Rotational Laxity of the Knee following Reconstruction of the Anterior Cruciate Ligament using Single vs Double-Bundle Surgery

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Declaration

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I, ANDREA HEMMERICH, hereby declare that:

1. the above thesis is my own unaided work both in concept and execution, and that apart from the normal guidance from my supervisor, I have received no assistance;
2. neither the substance nor any part of the above thesis has been submitted in the past, or is being, or is to be submitted for a degree at the University of Cape Town or any other university.

The thesis has been presented by me for examination for the degree of Doctor of Philosophy in Biomedical Engineering.

Signature

Date

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Abstract

Knee laxity following anterior cruciate ligament (ACL) injury may lead to long-term joint degeneration causing osteoarthritis. Although traditional surgical techniques are sufficient in providing anterior-posterior knee strength, laxity in remaining degrees of freedom persists. The double-bundle surgical technique, reconstructing both anteromedial and posterolateral bundles of the ACL, is maintained to provide superior rotational restraint; however, transverse plane kinematics have not been accurately assessed and clinical evidence is generally restricted to subjective qualitative measurements of laxity under passive loading conditions.

A magnetic resonance imaging (MRI) compatible device and three-dimensional image processing technique was therefore developed to assess passive knee laxity under known torsional loading *in vivo*, with repeatability results demonstrating a standard error of measurement of less than 0.75° in transverse plane rotational measures. A randomised control trial was conducted with 32 patients exhibiting isolated ACL rupture; subjects were allocated either a single or double-bundle reconstruction and tested prior to and approximately five months following surgery. Passive rotational laxity was quantified using the verified testing apparatus and dynamic kinematics of 22 of those subjects were measured using established gait analysis methods. Three-dimensional kinematics were concurrently assessed in left and right knees of a group of healthy control subjects under both passive and dynamic testing conditions to establish baseline data sets. Linear mixed model statistical analyses enabled a comparison of results across surgical

groups pre- and post-operatively, as well as with the contralateral uninjured knee and healthy control groups.

Passive laxity assessment of the 15 Control subjects demonstrated asymmetry in left and right rotational kinematics when evaluating internal and external rotation independently, thereby suggesting that control comparisons in pathological assessment should not be confined to contralateral knee data. A 2.4° increase in internal rotational laxity observed in the ACL-deficient relative to the normal knee in extension was restored by both single and double-bundle reconstructions. A significant interaction between single and double-bundle surgical techniques pre- to post-operatively was demonstrated when assessing internal rotation at 30° of knee flexion. With single-bundle knee rotation closer to that of the uninjured group mean, the decreased degree of internal rotation observed in the double-bundle knees indicated a propensity to overconstrain motion following reconstruction of isolated ACL tears.

While no difference in overall range of rotation was found under physiological loading conditions, a significant surgery by test-time interaction of the midpoint of the range of movement was observed during the high-demand activities. A greater external rotational shift in the single-bundle group following reconstruction suggested a muscle co-contraction stabilization strategy associated with ineffective internal rotational torque due to hamstrings tendon donor-site morbidity. The kinematics of the double-bundle patient group were closer to those of the control group, suggesting improved joint restraint.

The findings from passive laxity testing indicated that the contribution of the intact and reconstructed ACL to joint restraint is limited under isolated torsional loading. Divergent outcome following single and double-bundle surgical techniques under dynamic loading conditions suggests, however, that ACL deficiency significantly affects the functional capacity of those structures that are primarily responsible for rotational constraint of the knee. While the double-bundle reconstruction provides physiologic control resulting in knee kinematics closer to normal than the single-bundle surgery, it is proposed that the improved stability is due to loading restraint in another degree-of-freedom, rather than specifically axial rotation. Furthermore, it should be cautioned that this restraint may be a consequence of excessive graft tensioning, which could simultaneously account for the outcome of the passive rotational laxity study.

Further improvements to ACL surgical techniques are required to better reproduce passive and weight-bearing kinematics of the uninjured knee and to prevent long-term joint degeneration. While the double-bundle technique demonstrates superior constraint in dynamic loading situations, care must be taken by orthopaedic surgeons to avoid excessive graft tension and overconstraint of joint motion. Consideration should be given to treating the primary restraints of rotation and to high-quality procedures, rather than simply relying on the double-bundle reconstruction to provide sufficient joint stability.

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Nomenclature

2D	two-dimensional
3D	three-dimensional
ACL	anterior cruciate ligament
ACLD	anterior cruciate ligament deficient
AM	anteromedial
AP	anterior-posterior
BPTB	bone-patellar tendon-bone
CAS	computer assisted surgery
CT	computed tomography
DB	double-bundle
DOF	degree-of-freedom
DP	distal-proximal
ICC	intraclass correlation coefficient
IE	internal-external
LCL	lateral collateral ligament
MCL	medial collateral ligament
ML	medial-lateral
MRI	magnetic resonance imaging
PCL	posterior cruciate ligament
PL	posterolateral
RCT	randomised control trial
RSA	roentgen stereogrammetric analysis
SB	single-bundle
SEM	standard error of measurement

Chapter 1

Introduction

1.1 Background

Treatment of the ruptured anterior cruciate ligament (ACL) has improved greatly over the last twenty years with emerging research in this field of orthopaedic medicine. This has resulted in increased success at restoring normal function to the knee joint and allowing patients to return to their activities of daily living in the months and years following injury.

However, long-term complications subsequent to surgical reconstruction have clinicians questioning in greater depth the role of the ACL and, accordingly, how to improve methods of treatment. The function of the ACL in stabilizing the joint in the sagittal plane has been understood for some time since the most predominant direction of tibial laxity following an isolated ACL injury is anteriorly. Due to the emphasis placed on restoring anterior-posterior (AP) translational constraint, the potential supplementary capacity of this ligament was often overlooked; its contribution to rotational constraint is, therefore, still uncertain. Not until patients with seemingly stable knees developed additional knee deficiencies or re-injured themselves, did clinicians begin associating the injured ACL with rotational laxity.

With the aspiration to improve long-term knee biomechanics following ACL reconstruction, surgeons have increasingly endeavored to recreate the native ligament properties. One aspect that has been closely examined in recent years is the significance of the two bundles – anteromedial and posterolateral – of the ACL, with scientific studies supporting the theory that the two bundles function discretely to help constrain the loads experienced at the joint. Research has shown that ligament bundle position, orientation, and tension are not only different from one another, but also vary with knee flexion angle and specific loads applied to the joint. One attempt by which to improve the outcome of ACL surgery, has therefore been to reconstruct both bundles (double-bundle technique), rather than just one bundle (single-bundle technique) of the ligament.

Investigations comparing these two surgical techniques have primarily been conducted on cadaver knees. To eliminate the effects of transformed joint tissue properties on biomechanical outcome, more *in vivo* research is required; however, limitations with existing, non-invasive laxity measurement tools have hindered this field of study.

In order to determine the function of the ACL in the transverse plane, a reliable method of measuring the kinematics is required. Devices used by clinicians to diagnose ACL injury are currently limited to measurement of static AP translation since this is the direction in which the most severe laxity is observed. Measurement of rotational laxity of the joint is therefore simply based on the subjective assessment of the surgeon. Advanced methods of laxity measurement in all three planes of knee motion are required to improve diagnosis and assess treatment.

