantation of TKR.

As previously reported, the ideal position for PFJ kinematic measurements would be erect (Section 5.2.1), performing a squat or walking down stairs. The majority of CT and MRI scanners (including those available at the host institution) are orientated horizontally, requiring the subject to be lying prone or supine. Having the subject orientated in this manner introduces the added methodological complication of simulating the conditions for upright squatting/walking under normal gravity. Previous studies have used techniques including: the subject lying face down whilst flexing the knee [168], sitting whilst extending against a footplate [169] and using weights/pulleys in an attempt to recreate the forces through the knee joint. It is unlikely, however, that these will be a true reflection of activities of daily living [170].

A methodological approach was trialled using an implanted cadaveric knee (Section 5.3) and a normal subject in both a 1.5 T MRI scanner (Siemens Magnetom, Phillips, Guildford, UK) and a 64-slice CT scanner (Lightspeed, GE Healthcare, Bucks, UK). The experiments addressed two considerations:

- 1) protocol required to obtain images of the patella and
- 2) the spatial constraints of scanners when measuring knee flexion.

An implanted cadaveric knee was used to assess the CT and MR dose required to obtain satisfactory images using the equipment available at our institution.

A normal subject was used to assess the spacial constraints of knee flexion within the confined space of the CT and MRI coils. Whilst previous authors have successfully demonstrated knee kinematics using MRI with subjects lying face down, effectively isolating the femur, this bears no relation to functional activity. The protocol required the foot to be static during flexion, i.e. recreating a functional stance (section 5.2.1). The subject was seated with the foot planted against a fixed footplate. A progressive movement towards the coil was performed from extension to 90° flexion, in 10° increments, recreating a ‘horizontal squatting’ movement. Throughout the exercise, an attempt was made to keep the knee joint in the target region of the scanning coil (Figure 5.4).

There were two major issues encountered which negated the use of CT or MRI.

1. For suitable data to be obtained from the CT studies, a minimum of 9 images were required (at 10° intervals) ranging from extension to high flexion. In the native knee it is possible to acquire all the required information regarding the

position of the femur, tibia and patella from a ‘scanogram’, the long leg scout radiograph taken to assess alignment prior to a full CT. In this assessment, the dose required to obtain a scanogram of the cadaveric knee was 4.06 mSv. There is considerable variation in the literature regarding the realistic minimum dose of radiation to which the investigation can be reduced. The Imperial protocol [171] has a reported dose of 0.74 mSv whilst the Perth protocol [172] has a reported dose of 2.7 mSv. This implies that if a scanogram was performed every 10° of flexion, then for each study, a subject would be irradiated with a minimum of 10 mSv to 27 mSv per leg. A typical CT scan of the pelvis gives around 10 mSv, which is equivalent to about 4.5 years of United Kingdom background radiation [173].

2. The diameters of CT and MRI scanning coils vary between scanners but at the host institution were between 57 cm and 60 cm (Phillips, Guildford, UK; Seimens, Hayes, UK). In the latter stages of flexion it was difficult to orientate the leg such that the knee joint was held within the target zone for the coil.

Experimental techniques using MRI are prone to the same spacial constraints of aligning the knee in the target area, but have the advantage of not requiring ionising radiation. The time required to acquire the image data in one position of flexion is approximately 10 to 20 minutes. Dynamic MRI techniques have been successful in quantifying the coronal plane movements of the patella in the native knee [8]. However, despite advances in metal suppression image manipulation techniques there is a paucity of literature regarding MRI measurements of coronal plane movements of the patella following knee replacement arthroplasty. At the start of this work there was no commercially available software to provide suitable images of the patella on MRI following TKR.



Figure 5.4: Illustrating the practical considerations of obtaining satisfactory position inside a CT or MRI scanner.

Conclusions

CT and MRI techniques for imaging PFJ were tested and ultimately rejected due to: the large radiation dose required to obtain satisfactory images; the time taken to scan patients; the time scale and cost of creating, validating and implementing software to reduce metal artefact; the necessity to lie flat (not a physiological position); and the inability of subjects to flex beyond 60° due to space restrictions within the coil and target acquisition.

5.3.3 Ultrasound

Ultrasound has been used for assessing anatomical abnormalities of the knee and PFJ for over 10 years [6, 174, 175, 176]. The advantages of ultrasound over other imaging modalities are that it does not involve ionising radiation and can be used to image the patella in the presence of metalwork [177]. Previous attempts have been made to quantify coronal plane patella movements during flexion using ultrasound, however these were compromised by measurement errors of up to 6.4 mm (2 x SD) due to issues with securing the probe to the leg via bracing [6, 178].

A major part of the work of this thesis involved developing a novel method to address the limitations of previous uses of ultrasound for measuring PFJ kinematics. Over a period of 2 years, the development, validation and subsequent implementation of a new measurement concept was performed, termed the MAUS method that combines Motion Analysis with UltraSound.

5.4 MAUS: Motion Analysis in combination with Ultra-Sound

5.4.1 Part 1: Development of Technique

The errors associated with skin movement artefact over bony landmarks in motion capture systems are well known [179]. However, the accuracy of motion capture systems to track rigid objects is far superior, the technique being commonly used in contemporary surgical navigation systems [180].

To date, no study has accurately measured *in vivo* measurements of the patella in three dimensions (especially in the coronal plane) relative to the femur and tibia in the presence of a knee joint replacement. A novel technique was developed during the current work combining an ultrasound scanner with a motion capture system, which reduced previous measurement errors.

Aim

The aim of this work was to develop and validate a novel technique using a Motion Analysis and UltraSound (MAUS) system to measure 3D kinematics of the PFJ and the movement of the patella relative to the femur and tibia.

What follows here are details of the investigations used to assess the feasibility of combining ultrasound and motion capture in the measurement of PFJ kinematics, followed by a description of the concept, and finally an outline of the practical considerations for the technique.

5.4.2 Assessment of distortion/accuracy of ultrasound with depth change, lateral displacement and signal strength change

A description of the principle components of ultrasound and how an ultrasound image is formed can be found in Section 2.6.3.

Introduction

For reliable measurements to be taken, an assessment to identify and quantify the amount of distortion within the ultrasound image was required. It is possible that when using ultrasound as a measurement tool, the process of generating an ultrasound image may introduce a spatial distortion. This might not create a problem for the clinical use of ultrasound, where it is often used as a qualitative identification tool, but would be an issue for the proposed use as a quantitative measurement tool. It is possible that ultrasound measurement of the width of an object might be correct when the object is close to the probe, but be different when measured at extremes of the ultrasound image. Therefore, an experiment to test these possible effects was deemed necessary. An experiment was performed to measure a known object of fixed dimensions at different distances from the probe, lateral displacements and at different signal strengths.

Method

A calibration box was constructed that consisted of a grid of pre-drilled holes at 10 mm intervals in the baseplate, milled using a CNC (Computer Numerical Control) machine (Bedmill CNC400, Lilian, Taipei, Taiwan, precision 0.04 mm). The drill holes received a calibration block (20 mm x 40 mm) that could be inserted in 9 different positions (Figure 5.5 (a) and (b)). The calibration box featured a window on the front surface across which a transparent membrane was attached, through which ultrasound could pass. A Voluson 2D ultrasound probe (SP10-16-RS, GE Healthcare, Bucks, UK) was clamped in a fixed position relative to the calibration box window, set at an image depth of 38 mm (corresponding to a frequency of 18 Hz)

and at a power setting of 50%. The box was then filled with water, an echo-neutral medium. The block was placed in position one (Figure 5.5) and an ultrasound image acquired. Without moving the probe, the block was then moved to a second position (position 2), and a further measurement of block width taken. This was continued until all 9 positions had been measured. Next, the signal strength was increased by 50% to 100% and the same 9 positions were then repeated and measurements of the block face taken, totalling 18 total positions (Table 5.4). The cartesian coordinates of the edges of the block in the different positions were measured from the the ultrasound images using Matlab (Version 7.4.0 (R2007a), The MathsWorks, Natick, MA, USA). Figure 7.9(c) shows a typical example of a Matlab image from which measurements were taken. Readings in different positions, at different strengths were then compared to assess the amount of distortion when the position in the grid was altered, or when the signal strength was changed.

Conditions for Testing Properties of Ultrasound	
Parameter	Number of measurements
Left to Right positions	3
Depths	3 (x 3 = 9)
Signal strengths	2
Total	18 measurements

Table 5.4: Quantifying recorded conditions and positions of block to assess for image distortion.

Measurements of width of the block face in the 18 positions were analysed to establish the degree of potential distortion within the image space.

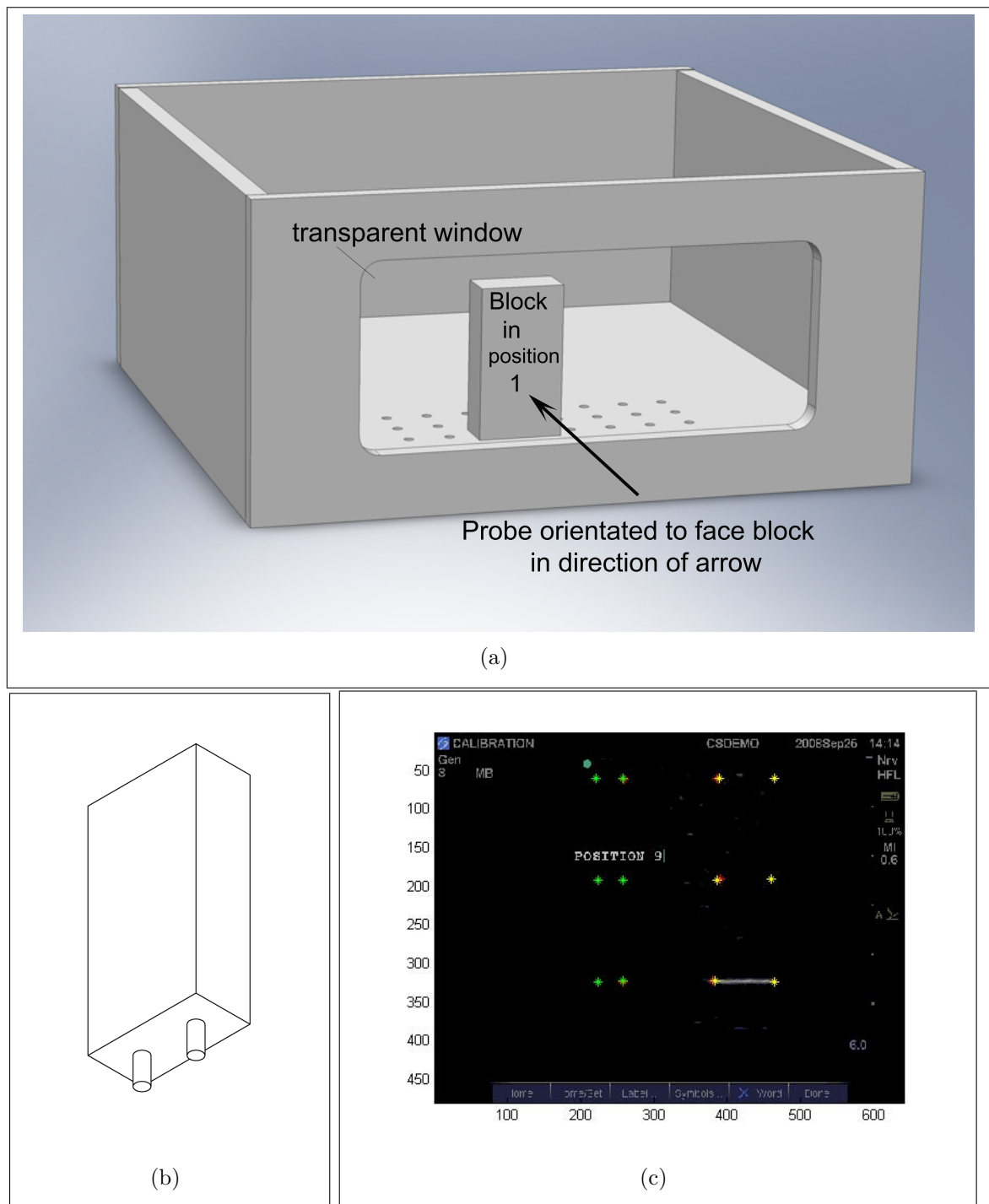


Figure 5.5: Showing (a) Calibration box, (b) Calibration block and (c) Image measurements of block in different positions.

Measurements of Length (L) of block face		
	L at signal strength 50% (mm)	L at signal strength 100%
Mean	20.4 mm (SD 0.4 mm)	20.2 mm (SD 0.5 mm)

Table 5.5: Measurements of block face used to assess changes with depth, displacement and signal change.

Results

The mean block width in all positions and at all signal strengths measured 20.3 mm (Table 5.5).

Conclusions

There were no significant ultrasound image distortions associated with altering depth, lateral displacement or signal strength.

5.4.3 Concept of combining Ultrasound with Motion Capture

For the purposes of description, attention will be focussed on describing movements of the patella relative to the femur, but the exact same principles can be applied to the tibia.

To describe patellofemoral kinematics, the 3D locations of both patella and femoral segments are required. These can be provided by 3D coordinates (x,y,z) of bony landmarks; a minimum of 3 non-colinear femoral landmarks are required to define a femoral coordinate system (eg. greater trochanter, hip joint centre, lateral condyle) and similarly 3 non-colinear landmarks are required to define the patella coordinate

system (e.g. inferior patella pole, medial and lateral corners of the patella) as shown in Figure 5.6. Landmark locations can be provided by physical retro-reflective skin markers, or as in this study, virtual markers defined in the ultrasound image. It is commonplace in motion capture studies to manually palpate a bony landmark and place a reflective marker on the skin surface directly over the landmark to identify it. Whilst the subject remains stationary, theroretically the position of the marker does not change. However, when the patient moves (eg. from standing to crouching) the position of the bony landmark may have altered relative to the overlying skin marker, a phenomenon known as *skin movement artefact*. This will be further influenced by the depth of the bony landmark from the skin due to interposition of muscle, soft tissue and adipose tissue. By using ultrasound to locate a bony landmark, the effect of skin movement artefact can be avoided. There is no necessity to rely on skin markers as, provided that the position of the probe is known, ultrasound can be used to directly image the bony landmark.

5.4.4 Components of the MAUS system

Motion Analysis - capture and filtering or processing

Subjects were scanned using a VICON motion capture system with 12 infra-red cameras (370 Motion System, VICON, Oxford, UK). The VICON system is a digital video based optical tracking system. The twelve cameras record the three dimensional locations of reflective markers at frequencies up to 200 Hz (typical measurement frequency for these experiments was 100 Hz). Provided more than one camera can visualise a marker, its three dimensional position can be described using the principles of stereophotogrammetry [181].

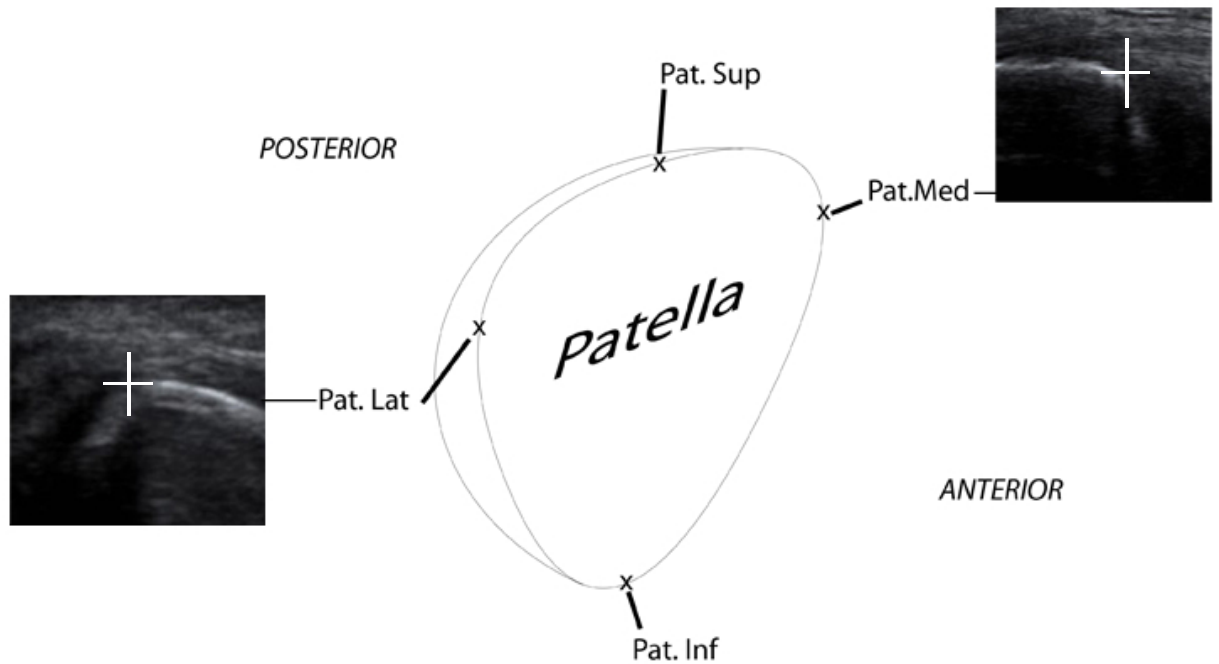


Figure 5.6: Showing bony landmarks on the patella and examples of ultrasound images acquired. Pat Inf = inferior pole of patella; Pat.Sup = most superior part of patella, Pat.Med/Lat = Medial/Lateral corners of the patella.

Marker positions are expressed in as real-time three-dimensional coordinates by the VICON system. By a process of triangulation, the positions of each marker are calculated, and given as a displacement from the origin. A graphical illustration of the output from the VICON system in the x direction for a single marker is shown in Figure 5.7.

Of note is the oscillation of the position in Figure 5.7, between 130.08 mm and 130.11 mm. To enhance ‘noise reduction’ of the signal, a filter needs to be applied to the data. A filter is a signal processing tool that analyses the signal and removes unwanted noise.

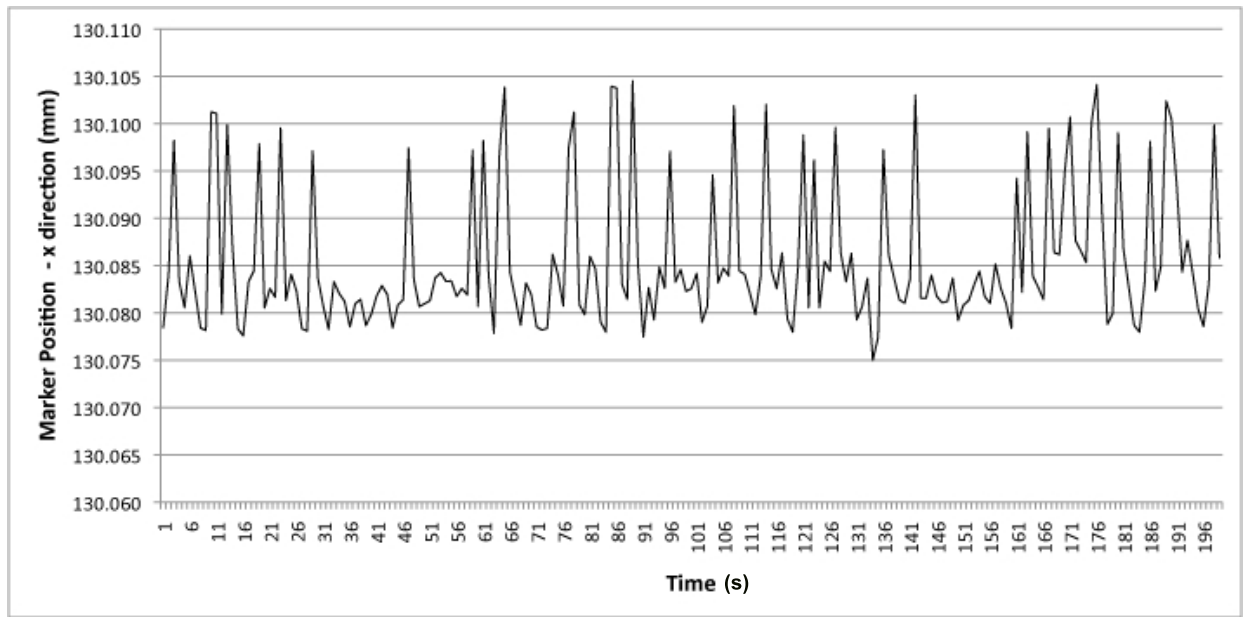


Figure 5.7: Showing the real-time output of VICON calculated position of a stationary marker in the x direction.

Two types of filter were tested, the low pass Butterworth Filter (designed to produce as flat a frequency response as possible) and the Median Filter (a filter which runs through data entry by entry, replacing each entry with the median of neighbouring entries).

Using raw data from a single marker the effect on the signal of two filters were compared (Butterworth and Median). The x, y and z coordinate positions of a single stationary marker were collected over one second (100 frames). The two filters were applied to the raw data. The mean x, y and z coordinates of the marker were then calculated using the three methods: Raw, Butterworth and Median Filters. The differences between the Raw and Butterworth Filter (Butter), and Raw and Median Filter (Median) estimations were then measured for the x, y and z components of

the marker.

The process was repeated over five further one second time intervals and results are given in Table 5.6.

Differences in Displacement Error of Filters						
	Raw - Butterworth Filter (mm)			Raw - Median Filter (mm)		
Trial	Butter X	Butter Y	Butter Z	Median X	Median Y	Median Z
1	0.73	0.15	0.01	0.01	0.001	0.003
2	0.91	0.04	0.76	0.004	0.01	0.01
3	0.64	0.20	0.43	0.002	0.007	0.007
4	0.63	0.07	0.53	0.008	0.007	0.0004
5	0.19	0.16	0.72	0.01	0.005	0.003
Mean	0.62	0.12	0.49	0.005	0.007	0.004
	Butter = 0.41 mm			Median = 0.02 mm		

Table 5.6: Displacement of position of marker following use of filters. Butter = Butterworth Filter, Median = Median Filter.

The mean error across all trials for the Butterworth Filter was 0.41 mm, the Median Filter produced a mean error of 0.02 mm.

For consecutive experiments, the Median Filter was used on all VICON data prior to further processing.

Ultrasound

A Voluson-i ultrasound scanner (GE Healthcare, Waukesha, WI, USA) with a 2D probe (frequency range 4.7 - 13 MHz) was used to collect still images as shown in Figure 5.8.

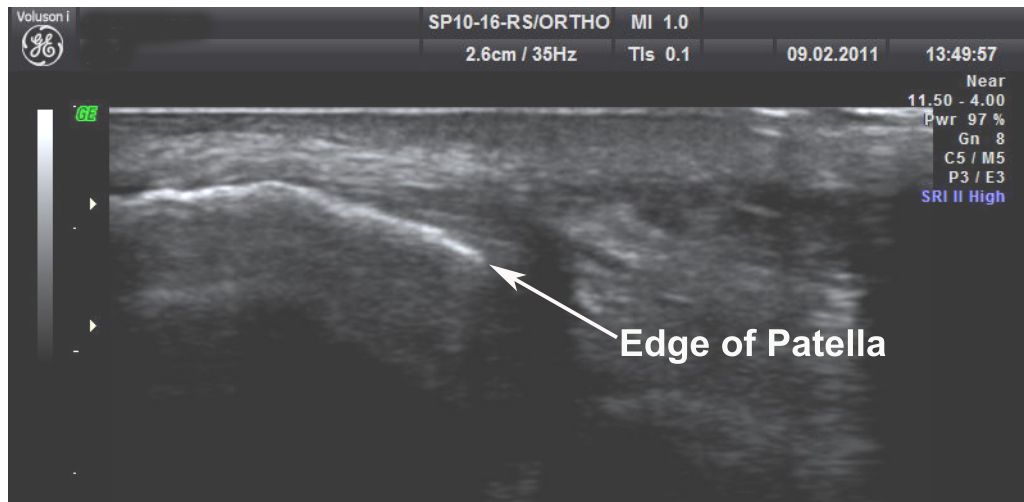


Figure 5.8: Showing a typical image of the lateral aspect of the patella using a GE Voluson-i ultrasound scanner.

By tracking the ultrasound probe in the motion capture laboratory using reflective markers, its position can be described accurately in the three dimensional laboratory space. Whilst it is not possible to see the ultrasound image directly in the motion capture system, the markers mounted on the probe accurately describe the position and orientation of the probe when the ultrasound image of the bony landmark was being captured (Figure 5.9).

Defining Ultrasound Points in the Motion Capture System

During the scanning procedure, landmarks were located using ultrasound, and by small adjustments of the probe position on the leg, manoeuvred into the centre of a target point, provided by a set of cross-hairs marked onto a transparent overlay on the ultrasound screen (Figure 5.10). The target point provided a fixed location in the image which could then be transformed from the image based coordi-

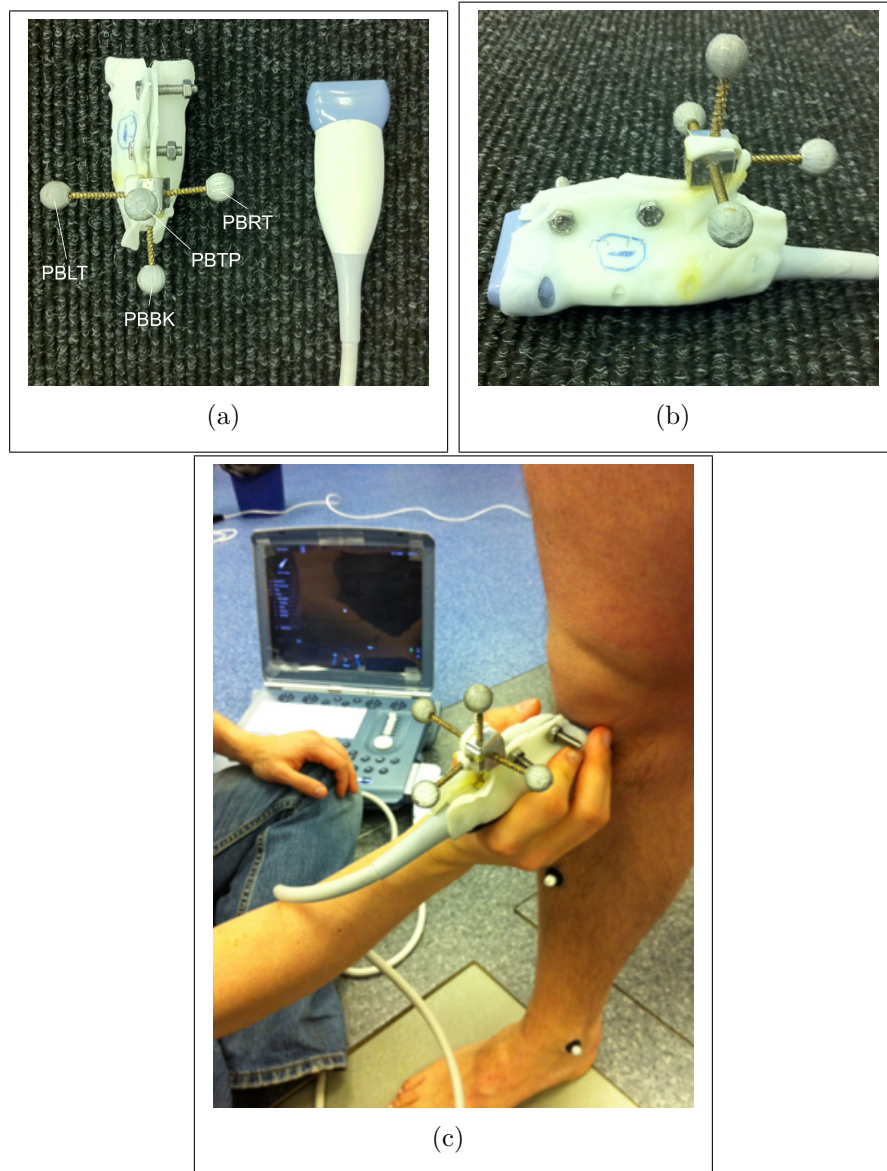


Figure 5.9: Showing (a) probe jacket with mounted reflective markers and probe, (b) jacket mounted on probe and (c) in use.

nates to the body fixed coordinates of the probe using a mathematical transformation.