Advancement in medical imaging has made it a more accessible resource for use in the diagnosis of musculoskeletal injury. Imaging techniques, such as magnetic resonance imaging (MRI), have similarly been applied by biomechanists to develop more accurate methods of measuring joint kinematics *in vivo*. The advantage of MRI is that it is non-invasive while having the ability to determine the precise three-dimensional position and orientation of the underlying bone, thereby avoiding soft tissue artefact associated with skin-based measurements.

While static passive laxity measures are the simplest clinical method of determining knee pathology, gait analysis has been beneficial in demonstrating subtle distinctions between the healthy and compromised limb under dynamic, physiological loading conditions. With improved technology enabling more accurate three-dimensional measurement of tibiofemoral kinematics, this tool is becoming an indispensible means by which to evaluate *in vivo* knee laxity in patient groups.

Both medical imaging and gait analysis provide accurate and objective methods by which to determine changes in knee joint laxity resulting from treatment such as single or double-bundle surgical reconstruction of a ruptured ACL. Scientific research generated from the implementation of these devices promises to be a valuable contribution to the pursuit of improved methods by which to treat ACL deficiency and avoid long-term joint degeneration.

1.2 Objectives

The objectives of this thesis were to determine the role of the native anterior cruciate ligament (ACL) and the effects of ACL reconstruction on *in vivo* rotational laxity of the knee joint. In particular, the outcome of single and double-bundle reconstruction in constraining rotational laxity was of interest.

The first specific goal was thus to design a device to accurately apply a known torsional load to the knee while being scanned using MRI. A procedure by which the MR images could be analysed to determine the precise 3D position and orientation of the tibia with respect to the femur was furthermore required to accurately describe the joint laxity under the specific loading conditions.

The next objective was to use the torsional laxity apparatus to determine the rotational knee laxity of healthy individuals, patients with isolated ACL-rupture, and patients with single-bundle and double-bundle reconstructions. We furthermore wished to compare the laxity of patients' contralateral knees with that of the healthy, uninjured population in order to examine the possibility of inherent knee laxity in people with ACL injury and to assess whether the contralateral limb may be used as a control when testing for rotational laxity. Effectively, our intention was to gain a better understanding of the behaviour of the knee under internal versus external torsional loads at full extension and 30° of flexion in the specified functional status groups.

The purpose of the final study was to evaluate the functional laxity outcome of the single and double-bundle surgical techniques under physiological loading conditions, focussing on the transverse plane rotational knee kinematics. Using established gait analysis methods, the aim was to determine whether the double-bundle procedure demonstrated superior rotational constraint to the single-bundle reconstruction with the knee experiencing realistic forces from everyday activities.

The knowledge gained from this thesis is intended to extend our comprehension of the effects of single and double-bundle surgical reconstruction techniques on *in vivo* knee biomechanics and to improve methods of treatment of ACL injury.

1.3 Document overview

Chapter 2 reviews the relevant literature in the field. The function of the anterior cruciate ligament in transverse plane rotational constraint is addressed, focussing on methods of assessment, in addition to passive and dynamic joint laxity in the ACL-deficient and reconstructed knee. In particular, the investigations comparing single and double-bundle ACL-reconstruction are critiqued.

Chapter 3 describes the MRI-compatible device, data collection and image processing methodology developed to measure three-dimensional kinematics of the knee under torsional loading. Results from feasibility and *in vivo* repeatability studies are presented.

Chapter 4 investigates the three-dimensional knee kinematics of a group of 15 healthy subjects under passive torsional loading using the methodology presented in Chapter 3. Left-right symmetry is analysed and the significance of coupled motion is discussed.

Chapter 5 presents the findings of a randomised control trial in which 32 patients with isolated ACL rupture were allocated either a single or double-bundle surgical reconstruction. Patients were tested pre- and post-operatively under the same passive torsional loading conditions as the healthy subjects in Chapter 4. The interaction of surgical technique by test time (preceeding and following surgery) is investigated.

Chapter 6 presents the outcome of dynamic joint laxity testing in 22 patients allocated either single or double-bundle surgical procedures to reconstruct the injured ACL. Three-dimensional knee kinematic data were collected prior to and following surgery during low and high-demand gait activities. The discussion emphasizes the significance of the contribution of muscles as dynamic stabilisers on rotational kinematics of the knee.

Chapter 7 qualitatively compares the conclusions of the passive and dynamic rotational laxity studies. Final conclusions and recommendations for future work are discussed.

1.4 Publications originating from this PhD thesis research

An expanded edition of the following journal publication is presented as Chapter 3:

- Hemmerich A, van der Merwe W, Vaughan CL. (2009). Measuring three-dimensional knee kinematics under torsional loading. *Journal of Biomechanics* **42**, 183-186.

The following peer-reviewed abstracts are also direct outcomes of this thesis and have been presented as seminars or poster exhibits at international conferences:

- Hemmerich A, van der Merwe W, Batterham M, Vaughan CL. (2009). Rotational laxity in anterior cruciate deficient and reconstructed knees: A prospective randomised control trial comparing single and double-bundle surgical techniques. *22nd Congress for the International Society of Biomechanics.* Cape Town, South Africa
- Hemmerich A, Vaughan CL, van der Merwe W. (2009). Prospective randomised study to compare single-bundle versus double-bundle ACL reconstruction in restoring rotational 3D kinematics of the knee. *7th Biennial Congress of the International Society of Arthoscopy, Knee Surgery & Orthopaedic Sports Medicine.* Osaka, Japan

- Hemmerich A, van der Merwe W, Vaughan CL. (2007). Repeatability of 3D knee joint kinematic measurements in vivo under torsional load. *21st Congress for the International Society of Biomechanics*. Taipei, Taiwan

- Hemmerich A, van der Merwe W, Vaughan CL. (2006). Three-dimensional in vivo knee joint laxity under torsional loading. *5th World Congress of Biomechanics*. Munich, Germany

- Hemmerich A, van der Merwe W, Vaughan CL. (2006). Three-dimensional in vivo motion analysis of knee joint laxity under torsional loading. *9th Symposium on 3D Analysis of Human Movement*. Valenciennes, France.

Chapter 2

Literature review

2.1 Rotational laxity of the knee joint

2.1.1 Knee anatomy and rotational restraint

Several structures have been shown to limit rotational laxity of the knee joint including the joint capsule, the collateral and the cruciate ligaments (Fuss, 1991; Markolf *et al.*, 1976; Wang & Walker, 1974); however conflicting reports as to the degree to which each of these structures, in particular the anterior cruciate ligament (ACL), contribute to knee rotational restraint have been found in the literature. The primary role of the ACL is to restrain anterior displacement of the tibia relative to the femur (O'Connor & Zavatsky, 1993); with research focussing on the ACL and anterior-posterior (AP) translation of the joint, its role in preventing rotational laxity has been largely overlooked (Zaffagnini *et al.*, 2000).

Although some researchers have concluded that the ACL does not play a major role in rotational constraint (Lane *et al.*, 1994), more recent studies have demonstrated a significant increase in rotational laxity following ACL injury (Georgoulis *et al.*, 2003; Tashman *et al.*, 2004; Yagi *et al.*, 2002; Zaffagnini *et al.*, 2000). The capacity of the ACL to restrain rotation in the transverse plane has been attributed to the location of its tibial and femoral insertion sites and its orientation within the joint.

The ACL extends from the anterior part of the tibial plateau to the medial side of the lateral femoral condyle. It is comprised of a multitude of fibers that are commonly considered to be divided into two bundles – anteromedial (AM) and posterolateral (PL) – whose nomenclature is based on their tibial insertions (Figure 2.1). Its oblique orientation causes ligament tensioning and resistance as the tibia pivots about the transverse plane axis of rotation, due to an increase in the distance between tibial and femoral insertion sites. With internal rotation, twisting of the ACL about the posterior cruciate ligament (PCL) precipitates further tensioning of the ligament (Blankevoort & Huiskes, 1996). It has been moreover suggested that the PL bundle has a greater mechanical advantage in restraining rotation with its femoral insertion further from the axis of rotation than that of the AM bundle (Yagi *et al.*, 2002). However, these suggestions are speculative; there is no objective data to support this.

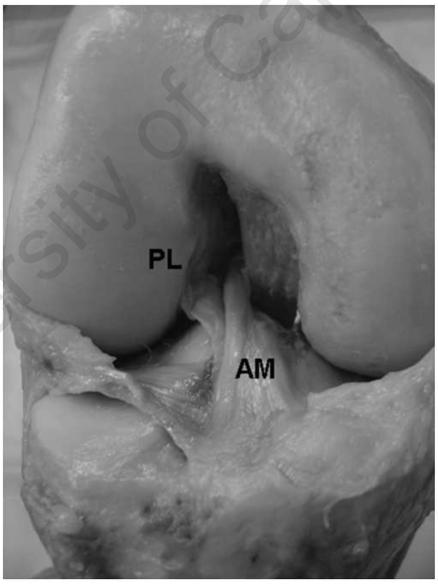

Figure 2.1: Cadaver knee showing AM and PL bundles (Petersen *et al.*, 2006).

Cadaveric and computational studies investigating the mechanics of the ACL bundles during passive flexion-extension have shown that the anteromedial bundle is taut in flexion while the posterolateral bundle is taut in extension, this is accredited to the position of the femoral insertions and resulting orientation of the individual bundles as illustrated in Figure 2.2 (Amis & Dawkins, 1991; Chhabra *et al.*, 2006; O'Connor & Zavatsky, 1993). Due to its relative enhanced tension at lower flexion angles, it has been found that the PL bundle is of greater importance in limiting joint laxity between 0° and 30° (Amis & Dawkins, 1991; Markolf *et al.*, 2009; Yagi *et al.*, 2002).

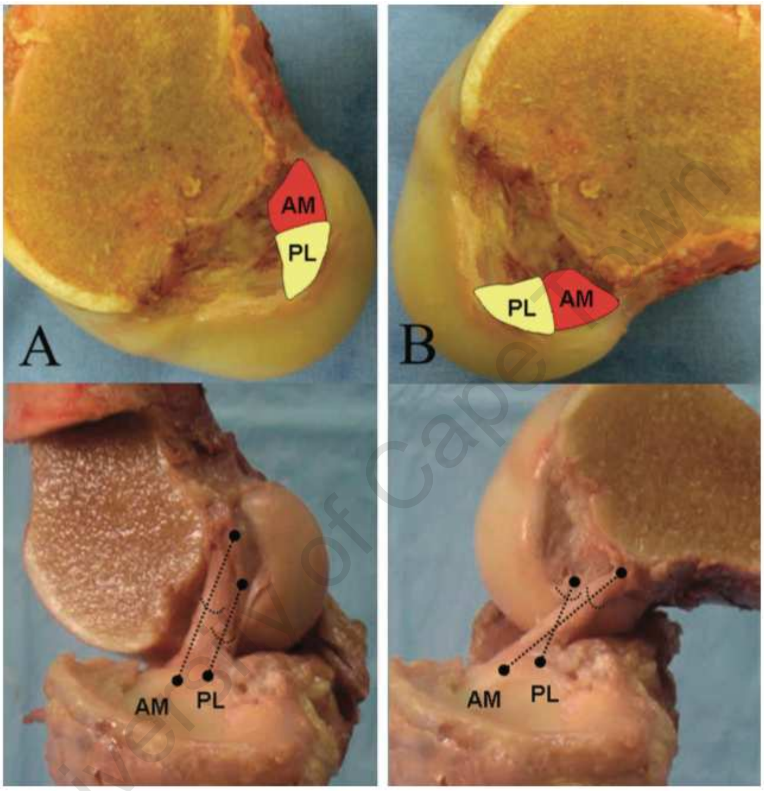

Figure 2.2: Sagittal sectioned view of tibiofemoral joint. A: Anteromedial and posterolateral bundles are parallel with knee in extension. B: Bundles are crossed at 90° of flexion and femoral insertion sites are now horizontal (Chhabra *et al.*, 2006).

The ACL is, nonetheless, only a secondary restraint to axial rotation. It has been extensively maintained that the medial collateral ligament (MCL) is the primary restraint to external rotation (Csintalan *et al.*, 2006; Harfe *et al.*, 1998; Meyer & Haut, 2008; Nordt *et al.*, 1999); however, the lateral collateral ligament (LCL) and posterolateral structures have also been shown to contribute to external rotation restraint (Blankevoort *et al.*, 1991; Kaneda *et al.*, 1997; Markolf *et al.*, 1976). Active joint rotational stability is provided by the hamstrings and iliotibial band, which externally rotate the tibia due to greater influence of the biceps femoris than the medial hamstrings (Kwak *et al.*, 2000). Passive constraint of internal rotation is not only provided by the collateral ligaments, but also by the tibiofemoral contact surfaces (Blankevoort & Huiskes, 1996; Wang & Walker, 1974). The greater posterior slope of the medial tibial surface provides an appreciable contribution to internal rotation restraint at 90° of flexion (Blankevoort & Huiskes, 1996). The menisci are also considered to be secondary joint stabilisers, with the medial maintaining greater stiffness than the lateral meniscus (Masouros *et al.*, 2008; Wang & Walker, 1974).

2.1.2 Importance of maintaining normal knee kinematics

Although it is often possible for a patient with a deficient ligament to perform the majority of activities that he or she conducted on a daily basis prior to injury, treatment to restore normal knee kinematics is important for various reasons. Restoration of joint stability and the elimination of 'giving way' symptoms can allow patients to return to higher intensity level activities within a year of surgery in most cases.

Long-term laxity associated with ACL deficiency is less well understood. Pathological knee kinematics result in changes in positions of tibiofemoral contact points and, consequently, altered stress distributions in the articular cartilage and greater loads on the surrounding joint structures (Chaudhari *et al.*, 2008; Li *et al.*, 2006; Stergiou *et al.*, 2007). Meniscal tears, damage to cartilage, as well as excessive ligament loading may be consequences of chronic ACL deficiency; resulting degeneration of these joint structures may lead to osteoarthritis (Chaudhari *et al.*, 2008; Li *et al.*, 2006; Masouros *et al.*, 2008; Sharma *et al.*, 1999; Shefelbine *et al.*, 2006; Stergiou *et al.*, 2007).

2.2 The ACL-deficient knee: Methods of treatment

An ACL injury may result in a partial or complete rupture of the ligament with possible damage to surrounding structures such as collateral ligaments and the meniscus. Treatment should be based on the extent of the injury in addition to the level of activity to which the patient intends to return and his or her individual characteristics such as age and medical condition (Miller, 2004).