Any point described in terms of a given cartesian coordinate system can be represented in another coordinate system by the process of transformation. In order to perform the transformation the relationship between the two coordinate systems

must be established - usually performed by a calibration procedure. Transformation from one coordinate system to another is usually performed using matrix operations, multiplying the coordinates of a point in one coordinate system by a transformation matrix to give the coordinates of the same point in the other coordinate system [182]. Similarly, a point in a 2D system, e.g. an image, can be represented as a point in a 3D coordinate system by a 2D to 3D transformation matrix. A calibration procedure was performed (mathematical details in Section 5.5.3) to obtain the transformation matrix from the ultrasound image coordinate system to the body fixed coordinate system of the probe (Figure 5.10).

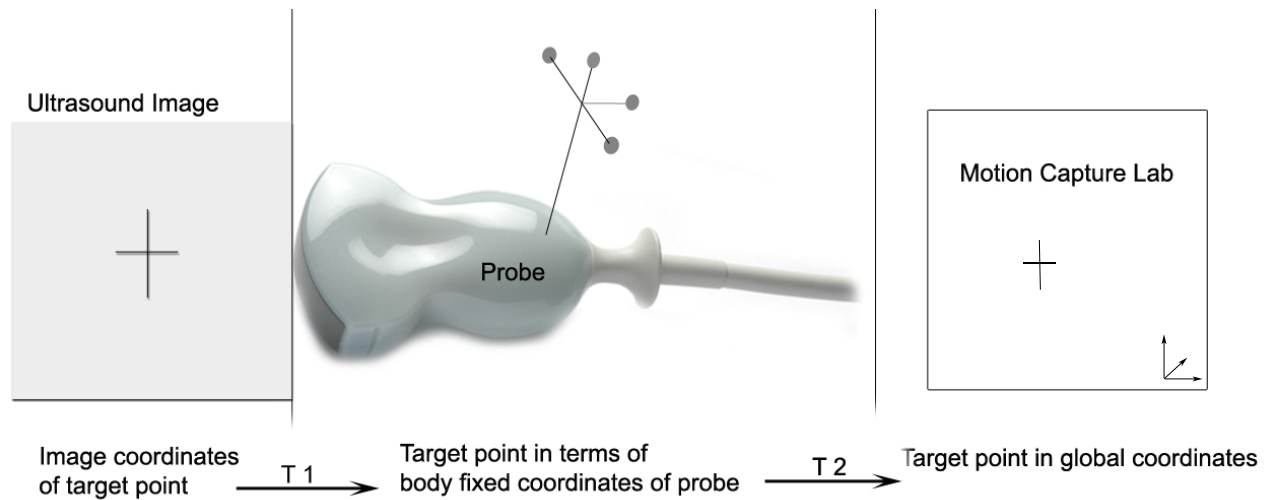


Figure 5.10: Showing transformations T_1 , between the image coordinate system and the probe, and T_2 , between the probe and the laboratory.

In this way, the ultrasound probe was employed as a surrogate pointer, identifying a landmark under the skin, thereby circumventing skin movement artefact, directly providing 3D coordinates of bony landmarks. With the leg at different degrees of flexion, the precise location of the femur, patella and tibia could then be reconstructed in 3D.

A clinical study comparing normal and replaced knees can be found in Chapter 6. Details of the calibration technique and mathematical explanation of the rigid body transformation between the image coordinate system and the global coordinate systems are given below in Section 5.5.

5.5 Technical description of MAUS method

5.5.1 Definitions

The terminology used in the description of the technique are defined here.

Global Coordinate System

This is provided by a Vicon Nexus Motion Capture System comprising 12 infra-red cameras. This was used in combination with Nexus software (Vicon Motion Systems, Oxford, UK) providing a 3D space within which coordinates for reflective markers (attached to skin and the ultrasound probe) could be tracked.

Ultrasound coordinate System

This was a two-dimensional (2D) space provided by raw, static 2D ultrasound images that were obtained, in DICOM (Digital Images and Communication in Medicine) format and analysed in Matlab (Version 7.4.0 (R2007a), The MathsWorks, Natick, MA, USA).

Local Coordinate Systems

Embedded co-ordinate systems were established for the three individual segments of the lower limb: femur (F), tibia (T) and patella (P), shown in Figure 5.11.

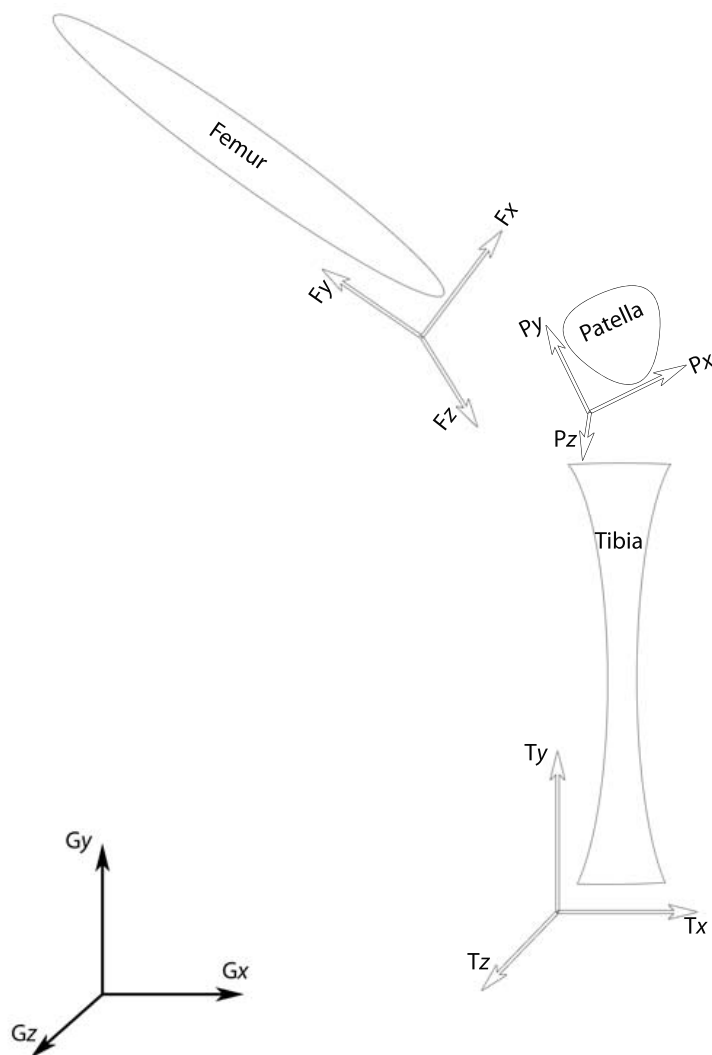


Figure 5.11: Showing individual coordinate frames for Global coordinate system (G), Tibia (T), Patella (P) and Femur (F).

Orthogonal unit vectors $\vec{x}$, $\vec{y}$, and $\vec{z}$ were used to define axes in each limb segment coordinate frame. The axis definitions are given in Table 5.7, derived from points (markers) shown in Figure 5.14. Ascribing an individual coordinate frame and origin

to the segments allowed for separate descriptions of patella movement relative to the tibia and femur.

Segment Axes Descriptions			
Segment	Plane	Vector Providing Axis	Vector Description
Femur	Coronal	$F\vec{x}$	Grt. Troch to HJC
	Axial	$F\vec{y}$	KJC to HJC
	Sagittal	$F\vec{z}$	Cross product of $F\vec{x}$ and $F\vec{y}$
		Origin	KJC
Patella	Coronal	$P\vec{x}$	Pat. Lat to Pat. Med
	Axial	$P\vec{y}$	Pat. Inf to Mid-point of Pat
	Sagittal	$P\vec{z}$	Cross product of $F\vec{x}$ and $F\vec{y}$
		Origin	Mid-point of patella
Tibia	Coronal	$T\vec{x}$	ANK to MMA
	Axial	$T\vec{y}$	AJC to KJC
	Sagittal	$T\vec{z}$	Cross product of $F\vec{x}$ and $F\vec{y}$
		Origin	Tib.Tub

Table 5.7: Showing description of segment axes

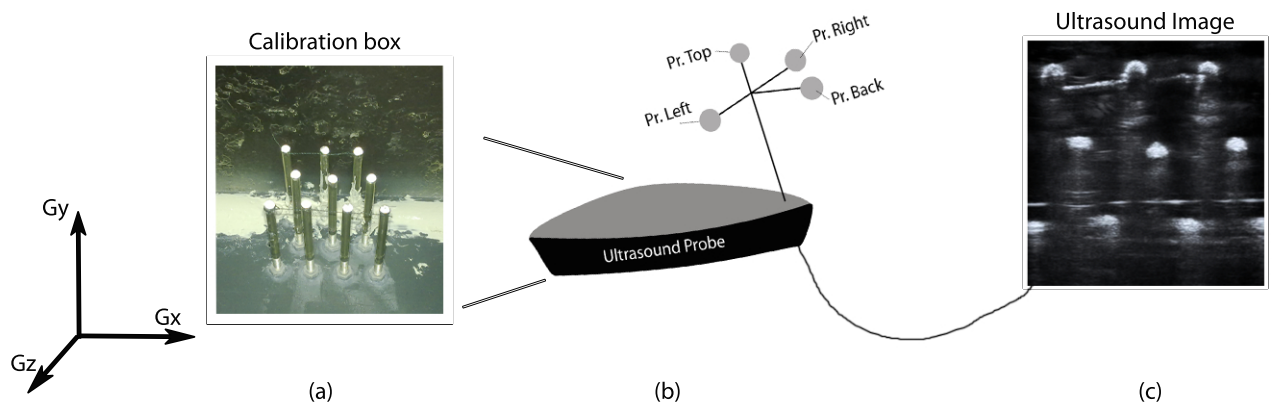


Figure 5.12: Showing (a) Phantom with 10 co-planar 3 mm markers, (b) Schematic of markers used to track ultrasound probe and (c) Ultrasound image of phantom markers.

5.5.2 Calibration Algorithm - Transforming Ultrasound Image Coordinates into the Global Coordinate System

The fundamental principle of this technique is to describe bony landmarks located on an ultrasound image in the global coordinate system. Whilst the ultrasound probe itself is not visible in the motion capture system, retro-reflective markers can be attached as shown in Figure 5.12 (b), allowing precise tracking of the probe's position in the global coordinate system. However, the relationship of the image to the probe markers remains unknown.

The calibration algorithm described here comprises two separate mathematical functions: the transformation between points in the image coordinate system and the global coordinate system and the transform between the probe coordinate system and the image coordinate system. When the resulting rigid body transformation is applied to the coordinates provided by the probe markers, the offset to origin the ultrasound image can be calculated, providing 3D coordinates of the position in the correct orientation.

This, the first step is calibration of the image space; to do this, coordinate data of a series of points that can be seen in both the image coordinate system and the global coordinate system must be acquired.

A phantom was constructed as shown in Figure 5.12 (a) using 10 co-planar retro-reflective markers, contained within a rigid box, one wall of which had a membranous window allowing for ultrasound measurements. The phantom markers were first imaged using the motion capture system, to establish the individual coordinates of each marker in the global coordinate system. Without adjusting the position

of the box, the phantom was then submerged under water and the position of the ultrasound probe (Figure 5.12(b)) was adjusted until an image of all the markers was obtained. This was only possible when the plane of the ultrasound beam was aligned with the plane of the phantom markers. The pose of the ultrasound probe markers in the global coordinate system was also recorded at this point.

Knowing the position of the markers in both the ultrasound image and the global coordinate system and the position of the probe when the image was taken, it was possible to calculate a calibration matrix (T_i). The second function required was a transformation matrix providing the offset between the probe markers and the image. The probe has its own embedded coordinate system provided by the markers, as does the ultrasound image, the orientation of which was calculated during the calibration algorithm. A transformation matrix was therefore required to provide a rigid transformation between the probe markers and the image comprising three rotations, a translation and a scaling matrix (converting pixels to mm). When applied to the origin of the probe's rigid body, it will provide the coordinates of the origin of the image.

5.5.3 Creation of Virtual Ultrasound Point in Motion Capture System

Contained here is the theoretical background behind a customised Bodybuilder programme (Vicon Systems, Oxford, UK), written to create a target ultrasound point (T) as a virtual marker in the motion capture space.

Using the phantom apparatus (described in Section 5.4.2), the ultrasound probe was set up such that cross hairs marked on the centre of the ultrasound image were

aligned over the central phantom marker, or target marker. In this arrangement, global coordinates of both the probe markers and T (Figure 5.13) are known, and hence the offset between the two can be measured.

Applying the Offset to the Probe Markers

Firstly, the four probe markers are used to form an individual segment (Probe) comprising three axes, with PBLT as the origin. This constitutes the local coordinate system of the probe. The global coordinate value of the offset from the origin to T is then used to calculate a transformation matrix (M) which describes the transform between T and the origin (PBLT) in the local coordinate system.

When the probe is moved to a new position in the motion capture space, the location of T in terms of the global coordinate system can be calculated by applying the inverse matrix of M to the new coordinates of the probe origin (PBLT).

The transformation matrix used can be thought of as the combination of a series of manoeuvres, sequentially transforming the ultrasound point to the probe markers, or vice versa. Firstly, three rotation matrices are calculated, representing Euler angles around the three axes are applied, representing the change in orientation of the probe in the global coordinate system.

Next, a translation matrix (O) is used to provide the offset in the local coordinate system. A scaling matrix (S) was not required as the unit distance between points was fixed as demonstrated in Section 5.4.2. The matrices required are illustrated

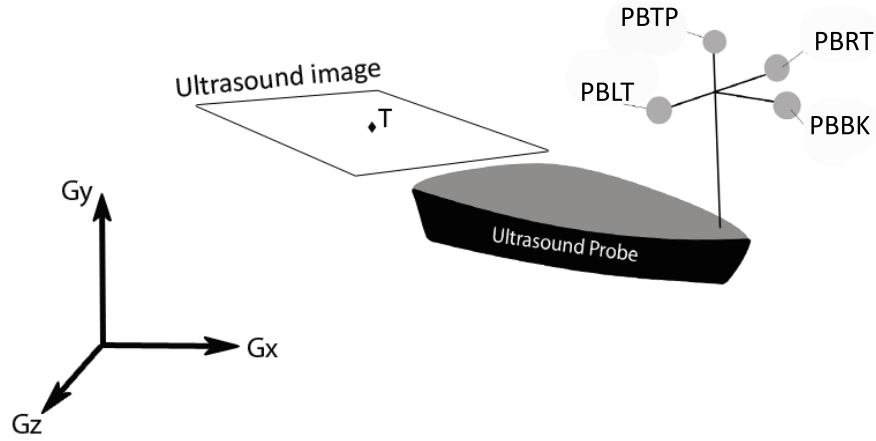


Figure 5.13: Showing target point (T) and the probe markers (Left; PBLT, Right; PBRT, Top; PBTP, Back; PBBK) mounted on the ultrasound probe, in relation to the Global (Motion Capture) coordinate system (Gy, Gx, Gz).

below:

Translation Matrix used to define the image point relative to the fixed point defined as O;

$$\mathbf{O} = \begin{pmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ dx & dy & 1 \end{pmatrix}$$

Rotation Matrix 1 about x-axis is in yz plane and defined as R1;

$$\mathbf{R1} = \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos\Theta & \sin\Theta \\ 0 & -\sin\Theta & \cos\Theta \end{pmatrix}$$

Rotation Matrix 2 about y-axis is in xz plane and defined as R2;

$$\mathbf{R2} = \begin{pmatrix} \cos\alpha & 0 & -\sin\alpha \\ 0 & 1 & 0 \\ \sin\alpha & 0 & \cos\alpha \end{pmatrix}$$

Rotation Matrix 3 about z-axis is in yx plane and defined as R3;

$$\mathbf{R3} = \begin{pmatrix} \cos\beta & -\sin\beta & 0 \\ \sin\beta & \cos\beta & 0 \\ 0 & 0 & 1 \end{pmatrix}$$

Now a Final Transformation Matrix (M) can be applied combining all the above functions;

$$\mathbf{M} = \mathbf{O} \times \mathbf{R1} \times \mathbf{R2} \times \mathbf{R3}$$

where

$O = \text{Translation Matrix}$

$R1 = \text{Rotation Matrix in yz plane}$

$R2 = \text{Rotation Matrix in xz plane}$

$R3 = \text{Rotation Matrix in yx plane}$

$$\mathbf{M} = \begin{pmatrix} 1 & 0 & 0 \\ 0 & 1 & 0 \\ dx & dy & 1 \end{pmatrix} \begin{pmatrix} 1 & 0 & 0 \\ 0 & \cos\Theta & \sin\Theta \\ 0 & -\sin\Theta & \cos\Theta \end{pmatrix} \begin{pmatrix} \cos\alpha & 0 & -\sin\alpha \\ 0 & 1 & 0 \\ \sin\alpha & 0 & \cos\alpha \end{pmatrix} \begin{pmatrix} \cos\beta & -\sin\beta & 0 \\ \sin\beta & \cos\beta & 0 \\ 0 & 0 & 1 \end{pmatrix}$$

5.5.4 Landmarks and Points Used To Define Probe and Limb Segments

A description of the location of markers is given below and summarised in Table 5.8, and illustrated in Figures 5.6, 5.9 and 5.14.

Probe Markers

The probe had 4 markers mounted to a bespoke, removable, thermoplastic jacket (Figure 5.9). Previous attempts to attach markers directly to the probe had been hampered by the view of the markers being obstructed by the leg or other body parts.

Femur

The skin markers used can be seen in Figure 5.14. There is a paucity of distinct identifiable anatomical features on the bony femur that can be found with ease using ultrasound. For the purposes of this experiment, three bony landmarks were required and the following were used;

Greater trochanter - the apex of the supero-lateral tip of the greater trochanter found by moving the probe from distal to proximal over the greater trochanter.

Lateral condyle - the peak of the bony prominence of the lateral epicondyle found when moving the probe from anterior to posterior.

Hip joint centre - Ehrig et al describe a functional method (the SCorE method) for hip joint centre estimation [76]. With markers attached to the thigh and lower leg, subjects perform a leg circumduction exercise in the motion capture system. The apex of a cone formed by the movement of the markers represents the hip joint centre [76]. Following a systematic review of all methods of measuring the hip joint centre the SCorE method was used as it provides the lowest error, at 1.3 mm, compared with up to 20 mm with other regression methods [183, 184].

Tibia

The skin markers used can be seen in Figure 5.14. The following ultrasound markers were used:

Tibial tubercle - the apex of the bony prominence when moving from distal to proximal.

Medial and lateral malleoli: retro-reflective skin markers were used to represent the malleoli. Due to their distal location relative to the patella and the lack of movement at the ankle, the error associated with skin movement at the ankle was considered acceptable. Together the medial and lateral maleoli were used to calculate the ankle joint centre (AJC). For the purposes of this experiment, the three dimensional position of the AJC was calculated using the automated kinematic modelling software provided by the VICON system (ANYBODY, VICON, Oxford, UK).

Patella

The landmarks used to represent the patella are shown in Figure 5.6.

Inferior patella (Pat Inf) - the most distal point on the apex of the patella.

Superior patella (Pat.Sup) - the superior pole of the patella.

Medial patella (Pat.Med) - the supero-medial pole of the patella.

Lateral patella (Pat.Lat) - the supero-lateral pole of the patella.

Marker Set

A modified Helen Hayes [185] marker set used for the experiments can be seen in Figure 5.14. Additional skin markers were required (a) for the SCorE, (centre of the femoral head) calculation and (b) for registration of the ultrasound data to the leg.

5.5.5 Measurement of Motion

The description of coronal plane translations of the patella (as described in section 2.5) relative to the femur and tibia, requires quantification of the relationship of the patella to a fixed point (origin) and its movement along a fixed line during the squatting movement performed by the subjects.

The origin for the femur was the lateral condyle and for the tibia, the tibial tubercle. Both were located using ultrasound and could be found with relative ease.

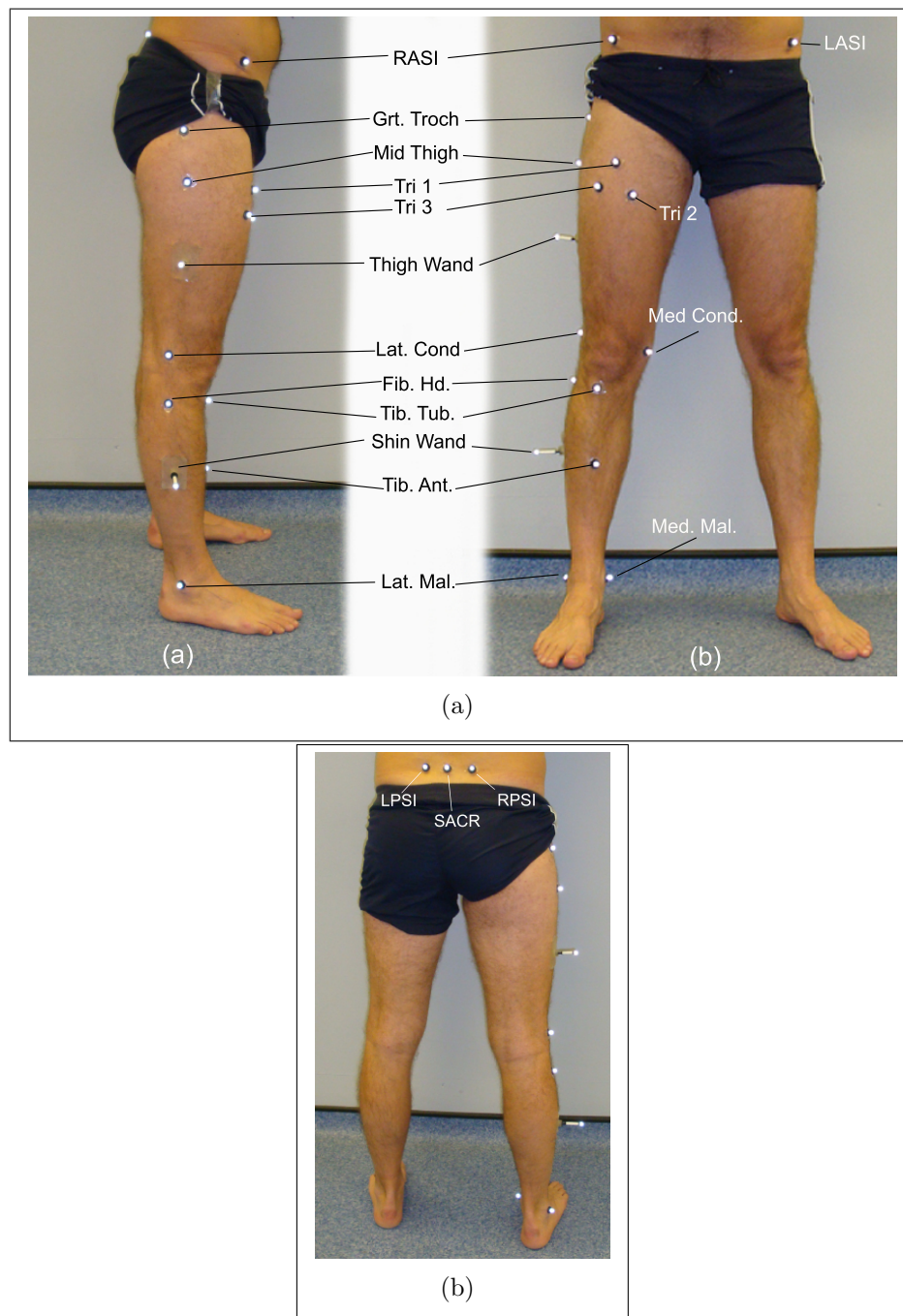


Figure 5.14: Showing skin marker set-up from the (a) front (b) side and (c) rear.

Axes

Whilst intuitively, the medio-lateral axis of the knee joint itself, or indeed that of either the femur or tibia (eg. medial and lateral epicondyles) present possible options

Markers		
Segment	Marker name	Description
Pelvis	LASI	Left anterior superior iliac spine
	RASI	Right anterior superior iliac spine
	SACR	Sacral marker, placed at same height as iliac spine markers
	LPSI	Left posterior superior iliac spine
	RPSI	Right posterior superior iliac spine
Femur	Grt. Troch	Greater Trochanter
	Mid Thigh	Lateral Thigh Marker
	Tri 1,2,3	Anterior thigh triangle for Ehrig calculation
	Thigh Wand	Midpoint of Grt. Troch and Lat. Condyle markers
	Lat. Cond	Lateral Femoral Condyle
	Med. Cond	Medial Femoral Condyle
Tibia	Fib. Hd	Fibula Head
	Tib. Tub	Tibial Tubercle
	Shin Wand	Midpoint of Fib Head and Lat. Malleolus
	Lat. Mal	Lateral Malleolus
	Med. Mal	Medial Malleolus
Patella	Pat. Sup	Most superior point of patella
	Pat. Inf	Most inferior point of patella
	Pat. Med	Most medial point of patella
	Pat. Lat	Most lateral point of patella
Probe	Pr. Top	Superior Probe Marker
	Pr. Back	Posterior Probe Marker
	Pr. Right	Right Hand Probe Marker
	Pr. Left	Left Hand Probe Marker
Virtual	HJC	Hip Joint Centre
	KJC	Knee Joint Centre
	AJC	Ankle Joint Centre

Table 5.8: Showing description of markers for each segment.

for a medio-lateral reference line, the tibia and femur undergo axial rotation and varus-valgus displacements during knee flexion. Such movements would alter the orientation of the axis and potentially mask subtle underlying movements.