2.2.1 Nonoperative

Nonoperative treatment is usually reserved for those patients who are satisfied to limit their activity level following injury (often older people); it is seldom recommended for those who wish to return to competitive sport, especially activities involving pivoting. Nonoperative treatment is typically limited to a rehabilitation program involving exercise to strengthen muscles. In order to prevent repetitive impact loading or situations that may cause further injury, patients are taught to modify the ways in which activities are conducted. Knee braces can also provide additional stability for the joint (Larson, 1993).

2.2.2 Surgical reconstruction

Surgical reconstruction of the torn ACL was first successfully achieved in the late 19th century (Colombet *et al.*, 1999). The procedure is now common practice with an estimate of over 100,000 reconstructions performed every year (Lewis *et al.*, 2008). The remnants of the native ACL are excised and a replacement graft is extended from the tibial to the femoral insertion site of the native ACL. The graft is fixed to the bone through tunnels extending from the anterior side of the tibia through the tibial plateau and from the medial side of the lateral femoral condyle passing proximally towards the lateral side of the femur.

Fixation devices include the interference screw, endobutton, and staple; choice of hardware is partially dependant on the type of graft used. The most common types of autografts used for ACL reconstruction include the bone-patellar tendon-bone (BPTB) and the four-strand semitendinosus/gracilis tendon autografts (Fu *et al.*, 2000).

The BPTB graft, which harvests the middle third (approximately 10mm in width and 100 mm in length) of the patellar tendon including sections of bone from the insertion sites at the patella and tibia, has been a preferred technique because the bone interference fit within the tibial and femoral tunnels permits bone to bone healing and consequently more rapid recovery. Its high stiffness and ultimate tensile strength, furthermore, limit graft failure (Fu *et al.*, 2000).

The disadvantage with this procedure, however, is that the bone tunnel placement is dependant on the length of the graft: the tunnel must be located so that a sufficient tunnel length is available to adequately tension the graft (Jackson & Lemos, 1993). Furthermore, due to long-term morbidity in the donor knee, such as pain in the patellar region and flexion contracture, as well as decreased joint power, alternative surgical techniques are often preferred (Feller *et al.*, 2001; Kowalk *et al.*, 1997; Marcacci *et al.*, 2003).

Initially, ACL reconstruction was performed using a single-bundle (SB) technique. As the significance of the two separate bundles was not fully understood, a graft simulating only the AM bundle of the ACL was used, which restricted AP translation of the knee. Since then, ACL reconstruction has been improved with graft construction now accounting for both bundles of the native ligament. This double-bundle (DB) approach is another advantage of the hamstrings graft procedure, in which the four strands of the graft are typically composed of the semitendinosus and gracilis tendons with each folded back on itself. Whereas the single-bundle technique has only one fixation site at both the femur and tibia, the double-bundle technique customarily requires two tunnels in both bones as shown in Figure 2.3. Both the AM and PL bundles are reconstructed and fixed independently in the separate tunnels. Although the DB technique was shown to have better post-operative functional results in certain studies, the procedure is more complex, time-consuming, and expensive to perform than the SB technique (Brophy *et al.*, 2009). Additional complications concerning revision surgery due to a second tunnel in each bone have also been intimated (Harner & Poehling, 2004). (A detailed comparison of the functional outcome of the SB and DB techniques is given in section 2.4.1.)

Bone-patellar tendon-bone graft procedures have generally been associated with increased donor-site morbidity than hamstrings grafts (Fu *et al.*, 2000). Not only do patients more often complain of anterior knee pain following BPTB graft reconstruction, in particular during kneeling (Feller *et al.*, 2001), Kowalk *et al.* (1997) also demonstrated reduced flexion moment and power at the injured knee during stair ascent that had not been observed pre-operatively. Simultaneous

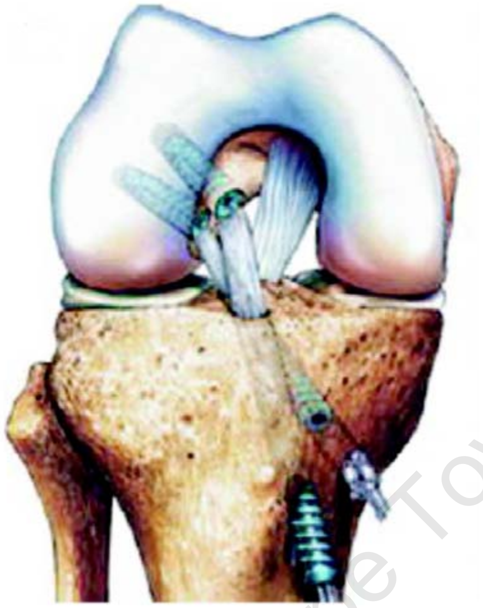

Figure 2.3: Schematic of double-bundle reconstruction (Järvelä, 2007).

increases in joint moment and power of the contralateral ankle suggested a compensation mechanism due to morbidity of the graft harvest site (Kowalk *et al.*, 1997).

Morbidity associated with hamstrings graft reconstruction includes weakening of knee flexion and internal rotation strength (Aune *et al.*, 2001; Viola *et al.*, 2000). Figure 2.4 illustrates the mechanical advantage of the semitendinosus and gracilis muscles in flexion and rotation due to the positions of their distal insertion on the medial aspect of the tibia. Functional deficit is generally minimal with only a 5% decrease in muscle strength reported three years post-operatively (Fu *et al.*, 2000).

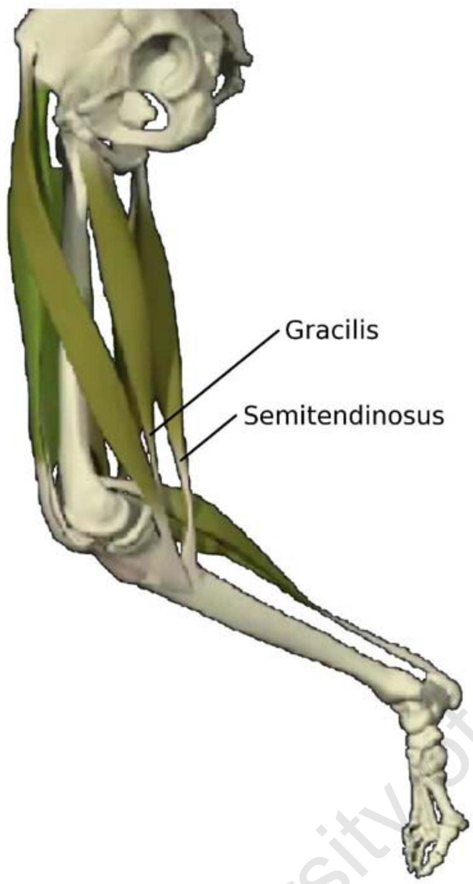

Figure 2.4: Semitendinosus and gracilis tendons shown with knee in flexion (adapted from Moore & Dalley (2005)).