Patellofemoral descriptions are given here relative to the pelvic coordinate system, specifically along the line indicated by the anterior superior iliac spine skin markers,

LASI-RASI whilst patellotibial movements are measured along the line formed by the medial and lateral maleoli. The orientation of these lines in the laboratory space remained unchanged throughout the squatting exercise.

All subjects (n= 60) performed a squatting movement. With both feet shoulder width apart, recordings of the above landmarks at 6 distinct values of knee flexion angle during the downward phase of the squat were collected; full extension, 20°, 30°, 45°, 60°, 90°. A Zimmer frame was used by the subject to steady themselves where required. Coordinate data from the motion capture system was recorded at a frequency of 100 Hz. Each bony landmark was found using ultrasound. When the operator had positioned the bony landmark in the cross-hairs of the ultrasound image, a handheld trigger switch was deployed. The function of the trigger switch was to mark the time when the ultrasound probe was orientated over the landmark by generating a spike signal. The spike represented 0.1 seconds or 10 frames.

5.6 Part 2: Validation

Accuracy of a system is defined as the closeness of the measurement to its true value. The precision of a system is the degree to which measurement under unchanged conditions, returns the same results. The following potential sources of error were studied: Vicon Error, Calibration Error, Target Acquisition Accuracy, Assembly Errors and Inter- / Intra- observer Error. The accuracy and precision of this system was then compared to previous established motion capture methods, including skin markers and thermoplastic patella clamps.

5.6.1 VICON Error

The VICON system introduces an error in locating the reflective markers. The accuracy of the motion capture system was calculated as the combined residual error of all 12 cameras (Vicon Residual) prior to every experiment from the calibration procedure. The VICON error was calculated using the static and dynamic calibration data collected at the beginning of each experiment. All cameras were calibrated simultaneously using a wand with a fixed array of retro-reflective markers. By using a combination of static measurements and movements of the wand markers in the lab space via triangulation algorithms, the measurement errors of the infra-red cameras were calculated. The mean results across all sessions of data collection during 19 sessions, (Chapter 6) was 0.28 mm (SD 0.05 mm). A sample set of camera errors following dynamic and static calibration are shown in Table 5.9.

VICON Camera Calibration		
Camera	Wand Count	Image error (mm)
1	4166	0.24
2	4095	0.25
3	4733	0.27
4	1004	0.28
5	1994	0.45
6	6400	0.22
7	1330	0.34
8	2277	0.39
10	4332	0.21
11	3776	0.25
12	4167	0.20
	Mean	0.28 mm (SD = 0.05 mm)

Table 5.9: Showing example of VICON calibration data.

5.6.2 Ultrasound Calibration Error

This was assessed by measuring the accuracy of the ultrasound calibration technique which was used prior to each of 19 scanning sessions. A phantom was constructed containing 10 reflective markers of 2 mm diameter which provided 10 points which could be imaged by both the motion capture system and the ultrasound (Figure 5.15). A comparison was made of the predicted positions of the phantom markers from the ultrasound technique (provided by the calculated transform) and the actual positions of the markers taken to be that provided by the motion capture system from each of the 19 scanning sessions (see Appendix 5.5.2). Error was represented as maximum error for all 10 points over all sessions. The maximum error was 0.52 mm (range 0.18 mm - 0.79 mm).

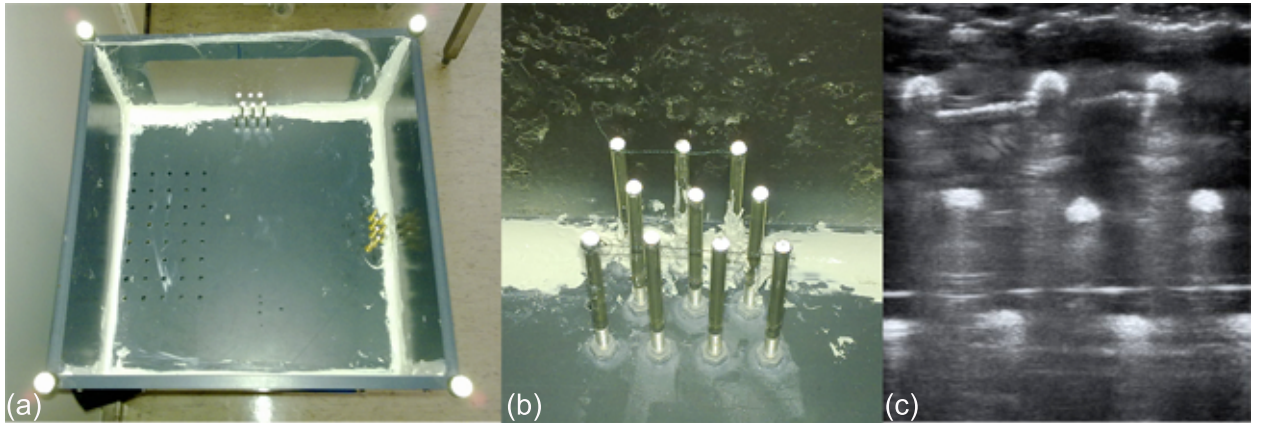


Figure 5.15: Showing (a) the phantom box, (b) close up of markers in phantom and (c) ultrasound images acquired when testing the accuracy of the MAUS system.

5.6.3 System Accuracy including Assembly Errors

The aim of this experiment was to establish the accuracy and precision of the MAUS system in terms of setting up all the required equipment and acquiring the desired

point.

Sixty subjects were scanned during nineteen sessions spanning a period of two years. During this time the target screen and the ultrasound probe jackets were applied and removed on each occasion, introducing a potential source of error. It is possible that by removing and re-applying the probe jackets, the reflective markers might change their orientation relative to the probe. The central marker from the phantom constructed for calculating calibration errors (section 5.6.2) was used as the target marker for the purposes of this experiment. The target marker's position could be described in x,y,z coordinates within the motion capture laboratory space (termed the TRUE coordinates as this was considered the gold standard measurement) and via the ultrasound system (termed the CALCULATED coordinates).

The phantom was positioned in the laboratory (position 1) and the TRUE position of the markers was recorded, using the VICON system. Next the phantom was filled with water, providing a transmission medium and making it possible to detect the marker using ultrasound. The ultrasound equipment was assembled; the target point applied to the ultrasound unit image, the probe jacket fitted and the data recorded to enable the CALCULATED position to be established using the post-processing previously described. The difference between the TRUE and CALCULATED coordinates in terms of x,y,z, displacements was established. The target point and probe jacket were then removed and re-applied and the CALCULATED coordinates re-established a total of four times for position 1. The above process was repeated in three further different positions (positions 2 - 4) within the motion capture laboratory to ensure there was no measurement bias dependent on location.

100 frames of data were collected and the mean position of all points calculated.

Results are displayed in Table 5.10 as the mean displacement in x,y,z directions.

RMS errors			
Position	X	Y	Z
1	0.32	0.77	1.17
	5.44	0.39	1.20
	2.09	1.01	3.84
	0.14	0.09	0.85
2	0.54	0.49	4.67
	0.47	0.04	4.82
	2.05	0.30	6.50
	0.05	0.27	1.59
3	1.05	2.37	1.33
	0.64	2.72	0.74
	0.68	0.32	2.15
	0.84	0.43	1.64
4	2.18	0.44	3.41
	3.16	0.41	2.79
	0.30	0.32	0.81
	0.24	0.26	0.23
Mean	1.27	0.66	2.36
Total Error = 1.43 mm			

Table 5.10: Root Mean Square (RMS) error of phantom point following removal and re-application of probe jacket.

The RMS error of the system across all 16 trials in all directions was 1.43 mm (SD 1.55 mm). Further analysis revealed that the least accuracy was seen in the Z plane (positive direction, floor to ceiling) with a mean RMS error of 2.36 mm. It is possible that this might have been influenced by movements of the phantom about the laboratory, and uneven flooring.

The RMS error of the system was 1.43 mm.

5.6.4 Intra- and Inter-Observer Error

Assessment was made of the intra- and inter-observer error for measurement of coronal plane patella movement. Bony landmarks (Tib. Tub, Lat. Pat, Sup. Pat, Med. Pat, Inf. Pat, Lat. Cond and Grt. Troch) were found using the ultrasound method for 4 knees (2 male and 2 female) by 2 independent observers. Landmark locations were established at each flexion angle for each subject. These landmarks were used to provide a *coronal profile*. The *coronal profile* characterises the displacements of the patella throughout the flexion range and is represented as a single value of mm/degree, representing the gradient of the regression line fitted to a graph of medio-lateral displacement versus flexion angle. A representative case can be seen in Figure 5.16.

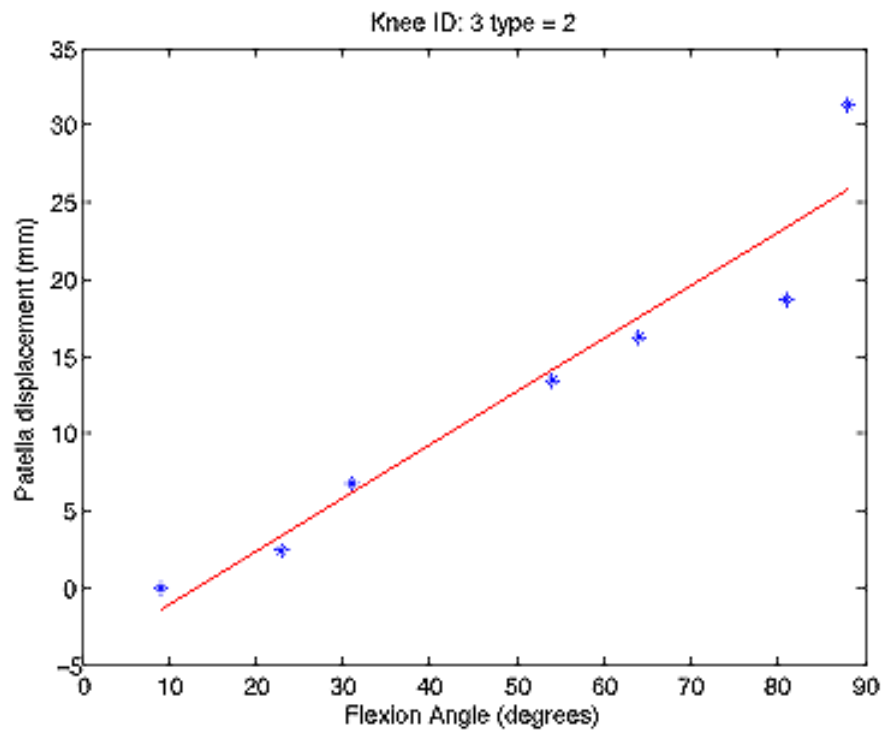


Figure 5.16: Showing a regression line fitted to a representative case of coronal plane patella displacements during flexion of the knee.

The same set of landmarks were then obtained for a second time at each flexion

position by each of the observers at a later date. The gradients of these graphs (coronal profiles) were used to assess intra-observer error. The errors were expressed as the standard deviation (SD) of the measured difference, as described by Altman. [186]. Results are shown in Table 5.11. The inter-observer error was 0.0092 mm/degree of flexion.

Inter and Intra-observer Errors (mm/°)		
Trial	Investigator 1	Investigator 2
1	0.0011	0.0057
2	0.0069	0.0134
3	-0.0062	0.0133
4	0.0095	0.0070
Mean Intra Observer Error	0.0033 mm/°	0.0109 mm/°
Inter-Observer Error = 0.0092 mm/degree		

Table 5.11: Showing the Intra- and Inter- Observer Errors for the MAUS Technique.

5.6.5 Comparison of MAUS system with previous techniques

Following the intra- and inter-observer error study a direct comparison was made to see whether there was any statistically significant differences between the measurements of coronal plane displacements made using the ultrasound system compared with two established motion capture methods of measuring patella displacements; skin markers and patella clamps.

Skin markers have often been used to represent bony landmarks and are placed over a bony prominence [179]. A more recent approach has been the use of thermoplastic patella clamps, which are moulded to the subject's patella and sit directly on the skin overlying the patella (Figure 5.17) [187]. Prior to the experiment, the clamp was moulded to the underlying patella. During the subsequent exercise, the clamp is

believed to follow the patella and so used as a surface representation of the patella's position.



Figure 5.17: Showing patella clamp attached to subject's left knee.

The same subjects (4 knees; 2 male, 2 female) used for the inter- and intra-observer error studies then performed the same squatting movements on 2 separate occasions using first the skin markers, then the patella clamp. The outcome variable used was the *coronal profile* (mm/°). Each of the techniques was assessed for statistically significant differences in the measurement of coronal plane displacements using the Students t-test, as a two tailed test and assuming groups of unequal variance.

Results

There was no statistically significant difference between ultrasound observers 1 and 2 ($p = 0.132$). However, ultrasound was shown to be statistically different from both

skin markers ($p = 0.016$) and thermoplastic clamp ($p = 0.024$) methods (Table 5.12). Of note is the much larger spread of data for repeated trials using the skin marker and thermoplastic clamp methods compared with the ultrasound data (Figure 5.18).

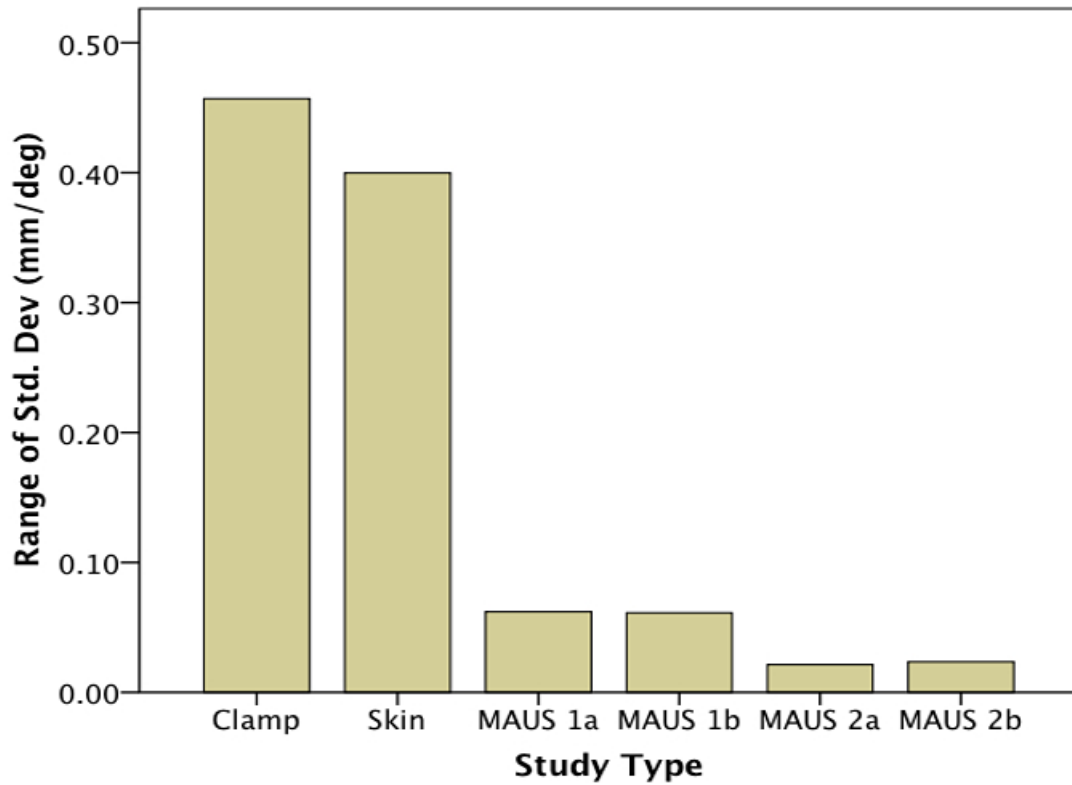


Figure 5.18: Showing reproducibility of methods. MAUS = Motion Analysis with Ultrasound, Clamp = Thermoplastic clamp.

Technique Compared	Significance
Ultrasound 1 vs Ultrasound 2	$p = 0.132$
Ultrasound vs Skin Markers	$p = 0.044$
Ultrasound vs Clamp	$p = 0.024$
Ultrasound vs Skin + Clamp	$p = 0.002$

Table 5.12: Showing the statistical differences between different techniques for measuring patella displacements.

5.7 Discussion

This chapter chronicles the development and testing of a new tool (MAUS) which can be used for the *in vivo* measurement of patella movements in normal knees and following knee arthroplasty. The over-riding design principle driving all methodological considerations was that the measurement system should be designed such that it could be used in subjects (including post-arthroplasty patients) performing an activity of daily living that is likely to cause pain following knee replacement. Such a test would not only provide a relevant insight into the position and orientation of the knee joint during pain, but also a potential investigative clinical tool. The MAUS technique was used here standing upright with the foot in the stance position. Hill et al (2000) demonstrated longitudinal rotations of 15° to 20° that occur during loaded and unloaded flexion at the knee joint [188]. This is important when considering movements of the patella which has its distal fixation at the tibial tubercle. It is reasonable to assume that if the foot is stationary throughout the exercise then tibial rotation will be minimised and the tibial tubercle position will be stationary. Conversely, if the patient were lying prone or supine with unrestricted foot movements, patella displacements may be affected by uncontrolled movements of the tibia. For the purposes of this study a squatting exercise was therefore chosen.

Various potential imaging modalities were considered during this study including plain radiographs, fluoroscopy, flat bed CT and MRI (Section 5.3). The primary difficulty with C-arm fluoroscopes, CT and MRI were obtaining satisfactory images throughout the full range of motion. Orientating the fluoroscopy C-arm to obtain images during the early range of motion (0° to 30°) was impractical. Similarly, it proved cumbersome to maintain the position of the knee in the centre of the CT and

MRI coils during the latter phases (70° to 90°) of flexion.

Previous studies have attempted to use motion capture or ultrasound to measure *in vivo* patellofemoral kinematics following knee replacement. To date, errors have been reported between 6 to 10 mm in some cases [6]. There are no previous studies that have combined the two modalities. Section 5.4.2 revealed that ultrasound shows no distortion with depth or displacement whilst scanning to depths of 50 mm. It also has the benefits of being both non-invasive and non-ionising.

Comparison of this technique with previous methods reveal that it is more reproducible, more precise and more accurate. The errors associated with the current technique are a significant improvement on the best efforts of previous techniques using only either ultrasound or motion capture analysis. Figure 5.18 graphically illustrates the range of results obtained when directly comparing the MAUS technique to previous attempts to quantify coronal plane movements of the patella, namely skin markers and thermoplastic clamps.

Limitations

The ideal movement to be investigated would have been stair climbing/descent, an activity which can cause significant PFJ discomfort. Whilst this was attempted, it was later abandoned, due to the difficulty of locating a moving target and the time required to acquire the ultrasound images.

In each position of flexion, seven ultrasound points must be found, each of which

requires 20 to 45 seconds, averaging around 2.5 to 5 minutes per position. It was impractical to keep subjects in a position of discomfort (often on one leg when climbing/descending stairs) for such a length of time. Instead the squat was used, which allowed for bilateral stance and a side support to be used. Additional limitations of the chosen technique were that recorded positions were not a continuous movement, but six measurements taken in isolation (full extension, 20°, 30°, 45°, 60° and 90°). Furthermore, the positions that were recorded showed slight variation between individuals in that whilst the desired position might have been 90°, the achieved positions ranged from 87° to 94°.

Ultrasound scanning of bony landmarks is a user dependent process, which has been shown to be influenced by factors such as training and skill fade over time. Whilst this study has assessed the inter- and intra-observer variability of two investigators, a further study is required to formally assess the repeatability of the MAUS technique using a larger cohort.

Subsequent measurements were performed by a single observer with the least intra-observer error. Ideally, a multiple observer trial should be carried out on subjects to highlight the true range of variation of measurement.

Section 5.4.4 describes the trials of different signal filters to smooth data from VICON measurements of marker position. Despite filtering, the recorded position of markers still showed oscillations of up to 0.02 mm. As a potential source of error, improvements might be made by applying alternative low-pass filters to the VICON data prior to processing.

5.8 Summary

The MAUS technique combines the use of motion capture tracked skin markers and ultrasound to provide the 3D coordinates of bony landmarks for the femur, tibia and patella with an overall error of 1.84 mm (2 x SD).

Chapter 6

PFJ Kinematics: Normal versus Replaced Knee

There is a wide range of outcome following knee arthroplasty. The primary motivation behind the current work was to obtain a greater understanding of why this occurs. Whilst a large body of work exists describing knee kinematics, the vast majority have been restricted to descriptions in the sagittal plane. A description of experiments measuring coronal plane patellofemoral kinematics in normal and replaced knees using the MAUS technique (developed in Chapter 5) is included here.

6.1 Introduction

There are many descriptions of sagittal plane kinematics following knee arthroplasty, using measures such as the patellar tendon angle (PTA) throughout flexion, as an indicator of patellar displacement, and movements of the patellofemoral contact point [46, 145, 147, 146]. The majority of these studies conclude that the kinematics of the replaced knee are different from that of the native knee. Similarly, sagittal plane

methods have highlighted differences between prosthetic designs [46, 147]. However, sagittal plane kinematic findings alone do not explain the observed differences between satisfied and non-satisfied patients.

To date, there are no reliable descriptions regarding coronal plane PFJ kinematics following joint replacement. Previous attempts to study *in vivo* coronal plane patellofemoral kinematics following joint replacement have suffered from difficulties with the patella being obscured by the components, metal artefact and/or large measurement errors [6, 161]. Further details of previous approaches can be found in Section 2.6.3. Consequently, the most incomplete understanding is of the coronal plane kinematics. It is possible that abnormal movements in the coronal plane may represent a source of anterior knee pain. The assessment of kinematic differences between painful and non-painful knees is beyond the scope of this thesis. This work is concerned with the implementation of a measurement tool (MAUS) to assess coronal plane kinematics in normal and replaced knees.

6.1.1 Aim

The aim of this study was to apply the measurement concept underpinning the MAUS technique, developed in Chapter 5, to investigate whether it is suitable to assess differences in the coronal plane patellofemoral and patellotibial kinematics of normal and replaced knees.

6.2 Materials and Methods

6.2.1 Study Design

This Ethics Committee approved (Oxford Research Ethics Committee B: 08/H0605/140/Version 4), prospective, case-controlled study involved three different groups: normal healthy subjects, post-operative TKR and post-operative patellofemoral joint replacement (PFJR) patients.

6.2.2 Power Estimation and Sample Size

The required sample size was derived using Altman's technique [186], using the *coronal profile* (mm/°) outcome measure as the primary variable. This was defined as the gradient of the graph of coronal plane displacement versus flexion angle (see Section 5.6.4). From a pilot study of six normal knees, the mean coronal profile was 0.088 mm/° and the standard deviation (sd) was 0.12mm/°. The standardised difference, $\Phi = 1$, was calculated using δ , the value of relevant clinical differences in *coronal profile* defined as a measurement that lay one standard deviation outside of the mean (0.12mm/°) as shown in equation 6.1.

$$\phi = \frac{\delta}{sd} \tag{6.1}$$

Altman's nomogram revealed that the minimum sample size required to compare two cohorts of knees (e.g. normal subjects vs TKR) was 18 in each group (i.e. $2n = 36$) for a power of 85% using an alpha significance level of 0.05.

Based on a minimum of 18 knees being required for each of two groups, the recruitment of 20 knees to each of the three groups of this experiment was considered appropriate.

Sixty knees were recruited to three groups:

1. Twenty healthy volunteers (14M: 6F, age 23 to 36 years) from the host research centre, this group will be referred to as the 'Normal' group.
2. Twenty knee replacement patients implanted with the AGC TKR (Biomet Ltd., Swindon, UK) patients (6M:14F age 49 to 72 years), the 'TKR' group.
3. Twenty knee replacement patients implanted with the FPV PFJR (Wright Medical Technology, Arlington, TN, USA) patients (10M:10F age 42 to 72 years), the 'PFJR' group.

The normal group comprised subjects with healthy knees. Exclusion criteria included previous knee surgery, anterior knee pain and any developmental abnormality of the knee. Subjects included in the TKR and PFJR groups comprised those with an underlying diagnosis of primary osteoarthritis who had undergone primary arthroplasty at least 12 months previously. By 12 months post-operation most issues regarding residual swelling, stiffness or transient pain have been resolved in the majority of cases [189]. The TKR and PFJR groups were recruited at random from the theatre lists at the host institution. No filtering other than the exclusion criteria was applied for patient selection. All patients underwent the procedure via a medial parapatellar approach. No record of post-operative pain or function was available in the patient groups.



Figure 6.1: Showing the use of the tracked ultrasound probe to locate bony landmarks.

Equipment and Subject Preparation

Prior to each experiment all VICON and MAUS equipment was calibrated, as described in Sections 5.6.1 and 5.6.3 respectively.

Each subject was given an explanatory tour of the motion capture laboratory, informed of all forthcoming procedures and had any questions addressed. Informed consent was obtained. The subjects then changed into clothing allowing visualisation

of the waist and legs. All measurements were taken with the subject barefoot.

The following physical parameters were measured and represent essential inputs for the automated kinematic models used to define the hip and ankle joint centres (Section 5.5.4):

1. Patient height when standing with their back against the wall (mm)
2. Leg length measured from ASIS to medial malleolus with leg in extension (mm)
3. Inter ASIS distance (mm)
4. Ankle width measured as greatest distance between medial and lateral malleoli (mm)

Reflective markers measuring twenty-five millimetre diameter were attached using adhesive tape over the landmarks given in Table 5.8 as shown in Figure 5.14. The same investigator applied all markers to all subjects and the probe to ensure repeatability as established in Section 5.6.4.

6.2.3 MAUS Technique

The MAUS technique, developed as part of the current thesis, was used to locate the same landmarks. The technique combines the use of motion analysis with ultrasound, using a tracked probe (Section 5.4.4) to locate bony landmarks which are represented as virtual points in the VICON system (Figure 6.1). A three dimensional representation of the landmark positions is given in Figure 6.2.

6.2.4 Landmarks used for the creation of femoral, tibial and patella segments

The landmarks used to define the femur, tibia, and patella, and the method used to locate them (MAUS vs VICON motion capture system) are shown in Table 6.1. A description of the creation of segments can be found in Section 5.5.4. The centre of the patella (COP) was calculated as the centroid of four points on the border of the patella (Pat. Sup, Pat. Inf, Pat. Med and Pat. Lat)

Methods Used To Locate Landmarks			
Segment	Name	Description	MAUS (M)/ VICON (V)
Femur	HJC	Hip Joint Centre	V
	GTR	Greater Trochanter	M
	LCON	Lateral Femoral condyle	M
Tibia	TITU	Tibial Tubercle	M
	LAMA	Lateral Malleolus	V
	MEMA	Medial Malleolus	V
Patella	PATsup	Most superior point of patella	M
	PATinf	Most inferior point of patella	M
	PATmed	Most medial point of patella	M
	PATlat	Most lateral point of patella	M

Table 6.1: Showing methods used to locate landmarks. VICON = motion analysis system, MAUS = combined motion analysis and ultrasound system.

6.2.5 Subject Movements

Each subject performed a squatting movement. The starting position was with the subject standing with their feet shoulder-width apart. Measurements of all landmarks (MAUS and VICON) were taken at a series of sequential positions of increasing knee flexion: at full extension, 20°, 30°, 45°, 60° and 90° flexion, indicated by a goniometer, using the technique described in Section 5.5.5. The goniometer was used for approximate positioning purposes only, later reconstruction of the leg segments

gave the true flexion angle.

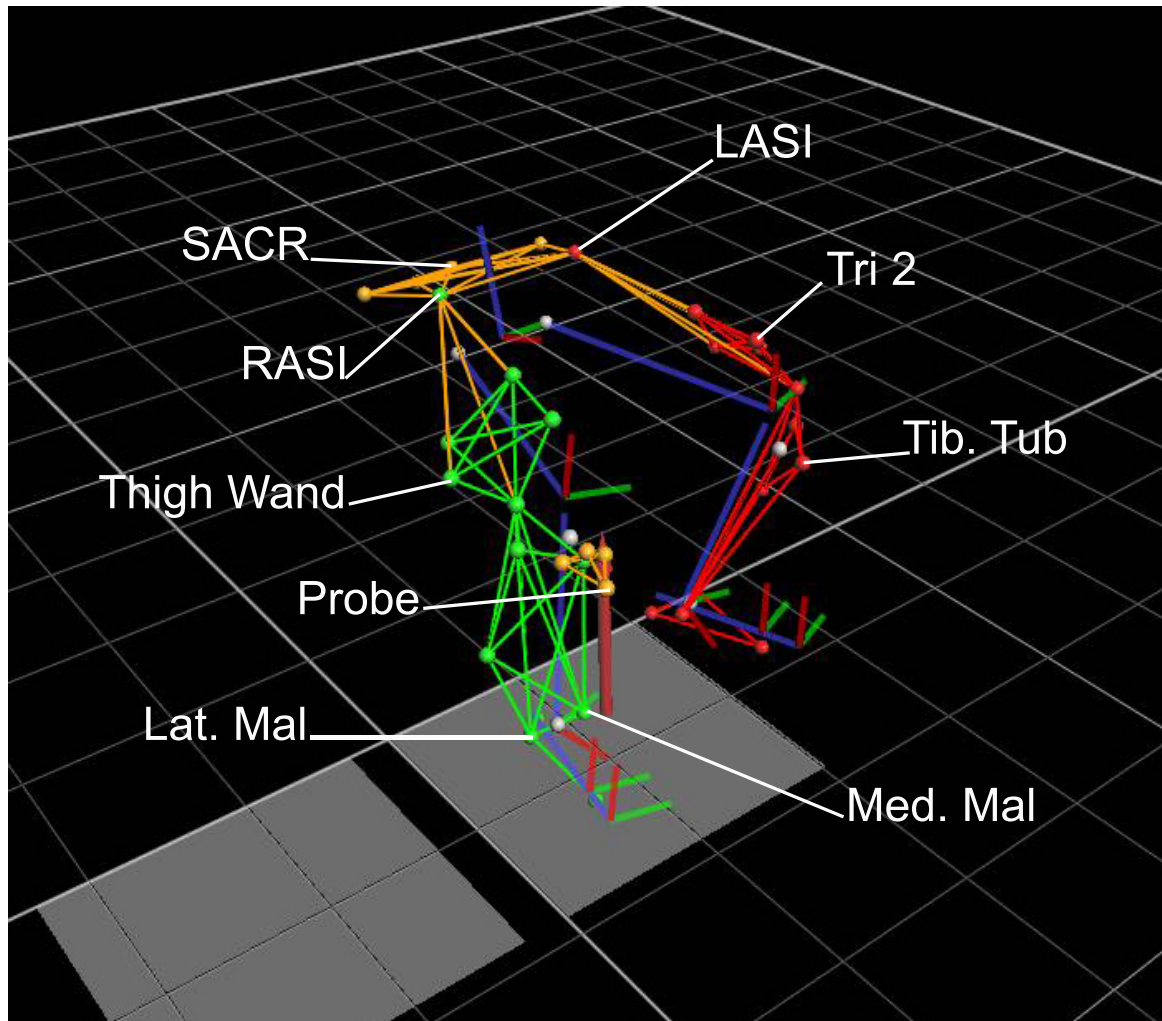


Figure 6.2: Showing a typical image of marker locations from VICON system.

6.2.6 Data Processing

This study was concerned with measuring patella movements in the medio-lateral (coronal) plane in both healthy normal knees and following knee arthroplasty.

In essence, the coronal plane displacement of the COP relative to the origin of the femur or tibia was required. A description of the processes used to combine data from the five different positions of flexion is included here, using the tibia as an example, (however, the process can equally be applied to the femur). The software required to apply the mathematical transformations to the coordinate data from the VICON system (outlined below) was written by enlisting the help of a computer programmer. Details of the programme architecture can be found in Appendix B.

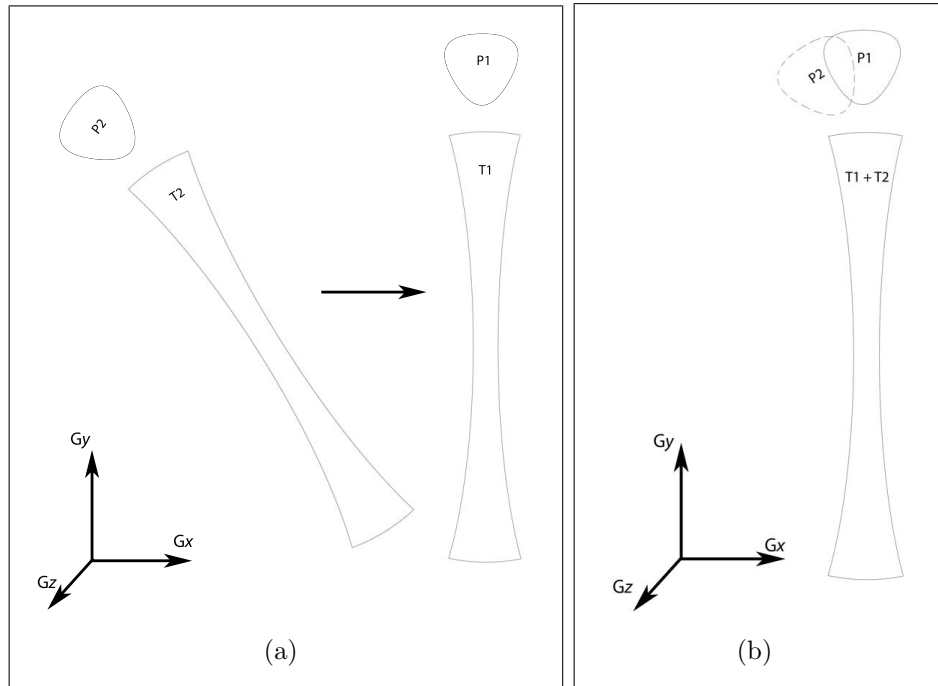


Figure 6.3: Showing Tibia, T and Patella, P in (a) the index (1) and flexed (2) positions, (b) with the flexed position transformed back into the index coordinate system of the tibia ($T1$).

In these experiments the lower limb bones (eg. tibia and patella) are represented as rigid bodies, defined by 3D coordinates of bony landmarks. Each rigid body has its own body fixed coordinate system. The position and orientation of the tibia and patella in extension were defined as the *index* positions (T1 and P1). When the leg moved into a position of flexion, the location and orientation of the the tibia and patella change: T2 and P2 in Figure 6.3a. To calculate the relative change in position with respect to the tibia, of the patella from index position (P1) to flexed (P2), all markers from the flexed position were transformed into the index coordinate system of the tibia (T2 to T1) as shown in Figure 6.3b. The displacement of the patella could then be directly measured relative to the tibial coordinate system.

Centre Of Patella (COP)

The position of the COP was calculated as the centroid of the four patella points shown in Table 6.1. The displacement of the COP was calculated relative to the origin of the femur and tibia for each position of flexion. Lateral displacement is defined as movement in the positive direction.

Origins and definition of coronal plane

The origin of femoral segment was the lateral condyle, the direction of measurement of coronal plane displacements, along the line formed between the LASI and RASI pelvic markers. For the tibia, the origin was defined as the tibial tubercle with the coronal plane being that of the medial and lateral malleoli.

6.2.7 Analysis of movements

It was noted that between full extension and 20° of flexion, all patellae underwent a sharp movement from a lateral position to a more medial one. This discontinuity was known to be representative of engagement of the patella as it enters the proximal trochlear groove in early flexion. The ‘engagement discontinuity’ measured between 3.0 mm and 26.5 mm (Table 6.2). Graphical illustration of the variation between subjects can be seen in Figure 6.4.

Engagement Discontinuity (mm)			
Subject	Mean	Std Dev	Max
Normal	3.4	3.0	9.8
TKR	4.9	3.8	11.6
FPV	6.5	5.8	26.5

Table 6.2: Showing mean and standard deviations of engagement discontinuity.

The range of observed movement by the patella is likely to be a reflection of the resting position of the patella in extension and the relative position of the proximal end of the trochlear groove. When standing up, with the knee resting in full extension, the patella position is dictated largely by the surrounding soft tissue tension provided by the quadriceps muscles (notably vastus medialis obliquus) and surrounding ligaments (the patellofemoral ligament complex). Indeed, prior to the initiation of flexion, there might not be ANY contact between the patella and the trochlea groove.

Due to the cohort variation shown prior to engagement of the patella, the measurements of coronal plane displacement at flexion angles of 20° were taken as the reference (medio-lateral) position of the patella.

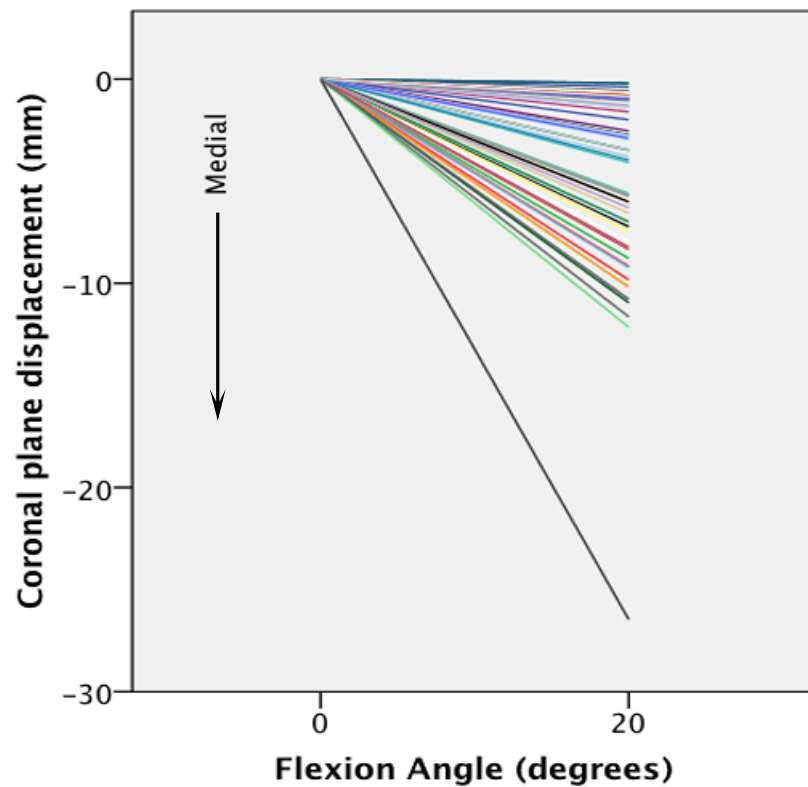


Figure 6.4: Showing the range of size of engagement flick between all subjects.

Outcome Measure

On the graph of medio-lateral displacement versus flexion angle, regression analysis was performed between 20° and 90° to obtain the gradient, measured in $\text{mm}/^\circ$ (Section 5.6.4). This provided the outcome measure of a *coronal profile* for each individual's knee.

Exploratory data analysis was performed to assess the distribution of results from measurements performed ($n = 60$), using the Shapiro Wilks test [190]. This indicated that the data were normally distributed ($p = 0.56$). Figure 6.5 shows Q-Q and box

and whisker plots of the data, confirming normal distribution.

Statistical Analysis

An analysis of variance (ANOVA) test was applied to the two sets of data, (patellofemoral and patellotibial coronal profiles) to assess whether statistical differences existed between the means of the three groups (Normal, TKR, PFJR). Whilst there was no statistically significant difference between the three groups for the patellotibial data, $p = 0.586$, there were highly significant differences between the three groups within the patellofemoral data, $p = 0.004$. Therefore, for the patellofemoral data, posthoc independent t tests were used to examine differences in the coronal profiles of each of the three groups: Normals, TKR, and PFJR using 2 way analysis of samples of unequal variance. Statistical tests were performed using SPSS version 18.0 (SPSS Inc, Chicago, IL, USA).

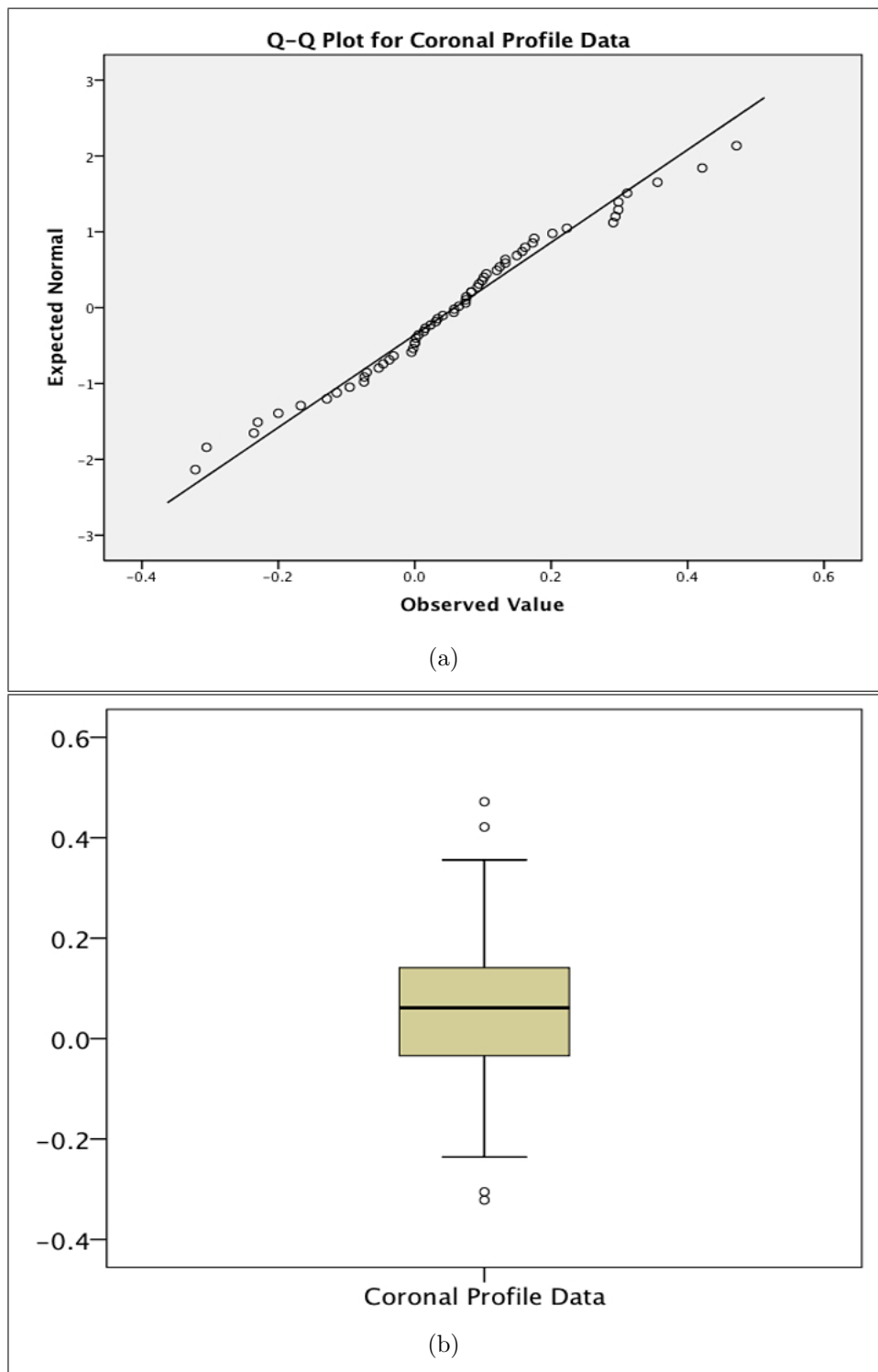


Figure 6.5: Showing results of Shapiro-Wilks tests of normal distribution across all three groups: Normal, TKR, PFJR ($n = 60$): (a) Q-Q plot, (b) box and whisker plot.

6.3 Results

6.3.1 Patellofemoral Kinematics

This section reports patellar displacements relative to the femur.

Normal subjects showed a trend of *lateral* displacement of the patella between 20° and 90° (mean 6.0 mm) as shown in Figure 6.6.

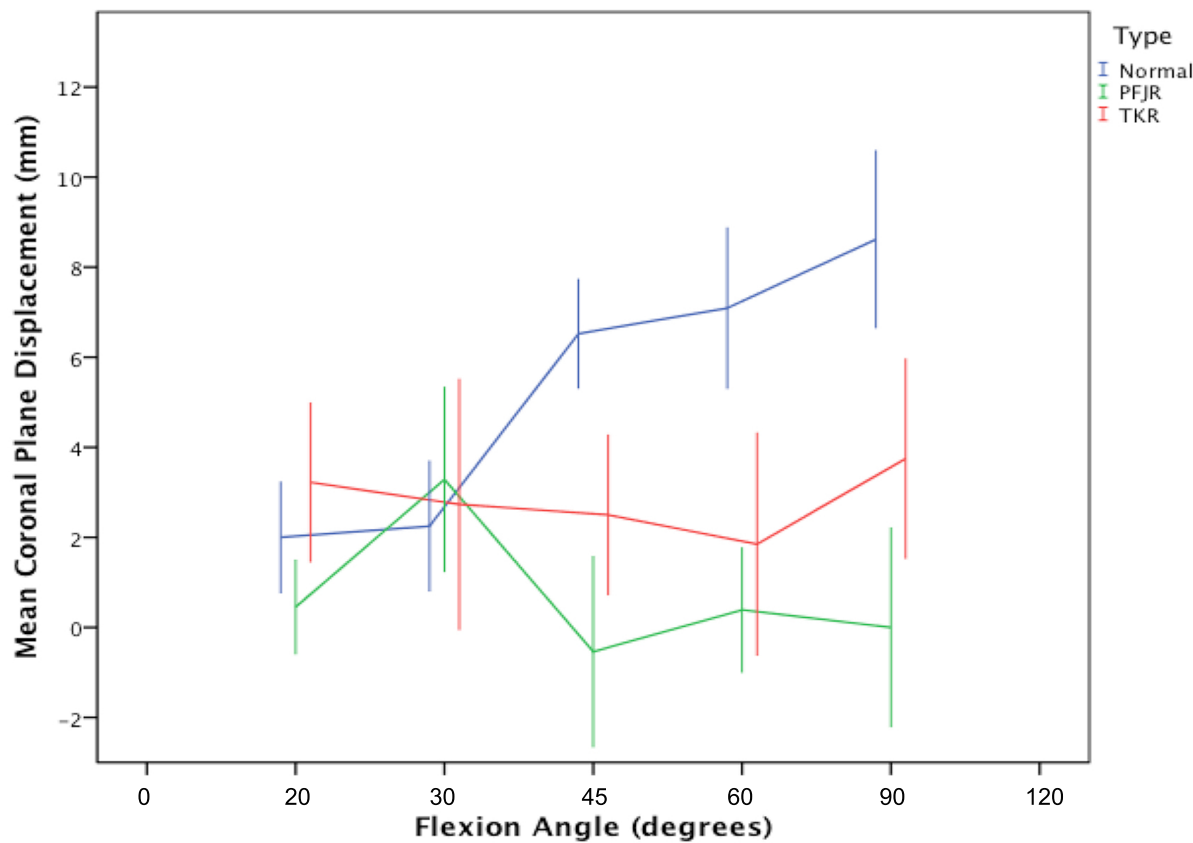


Figure 6.6: Showing typical coronal plane patellofemoral displacements during flexion for normal, TKR and PFJR patients.

An initial *lateral* movement is shown by PFJR patients occurring between 20° and 30°,

with a mean displacement of 2.9 mm. After a *Medial* displacement between 30° and 45° of flexion the patellae moved along a straighter trajectory to 90°, Figure 6.6. TKR patients, however, demonstrated the straightest trajectory with a net *medial* displacement of 1 mm between 20° and 60°, before a *lateral* shift between 60° and 90°, a mean displacement of 3.5 mm. Of note are that the TKR patients showed larger intersubject variation of measured data than either the PFJR patients or the normal subjects.

6.3.2 Patellotibial Kinematics

This section reports patella displacements relative to the tibia.

Figure 6.7 shows the movements of the patella relative to the tibia for the normal subjects. Between 20° and 60° there was a *medial* movement (mean 6.6 mm). This was followed by a *lateral* movement between 60° and 90° (mean 3.0 mm). Of note is the large range of measurements seen in normal subjects with standard deviations of up to 10.8 mm.

Figure 6.8 shows the trend of patellotibial movement in the replaced knee for the TKR and the PFJR patients. The trends for the TKR and PFJR groups are different from the *medial* - *lateral* trend of normal subjects and indeed different from each other. In early flexion TKR patients showed a *medial* movement to 30° (mean 2.9 mm), a *lateral* displacement between 30° and 45° (mean 5.1 mm) before a larger displacement *laterally* to 90° (mean 10.0 mm). The maximal standard deviation was 25.8 mm.

The PFJR group (Figure 6.8b) showed a *lateral* movement initially between 20° and 45° (mean 3.81 mm) followed by a *medial* movement between 45° and 90° (mean 7.9

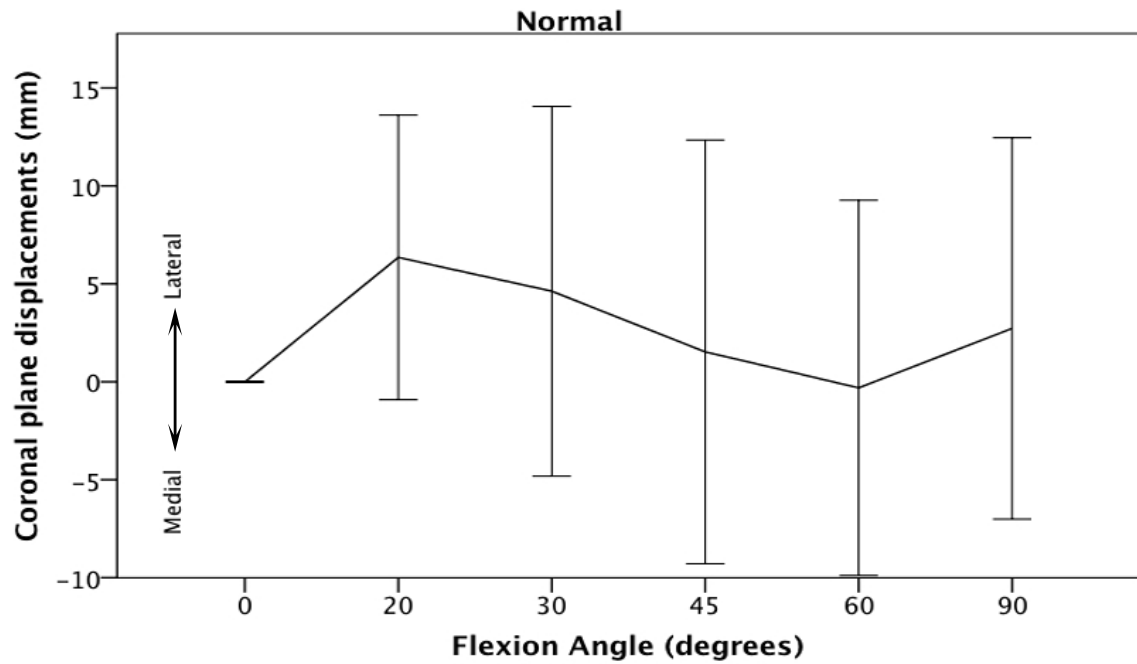


Figure 6.7: Showing typical coronal plane patellofemoral displacements during flexion for normal patients.

mm). The maximal standard deviation was 16.4 mm.

6.3.3 Coronal Profiles

The coronal profiles for each subject are given in Table 6.3. The mean coronal profile for normal subjects was 0.14 mm/°, 0.04 mm/° for TKR patients, and -0.02 mm/° for PFJR patients. A negative coronal profile indicates movements in a medial direction.

Significant differences were noted when comparing the *coronal profiles* of normal subjects and TKR patients ($p = 0.028$), normal subjects and PFJR patients ($p = 0.002$) but no significant difference between the PFJR and TKR patients ($p = 0.277$), as shown in Table 6.3.