2.3 Measuring joint laxity in the ACL-deficient knee

2.3.1 Concepts of joint motion

Descriptions of joint motion found in the literature are often ambiguous, making research outcomes unintelligible and comparisons between studies difficult. In order to standardize the conventions used in biomechanics research, the International Society of Biomechanics recommended the 'joint coordinate system' (Grood & Suntay, 1983) for the description of tibiofemoral kinematics (Wu & Cavanagh, 1995). This system presents the six degree-of-freedom (DOF) motion of the distal segment (e.g. the tibia) with respect to the proximal segment (e.g. the femur) as rotations about and translations along anatomical axes, making it easily understood by clinicians (Grood & Suntay, 1983). Segment axes are defined based on anatomical landmarks. Figure 2.5 illustrates the definition used by Hoshino *et al.* (2007) where the flexion-extension axis was embedded in the distal femur, internal-external (IE) rotation occurred about the longitudinal axis of the tibia, and ab-adduction took place about the floating axis which was defined by Grood & Suntay (1983) as perpendicular to the previous two axes.

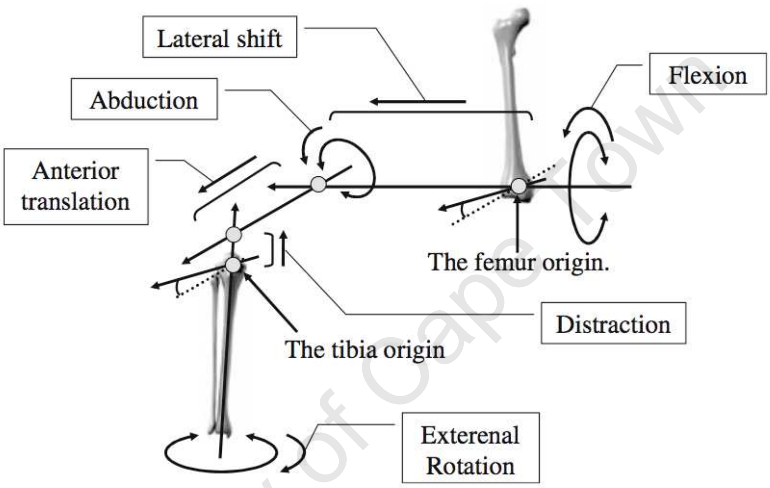

Figure 2.5: Grood & Suntay (1983) developed the joint coordinate system to describe the kinematics of the knee joint. This figure illustrates its application by Hoshino *et al.* (2007).

One limitation of the joint coordinate system convention is its susceptibility to kinematic crosstalk. This can occur when a defined axis of rotation is misaligned with the actual axis of rotation; the result is that rotation in one anatomical plane is misinterpreted for rotation in another (Piazza & Cavanagh, 2000). Crosstalk errors of up to 15° can easily transpire with proportional axis misalignment (Kadaba *et al.*, 1990; Piazza & Cavanagh, 2000).

Crosstalk and the alignment error that can occur from axes defined using anatomical landmarks have been motivation for some biomechanists to measure the 'helical axis' of rotation based on the three-dimensional (3D) tibiofemoral movement (Besier *et al.*, 2003; Dennis *et al.*, 2005; Mannel *et al.*, 2004; Marin *et al.*, 2003). Instead of defining three rotational and translational axes from which tibiofemoral motion is measured, the helical axis method describes segment motion as a rotation about and translation along a single axis which is defined from the tibiofemoral motion (Woltring, 1991).

The helical axis corresponds to the anatomical axis as long as rotation in a single anatomical plane occurs (e.g. flexion-extension); the difficulty arises with the physical interpretation of coupled motions (Woltring, 1991). An example was demonstrated by Dennis *et al.* (2005), who illustrated significant variation in position and orientation of the helical axis during a deep knee bend activity, thereby reflecting the complex motion of the knee. To describe this motion clinically in terms of the combined anterior-posterior translation and internal-external rotation, it was nonetheless necessary for the authors to use an anatomical tibial reference frame in which tibiofemoral contact points were described and subsequent clinical joint motion was measured (Dennis *et al.*, 2005).

Interpretation of anterior-posterior translation of the tibia with respect to the femur may seem simple; however, due to differences in measurement methods and/or segment coordinate systems, results from one study may not be comparable to those of another. Studies using cartesian coordinate systems, have reported AP translation measured along the tibial anterior axis (Robinson *et al.*, 2007; Yamaguchi *et al.*, 2009) as well as the floating axis of the joint coordinate system as illustrated in Figure 2.5 (Benoit *et al.*, 2007; Grood & Suntay, 1983; Hoshino *et al.*, 2007; Reed-Jones & Vallis, 2008; Woo *et al.*, 2006). Depending on the tibial IE rotation angle, this distinction could have significant effects on the translation measured.

The position of the origins of the respective segments can also affect measured translation (Roos *et al.*, 2006). In a study conducted by Beardsley *et al.* (2007), differences in AP translation were calculated using two different coordinate systems under anterior loading conditions. Considerable rotations (e.g. up to 31° in the sagittal plane) accompanied the translation in all three anatomical planes, which contributed to the discrepancies in translation measured between the two coordinate systems. A mean difference in AP translation of 4.2 mm was measured between coordinate systems and it was found that a 3° rotation in each anatomical plane would result in a 2 mm difference in AP translation; these values are routinely considered clinically significant (Beardsley *et al.*, 2007).

This section outlining the complexities involved with the description of joint motion was not intended as a complete review of the literature on this subject, but rather as a brief overview to enable a better understanding of the ensuing sections which critique scientific studies that have investigated joint laxity.

2.3.2 Instrumentation used to measure *in vivo* joint laxity

A clinician can perform numerous tests in order to assess the structural integrity of the knee ligaments. A positive outcome is commonly given by a qualitative description, rather than a quantitative value; for example, motion that varies from the contralateral knee or an audible clunk characteristic of subluxation of the tibial plateau on the femoral condyle (Magee, 1992). These tests examine AP, medial-lateral (ML), and rotational laxity and most commonly include the Lachman, the anterior drawer, and the pivot shift tests. Although these examinations, when performed correctly by an experienced clinician, can verify whether or not there is an ACL deficiency with high reliability, the extent of the injury can be difficult to determine (Magee, 1992).

Instrumented knee laxity measurement devices have been developed to quantify the extent of knee laxity. The most commonly used arthrometers are the KT-1000 and KT-2000 (MEDmetric Corp, San Diego, CA). Only passive AP laxity is tested by these devices; quantitative clinically accessible devices used to measure other degrees of joint laxity are not currently available. However, alternative methods have been developed by several researchers to measure joint motion in one or more planes of motion under various loading conditions (Koh *et al.*, 2005). These can, by and large, be divided into two categories: tracking marker devices and medical imaging techniques.

2.3.2.1 Tracking marker devices

Tracking marker measuring systems have traditionally been used in the gait analysis laboratory and may consist of optoelectric systems (Vaughan *et al.*, 1999) or electromagnetic tracking devices (Hemmerich *et al.*, 2006). In this environment, markers are assumed to be rigidly fixed to the segments of the joint in question. The position and orientation of the proximal and distal segments (e.g. femur and tibia) can then be determined under physiological loading conditions from the information provided by the markers.

Tracking devices have also been used to measure passive joint laxity, such as varus-valgus and IE rotation. Measurement accuracy was reported by Shultz *et al.* (2007) as generally within 2° under a 10 Nm varus-valgus load and 3° to 4° under a 5 Nm internal-external torsional load and by Tsai *et al.* (2008) as within 5° of rotation under 6 Nm of torque. In both of these studies, the error was attributed in part to skin motion artefact.