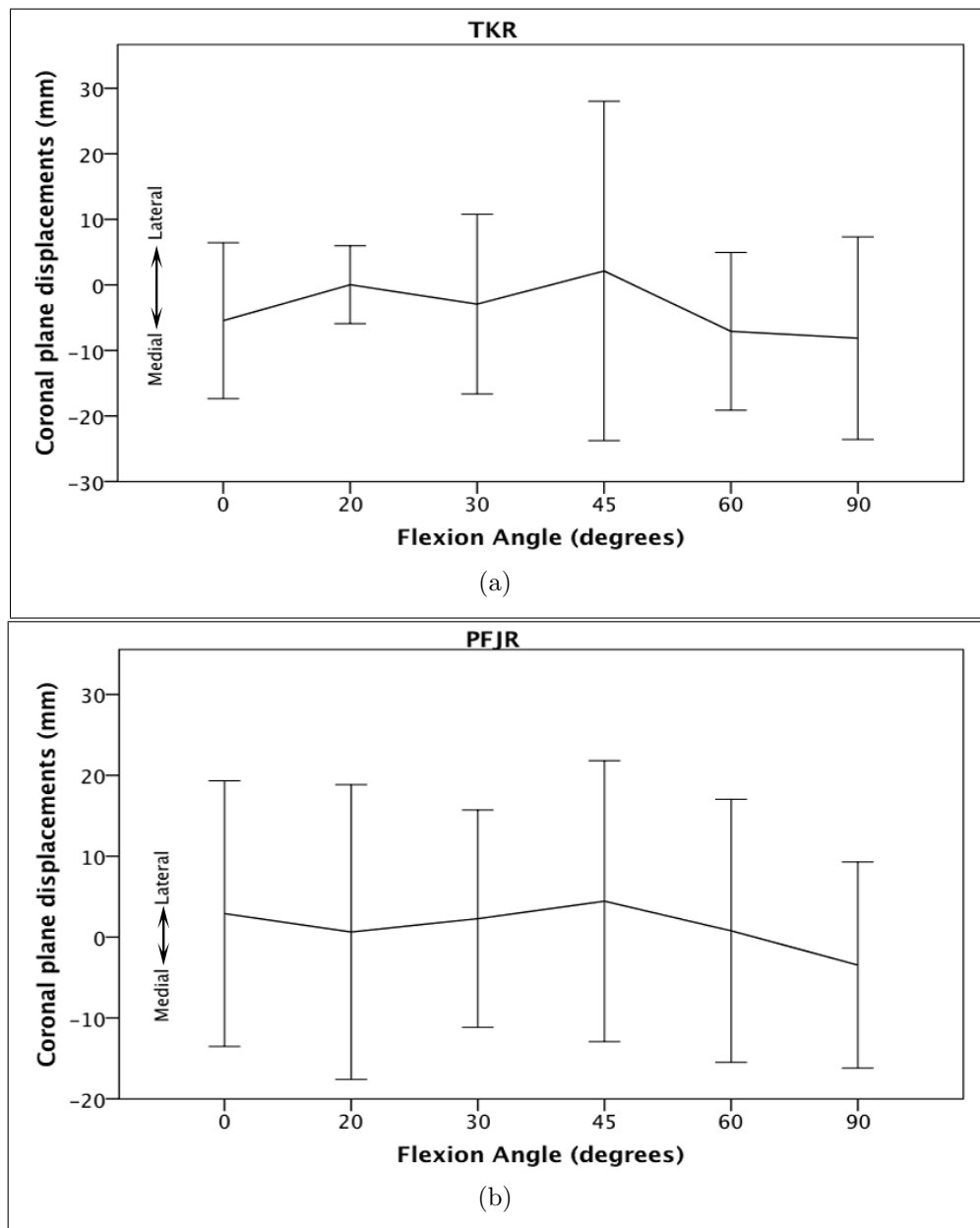


Figure 6.8: Showing typical coronal plane patellofemoral displacements during flexion for (a) TKR and (b) PFJR patients.

There were no significant differences between the coronal plane movements of the patella relative to the tibial segment between ANY of the groups (Table 6.4).

Measurements of Coronal Profile (mm/°)			
Subject	Normal	TKR	PFJR
1	0.3113	0.2942	0.1728
2	0.2018	-0.0706	-0.1289
3	0.1049	0.0337	-0.0745
4	0.0747	-0.3217	-0.2001
5	0.0941	-0.2302	0.3559
6	0.1573	0.0411	-0.0954
7	0.1618	-0.0022	0.0572
8	0.1328	0.2909	-0.0048
9	0.0989	0.0017	0.0156
10	0.0757	-0.0462	-0.1672
11	0.0828	-0.1145	0.0054
12	0.1326	0.0647	0.2987
13	0.1246	0.1495	-0.3053
14	0.1202	0.4717	-0.2359
15	0.4214	0.0306	0.0231
16	0.0828	0.092	-0.074
17	0.0745	0.0132	-0.0374
18	0.175	-0.053	-0.2273
19	0.2984	0.2229	0.0326
20	0.1012	-0.0312	-0.0819
Mean	0.14 mm/°	0.04 mm/°	-0.02 mm/°
Std. Dev	0.09 mm/°	0.18 mm/°	0.17 mm/°

Table 6.3: Showing results of coronal profile by subject type: Normal, TKR and PFJR.

Significance of Coronal Plane Movements of Patella		
	Femoral Coordinate System	Tibial Coordinate System
Normal vs TKR	p= 0.03	p= 0.54
Normal vs PFJR	p <0.01 (0.002)	p= 0.58
PFJR vs TKR	p= 0.27	p= 0.36

Table 6.4: Showing significance of coronal plane patella kinematics for normal subjects, FPV and TKR patients.

6.4 Discussion

Despite advances in operative techniques and component design, the issue of patient dissatisfaction with the outcome of their operation remains. The reason for this

is not clear. Whilst previous analyses have focused on sagittal plane kinematics, there is little reliable literature regarding coronal plane kinematics following arthroplasty. Understanding of the coronal plane kinematics in this group may help our understanding of the aetiology of anterior knee pain. Chapter 5 of this thesis documented the development of a technique to measure patellofemoral and patellotibial kinematics using ultrasound in combination with motion analysis.

The aim of this chapter was to assess whether the measurement tool could be applied to a measure coronal plane movements of the patella relative to both the femur and tibia in normal subjects and specifically following knee arthroplasty.

The results of this study demonstrate the successful implementation of the MAUS technique, presenting patella kinematic data regarding equal numbers of men and women in three groups of normal subjects, TKR and PFJR patients. Whilst similarities in the trends of movements were seen, significant differences between normal and replaced knees were highlighted.

Trends

The assessment of the patellofemoral kinematics revealed different trends of movement between normal and replaced knees. The predominantly lateral movement shown by the normal subjects was not repeated in the TKR and the PFJR patients, who demonstrated a much more neutral/straight trajectory. Possible reasons for this observation might be related to implant design and/or positioning of implants at operation. If the design of the prosthetic trochlear or patellar articulating surface does not reflect that of normal subjects it is possible that this may adversely effect the kinematics of the replaced knee. Similarly, alteration of the desired position and

orientation due to inconsistencies in surgical implantation technique may lead to abnormal kinematic profiles.

Of note is the range of variation of measurements within all of the groups for both patellofemoral and patellotibial measurements. In the normal knee, these balanced movements of the patella during flexion are influenced by the subtle interaction between bony and soft tissue constraints. Implantation of a prosthesis that does not mimic the previous bony arrangement may alter the freedom of movement of the patella and the soft tissues will be put under tension, which may lead to pain.

Extension to Engagement

The results of this study revealed large variation (2.9 mm to 26.5 mm) in patellar displacements from extension to 20°. It is during this flexion range (0° to 20°) that the patella engages into the trochlea. When at rest, standing with the knee in extension, the posterior capsule is tight, and the quadriceps are at rest. This leaves the patella mobile with the patellofemoral joint reaction force being effectively zero. The degree of movement that the patella undergoes between extension and engagement will then depend on the relative position of the trochlear groove to the resting position of the patella, which is dictated by the resting soft tissue tension in the quadriceps and patellofemoral ligaments, the anatomical axis of the femur, the offset of the tibial tubercle relative to the groove, the version of the femur and the torsion of the tibia. The trochlear position following implantation will depend additionally on the mediolateral orientation of the implanted components.

Latter Stages of Flexion

Between 60° and 90° both the patellofemoral and patellotibial data showed a lateral shift in the position of the patella in the normal subjects that was not seen in either the TKR or the PFJR patients. The lateral shift is likely to represent the movement of the patella off the condyle and into the inter-condylar notch. This movement is not evident to such an extent following TKR and PFJR. This is likely to be a reflection of the design of the notch component of the prosthesis, which does not mimic that of the normal subjects. As such, any movements in the notch that deviate from the normal pathway are likely to give rise to abnormal tension in the soft tissues which might lead to pain.

Observed differences between femoral and tibial coordinate systems

Patellar kinematics differed depending on the coordinate frame from which the patella was viewed. The differences noted between normal and replaced knees from the femoral perspective were not repeated from the tibial perspective. For the purposes of this study, only patellar movements in the coronal plane have been calculated. Axial rotations between femur and the tibia have not been studied here. This study has shown that the trends of movements of the patella relative to the femur were not reproduced when assessing the patellotibial kinematics. It is possible that the movements of the patella are dictated primarily by the femur, following which there is a rotational movement of the tibia to maintain the patella position in line with the tibial tuberosity. Rotational movements of the tibia during flexion may represent positional change dictated by soft tissue tension created during patellofemoral movements. If patellofemoral movements were to dictate tibial movements, this might also explain the lack of trend in the patellotibial data.

Coronal Profiles

Using the *coronal profile* outcome measure, normal knees were shown to have significantly different patellofemoral kinematics from both TKR and PFJR. There was no significant difference between the two types of replacement. This implies that the kinematics of the normal healthy knee joint are not reproduced following implantation of components in the coronal plane. It is possible that this may in part be due to the shape of the components not being similar to the shape of the normal knee.

The mean patellofemoral coronal profiles for PFJR patients were $-0.02 \text{ mm}/^\circ$ compared to $0.14 \text{ mm}/^\circ$ for normal subjects and $0.04 \text{ mm}/^\circ$ for TKR patients. The negative value for the PFJR coronal profile indicates a net movement of the patella medially. This might be a reflection of the challenges of orientating the trochlea during implantation. Whilst during TKR referencing of position can be measured from and dictated by, the position of the femoral condyles, the same technical constraints are not present during PFJR. As such, the final position and orientation of the PFJR trochlear component is less controlled by instrumentation and more by the relative thickness of the articular cartilage on the medial and lateral femoral condyles at operation, the optimal position being decided by the operating surgeon. This might explain the more medial trajectory of the patella during flexion in the PFJR patients.

Limitations

The axis chosen for coronal plane measurements of patellar displacements relative to the femur was along a line joining LASI and RASI markers, and for the tibia; the medial and lateral malleoli. It is possible that these do not represent a true medio-lateral direction and as such will not be measuring the true line of displacement. As such, the line of displacement of the patella is likely to lie at an angle to the LASI-RASI line. This in turn will give rise to inaccuracies in the measurement of magnitude of coronal plane displacements. A similar problem will be present using the medial maleolus (tibia) and lateral maleolus (fibula). Future attempts to define this axis should aim to use landmarks situated on the relevant bone (femur and tibia).

In designing this experiment, it was important to consider a range of positions of flexion that could be examined in all subjects, regardless of whether the subject had a normal or replaced knee, and independent of demographics. The choice of range, 0° (full extension) to 90° proved attainable by all subjects. To represent this range the isolated positions of extension, 20°, 30°, 45°, 60° and 90° were chosen. However, following removal of data from the first 20° of flexion due to the observed engagement discontinuity, only five positions remained. Future studies will aim to increase the number of positions scanned.

Each subject was measured only once. The preparation of subjects (including physical measurements and marker placement) and scanning process took approximately 45 minutes to one hour for each leg. The squatting exercise was at times strenuous for some subjects, in particular the joint replacement groups. For this reason it was considered inappropriate to repeat the process. With further refinement to technique

(Chapter 8) the time required to complete the scanning process should decrease, thereby proving an opportunity for repeated measures.

The data presented here reveals a trend in patella displacements. Data from normal subjects suggests that the patella follows a curved path down the trochlea. The choice of a single outcome measure that can be applied to all data that adequately describes the characteristics of the curve, is difficult. For the purposes of this experiment a straight line was fitted to the curve (*coronal profile*). This will mask the subtle differences in curvature between subjects. Alternative outcome measures that could be trialled include polynomial curve fitting and regression algorithms. These mathematical algorithms describe the characteristics of curves and may provide a more descriptive measurement tool.

6.5 Summary

This chapter has shown the successful use of the MAUS technique in assessing normal and replaced knees. Significant differences were seen between the coronal plane kinematics of normal and replaced knees using the MAUS technique. This would imply that following implantation of knee arthroplasty the kinematics of the knee are altered.

Chapter 7

Anatomical Factors Affecting PFJ Kinematics

Techniques for measuring patellar displacements in the coronal plane and descriptions of patellarkinematics in normal and replaced knees have been addressed (Chapters 5 and 6) and the potential consequences of abnormal patellarkinematics after joint replacement on surrounding soft tissues discussed. One focus of the discussion has been considering how new designs might help address patellofemoral problems, however the investigation would be incomplete without examining anatomical factors which might also lead to pain and dysfunction in native knees. Whilst analysis of the pain pathways involving soft tissues of the knee are beyond the scope of this thesis, further analysis of the anatomical factors within the knee which maintain normal patellarkinematics and prevent abnormal movements, are included here.¹

¹The work described in this chapter has been published as Monk et al. The Pathoanatomy of Patellofemoral Subluxation. (2011) JBJS 93-B:10, p1341-1347

7.1 Introduction

Anterior knee pain is one of the commonest knee problems. In many cases it is thought to be related to patellofemoral subluxation and instability [191]. This condition represents a spectrum ranging from mildly abnormal patellofemoral tracking to complete dislocation where no part of the patella remains in the trochlear groove. Within this spectrum there exists patellar subluxation, where the apex of the patella is not moving down the trochlear groove. Subluxation can be assessed using dynamic Magnetic Resonance Imaging (MRI) [157, 191, 192]. Measurement of patellar subluxation from dynamic MRI scans, based on a cine-loop of sequential axial images, is semi-quantitative and has been shown to be reproducible and reliable [8, 157]. There is an established relationship between increasing subluxation and worsening pain [8, 157, 192]. O'Donnell et al [8] studied the relationship between pain and subluxation in asymptomatic volunteers and patients with anterior knee pain and found an association between pain and subluxation. They reported, in the asymptomatic group, 91% had no or mild subluxation, 9% had moderate subluxation and no subjects had severe subluxation; in contrast in the painful group, 17% had moderate subluxation and 7% had severe subluxation.

To date there have been few MRI studies exploring the relationship between bone and soft tissue abnormalities that might lead to abnormal kinematics of the patellofemoral joint. In addition, whilst studies allude to the contribution of trochlea and quadriceps dysplasia to patellar maltracking [123, 193], few have quantified their relative importance.

The aim of this study was to examine the relationship between specific anatomical

variables measured on MRI scans and patello-femoral subluxation, and to quantify their importance. The variables studied were those thought give a predisposition to subluxation and include: position of the patella relative to the trochlea in extension, the length of the vastus medialis tendon, bony dysplasia in both the trochlea and patella, and cartilage thickness.

7.1.1 Patients and Methods

Between April 2006 and February 2008, 270 patients with anterior knee pain with possible patellofemoral maltracking were referred by specialist orthopaedic surgeons for MRI at our institution. Magnetic resonance scans were carried out using a Siemens 1.0 Tesla Expert Magnetic Resonance Image system (Siemens AG, Munich, Germany). The MRI examination consisted of routine 3D acquisition sequences with the knee in extension, including fat suppression images, a sagittal orientated T2 gradient sequence, and axial sequences (inter-slice distance 1.6 mm). An additional patellofemoral dynamic MRI examination was also performed using a previously developed technique [157]. Exclusion criteria included any previous surgery to the knee, any bony abnormality (including previous fracture or arthritic change) and any major cartilage defects (greater than 10 x 10 mm). The principal inclusion criterion was that there were adequate quality axial, coronal, sagittal and dynamic films for the purposes of assessment. A power calculation was performed using a commonly measured variable: the Tibial Tubercle to Trochlear Notch Distance (TTD). The calculation was based on differences seen at full extension. To identify a difference of 1 mm using Altman's method, 2n was found to be 30 for 90% power (Standard Deviation or SD = 3.3, Standardised Difference = 1.1). Sixty patients were randomly selected from the referral group that met the inclusion criteria. There were 37 females

and 23 males, with 30 right and 30 left knees. The dynamic MRI examination was carried out with the knees raised on a cushion and strapped together (Figure 7.1).



Figure 7.1: MRI setup for dynamic scanning of knee joint.

An inflated plastic ball was then placed between the distal tibiae and the upper part of the magnet. During the scan, the patient operated a hand-held valve which released air from the plastic ball, whilst simultaneously extending their knee against the resistance of the deflating ball (Figure 7.2).

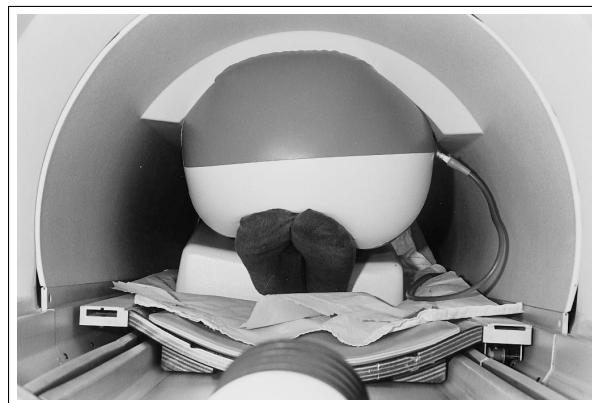


Figure 7.2: Patient deflating the beach ball during dynamic scanning of knee joint.

A series of six axial slices, with an inter-slice spacing of 5 mm thickness, were ob-

tained for each patient, as well as one sagittal slice. The axial slices included the full excursion of the patella. The sagittal image was orientated along the long axis of the patella; this slice was used to ensure alignment. Over two minutes the sequence was repeated 15 times. A cine-loop feature then combined the axial slice closest to the centre of the patella from each of the 15 repeat sequences to provide a dynamic coronal plane evaluation throughout the range of extension [157]. Two independent consultant radiologists assigned a grade of radiological subluxation to each patient's dynamic MRI scan, with each knee being evaluated with respect to its degree of subluxation. Measurements were made of the maximum lateral movement of the apex of the patella in the trochlear sulcus. The lateral movement was graded using the following scale established by McNally et al, 2000 [157]:

- No/mild = 0 - 5 mm subluxation
- Moderate = >5 - 10 mm subluxation
- Severe = >10 mm subluxation

The study sample included 10 patients (16.7%) with mild clinical or asymptomatic subluxation, 32 patients (53.3%) with moderate subluxation and 18 patients (30%) with severe subluxation as established by the dynamic MRI assessment.

For each subject the following variables were measured using routine 3D MRI sequences obtained with the knee in extension. Measurements were made using 3D post-processing software (Centricity PACS-IW, GE Healthcare, New York, US).

Measures

1. Tibial Tubercle to Trochlear Notch Distance (TTD). This was defined as the distance between the mid-tubercle plane (mid-point of the patellar tendon) and the mid-sulcal plane located on axial MRI slices. Both lines were drawn perpendicular to a line joining the posterior aspect of the femoral condyles (Figure 7.3).

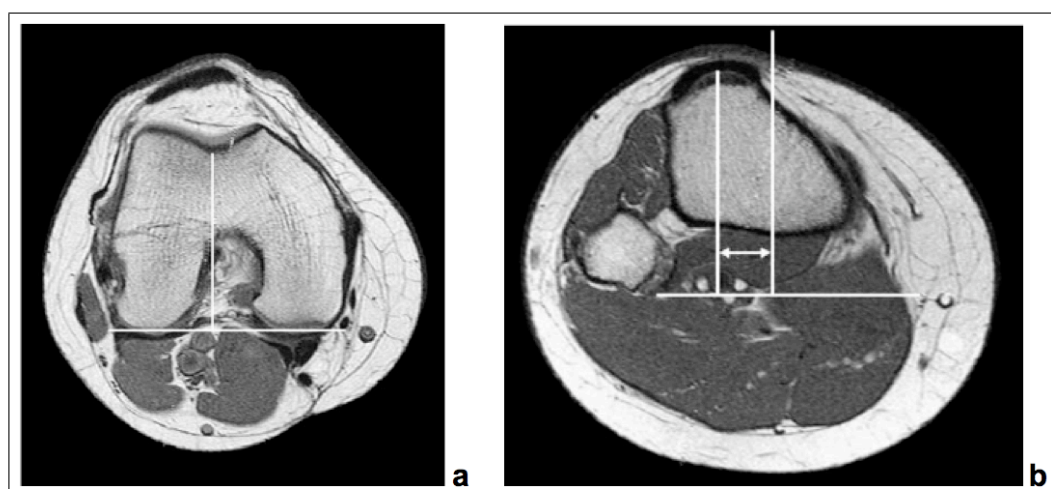


Figure 7.3: Measurement of Tibial Tubercle - Trochlea Notch Distance (TTD) from a (central position of trochlea sulcus, referenced from the posterior facets of the femoral condyles) to b (tibial tubercle).

2. Patella Type. Based on Wiberg Classification of patella shape and assessed on axial slices:

Type I: the facets are concave, symmetrical and of equal size.

Type II: the medial facet is rather smaller than the lateral facet. The lateral facet is concave.

Type III: the medial facet is markedly smaller than the lateral facet (Figure 7.4).

3. Patella Alta. A modified Insall-Salvatti ratio was used based on the length of the

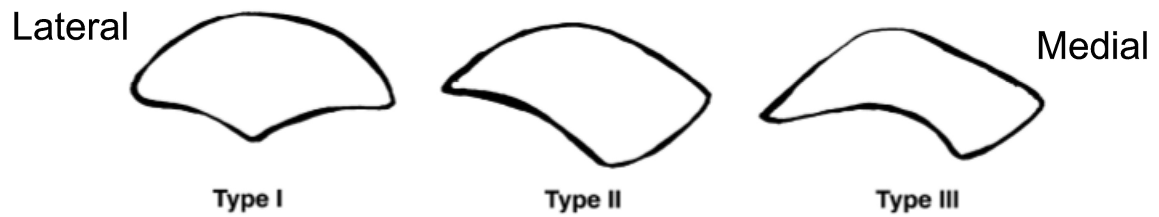


Figure 7.4: Showing Wiberg Classification of the Patella. Type I - equal sized facets, Type II - slightly larger lateral facet, Type III - much larger lateral facet. From Tecklenburg et al 2006.

patella divided by the length of the patellar tendon measured along its deep surface from sagittal slices, a value of less than 0.8 representing patella alta (Figure 7.5) [194, 195].

4. Patella Engagement. Engagement of the patella into the trochlear sulcus in full extension was expressed as a percentage. The highest point of the articular surface of the trochlear sulcus was identified using the highest axial slice on which cartilage could be identified. Using the multi-planar localisation tool, this point was identified on the sagittal view, allowing the corresponding point on the articular surface of the patella to be identified. The height of this point from the inferior aspect of the articular surface of the patella was then measured. From this measurement, the percentage of the articular surface engaged in the trochlear sulcus was calculated (Figure 7.6).

5. Trochlear sulcus angles. The angle subtended by the walls of the trochlear sulcus at the base of the groove was measured for both bone and cartilage. Whilst routine MRI utilises axial slices, perpendicular to the femoral axis, for the purposes of this study measurements were taken from reconstructed axial slices, to achieve the plane perpendicular to the articulating surface (Figure 7.7).

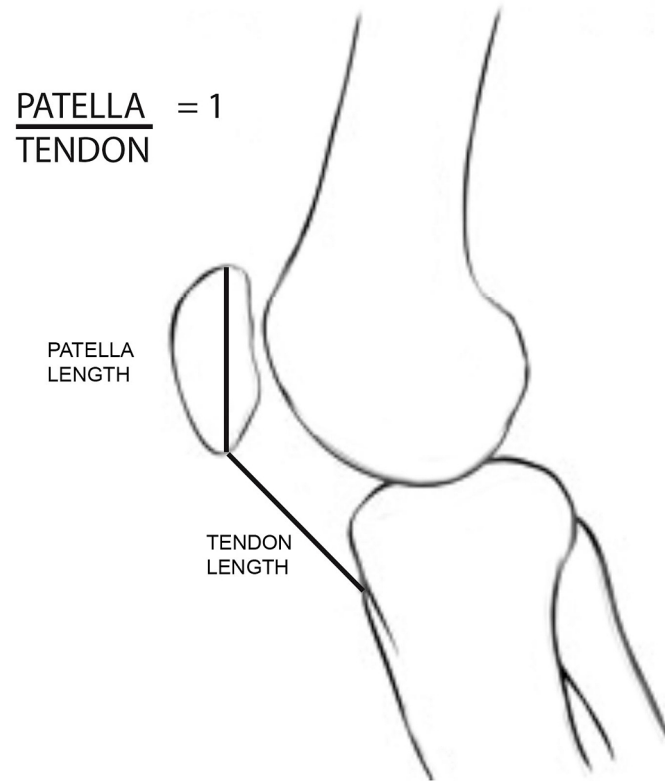


Figure 7.5: Insall Salvatti technique for measuring patella alta; ratio of patellar length to patellar tendon length. From Monk et al, 2011.

Both the bone and cartilage sulcus angles vary as a function of height within the trochlea, therefore sulcus angle measurements were taken at three reproducible points within the trochlea (Figure 7.8):

- (1). The axial slice 3.2 mm (i.e. two slices) distal to the most proximal axial slice showing articular cartilage in the trochlea. The measurements of sulcus angle at this level were termed the top bone sulcus angle (top bone S.A.) and the top cartilage sulcus angle (top cart S.A.).
- (2). The axial slice 1.6 mm (i.e. one slice) proximal to the first axial slice showing

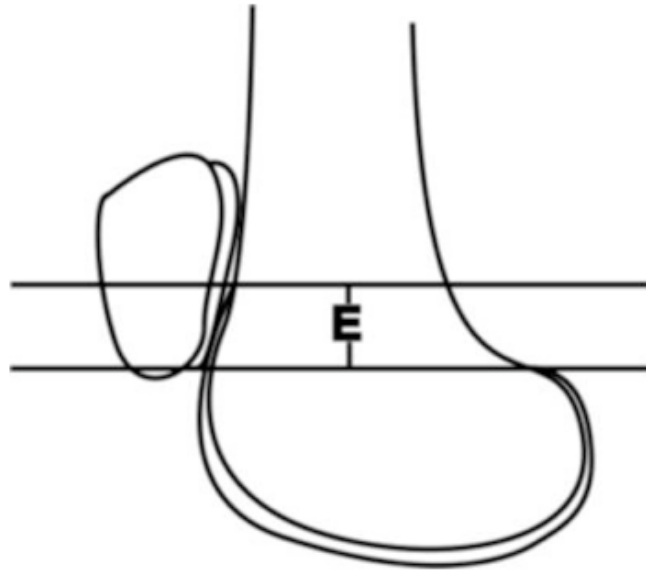


Figure 7.6: Showing the percentage overlap (E) of patella cartilage in the trochlear groove.

the inter-condylar notch. The measurements of sulcus angle at this level were termed the bottom sulcus angles (bot S.A.).

(3). Midway between the slices described in 1 and 2 (Figure 7.8) represented the middle sulcus angles for both the bone and the cartilage (mid S.A.).

6. Distance of Vastus Medialis Obliquus muscle from patella (VMOP). This was defined as the shortest distance between the supero-medial aspect of the patella, to the edge of the vastus medialis obliquus muscle.

7. Patella apex cartilage thickness. The cartilage thickness at the apex on the posterior aspect of the patella.

8. trochlear sulcus cartilage thickness. The maximum thickness of cartilage in the trochlear groove. Measured from axial slices.

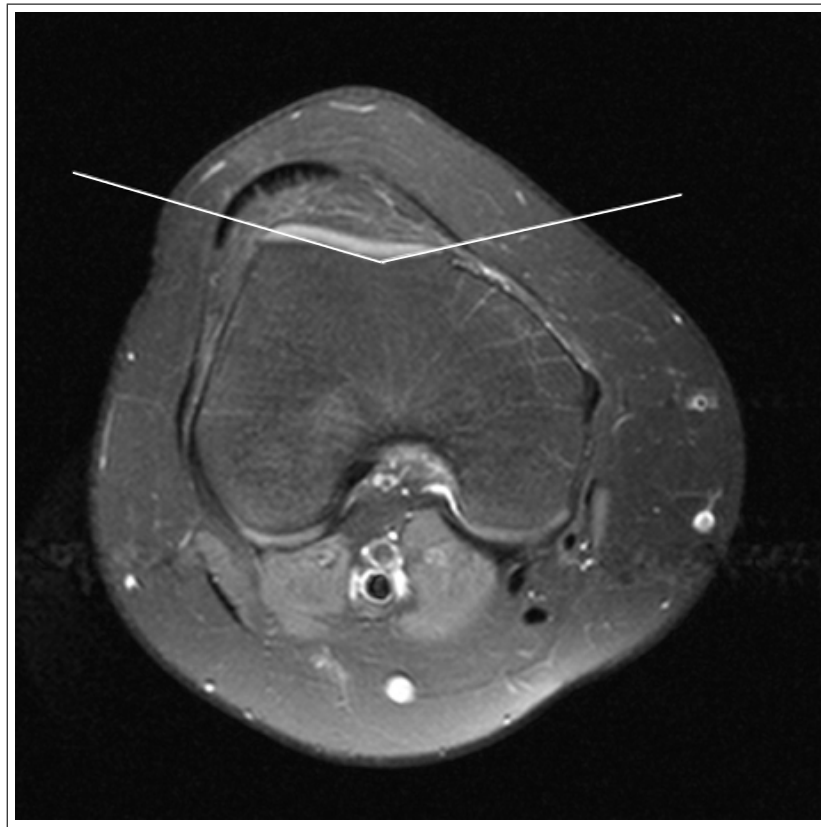


Figure 7.7: An example of sulcus angle measurement for bone.

Analysis

The data were analysed in SPSS (SPSS v18.0.1, Chicago, Illinois, 2010). To assess characteristics within the three radiological subluxation groups, descriptive statistics for each measured variable were calculated: mean with standard deviation (SD) for normally distributed variables and median with IQR for non-normally distributed variables. Analysis of variance (parametric and non-parametric Kruskal-Wallis) was used to determine the statistical significance of any group difference in individual variables, in particular whether there was any significant linear trend across the groups. A linear trend was taken to be significant as long as its accompanying departure from linear trend was non-significant.

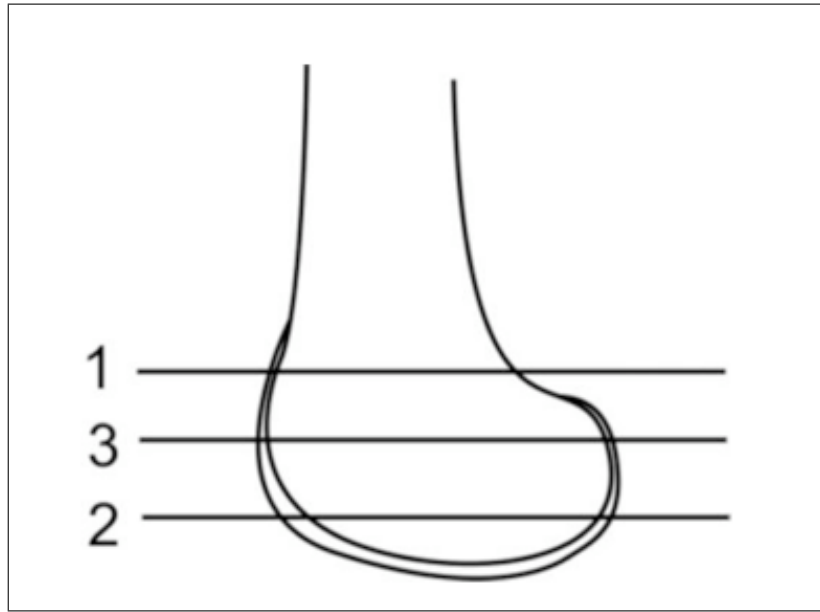


Figure 7.8: Showing the three levels at which the sulcus angles were measured.

Discriminant function analysis, a type of multivariate analysis of variance [196], was applied to establish whether the measured variables, when considered together, could discriminate between the three patellofemoral subluxation groups. The Wilks' lambda statistical test was used in an ANOVA (F) test of mean differences, such that the smaller the lambda for an independent variable, the more that variable contributes to the discriminant function. For any variable lambda varies from 0 to 1, with 0 meaning group means differ (thus the more the variable differentiates the groups), and 1 meaning all group means are the same. The F test of Wilks' lambda shows which variables contributions are significant [196].

In addition, an approach based on diagnostic test methodology Receiver Operator Characteristic (ROC) analysis was used to identify if any thresholds for radiological stability/instability were evident [197]. ROC curves plot sensitivity (y-axis) against

1-specificity (x-axis) for all possible cut-off points. Sensitivity is defined as the proportion of all patients with a positive outcome correctly identified (proportion of true positives identified), while specificity refers to the proportion of all patients with a negative outcome so identified (proportion of true negatives). The most efficient cut-off value, giving equal weight to specificity and sensitivity, is associated with the point closest to the top left hand corner of the ROC curve. The greater the area under the ROC curve (AUC) the greater the ability of the measure to differentiate. If the area under the curve is 0.5, this represents a test that is not predictive; the closer to 1.0, the better the differentiation of the measure.

7.1.2 Results

A summary of the variables measured for the patients in each of the three radiological subluxation groups is given in Table 7.1. There were statistically significant differences between the groups in terms of TTD, Patella Alta, Patella Engagement, Top bone S.A., Top cart S.A, VMOP, and Cart Thickness Troch, with there being linear trends across the groups for all characteristics but Patella Engagement for which the relationship was strong but non-linear.

The discriminant function analysis found that six of the measured variables in combination were statistically significantly able to discriminate between the three grades of subluxation (Table 7.2; variables given in order of the strength of the relationship): Patella Engagement (percent overlap of patella in trochlear sulcus); TTD (Distance between tibial tuberosity and apex of trochlear notch); VMOP; Patella Alta; Top bone S.A.; Top cart S.A. Patella engagement had a notably stronger relationship ($\lambda=0.6$, $p<0.001$) than the other variables ($\lambda=0.8$ to 0.9 , $p>0.001$). The measured variables that showed no discrimination included: Wiberg

Severity of subluxation				
Variable	Mild (n=10)	Moderate (n=32)	Severe (n=18)	ANOVA*
TTD	12.0(4.1) mm	15.4(5.5) mm	20.7(4.3)mm	p<0.001
Patella Type (Wiberg classification), n(%)				
Type 1	2(20.0%)	5(15.6%)	3(16.7%)	k-w p=0.82
Type 2	8(80.0%)	25(78.1%)	14(77.8%)	
Type 3	0(0%)	2(6.3%)	1(5.6%)	
Patella Altao	0.97(0.10)	0.85(0.17)	0.82(0.13)	p=0.013
Patella Engagement	41.4(7.5)%	20.5(12.5)%	16.2(8.4)%	p<0.001 p=0.005++
Trochlea sulcus angles				
Top bone S.A.	142.0(7.9)°	155.2(16.7)°	172.2(22.2)°	p<0.001
Mid bone S.A.	136.1(10.1)°	141.3(13.4)°	144.0(18.5)°	p=0.18
Bot bone S.A.	127.9(11.6)°	131.5(13.8)°	133.0(13.8)°	p=0.35
Top cart S.A.	150.2(10.9)°	163.0(20.7)°	175.3(18.5)°	p=0.001
Mid cart S.A.	143.8(11.8)°	155.1(15.4)°	150.7(12.9)°	p=0.22
Bot cart S.A.	134.6(13.2)°	138.4(13.4)°	143.0(16.6)°	p=0.15
VMOP	3.3(2.9) mm	6.2(4.2) mm	8.2(2.8) mm	p=0.001
median(IQR)	4.9(0-6.0)	5.6(3.7-8.5)	8.0(6.4-10.7)	k-w p=0.004
Cart Pat	4.8(1.3) mm	4.4(0.7) mm	5.2(1.1) mm	p=0.36
Cart Troch	3.0(0.8) mm	4.0(1.1) mm	4.3(0.8) mm	p=0.001

Table 7.1: Characteristics of the three radiological subluxation groups (mild, moderate, severe). Data are shown as mean (SD) unless otherwise stated. o pat length / tendon length *p-value for linear trend on one-way ANOVA, unless otherwise stated ++ deviation from linear trend

Classification of the Patella; Thickest cartilage on trochlea; Thickest cartilage on patella; Mid bone S.A.; Bot bone S.A.; Mid cart S.A.; Bot cart S.A.

The variable with the strongest association with subluxation, patella engagement ($\lambda=0.644$) had been observed to have a strong, although, non-linear relationship with subluxation group (p<0.001; Table 7.1), with mean engagements of: 41% (range 34% 48%) for no/mild subluxation, 20% (range 12% - 32%) for moderate and 16% (range 8% 24%) for severe subluxation. The non-linearity of the relationship can be seen in Figure 7.9 while there were a few patients with patella engagement greater than 30%

Output of Discriminant Analysis		
Variable	Wilks Lambda	F test of Wilks lambda (p value)
Patella Engagement	0.644	<0.001
TTD	0.828	0.001
VMOP	0.831	0.001
Patella Alta	0.843	0.001
Top bone S.A.	0.866	0.002
Top cart S.A.	0.877	0.006
Mid cart S.A.	0.952	0.096
Bot cart S.A.	0.985	0.348
Mid bone S.A.	0.989	0.435
Bot bone S.A.	0.993	0.525
Wiberg Patella Type	0.996	0.616

Table 7.2: Table showing output of the Wilks Lambda test of multivariate analysis (S.A. = sulcus angle, cart = cartilage).

who had moderate or severe subluxation, no patient with engagement of <30% had no/mild engagement.

On ROC analysis, a threshold of 30% was found to be the best cut-point for radiological stability, with values >30% having a sensitivity of 1.0 and a specificity of 0.86 (area under the curve, AUC = 0.94). Patella alta, although assessing similar features to patella engagement had less discriminatory power ($\lambda=0.84$) than patella engagement ($\lambda=0.64$) when considered in combination with it, and on individual analysis also had a weaker relationship ($p=0.013$; Table 7.1).

The trochlear sulcus angles for bone and cartilage in the superior trochlea were significantly associated with subluxation, both individually and in combination (Tables 7.1 and 7.2, respectively). With increasing sulcus angle (flattening of the trochlear groove) the degree of subluxation increases (Figure 7.10).

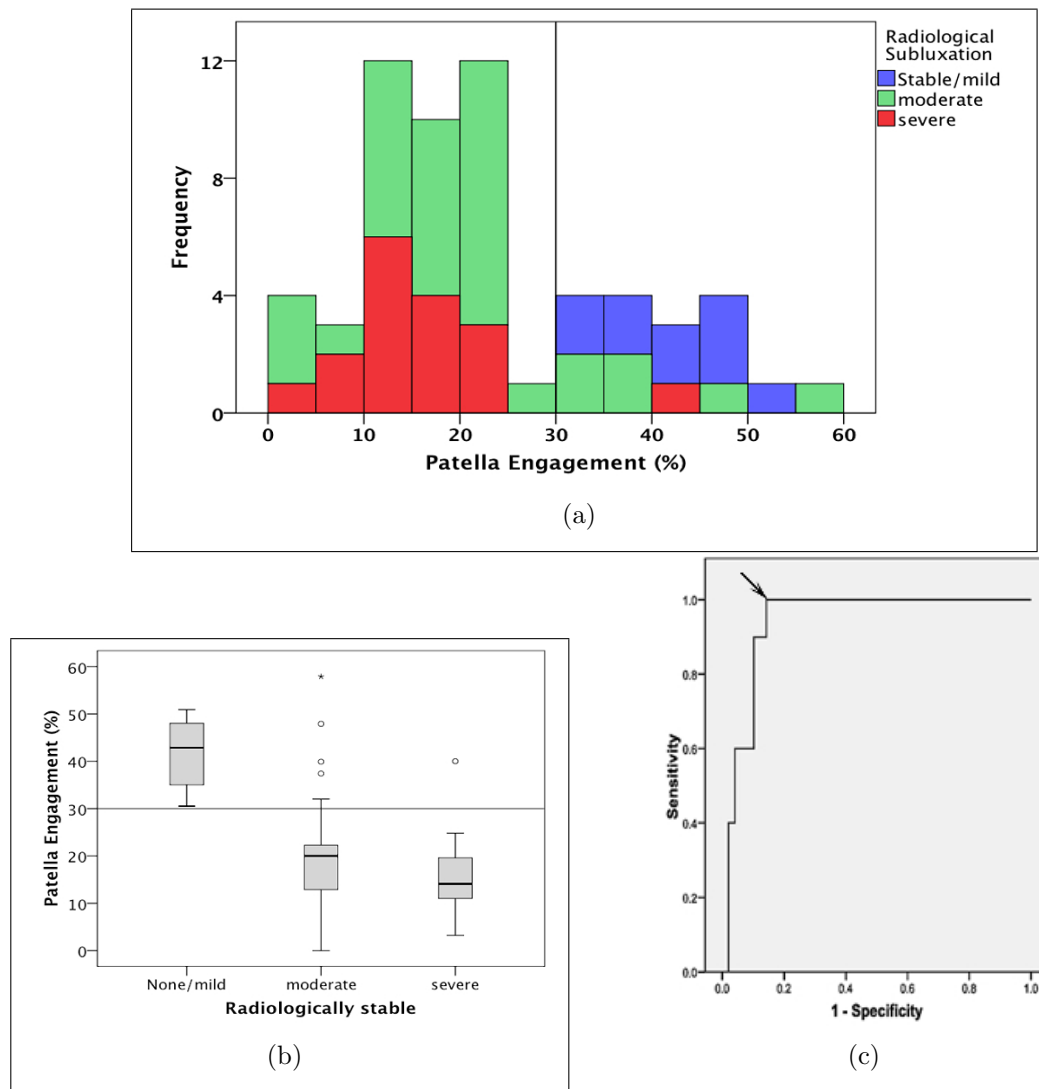


Figure 7.9: a) showing the distribution of subjects patella engagement, b) the relationship between patella engagement and radiological instability. Threshold indicated at 30 percent. c) ROC curve showing the relationship between patella engagement and radiological stability.

Whilst all stable subjects showed a sulcus angle less than 155° , there was a large degree of overlap between the subluxation grades, with a number of subjects in the moderate and severe subluxation groups also showing a sulcus angle of less than 155° . On ROC analysis, a threshold of $152^\circ (>152^\circ)$ had a sensitivity of 68% and specificity of 100% for detecting at least moderate radiological instability (AUC=0.82).

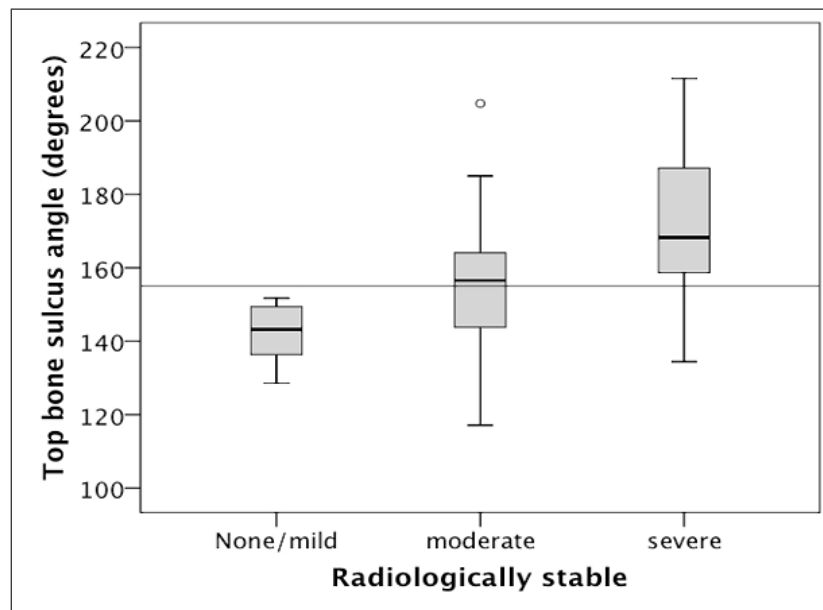


Figure 7.10: Showing the distribution of the sulcus angles at the top of the bony trochlea for the three radiological stability groups

With increasing grade of subluxation, the tibial tubercle to trochlear notch distance (TTD) increased from a mean of 12.0 mm (SD 4.1 mm) to a mean of 20.7 mm (SD 4.3 mm) (linear trend $p < 0.001$; Table 7.1). As with sulcus angles, there was overlap between the subluxation grades with the best cut-point from ROC analysis found to be a value of 14.5 mm: values greater than this had a sensitivity of 76% and specificity of 70% for detecting radiological instability (AUC=0.80) (Figure 7.11). Every subject with no/mild instability had a TTD of < 19 mm; the sensitivity of this threshold (< 19 mm for stability) was 40% and specificity 100%.

As the distance of the VMO muscle from the patella (VMOP) increased, overall subluxation increased ($p = 0.001$; Table 7.1 and Figure 7.12).

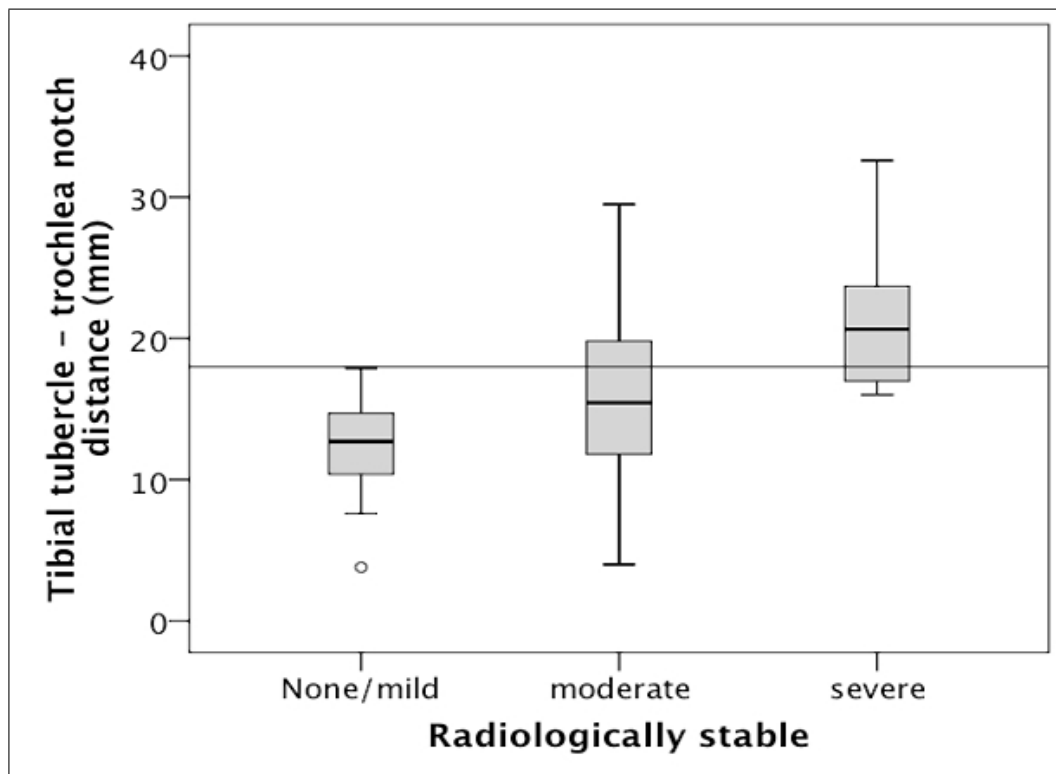


Figure 7.11: A box and whisker plot showing the relationship between radiological subluxation and tibial tubercle-trochlea notch distance (TTD).

7.1.3 Discussion

Abnormal patellofemoral tracking has been shown to contribute significantly to anterior knee pain, although the reason why maltracking causes pain is not well understood [9, 198]. During knee flexion, subluxation of the patella leads to abnormal stresses being transferred through surrounding soft tissues, the cartilage and subchondral bone. These factors may contribute to anterior knee pain. The causes of patellofemoral subluxation are highly likely to be multi-factorial. Although certain aberrant anatomical features are known to contribute to subluxation of the patella, to date there have been no studies that which have attempted to identify which anatomical features are most relevant.

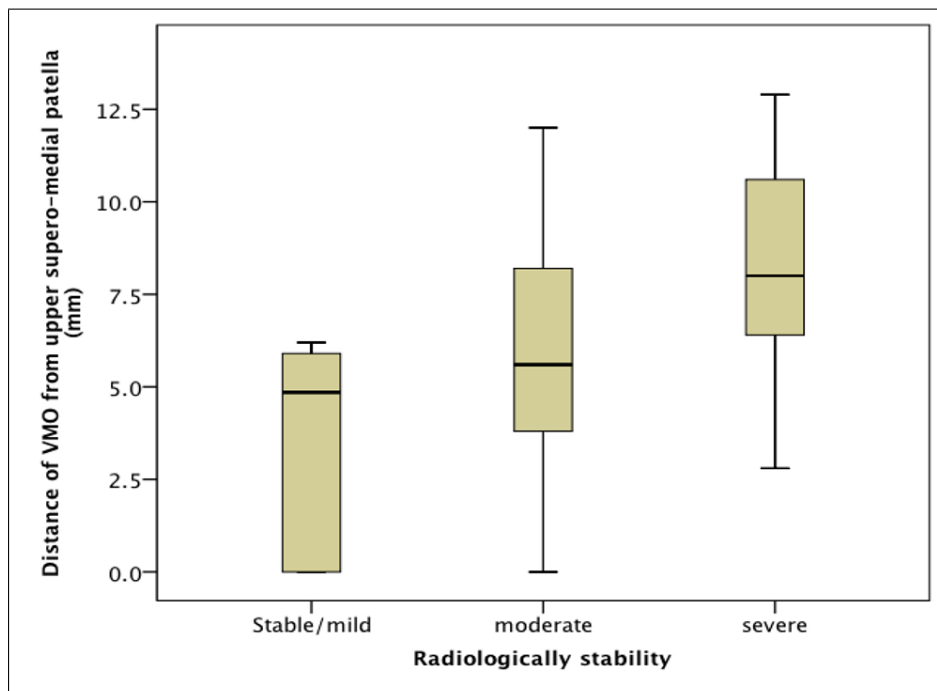


Figure 7.12: Showing the relationship between radiological stability and the distance of the Vastus Medialis Obliquus muscle belly from the edge of the bony patella.

Our analysis suggests that important anatomical factors associated subluxation include the position of the patella relative to the trochlea (Patella Engagement, Patella Alta), the position of the tibial tubercle (TTD), VMO position and the morphology of the proximal trochlea (proximal bony and cartilage sulcus angle). The factors which probably do not cause subluxation include: the shape of the patella, the morphology of the middle and lower parts of the trochlea (sulcus angles for cartilage and bone) and cartilage thickness of the trochlea and patella.

Patella engagement had the strongest association with patella subluxation; this variable effectively assesses the start position of the patella in relation to the trochlear sulcus in extension, expressed here as percentage overlap. This implies that as percentage engagement in the trochlea increased, the degree of subluxation

decreased. The histogram in Figure 7.9a shows that no subject within the mild group had a patella engagement below 30%. This implies that a threshold exists to distinguish between the no/mild radiological subluxation group and the moderate/severe groups; less than 30% engagement was associated with moderate or severe subluxation (Figure 7.9).

Previous authors have documented the relevance of a high riding patella as a prerequisite to subluxation, supported by the fact that patella alta is a feature present in over 50% of cases of dislocation [199]. In this study, patella engagement showed a more significant relationship with the degree of subluxation in the discriminant analysis test than patella alta. In addition, there was a clear threshold between no or mild subluxation and moderate or severe subluxation in terms of patella engagement but this was not evident for patella alta. Intuitively it seems likely that patella engagement will directly affect stability, whereas patella alta (calculated using the modified Insall-Salvatti method) will only affect it indirectly by its relationship to patella engagement.

Problems with the radiographic measurement of patella alta have been well documented, the most notable being the difficulty in establishing the origin and insertion of the patellar tendon [200, 201]. The measurement of patella engagement from MRI images provides an alternative measurement with well defined landmarks offering a more precise description of the patella's position relative to the trochlea [202].

The findings reported here suggest that if patella engagement can be optimised, then the risk of patella subluxation should be decreased. The data suggests a clear cut-off

at 30% engagement between subluxation and no subluxation. Therefore, surgery aimed at preventing subluxation should perhaps aim to achieve patella engagement between 30% and 50%.

Subluxation and dislocation have been shown to typically occur during 0° to 30° flexion and infrequently in the terminal phases of extension [9]. During the first 30° of flexion and approaching full extension, the patella will be in the superior aspect of the trochlea. This study shows that a relatively flat superior trochlea correlates with worsening subluxation (Figure 7.10). The shape of the trochlea lower down has no significant effect on subluxation.

The position of the belly of the VMO in relation to the patella was shown to be a significant feature associated with patellar subluxation. It follows, therefore, that interventions that enhance the action of VMO should improve symptoms related to PFJ subluxation. However, this study demonstrated no clear cut-off in VMOP between subluxation and no subluxation.

Previous assessments of the contribution of increasing TTD to patellofemoral subluxation have revealed a consistent relationship. McNally et al [157] noted that all but 1 of 44 knees with normal tracking had a TTD of less than 20 mm and that patients with severe maltracking had a TTD of greater than 20 mm. The data from the current study confirms this relationship, with a clear threshold evident at 19 mm although the moderate subluxation group overlapped both the no/mild subluxation and severe subluxation groups. Alteration of the position of insertion of the patellar tendon during surgery effectively reduces the TTD.

It is clear that the cause of subluxation is multifactorial, although engagement was the most important factor from the current analysis. Closer analysis of the engagement data reveals insights into the key elements leading to subluxation. All patients with no or mild subluxation had $>30\%$ engagement and all but 5 (10%) with moderate or severe subluxation had $<30\%$ engagement. This would suggest that in patients with anterior knee pain resistant to conservative measures and moderate or severe subluxation, surgery directed at improving engagement should be considered. The five outliers of the moderate and severe subluxation groups with patellae above the threshold of 30% patella engagement all had dysplastic proximal trochleae (range of bony trochlear sulcus angles 158° to 205°) with only two having in addition, a TTD of >19 mm (20 mm - 21 mm). This suggests that altering the shape of the trochlea as well as improving engagement might be beneficial. However, this is probably unnecessary as there are many patients with dysplastic trochleae that do not sublux.