An even simpler method of tracking IE rotation using a protractor demonstrated limited accuracy; measurement errors determined by comparing the device results with those calculated using roentgen stereogrammetric analysis (RSA) showed a systematic overestimate of 100 % of the actual rotation angle (Almquist *et al.*, 2002). A further disadvantage of this system was the reliance on the precise alignment of the tibia with the external tracking device, i.e. the protractor.

To avoid soft tissue artefact, tracking markers are ideally fixed to the underlying bone. While ethically, this procedure is not authorized under most circumstances due to its invasive nature, several researchers have used these methods intraoperatively where computer-assisted surgical (CAS) instruments are already secured to the tibia and femur (Bull *et al.*, 2002; Ishibashi *et al.*, 2005; Martelli *et al.*, 2007; Robinson *et al.*, 2007). Although this technique has demonstrated greater precision of 3D joint measurements, the load was applied manually in most cases, relying on the investigator to accurately apply consistent quantities of force and/or torque to the joint.

2.3.2.2 Medical imaging techniques

The main advantage of medical imaging techniques used to measure knee joint laxity is the elimination of soft tissue artefact. The Telos stress device was developed to determine the position of the tibia with respect to the femur in the sagittal plane using X-ray under anterior-posterior loading (Schulz *et al.*, 2005). A similar method using X-ray was developed by Sawant *et al.* (2004) to measure valgus laxity associated with combined cruciate and MCL injury. As with arthrometers, motion outside of a single plane cannot be measured and acquiring the transverse plane X-ray images that would be required to measure IE rotation is impractical.

The capability of magnetic resonance imaging (MRI) and computed-tomography (CT) to generate several two-dimensional (2D) slices has enabled investigators to overcome this limitation. In some studies, the three-dimensional nature of these imaging techniques was not fully exploited; rotation was simply determined by tracking the anterior-posterior translation of both the medial and lateral condyles on the tibial plateau in parallel sagittal plane slices (Iwaki *et al.*, 2000; Logan *et al.*, 2004; Okazaki *et al.*, 2007). This technique, although less sophisticated than more recent developments in 3D motion tracking using CT and MRI, was nevertheless able to illustrate the coupled internal rotation with flexion of the knee joint known as screw-home motion that results from medial side sliding and lateral side rollback of the femoral condyles (Hill *et al.*, 2000; Iwaki *et al.*, 2000).

By reconstructing 3D bone segments from several slice medical images, complete 6 DOF motion can be traced. Li *et al.* (2004b) superimposed the segment models generated from 3D fluoroscopy onto 2D X-rays taken at different angles of knee flexion. By correctly orienting the models in the orthogonal planes, 3D position and orientation were determined to accuracies of 0.1 mm and 0.1 degrees (Li *et al.*, 2004b).

This methodology was modified by the same research group for weight-bearing kinematic measurements. Three-dimensional images of the femur and tibia were generated from MR images and two orthogonal images were captured during a lunge activity using the 3D fluoroscope (Li *et al.*, 2004b). Using this technique, this research group was able to identify not only anterior-posterior movement of the ACL-deficient (ACLD) tibia, but also a significant lateral shift of the femur on the tibial surface (DeFrate *et al.*, 2006; Li *et al.*, 2006).

The technique employed by Fellows *et al.* (2005b) was slightly different in that low resolution 3D MRI scans were shape-matched to high resolution images taken at a neutral position. The advantage of these techniques that superimpose low resolution (or 2D) images onto high resolution 3D models is that the low resolution scans can be acquired in a relatively short period of time while the instrumentation used to generate the models may require high scan times for adequate resolution or may limit the position at which scans can be taken (e.g. full knee extension). While the short scan time of the low resolution images permits imaging of the knee in positions that could potentially not be maintained by the subject for the duration of the longer scan time required for the high resolution 3D images, they are nonetheless not instantaneous. These techniques are, therefore, still considered to be 'quasi-static' (Li *et al.*, 2004b).

Roentgen stereogrammetric analysis (RSA), in which small tantalum markers are surgically implanted into the bones, is employed similarly to optoelectric systems, except it uses radiographic images to track the markers. Brandsson *et al.* (2002) demonstrated the value of this technique to track dynamic motion while patients ascended an 8 cm high platform. With this biplane radiographic imaging technique, kinematic data was collected at 2 to 4 exposures per second. RSA was also used by Tashman *et al.* (2004). With a much greater capture rate of 250 Hz, 6 DOF kinematic data could be collected during downhill treadmill running with an accuracy within 1° for IE tibial rotation (Tashman & Anderst, 2003).

2.4 Surgical techniques and rotational laxity outcome

2.4.1 Single versus double-bundle reconstruction

At the commencement of this PhD thesis in early 2005, no journal publications could be found directly comparing SB (one tibial and one femoral tunnel) and DB (two tibial and two femoral tunnels) surgical techniques in a controlled investigation. Joint laxity following SB and 'non-anatomic' DB (one tibial and two femoral tunnels) had been investigated (Adachi *et al.*, 2004; Yagi *et al.*, 2002); however, these studies did not adequately address the issue of rotational laxity as only anterior translation was measured under the specific loading conditions. Nevertheless, in the cadaveric study conducted by Yagi *et al.* (2002), the combined valgus and internal torsional loading conditions did indicate a difference in joint laxity between the two reconstructive techniques. No difference between SB and DB techniques was found with anterior loading by Adachi *et al.* (2004).

Most of the studies comparing SB and anatomic DB reconstruction listed in Table 2.1 suggested superior outcome using the DB technique; however, the specific evidence may not make this general conclusion as elementary as the earlier literature proclaimed. Only six studies actually measured transverse plane rotation (Ferretti *et al.*, 2008; Ishibashi *et al.*, 2005; Markolf *et al.*, 2008b, 2009; Seon *et al.*, 2009; Steckel *et al.*, 2007a), while the remaining studies acquired other measures of restraint, primarily anterior-posterior laxity. Of those that measured IE rotation, only one applied a quantified internal torque to the knee joint and this was in concurrence with a valgus torque to simulate the pivot shift (Markolf *et al.*, 2009). Those that administered isolated torsional loading typically used 'manual [maximum] force' (Ferretti *et al.*, 2008; Ishibashi *et al.*, 2005; Seon *et al.*, 2009).

The other studies in Table 2.1 that formed conclusions regarding rotational laxity following SB or DB reconstruction did so based on subjective measures of the pivot shift test (Asagumo *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Markolf *et al.*, 2008b; Muneta *et al.*, 2007; Siebold *et al.*, 2008; Streich *et al.*, 2008; Yagi *et al.*, 2007). The pivot shift was typically evaluated on a positive-negative or four-point grade indicating the examiner's estimate of the degree of instability. There was no quantitative measure of the kinematics (either tibiofemoral rotation or translation in any anatomical plane) and it has been shown that the clinical assessment varies between examiners (Bull & Amis, 1998). The only distinct association to rotational restraint is the fact that internal torque is one component of the applied load, but even that has not been quantitatively measured.

An attempt to quantify *in vivo* kinematics resulting from the pivot shift has, in fact, been made by several researchers (Bull *et al.*, 2002; Hoshino *et al.*, 2007; Kubo *et al.*, 2007; Lane *et al.*, 2008; Lopomo *et al.*, 2009; Robinson *et al.*, 2007; Yagi *et al.*, 2007). These studies have used electromagnetic or optoelectric systems to track tibiofemoral movement in 3D space. Although measures of tibial translation, rotation, velocity, and acceleration during pivot shift contribute to its objective evaluation and understanding of the 3D kinematic laxity, measuring the magnitude of the applied loads is not possible with these instruments.