7.1.4 Conclusion

Anterior knee pain is a common and poorly understood problem. This study ranked the relative importance of bony, cartilaginous and soft tissue features, measurable from MRI, in causing patellofemoral subluxation. The degree of overlap of the patella in the trochlear groove was the most important factor and suggested that the overlap in extension should be $>30\%$ to avoid subluxation occurring. Other factors related to subluxation include the distance of the VMO muscle from the patella, tibial tubercle trochlear notch distance and the sulcus angle at the top of the trochlea. This study suggests that if patients, for whom conservative measures

prove ineffective, have symptoms related to subluxation, surgery should be focussed on improving patella engagement. A clinical study is required to test this hypothesis.

Chapter 8

Discussion and Clinical Implications

This chapter summarises the key findings of this thesis, highlighting the relevance and limitations of the work, proposes areas for further study and suggests clinical implications of the findings.

8.1 Phylogeny of the PFJ - An Evolutionary Perspective

The evolutionary development of the PFJ indicates that the adoption of an upright stance has led to dramatic changes in the PFJ, in particular the shape of the trochlear groove. With the advent of bipedal gait strategies (albeit part-time at first) an associated decrease in the sulcus angle of the trochlear groove (deepening of groove) was noted which coincided with the presence of a developing bony patella not noted in the lower order species [39]. The process of weight-bearing on two legs transformed the loading environment of the knee. During resting stance, the centre of gravity moved from a point anterior to the hindlegs during quadrupedal stance to a more posterior position, roughly through the centre of the knee in bipedal stance [203]. The conversion from four legs to two effectively doubled the weight being

passed through the knee joints, profoundly altering the loading pattern of the knee. It is possible that the rapid increase in magnitude of the forces acting through the PFJ accelerated the changes in the shape and orientation of the joint, illustrated by the sharp change in trends illustrated at the time of the African ground apes who were able to stand erect, around 13 Mya (Figure 3.7).

The overall shape of the knee was seen to change from a flat rectangular shape in the sprawling ancestors to a more rounded shape by the time of the bipedal and human groups. The reasons for this are not clear, however, the shape change coincides with a change in tibiofemoral kinematics. In the sprawling group, movements at the tibiofemoral articulation are uni-planar, the joint acting largely as a hinge allowing flexion and extension in the antero-posterior direction. Bipedal and human knee joint movements employ a modified hinge incorporating additional axial rotation. It is possible that the resultant rolling and sliding of the femur on the tibia have led to forces being distributed over a larger tibial area creating a larger plateau of bone distribution and a more rounded morphology.

Of particular note from this data is the alteration in the three dimensional position of the trochlea over time. Through the sprawling, climbing and erect phases of our evolution the trochlea was seen to move from a central position to a more medial location. Whilst the gait strategy of this era was primarily quadrupedal, the latter stages included the climbing and erect ancestors, who demonstrated a more medially located PFJ. Much of the extant African ape locomotor anatomy has been derived to support vertical climbing and above branch pronograde running and leaping [130]. To maintain balance during narrow-based quadrupedal, arboreal locomotion a combination of inversion of the foot with varus orientation of the knee, and a flexed

and externally rotated hip (Figure 8.1). It is likely that the medial position of the PFJ at this time reflects an adaptation to the medially placed axis of loading through the varus knee. Allied to this is the observation that climbing ancestors required loading during extreme hip flexion with the knee close to the chest. In this position any valgus alignment at the knee would cause the knee to abut against the chest. With the advent of upright bi-pedal gait, loading through the knee joint changed dramatically. The majority of bi-pedal gait is with the knee at near full extension (0-20° flexion), maintaining the centre of gravity over the midline. To accommodate this the knee has moved to a more valgus orientation, moving the biomechanical axis laterally, reflected in the lateral position of the PFJ.

These changes have highlighted anatomical features that are relevant to stability of the PFJ and normal kinematics.

Further Work

Of particular interest are the future trends in the 3D movement of the PFJ. Should the current trend continue, then the PFJ will move laterally and the trochlear groove will become progressively more shallow, further predisposing the human PFJ to lateral subluxation/dislocation. Further study is required to model the predicted 3D position and orientation of the trochlea should human evolution continue along the current profiles.



Figure 8.1: Showing the lower limb arrangement during vertical tree climbing; with foot inversion, varus knee position and external rotation of the hip.

Clinical Implications

This study demonstrates how the adoption of the upright, bipedal hominin stance has been accompanied by a progressive flattening of the trochlear groove and a lateralisation of the trochlear position relative to the tibiofemoral compartments. To accompany the range of flexion/extension and axial rotation associated with normal human gait, the patellofemoral joint has sacrificed stability for mobility. This helps explain the anatomical variability of the human trochlear groove (sulcus angles, shape, height) and why the patellofemoral joint is a source of pain.

8.2 Assessment of the normal shape and orientation of the PFJ in comparison with contemporary knee arthroplasty

This study has provided insights into the alignment of the trochlea in the normal bony and cartilaginous knee, allowing for comparison with current popular TKR and PFJR prostheses. Significant differences in the replaced knee were noted in both the trochlear groove orientation throughout the length of the trochlea and the magnitude of the displacements of the apex position when compared with normal subjects.

Comparing Form with Function

At the end of these studies it is now possible to speculate further on the relationship between Form and Function. Using the trajectories of the trochlear grooves in normal bone and cartilage, and also TKR and PFJR groups (Chapter 4), a direct comparison can be made with the coronal plane displacement profiles obtained using the MAUS technique on subjects (Chapter 6).

The comparison between the two sets of data is shown in Figure 8.2.

The qualitative movements of the patella are seen to closely mimic the underlying shape of the trochlea. The lateral movement of the patella is similar to the underlying bony and cartilaginous trochlear groove shape. Similarly, the more neutral and medial movements of the TKR and PFJR groups respectively, are reflected in the neutral and medially orientated trochlear grooves.

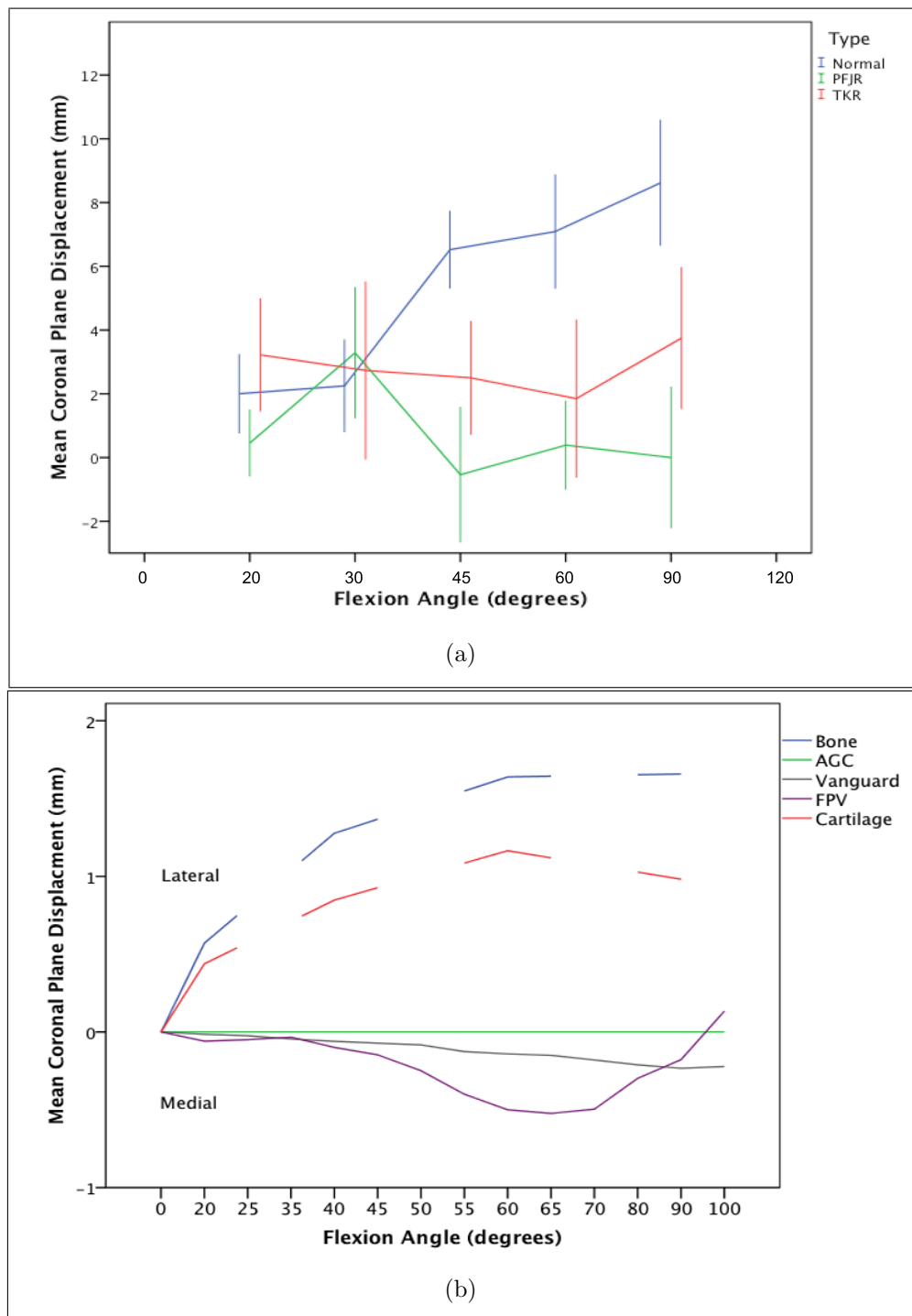


Figure 8.2: Showing (a) typical coronal plane patellofemoral displacements during flexion for normal, TKR and PFJR patients and (b) the trochlear groove position throughout the flexion arc.

Whilst the trajectory of the trochlear bone and cartilage is similar in shape to the recorded movements of the patella, the magnitude of the recorded movements is greater than five times than the excursion of the trochlear trajectory. This implies that the patellar apex is not guided directly over the apex of the trochlea during flexion of the normal knee. This may be in part be due to the subtle interaction between the bony and soft tissue constraints to movements of the patella and suggests that there might be a degree of *normal subluxation* associated with normal knee flexion.

Further Work

To further investigate this area a study is required which involves patients with planned knee arthroplasty. The study should compare the bony and cartilage trochlear groove morphology of patients pre-operatively, and using the MAUS technique, assess the pre-operative kinematics. This would allow direct comparison of the trochlea groove apex trajectory with the actual path of the patella. The process could then be repeated for the same cohort following knee arthroplasty, comparing the trajectory of the femoral component trochlear groove with the coronal profile of the patella during flexion. Regression analysis would establish whether statistically significant differences were present between the shape of the trochlear groove and the movements of the patella.

Chapter 4 investigated the shape of the trochlear groove in normal and replaced knees. Whilst this goes some way to explain the range of movements of the patella during flexion, it is likely that *patella* morphology is also important, and is not addressed in this thesis. Whilst patellofemoral articular contact patterns reveal that

the trochlea and patellar positions during flexion, excursions (subluxation) of the patellar apex from the trochlear apex will be dictated by the degree of entrapment of the patella in the groove. The morphological factors influencing entrapment in the trochlear groove include trochlear sulcus angle and patellar apex angle. Previous patella classification systems have relied upon medial and lateral facet size differences (Wiberg and Baumgartl systems) and ratios of length and depth (Greslamer system) as described in Section 2.3.3. A study is underway to develop a morphological patella classification system based on the articulating surface shape and its relationship to the underlying trochlear groove articulation.

Clinical Implications

Comparison of the 3D kinematics of normal and replaced knees has provided insights into how effective the replaced knee is at reproducing normal kinematics. There were significant differences in the patellar tracking profiles of both patient groups (PFJR and TKR) when compared with normal patellar movements. In particular, the coronal plane *patellofemoral* kinematics of TKR and PFJR were grossly abnormal. Whilst the normal coronal profiles (Figure 6.6) showed a predominantly lateral shift between engagement (20°) and 90° , the lateral shift of the replaced knees profiles was much less, and on occasion displayed a medial shift. This implies that the anatomy of the normal knee is not recreated following knee joint replacement surgery. This suggests that the PFJ shape and orientation in knee replacement should be re-considered.

8.3 The MAUS technique and measurement of coronal plane PFJ movements

As movements of the PFJ are inherently linked to the underlying shape, a technique able to measure PFJ kinematics in both normal and replaced knees that was both reliable and reproducible was required. Whilst previous studies have addressed the changes in sagittal plane kinematics in normal and replaced knees, to date, no authors have described a technique with accuracy of better than 2-3 mm. The validated MAUS technique (Chapter 5) was developed with the aim of assessing coronal plane movements of the patella in all knees. This has been shown in this thesis not only to be possible but also able to be sufficiently refined to be applied to the clinical population. Whilst the clinical study in Chapter 6 compared normal native knee joints with PFJR and TKR patients, it demonstrated the ability of the technique to be applied to a range of patients. It is likely that this technique will prove useful in the assessment of patients with suspected patellofemoral maltracking and those with poor results after knee replacements.

Further Work

Further work will include the use of the MAUS technique to investigate the range of outcome following TKR. It is possible that differences in coronal plane kinematics between patients may help explain the range of outcome experienced by patients. A formal study is planned comparing the coronal profiles of patients with and without post-operative anterior knee pain following TKR.

MAUS for measuring orientation of leg bones

Previous attempts (such as the Imperial technique [171]) have been made to describe the orientation of a leg using a minimal number of CT slices focussed around the hip, knee and ankle. From these few slices, the landmarks required to reconstruct femoral and tibial segments can be extracted [171]. The three dimensional images are then co-registered with post-operative CT data including the TKR. Measurements of the orientation of the prosthesis relative to the femur can then be made. It is entirely feasible that this process could be performed using the MAUS technique. Individual landmarks could be located on the femur and tibia, as shown in Chapter 5, which can be used to define the femoral and tibial segments. Similarly, it is possible that the outline of the femoral and tibial components could be defined using ultrasound. Further work should be directed at establishing the efficacy of ultrasound in describing the orientation of prosthetic components.

Six degrees-of-freedom studies

One of the primary drivers behind this thesis was to develop a method (MAUS technique) that could be applied to all knees to measure medio-lateral displacements of the patella relative to the leg. Because the MAUS technique provides the 3D position of the patella, femur and tibia, the same technique could be refined to allow measurements incorporating all 6 degrees of freedom.

Clinical Implications

The MAUS method provides a measurement tool of six degrees of freedom kinematics which can be applied to normal and replaced knees. With further refinement it is hoped that MAUS will be developed into a clinical assessment tool for investigating anterior knee pain. The method may also prove effective in the assessment of other joints including the shoulder, wrist and ankle.

8.4 Anatomical abnormalities causing PFJ subluxation

The hierarchy of relative importance of surgically correctable, anatomical variables around the PFJ and their influence on PFJ kinematics has been studied. It appears that the single most important factor is the degree of engagement of the patella in the trochlea in extension. This single factor that had the greatest influence on pathological subluxation.

Further Work

The degree of movement of the patella during engagement appears to be entirely dependent on the individual. Figure 6.4 illustrates the variation in shift of the patella during the early stage of flexion. To date, few studies have investigated the relevance and effects of this engagement phenomenon. Future studies should investigate the influence of engagement on normal gait. Between heel-strike and the ipsilateral stance phase, there is often a flexion movement of the knee observed prior to loading of that leg, the size of which is highly variable between individuals. This flexion movement may represent the flexion required to engage the patella in the trochlear

groove (providing stability) prior to weight-bearing on that leg. Such a study would investigate the flexion angle required to engage the patella in the trochlear groove (using MAUS) and compare it with the flexion angle demonstrated immediately prior to the stance phase during normal gait of that individual.

Clinical Implications

It is possible that surgical alteration of the degree of engagement of the patella in the trochlear groove may improve pathological subluxation and dislocation. A clinical study is required to assess the effect of altering patella engagement.

8.5 Conclusions

- The evolutionary study of the knee including specimens from the Devonian period (400 million years ago) to the human knee joint revealed progressive flattening of the trochlear groove and lateralisation of the patellofemoral joint relative to the tibiofemoral compartments. To accommodate the increased range of flexion, extension and axial rotation seen at the tibiofemoral joint in humans compared with earlier ancestors, the patellofemoral joint has sacrificed stability for mobility. This helps explain the great variability in modern patellofemoral shape and clinical subluxation.
- The study of shape of the patellofemoral joint has shown that the trajectory of the apex of the trochlear groove of the normal knee (bone and cartilage) is orientated in the opposite direction to that found in current TKR and PFJR designs. The trochlear groove of the patellofemoral joint in knee arthroplasty needs to be re-designed.
- An accurate method for assessing PFJ kinematics in normal and replaced knees has been developed combining Motion Analysis with UltraSound (MAUS). The technique has been validated and shown to be applicable for use in assessing native and replaced knees.

Further work is underway to develop the MAUS method into a powerful clinical investigative tool for assessment of the PFJ and potentially, all other joints.

- Kinematic analysis of the PFJ using the MAUS method demonstrated that the coronal plane pathway of the patella was opposite in normal and replaced knees. In the normal knee the patella tracks predominantly laterally whilst in the replaced knee it is medial.
- Assessment of bony and soft tissue contributions to abnormal PFJ kinematics revealed that the degree of engagement of the patella in the trochlea in extension has the greatest influence on subluxation. Clinical studies are required to assess the feasibility of surgical alteration of patella engagement.

Appendix A

Patient Forms

- Consent Form
- Invite Letter
- Participant Information Sheet
- Study Protocol
- Notice of Ethical Approval
- Nuffield Orthopaedic Centre Research and Development Application



**Oxford Orthopaedic Engineering Centre
Nuffield Department of Orthopaedic Surgery**

CONSENT FORM

Knee Motion Study

Researchers: Mr. Paul Monk Professor David Murray Dr. Richie Gill Dr. David Beard	Research Fellow Professor of Orthopaedic Surgery University Lecturer, University of Oxford University Lecturer, University of Oxford
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(Please initial box)

1. I confirm that I have read and understood the Participant Information sheet, dated 8/9/08, version 4 for the above study. I have had the time to consider it and the opportunity to ask questions about the project. ☐
2. I understand that my participation is voluntary and that I am free to withdraw at any time, without giving any reason, without my medical care or legal rights being affected. ☐
3. I have received satisfactory answers to all my questions? ☐
4. I understand that this study will not affect my clinical care in any way. ☐
5. I understand that relevant sections of my medical notes and data collected during the study, may be looked at by individuals from the Nuffield Department of Orthopaedic Surgery, from regulatory authorities or from the NHS Trust, where it is relevant to my taking part in this research. I give permission for these individuals to have access to my records. ☐
6. I agree to take part in the above study. ☐

<i>Name of Subject</i> <i>Please fill in box</i>
<i>Date</i>
<i>Signature</i>
<i>Name of Researcher</i>
<i>Date</i>
<i>Signature</i>

When completed, 1 for patient; 1 for researcher site file; 1 (original) to be kept in medical notes



Oxford Orthopaedic Engineering Centre
Nuffield Department of Orthopaedics, Rheumatology
and Musculoskeletal Sciences
University of Oxford
Botnar Research Centre
Nuffield Orthopaedic Centre
Headington
Oxford, OX3 7LD
Telephone: +44(0) 1865 227610 Fax: +44(0) 1865 227671
Oxford Research Committee B: 08/H9605/140

«Title» «first_name» «surname»
«Add1»
«Add2»
«Add3»
«Add4»
«Add5»
«Postcode»

2 October 2011

Dear «Title» «surname»,

Invitation to Participate in Research at the Nuffield Orthopaedic Centre

We would like to invite you to take part in a research study being carried out at the Nuffield Orthopaedic Centre, Oxford. The study is entitled: **Knee Motion Study - patellofemoral kinematics following knee joint replacement**. You have been asked to participate because you have had a knee replacement operation at the Nuffield Orthopaedic Centre, Oxford.

Our research aims to investigate how knee joints function after a knee joint replacement compared with normal knees. A small number of individuals experience pain and difficulties following knee replacement. It is possible that these problems originate from abnormal movements of the kneecap (patella). We aim to measure how the patella moves in normal knees, and compare this to how it moves in participants who have had knee replacement surgery.

We would be grateful if you could please take the time to read the enclosed Patient Information Sheet. This will explain what you need to know before deciding whether you wish to participate in the study. Could you then complete the Research Participant Response Sheet (indicating whether or not you wish to take part) and return it in the enclosed stamped addressed envelope.

If you would like further information or have any questions, you are welcome to get in touch with the Oxford Orthopaedic Engineering Centre using the contact details provided above.

Thank you for taking the time to consider participating in our study.

Yours sincerely,

Professor D W Murray MD, FRCS(Orth)
Consultant Orthopaedic Surgeon
Professor of Orthopaedic Surgery



Oxford Orthopaedic Engineering Centre
Nuffield Department of Orthopaedics, Rheumatology
and Musculoskeletal Sciences

PARTICIPANT INFORMATION SHEET

Knee Motion Study

You are invited to take part in a research study. Before you decide it is important for you to understand why the research is being done and what it involves. The following information outlines the study details. Please take time to discuss it with your GP friends and family. We would be happy to provide any further information you require.

Part 1 outlines the study and what happens if you take part.

Part 2 gives more detail about the conduct of the study.

Part 1 – Study Details

What is the purpose of the study?

Our research aims to investigate how knee joints function after a knee joint replacement compared with normal knees. A small number of individuals experience pain and difficulties following knee replacement. It is possible that these problems originate from abnormal movements of the kneecap (patella). We aim to measure how the patella moves in normal knees, and compare this to how it moves in participants who have had knee replacement surgery.

Why have I been chosen to take part?

You have been chosen to take part because our records show that you either have a completely normal knee or have had knee replacement surgery at the Nuffield Orthopaedic Centre.

Do I have to take part?

You do **not** have to take part in this investigation. If you decide to take part you are still free to withdraw at any time and without giving a reason. A decision to withdraw at any time, or a decision not to take part, will not affect the standard of care you receive; your treatment will proceed as normal.

What would happen to me if I take part?

If you decide to take part you will be given this information sheet to keep and be asked to sign a consent form. An appointment will be made for you to attend the Nuffield Orthopaedic Centre at your convenience. We will assess how your knee replacement is functioning by checking its movements in a gait laboratory (please refer to the section '**What tests would be performed?**' below). You will also be asked to fill in a knee questionnaire. The tests will be scheduled during a 1-2 hour visit. Once you have completed the tests, you do not need to return for any further visits.

Your lifestyle before and after the study tests will not be modified and no change in your medication, if you are taking any, will be necessary.

Will I have my travel expenses reimbursed?

We will reimburse your travel and parking expenses.

Part 2 – Detailed Information

If I take part, do I need to bring anything with me?

We suggest you bring comfortable shorts and a t-shirt with you. You will not require any footwear. There is a changing room and toilet that you may use. We have clean spare t-shirts and shorts that you can use, if you require.

What tests would be performed?

If you agree to take part in the study, we would ask you to undergo the following tests, which would be scheduled during a single 1-2 hour visit:

1. Ultrasound scan of the knee

Ultrasound scans of your knee will be performed by a healthcare professional. The equipment used is identical to that used when imaging foetuses in pregnant women. The ultrasound probe will be placed on the skin of your knee and we will take images of your kneecap.

2. Gait Laboratory

The gait laboratory uses motion capture analysis. You will have reflective marker balls attached to your legs which can be seen by cameras placed around the laboratory. We will ask you to sit on the edge of a chair or bed whilst we take two sets of images of your knee. For each set of images, the first image will be taken with your leg out straight. For the first set of images, the researcher supports your leg whilst images are taken in a series of different positions. For the second set of images, the process will be repeated with you moving the leg freely. In the unlikely event that movements of the leg are painful, the study will be stopped immediately.

Are there any side effects?

No. There is no radiation involved with either ultrasound scanning or motion capture analysis.

Will taking part change anything about my treatment?

No, your treatment will be exactly the same.

What are the risks of taking part?

There are no risks to you if you participate in the study.

What are the benefits of taking part?

We hope that this research will help increase our understanding of patients with knee pain and dysfunction in the future.

Will my taking part in this study be kept confidential?

All information collected about you during this study will be kept strictly confidential. Any information about you which leaves the hospital will have your name and address removed so that you cannot be recognised from it.

What will happen to the results of the research study?

The results of the study we hope would be published in scientific journals.

Who has reviewed the study?

It has been reviewed by experts in the field and has been approved by the Oxfordshire Clinical Research Ethics Committee (OXREC:)

Who is organising this study?

The study is being organised by the Oxford Orthopaedic Engineering Centre at the Nuffield Department of Orthopaedic Surgery, University of Oxford.

Where can I get more information?

If you have more questions about this study, please contact:

Professor David Murray

Botnar Research Centre, Nuffield Orthopaedic Centre, Headington, Oxford, OX3 7LD

Email: barbara.marks@ndos.ox.ac.uk or Telephone: 01865 227610

CERES (consumers for ethics in Research) produce a leaflet about participating in research, called '*Medical Research and You*' which you may find helpful. This is available from:

Postal address: CERES, PO Box 1365, London N16 0BW

e-mail: info@ceres.org.uk

web-site: www.ceres.org.uk

What do I do next?

- If you agree to take part in the study please complete and return the **Research Participant Response Sheet** in the enclosed stamped addressed envelope and we will contact you to arrange an appointment.
- If you do not wish to take part please tick the "I do not wish to participate" box and return the **Research Participant Response Sheet** in the enclosed stamped addressed envelope.

If you have doubts or queries, please do not hesitate to contact us on **01865 227610** or e-mail or write to:

Professor D W Murray
Consultant Orthopaedic Surgeon
Nuffield Orthopaedic Centre
Headington
OXFORD OX3 7LD

RESEARCH PARTICIPANT RESPONSE SHEET

Professor David Murray
Consultant Orthopaedic Surgeon
Botnar Research Centre
Nuffield Orthopaedic Centre
Oxford
OX3 7LD

Study Title: Knee Motion Study

Dear Professor Murray,

I have read your invitation letter and Patient Information Sheet.

I do not wish to participate in the above-mentioned study

☐

I would like to participate in the above-mentioned study

☐

My preferred contact method is: E-mail ☐, Telephone ☐ (morning/afternoon/evening), or Post ☐

I am available from _____ (day/month/year).

My contact details are provided below.

Please contact me to arrange an appointment to participate in the study.

NAME	
POSTAL ADDRESS	
E-MAIL	
PHONE (MOBILE)	
PHONE (WORK/HOME)	
SIGNATURE	
DATE	196

Study Protocol
**Patellofemoral Kinematics Following Knee
Replacement**

Investigators: P Monk
D W Murray
H S Gill
D J Beard

Ethics Reference: 08/H00606/114

Sponsor: Oxford Orthopaedic Engineering Centre

Funding: Nuffield Orthopaedic Centre General Charity

Background

Total knee replacement (TKR) has become the standard operative treatment for advanced joint diseases of the knee when conservative measures have failed. By 2030, the estimated number of total knee replacements being performed in the US alone is 3.48 million¹. Despite significant advances in the surgical technique and implant design, previous studies report anywhere between 10-50% of patients after knee replacement continue to have poor functional outcome or pain². The reason for this is not well defined, but is likely in part to be related to abnormal patello-femoral kinematics³⁻⁵. Although the replaced knee does not have normal kinematic function, it is important to understand how far it departs from normal⁶. Therefore, characterising the normal dynamic behaviour and the variance in patello-femoral kinematics in both normal and replaced knees is essential to facilitate further improvements in knee replacement design and surgical techniques⁷.

We and others have studied extensively tibio-femoral and patellofemoral kinematics in the sagittal plane, and have found marked differences between normal and replaced knees and also between different types of knee

replacement⁸⁻¹⁰. These differences, however, only partly explain the difference in outcomes between different knee replacements. It is therefore likely that coronal or transverse plane patello-femoral kinematics are also very important. There is very little information in the literature regarding transverse and coronal plane patellofemoral kinematics after knee replacement as this is difficult to image.

We have recently undertaken a study to establish the most suitable imaging modality to assess coronal plane patello-femoral kinematics after knee replacement. A cadaveric knee with a total knee replacement in situ was imaged using fluoroscopy in two planes, Computed Tomography (CT), Magnetic Resonance Imaging (MRI) and Ultrasound. Whilst an adequate view of the knee was seen from the lateral plane image with fluoroscopy, the antero-posterior view of the patella was obscured by the femoral component. A significant degree of metallic artefact was seen on MRI, and a high radiation dose and extensive software engineering for image processing required was required to provide adequate CT images. With ultrasound however, the position of the patella could be relatively easily defined.

In order to use ultrasound to assess kinematics we have combined it with our motion analysis system. We have attached retro-reflective markers to the ultra-sound probe and, following calibration, can determine the position of a bony land-mark in three dimensional (3D) space (fig 1). Using skin markers and ultrasound to determine the site of a series of land marks we are able to determine the position of the patella relative to femur and tibia. By repeating this at a series of flexion angles under different loads we will assess patello-femoral and patello-tibial kinematics in 3 dimensions.

Aims of Study

1. To develop a non-invasive system for measuring 'in vivo' patellofemoral kinematics (in both coronal and sagittal planes) using a combination of ultrasound and motion analysis.
2. To characterise patellofemoral kinematics in the normal knee and following knee joint arthroplasty.

Methodology**Part 1 – Refinement of Measurement Technique**

We initially will optimise our ultrasound measurement system for determining patella position and then will assess its accuracy.

A pilot study comprising 5 participants with normal knees will be performed. A Vicon Nexus Motion Capture System comprising 12 Nexus Cameras, an MA-300 EMG system and two AMTI force platforms will be used to provide 3D kinematic data. The data will be analysed using Bodybuilder (Vicon Motion Systems) software providing a coordinate system and trajectories for the tibia and femur. Knee ultrasound will be performed simultaneously to establish the position of the patella. The ultrasound probe will be prepared with retro-reflective markers so that it is visible to the Motion Capture system (Figure 1) so that the site of the patella can be determined in the motion capture coordinate system. To reduce variability in creating patellar coordinate systems, a theoretical square will be fitted to each patella touching the superior-inferior and medial-lateral edges of the patella. The centre of the box will be defined as the centre of the patella. Next, the kinematics of the patella will be measured relative to both the tibia and the femur. At each flexion angle the medial-lateral movement and tilt of the patella will be reported.

We will assess different ultrasound probes and determine the optimal number of probes. We will determine which landmarks can be adequately identified by skin markers and which need to be identified by ultrasound. We will assess whether the kinematics can be assessed dynamically or only statically. Repeatability and reliability assessments will be made.

Part 2. Study of patients with and without knee replacements

Early data suggests that greater than 5 mm medio-lateral subluxation of the patella might cause anterior knee pain and problems with patellofemoral instability¹¹. Based on this and information from our pilot studies we will undertake a power calculation to determine the number of patients in each study group and therefore the total number of groups that we can study.

We plan to study 5 different groups of subjects listed below.

The initial estimate is that we will require 20 participants in each group.

We will then address the following hypothesis:

Null Hypothesis: There is no difference in the medio-lateral movement of the patella during flexion and extension when comparing the following five groups;

Group 1: Normal knees with no history of patellofemoral instability or surgery

Group 2: Subjects with TKR with resurfaced patella

Group 3: Subjects with TKR without resurfaced patella

Group 4: Subjects with TKR and post-operative anterior knee pain

Group 5: Subjects with isolated patello-femoral replacement

Group 6: Subjects with isolated unicompartmental knee replacement

We do expect that there will be differences between these groups, and that a detailed study of these differences will give important insights into knee replacement design and function.



Figure 1. Showing reflective markers on both subject and ultrasound probe.

References

1. Kurtz S, Ong K, Lau E, Mowat F, Halpern M. Projections of primary and revision hip and knee arthroplasty in the United States from 2005 to 2030. *J Bone Joint Surg Am.* 2007 Apr;89(4):780-5.
2. Boyd AD Jr, Ewald FC, Thomas WH, Poss R, Sledge CB. Long-term complications after total knee arthroplasty with or without resurfacing of the patella. *J Bone Joint Surg Am.* 1993 May;75(5):674-81.
3. Eisenhuth SA, Saleh KJ, Cui Q, Clark CR, Brown TE. Patellofemoral instability after total knee arthroplasty. *Clin Orthop Relat Res.* May 2006;446:149-160.
4. Sharkey PF, Hozack WJ, Rothman RH, Shastri S, Jacoby SM. Insall Award paper. Why are total knee arthroplasties failing today? *Clin Orthop Relat Res.* Nov 2002(404):7-13.
5. Fehring TK, Odum S, Griffin WL, Mason JB, Nadaud M. Early failures in total knee arthroplasty. *Clin Orthop Relat Res.* Nov 2001(392):315-318.
6. Iwaki H, Pinskerova V, Freeman MA. Tibiofemoral movement 1: the shapes and relative movements of the femur and tibia in the unloaded cadaver knee. *J Bone Joint Surg Br.* Nov 2000;82(8):1189-1195.
7. Patel VV, Hall K, Ries M, et al. A three-dimensional MRI analysis of knee kinematics. *J Orthop Res.* Mar 2004;22(2):283-292.
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Oxfordshire REC B

2nd Floor, Astral House
Chaucer Business Park
Granville Way
Bicester
OX26 4JT

Telephone: 01869 604047
Facsimile: 01869 604055
Email: scsha.OxfordRECB@nhs.net

22 October 2008

Prof David Murray
Professor of Orthopaedic Surgery
Nuffield Orthopaedic Centre
Windmill Road
Oxford, OX3 7LD

Dear Prof Murray

Full title of study: Patella tracking before and after knee joint arthroplasty
REC reference number: 08/H0605/140

Thank you for your email of 21 October 2008, responding to the Committee's request for further information on the above research and submitting revised documentation.

The further information has been considered on behalf of the Committee by the Chair.

Confirmation of ethical opinion

On behalf of the Committee, I am pleased to confirm a favourable ethical opinion for the above research on the basis described in the application form, protocol and supporting documentation as revised, subject to the conditions specified below.

Ethical review of research sites

The Committee has designated this study as exempt from site-specific assessment (SSA). The favourable opinion for the study applies to all sites involved in the research. There is no requirement for other Local Research Ethics Committees to be informed or SSA to be carried out at each site.

Conditions of the favourable opinion

The favourable opinion is subject to the following conditions being met prior to the start of the study.

Management permission or approval must be obtained from each host organisation prior to the start of the study at the site concerned.

Management permission at NHS sites ("R&D approval") should be obtained from the relevant care organisation(s) in accordance with NHS research governance arrangements. Guidance on applying for NHS permission is available in the Integrated Research Application System or at <http://www.rdforum.nhs.uk>.

Approved documents

The final list of documents reviewed and approved by the Committee is as follows:

<i>Document</i>	<i>Version</i>	<i>Date</i>
Covering Letter		23 September 2008
Protocol	3	08 September 2008
Investigator CV	David Murray	
Application		11 September 2008
Questionnaire: Oxford Knee score		
Response to Request for Further Information		
Participant Consent Form	4	08 September 2008
Participant Information Sheet	4	08 September 2008

Statement of compliance

The Committee is constituted in accordance with the Governance Arrangements for Research Ethics Committees (July 2001) and complies fully with the Standard Operating Procedures for Research Ethics Committees in the UK.

After ethical review

Now that you have completed the application process please visit the National Research Ethics Website > After Review

You are invited to give your view of the service that you have received from the National Research Ethics Service and the application procedure. If you wish to make your views known please use the feedback form available on the website.

The attached document "After ethical review – guidance for researchers" gives detailed guidance on reporting requirements for studies with a favourable opinion, including:

- Notifying substantial amendments
- Progress and safety reports
- Notifying the end of the study

The NRES website also provides guidance on these topics, which is updated in the light of changes in reporting requirements or procedures.

We would also like to inform you that we consult regularly with stakeholders to improve our service. If you would like to join our Reference Group please email referencegroup@nres.npsa.nhs.uk.

08/H0605/140

Please quote this number on all correspondence

With the Committee's best wishes for the success of this project

Yours sincerely

Mr Robert King

Vice-Chair

Enclosures: "After ethical review – guidance for researchers"

Copy to: R&D Dept, Nuffield Orthopaedic Centre
Paul Monk, Research Fellow, Botnar Research Centre, NOC

R&D Management Approval Template			
Full title of project			
Patellofemoral Kinematics Following Knee Replacement			
Investigators: AP Monk, HS Gill, DJ Beard, DW Murray			
Ethics number:		Approved or submitted	Sponsor: <i>As agreed from the Ethics application</i>
08/H0605/140		Approved	Oxford Orthopaedic Engineering Centre
Funding Details: <i>Including grant number if appropriate.</i>	Nuffield Orthopaedic Centre General Charity (Hip and Knee Fund)		
Substantive/Honorary Contract with NOC. <i>Required if you have direct contact with NHS patients or staff. There may be a charge for non-NHS staff from HR dept</i>			
Yes, already in place-please specify type of contract	Yes. Research Fellow	Applied for-please give details	
GMC registration number:		Professional registration number:	
6079274			
<p>Background/Introduction:</p> <p>Total knee replacement (TKR) has become the standard operative treatment for advanced joint diseases of the knee when conservative measures have failed. By 2030, the estimated number of total knee replacements being performed in the US alone is 3.48 million¹. Despite significant advances in the surgical technique and implant design, previous studies report anywhere between 10-50% of patients after knee replacement continue to have poor functional outcome or pain². The reason for this is not well defined, but is likely in part to be related to abnormal patello-femoral kinematics³⁻⁵. Although the replaced knee does not have normal kinematic function, it is important to understand how far it departs from normal⁶. Therefore, characterising the normal dynamic behaviour and the variance in patello-femoral kinematics in both normal and replaced knees is essential to facilitate further improvements in knee replacement design and surgical techniques⁷.</p> <p>We and others have studied extensively tibio-femoral and patellofemoral kinematics in the sagittal plane, and have found marked differences between normal and replaced knees and also between different types of knee replacement⁸⁻¹⁰. These differences, however, only partly explain the difference in outcomes between different knee replacements. It is therefore likely that coronal or transverse plane patello-femoral kinematics are also very important. There is very little information in the literature regarding transverse and coronal plane patellofemoral kinematics after knee replacement as this is difficult to image.</p> <p>We have recently undertaken a study to establish the most suitable imaging modality to assess coronal plane patello-femoral kinematics after knee replacement. A cadaveric knee with a total knee replacement in situ was imaged using fluoroscopy in two planes, Computed Tomography (CT), Magnetic Resonance Imaging (MRI) and Ultrasound. Whilst an adequate view of the knee was seen from the lateral plane image with fluoroscopy, the antero-posterior view of the patella was obscured by the femoral component. A significant degree of metallic artefact was seen on MRI, and a high radiation dose and extensive software engineering for image processing required was required to provide adequate CT images. With ultrasound however, the position of the patella could be relatively easily defined.</p>			

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Consenting process: *To include details of who will consent and when or where the patient will be consented.*

Consent will be obtained from subjects at the Nuffield Orthopaedic Centre by Mr. A P Monk on arrival of the subject for the study

Recruitment details

Patients will be identified from operating lists and sent a letter from Professor Murray inviting them to join the study and a copy of the participant information sheet.

Method/Protocol:

Part 1 – Refinement of Measurement Technique

We initially will optimise our ultrasound measurement system for determining patella position and then will assess its accuracy.

A pilot study comprising 5 participants with normal knees will be performed.

A Vicon Nexus Motion Capture System comprising 12 Nexus Cameras, an MA-300 EMG system and two AMTI force platforms will be used to provide 3D kinematic data. The data will be analysed using Bodybuilder (Vicon Motion Systems) software providing a coordinate system and trajectories for the tibia and femur. Knee ultrasound will be performed simultaneously to establish the position of the patella. The ultrasound probe will be prepared with retro-reflective markers so that it is visible to the Motion Capture system (Figure 1) so that the site of the patella can be determined in the motion capture coordinate system. To reduce variability in creating patellar coordinate systems, a theoretical square will be fitted to each patella touching the superior-inferior and medial-lateral edges of the patella. The centre of the box will be defined as the centre of the patella. Next, the kinematics of the patella will be measured relative to both the tibia and the femur. At each flexion angle the medial-lateral movement and tilt of the patella will be reported.

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We do expect that there will be differences between these groups, and that a detailed study of these differences will give important insights into knee replacement design and function.

References used:

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5. Fehring TK, Odum S, Griffin WL, Mason JB, Nadaud M. Early failures in total knee arthroplasty. *Clin Orthop Relat Res.* Nov 2001(392):315-318.
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Number of patients recruited:
 Nil to date. 120 to be recruited

Services/Departments used at NOC
 Gait Laboratory

Purpose/Aims:

1. To develop a non-invasive system for measuring 'in vivo' patellofemoral kinematics (in both coronal and sagittal planes) using a combination of ultrasound and motion capture analysis in a gait laboratory.
2. To characterise patellofemoral kinematics in the normal knee and following knee joint arthroplasty.

Lay summary :

Following knee replacement an appreciable proportion of patients experience poor function or have pain, which may be related to how the kneecap (patella) moves. In the past it has been

Appendix B

Programmes

Listed here are the Matlab programmes used to process the MAUS data.

B.0.1 Load trial

```
function SWITCH SIGNAL, GAIT DATA = LoadMonkey(filename)
```

```
if exist(filename, 'file')
error('File "s" not found.', filename);
end
```

```
fid = fopen(filename, 'rt');
```

```
gait temp file = './gait temp';
```

```
switch temp file = './switch temp';
```



```

fid gait = fopen(gait temp file, 'wt');
fid switch = fopen(switch temp file, 'wt');


line read = fgetl(fid);


while line read ~= -1
fprintf(fid gait, 's - n', line read);
line read = fgetl(fid);
end


line read = fgetl(fid);
while line read ~= -1
fprintf(fid switch, 's-n', line read);
line read = fgetl(fid);
end


fclose(fid);
fclose(fid gait);
fclose(fid switch);


SWITCH SIGNAL = importdata(switch temp file);
GAIT DATA = importdata(gait temp file);


end

```

B.0.2 Registration

```
function [frame] = FitMonkey(p1, p2, p3, p4)
```

```
axis x = p2 - p1;
```

```
axis x = axis x/norm(axis x);
```

```
axis y = p4 - p3;
```

```
axis y = axis y/norm(axis y);
```

```
axis z = cross(axis x, axis y);
```

```
axis z = axis z/norm(axis z);
```

```
axis x = cross(axis y, axis z);
```

```
axis x = axis x/norm(axis x);
```

```
frame = [axis x; axis y; axis z];
```

```
end
```

Merge Data

```
function [POINTS GLOBAL] = MergeMonkey(landmark samples)

sample size = size(landmark samples,2);
data size = size(landmark samples(1).data, 1);
POINTS GLOBAL = [];

model sample = landmark samples(1).data(1:data size-1,:);

POINTS GLOBAL = [POINTS GLOBAL; landmark samples(1).data(:,)];

for i = 2:sample size
    transformation, err = AlignSamplesSVD Iter(model sample, landmark sam-
ples(i).data(1:data size-1,:), 1);
    uspt = landmark samples(i).data(data size,:) 1;
    uspt world = uspt*transformation';
    POINTS GLOBAL = [POINTS GLOBAL; uspt world(1:3)];
end
end
```

Reconstruction

```
function GAIT MARKERS, US POINT = ReconMonkey(gait sample, switch sample,
landmark labs)
```

```

threshold = 100;
switch sample size = size(switch sample,1);
scan number = 0;
scan sample count = 0;
labels size = size(landmark labs,1);
gait sample size = size(gait sample.data, 1);
gait label index = [];

mis freq = switch sample size/gait sample size;

if mis freq == 2
j = 1;
for i = 1:2:length(switch sample)
a = switch sample(i,:);
switch sample(j,:) = a;
j = j+1;
end
switch sample((gait sample size+1):switch sample size,:) = [];
end

threshold = 100;
switch sample size = size(switch sample,1);
scan number = 0;
scan sample count = 0;
labels size = size(landmark labs,1);

```

```

gait sample size = size(gait sample.data, 1);
gait label index = [];

for i=1:labels size
    idx = strmatch(landmark labs(i), gait sample.textdata);
    if isempty(idx)
        error lab = sprintf('Marker error(error lab);
    end
    gait label index = [gait label index idx'];
end

for i=3:gait sample size
    gait sample.data(i,gait label index(1:labels size*3-3)) = mean(gait sample.data(i-2:i-
    1,gait label index(1:labels size*3-3)));
end

for i=1:switch sample size-1

    if switch sample(i+1) - switch sample(i) > 30
        scan number = scan number + 1;
        scan sample count = 0;
    end

    if switch sample(i) > threshold

```

```

scan sample(scan number).data(scan sample count+1,:) = gait sample.data(i,gait
label index);
scan sample(scan number).index range(scan sample count+1,:) = i;
scan sample count = scan sample count + 1;
end
end

for i=1:6 scan sample(i).data = Mean NONZERO(scan sample(i).data);
scan sample(i).data = reshape(scan sample(i).data', 3, labels size)';

end

scan sample.data = scan sample tmp;

Merged Points = MergeMonkey(scan sample(1:6));

GAIT MARKERS = Merged Points(1:labels size-1, :);
US POINT = Merged Points(labels size:18, :);

end

```

Output

```

function [GLOBAL COORDS, FEM COORDS ANA, FEM COORDS MECH, TIB
COORDS, FLEX ANGLE , ANGLES TIB, ANGLES FEM] = UltraMonkey()

```

```
[filename, path] = uigetfile('*.txt;*.csv', 'Select a monkey');
```

```
fullpath = strcat(path, filename);
```

```
[filename, path] = uigetfile('*.txt;*.csv', 'Select a patella file');
```

```
fullpath pat = strcat(path, filename);
```

```
side = questdlg('Which leg did you scan?',...
```

```
'Select a monkey leg', 'LEFT', 'RIGHT', 'LEFT');
```

```
landmarklabels = 0;
```

```
switch side
```

```
case 'LEFT'
```

```
landmarklabels = importdata('landmark left.dat');
```

```
case 'RIGHT'
```

```
landmarklabels = importdata('landmark right.dat');
```

```
end
```

```
labels size = size(landmarklabels,1);
```

```
GLOBAL COORDS = zeros(4,4);
```

```
FEM COORDS ANA = zeros(4,4);
```

```

FEM COORDS MECH = zeros(4,4);
TIB COORDS = zeros(4,4);

switch data = 0;
gait data = 0;

switch data, gait data = LoadMonkey(fullpath);
switch data pat, gait data pat = LoadMonkey(fullpath pat);

switch data col = strmatch('SWITCH', switch data.textdata);

if switch data col
else
switch data col = strmatch('SWITCH', switch data pat.textdata);
end

switch data tmp = switch data.data(:,switch data col);

[gait markers, us pts] = ReconMonkey(gait data, switch data tmp, landmarklabels);

switch data col = strmatch('SWITCH', switch data pat.textdata);
switch data tmp = switch data pat.data(:,switch data col);
us pts pat = ReconPatCentroid(gait data pat, switch data tmp, 'USPT');

```



```

us pat sample size = size(us pts pat, 1);
us pat centroid = sum(us pts pat, 1)/us pat sample size;

if num us pts == 7
us point labels = ['TTUB';'LPAT';'SPAT';'MPAT';'IPAT';'LCOND';'GTROC';'PATCENTRE'];
else
us point labels = ['TTUB';'LPAT';'SPAT';'MPAT';'IPAT';'LCOND';'PATCENTRE
NEW';'PATCENTRE OLD'];
end point labels = [landmarklabels(1:labels size-1);us point labels];

ajc = 0.5*gait markers(6,:) + 0.5*gait markers(7,:);
kjc = 0.5*gait markers(4,:) + 0.5*us pts(6,:);

FLEX ANGLE = dot( (gait markers(5,:) - kjc)/norm(gait markers(5,:) - kjc), (ajc -
us pts(1,:))/norm(ajc - us pts(1,:)));
FLEX ANGLE = 180 - acos(FLEX ANGLE)*180.0/pi;

if num us pts == 7
fem frame ana = eye(3); FitMonkey(gait markers(2,:), gait markers(1,:), us pts(6,:),
us pts(7,:));
end
fem frame mech = FitMonkey(gait markers(2,:), gait markers(1,:), kjc, gait mark-
ers(5,:));

```

```
tib frame = FitMonkey(gait markers(2,:), gait markers(1,:), gait markers(12,:), us
pts(1,:));
```

```
us pts homo = ones(7,4);
us pts homo(:,1:3) = [us pts;us pat centroid]
```

```
FEM COORD FRAME = eye(4);
TIB COORD FRAME = eye(4);
```

```
FEM COORD FRAME(1:3,1:3) = fem frame mech;
FEM COORD FRAME(4, 1:3) = kjc;
FEM COORDS MECH = us pts homo*inv(FEM COORD FRAME);
FEM COORDS MECH = FEM COORDS MECH(:,1:3);
```

```
FEM COORD FRAME(1:3,1:3) = fem frame ana;
FEM COORD FRAME(4, 1:3) = kjc;
FEM COORDS ANA = us pts homo*inv(FEM COORD FRAME);
FEM COORDS ANA = FEM COORDS ANA(:,1:3);
```

```
TIB COORD FRAME(1:3,1:3) = tib frame;
TIB COORD FRAME(4, 1:3) = us pts(1,:);
TIB COORDS = us pts homo*inv(TIB COORD FRAME);
TIB COORDS = TIB COORDS(:,1:3);
```

GLOBAL COORDS = [gait markers; us pts];

LPAT = FEM COORDS MECH(2,1:3);

SPAT = FEM COORDS MECH(3,1:3);

MPAT = FEM COORDS MECH(4,1:3);

IPAT = FEM COORDS MECH(5,1:3);

PATCENTROIDFEMUR = (LPAT+SPAT+MPAT+IPAT)/4;

FEM COORDS MECH = [FEM COORDS MECH;PATCENTROIDFEMUR];

LPAT = FEM COORDS ANA(2,1:3);

SPAT = FEM COORDS ANA(3,1:3);

MPAT = FEM COORDS ANA(4,1:3);

IPAT = FEM COORDS ANA(5,1:3);

PATCENTROIDFEMUR = (LPAT+SPAT+MPAT+IPAT)/4;

FEM COORDS ANA = [FEM COORDS ANA;PATCENTROIDFEMUR];

MLPAT = MPAT-LPAT;

MLPAT =MLPAT/norm(MLPAT);

LPAT = TIB COORDS(2,1:3);

SPAT = TIB COORDS(3,1:3);

MPAT = TIB COORDS(4,1:3);

IPAT = TIB COORDS(5,1:3);

PATCENTROIDTIBIA = (LPAT+SPAT+MPAT+IPAT)/4;

TIB COORDS = [TIB COORDS;PATCENTROIDTIBIA];

```

pta pfa tilt spin = MonkeyAngles(TIB COORDS(:,1:3));
ANGLES TIB = tilt spin pta pfa;
pta pfa tilt spin = MonkeyAngles(FEM COORDS MECH(:,1:3));
ANGLES FEM = tilt spin; pta pfa;

figure;
hold on;
plot3(gait markers(:,1), gait markers(:,2), gait markers(:,3), 'r. '); text(gait mark-
ers(:,1), gait markers(:,2), gait markers(:,3), landmarklabels(1:labels size-1));
plot3(us pts(:,1), us pts(:,2), us pts(:,3), 'bx ');
text(us pts(:,1), us pts(:,2), us pts(:,3), us point labels(1:6));
title('Global Plot, red dots are gait markers, blue crosses are US reconstructed
markers');
xlabel('X(mm)');
ylabel('Y(mm)');
zlabel('Z(mm)');
axis equal;

```

Plot Femoral Coordinates

```

figure;
plot3(FEM COORDS ANA(:,1), FEM COORDS ANA(:,2), FEM COORDS
ANA(:,3), 'b. '); text(FEM COORDS ANA(:,1), FEM COORDS ANA(:,2), FEM
COORDS ANA(:,3), us point labels);
title('Femoral Frame ANA');

```

```

xlabel('Mediolateral');
ylabel('Proximodistal');
zlabel('Anteroposterior');
axis equal;

figure;
plot3(FEM_COORDS_MECH(:,1), FEM_COORDS_MECH(:,2), FEM_COORDS_MECH(:,3), 'b. '); text(FEM_COORDS_MECH(:,1), FEM_COORDS_MECH(:,2), FEM_COORDS_MECH(:,3), us point labels);
title('Femoral Frame MECH');
xlabel('Mediolateral');
ylabel('Proximodistal');
zlabel('Anteroposterior');
axis equal;

```

Plot tibial Coordinates

```

figure;
plot3(TIB_COORDS(:,1), TIB_COORDS(:,2), TIB_COORDS(:,3), 'g. ');
text(TIB_COORDS(:,1), TIB_COORDS(:,2), TIB_COORDS(:,3), us point labels);
title('Tibial Frame');
xlabel('Mediolateral');
ylabel('Proximodistal');
zlabel('Anteroposterior');
axis equal;
end

```

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