Accordingly, it is possible to summarise the methodologies of those studies that have directly compared SB and DB surgical techniques (Table 2.1) into four categories: those that applied transverse plane IE torque and measured rotation in the same plane (Ferretti *et al.*, 2008; Ishibashi *et al.*, 2005; Markolf *et al.*, 2008b, 2009; Seon *et al.*, 2009), those that applied transverse plane torque and measured kinematics other than IE rotation (Asagumo *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Muneta *et al.*, 2007; Siebold *et al.*, 2008; Streich *et al.*, 2008; Yagi *et al.*, 2007), those that applied loads other than transverse plane torque but still measured transverse plane rotation (Steckel *et al.*, 2007a), and those that applied loads other than transverse plane torque and measured kinematics other than rotation (Seon *et al.*, 2007; Yasuda *et al.*, 2006). Of those that have applied transverse plane torsional loads, only four studies applied *isolated* torsional loads, and only one applied a *quantified* torsional load (Markolf *et al.*, 2009); that particular study was conducted on cadaveric specimens and was, therefore, unable to account for the healing process following reconstruction or differences in mechanical properties from living joint tissue. Still, the DB reconstruction is routinely cited as the technique that is able to control *rotational* laxity better than SB surgery following ACL injury.

In actuality, the lack of standardised loading criteria precipitates a more subjective evaluation of joint laxity based on the discretion of the investigator, makes it difficult to compare study outcomes, and may be the reason for some of the conflicting results in the literature; for example, while Ferretti *et al.* (2008) found no difference in either anterior translation or rotation between surgical techniques under maximal manual anterior force and IE torsion, respectively, Seon *et al.* (2009) demonstrated better DB laxity control in both directions and Ishibashi *et al.* (2005) found that only DB anterior laxity (not rotation) was significantly reduced when compared to the SB technique under similar loading conditions. Furthermore, some studies, such as those of Kondo *et al.* (2008); Muneta *et al.* (2007); Seon *et al.* (2009); Siebold *et al.* (2008); Steckel *et al.* (2007a), and Yasuda *et al.* (2006) were in agreement with Ishibashi *et al.* (2005) regarding anterior-posterior constraint under anterior loading conditions, whereas others found no significant difference in measured anterior-posterior translation between SB and DB techniques (Adachi *et al.*, 2004; Asagumo *et al.*, 2007; Ferretti *et al.*, 2008; Järvelä, 2007; Streich *et al.*, 2008; Yagi *et al.*, 2007).

Outcome of the clinical pivot shift test was only slightly less ambiguous with most studies in agreement that the DB reconstruction provided superior constraint (Järvelä, 2007; Kondo *et al.*, 2008; Muneta *et al.*, 2007; Siebold *et al.*, 2008; Yagi *et al.*, 2007), while only Asagumo *et al.* (2007) and Streich *et al.* (2008) showed no difference between surgical techniques. To complicate the matter further, both Markolf *et al.* (2008b) and Steckel *et al.* (2007a) showed that the DB reconstruction actually overcorrected joint laxity in rotation, while the SB technique produced results closest to normal.

Table 2.1: Studies comparing single-bundle and double-bundle surgical techniques under passive loading conditions.

Note: Only anatomic (two-tibial + two-femoral tunnel) double-bundle techniques were included; studies investigating non-anatomic (single-tunnel, double-bundle) techniques have, therefore, been excluded from this list. (* quantity of applied load was not indicated. RCT is abbreviation for randomised control trial.)

Author (Year)	**Loads Applied**	**Data Acquired**	**Study Design and Measurement Methods**	**Findings**
Ishibashi *et al.* (2005)	Manual force*: anterior, internal-external torque	Anterior-posterior translation, internal-external rotation	Intraoperative (32 patients); optoelectric bone markers	DB showed improved anterior-posterior constraint to SB; no difference in rotation.
Yasuda *et al.* (2006)	133 N anterior load	Side-to-side anterior laxity	Prospective study (72 patients); KT-2000	Anatomic DB produced better anterior laxity than SB.
Asagumo *et al.* (2007)	Manual force*: anterior drawer, Lachman, pivot shift	Side-to-side anterior and dynamic joint laxity	Retrospective study (123 patients); KT-1000	No differences between DB and SB outcomes.
Järvelä (2007)	134 N anterior force, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (65 patients); KT-1000	DB showed better pivot shift control than SB; no difference in anterior laxity.
Muneta *et al.* (2007)	Manual force*: anterior drawer, Lachman, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (68 patients); KT-1000	DB produced better anterior laxity and pivot shift results than SB.
Seon *et al.* (2007)	No load (passive flexion)	Anterior translation of medial and lateral tibiofemoral compartments	Retrospective study (20 patients); MR imaging	DB produced better lateral side anterior laxity than SB; no difference in medial side laxity.

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Table 2.1 – continued

Author (Year)	**Loads Applied**	**Data Acquired**	**Study Design and Measurement Methods**	**Findings**
Steckel *et al.* (2007a)	Manual force*: anterior drawer, Lachman	Anterior translation, internal-external rotation	Cadaver; optoelectric bone markers	DB produced better anterior laxity outcome than SB; DB overconstrained rotation.
Yagi *et al.* (2007)	Manual force*: Lachman, pivot shift	Side-to-side anterior and dynamic joint laxity, tibial acceleration	RCT (60 patients); KT-1000, electromagnetic skin sensors	DB showed better pivot shift control than SB; no difference in anterior laxity.
Ferretti *et al.* (2008)	Manual maximum force*: anterior, internal-external torque	Anterior-posterior translation, internal-external rotation	Intraoperative (20 patients); optoelectric bone markers	No differences between DB and SB in anterior laxity or rotation.
Kondo *et al.* (2008)	133 N anterior force, pivot shift	Side-to-side anterior and dynamic joint laxity	Prospective study (328 patients); KT-2000	DB showed better anterior laxity and pivot shift control than SB.
Markolf *et al.* (2008b)	Valgus torque*, pivot shift	Anterior-posterior translation, varus-valgus, and internal-external rotation; graft forces	Cadaver; robotic testing system	DB overcorrected, while SB adequately constrained pivot shift laxity.
Siebold *et al.* (2008)	Manual anterior force*, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (70 patients); KT-1000	DB showed better anterior and pivot shift control than SB.
Streich *et al.* (2008)	134 N anterior force, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (50 patients); KT-1000	No difference between DB and SB in any measured outcome.

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Table 2.1 – continued

Author (Year)	**Loads Applied**	**Data Acquired**	**Study Design and Measurement Methods**	**Findings**
Markolf *et al.* (2009)	100 N anterior-posterior force, 5 Nm internal, and 5 Nm valgus torque	Anterior-posterior translation, varus-valgus, and internal-external rotation; graft forces and length change	Cadaver; robotic testing system	SB produced graft forces and knee kinematics closest to normal; DB showed high forces in posterolateral bundle near full extension.
Seon *et al.* (2009)	Manual maximum force*: anterior, internal-external torque	Anterior-posterior translation, internal-external rotation	Intraoperative (40 patients); optoelectric bone markers	DB showed improved anterior-posterior and rotational constraint to SB.

2.4.2 The roles of the anteromedial and posterolateral bundles

Greater insight is gained by reviewing the studies that have addressed the functions of the different bundles of the ACL. It is widely accepted that the AM bundle contributes most to knee constraint at higher angles of flexion, while the PL bundle contributes significantly to joint restraint near full extension (Amis & Dawkins, 1991; Chhabra *et al.*, 2006; Fuss, 1989; Gabriel *et al.*, 2004; Jordan *et al.*, 2007; Mae *et al.*, 2006; Robinson *et al.*, 2007; Yasuda *et al.*, 2008; Zantop *et al.*, 2006). During passive flexion-extension both bundles lose tension from 0° to 30° of flexion; however, the decrease in PL bundle tension is more considerable than that of the AM bundle (Amis & Dawkins, 1991; O'Connor & Zavatsky, 1993; Yasuda *et al.*, 2008). While the PL bundle continues to shorten (i.e. slacken) beyond 30° of flexion, the AM bundle begins to tighten again. The AM bundle, is consequently allocated the greatest proportion of the load of the ACL at flexion angles greater than 30° (Amis & Dawkins, 1991).

With the introduction of an externally applied load, the PL bundle has demonstrated similar properties throughout the range of flexion as in the unloaded condition; the AM bundle, however, did not slacken over the first 30° of flexion when an additional load was applied (Gabriel *et al.*, 2004; Jordan *et al.*, 2007; Li *et al.*, 2004a; Mae *et al.*, 2006; Yagi *et al.*, 2002). Instead, the elongation and tension of the AM bundle remained relatively constant or increased between 0° and 60° of flexion and then decreased through 120° of flexion (Gabriel *et al.*, 2004; Jordan *et al.*, 2007; Vercillo *et al.*, 2007). Despite the drop in tension of the AM bundle at higher flexion in the loaded condition, its overall tension was still significantly greater than that of the PL bundle at respective flexion angles (Gabriel *et al.*, 2004).

Although it does not contribute significantly to anterior-posterior joint restraint at higher angles of flexion, the PL bundle is considered important in maintaining rotational constraint, primarily at lower flexion angles (Yasuda *et al.*, 2008). Robinson *et al.* (2007) found that the mean transverse plane rotation measured intraoperatively during the pivot shift was substantially less with an isolated PL bundle than with only the AM bundle. It has furthermore been shown that the graft forces experienced during a simulated pivot shift test are closer to those of the intact ACL when both AM and PL bundles are reconstructed separately as compared to a single-bundle reconstruction in which all four strands of the graft were fixed in a single femoral and single tibial tunnel (Yagi *et al.*, 2002). The ability of the PL bundle to control rotation has been attributed to its more horizontal orientation (Robinson *et al.*, 2007).

2.4.3 Alternative strategies for surgically constraining rotational laxity

Having considered the structures involved in rotational constraint of the knee joint, it would be naïve to surmise that the number of ACL bundles that are reconstructed would be the only surgical consideration to affect joint laxity. The debate as to which injured structures should be surgically mended has been long-established, with advantages including improved mechanical properties of one particular structure and disadvantages encompassing further trauma and possible damage to other joint tissues. Amirault *et al.* (1988) described a study in which Macintosh's lateral substitution reconstruction was performed on 27 patients with chronic ACL deficiency to reinforce the lateral collateral ligament; 75% of these patients showed subjective improvement in knee constraint.

Nordt *et al.* (1999) and Zaffagnini *et al.* (2007) also recognized that residual joint laxity remains with concomitant medial ligament injuries. In a study comparing 20 patients with combined ACL and MCL injury to 37 patients with isolated ACL injury intraoperatively, greater varus-valgus and AP laxity was found in the combined injury group (Zaffagnini *et al.*, 2007). This study supported the post-operative findings of Nordt *et al.* (1999) in which eight of the 21 knees studied in their acutely injured ACL patients accounted for the greatest difference in measured IE rotation between reconstructed and contralateral uninjured knees.

For this reason, it may be imprudent to simply accept the conclusions of those studies comparing SB and DB surgical techniques (Table 2.1) that did not address the ramifications of participants with concomitant ligament or meniscal injuries on their outcomes (Asagumo *et al.*, 2007; Ishibashi *et al.*, 2005; Järvelä, 2007; Muneta *et al.*, 2007). In fact, only two out of thirteen of these *in vivo* studies cited meniscal injury as an exclusion criterion (Seon *et al.*, 2007; Siebold *et al.*, 2008).

Several studies have also considered the effects of tunnel position and resulting graft orientation on rotational laxity following ACL reconstruction. Not only has it been demonstrated that a more oblique or horizontal femoral tunnel position results in better rotational constraint than vertical graft orientation (Scopp *et al.*, 2004; Stevenson & Johnson, 2007; Zaffagnini *et al.*, 2008); it was also found that there was no significant difference in rotational and AP laxity between DB and laterally oriented SB reconstructions at most flexion angles (Yamamoto *et al.*, 2004). Furthermore, anatomical DB surgical reconstructions also showed significantly superior anterior-posterior and rotational constraint when compared to reconstructions with deeper (i.e. non-anatomical and, effectively, more vertical) posterolateral bundle positioning (Zantop *et al.*, 2008).

This should also be taken into account when considering the results of the studies in Table 2.1 that compared the SB and DB techniques. In some cases the SB reconstructions were performed with the graft placed at either the AM or the PL position in order to facilitate further analysis of the DB reconstruction in the same knee (Ishibashi *et al.*, 2005; Steckel *et al.*, 2007a). This may not be the ideal or typical position for the SB graft and may have generated inaccurate results for the SB reconstruction.

Initial graft tension likewise affects knee laxity following both SB and DB reconstruction. In general, increasing initial graft tension was found to decrease translational and rotational kinematics (Markolf *et al.*, 2008b, 2009; Suggs *et al.*, 2003). When measuring laxity specifically in the DB reconstructed knee, the order in which the AM and PL bundles were tensioned, the amount of force in each bundle, and the flexion angle at which they were tensioned all distinctly influenced AP translation and transverse plane rotation (Cuomo *et al.*, 2007; Hoshino *et al.*, 2007; Markolf *et al.*, 2008b, 2009; Suggs *et al.*, 2003).

All of these studies demonstrated tensioning conditions in which either translation and/or rotation was restricted to less than that of the normal knee; suggestions to avoid overconstraint included tensioning both AM and PL bundles simultaneously at low flexion angles (10° to 20° of flexion (Cuomo *et al.*, 2007; Markolf *et al.*, 2008b) and applying moderate to minimal tension, generally less than 40 N, to both bundles (Hoshino *et al.*, 2007; Markolf *et al.*, 2008b, 2009; Suggs *et al.*, 2003). These studies were conducted on cadaveric or computational models, however, so it is not clear whether similar results would be found *in vivo* following a period of graft healing.

2.5 Dynamic joint constraint

Although loading conditions may not be as precise as certain passive joint laxity assessment techniques, the main advantage of dynamic weightbearing analysis is that both passive restraints and actively generated muscle forces interact throughout physiological movement tasks. Measuring stability of the ACL-deficient knee during gait activities that are performed on a regular basis is not only a means by which to determine abnormal laxity that may lead to long-term joint degeneration, but is also an approach that can help investigators to understand the mechanism which resulted in injury in the first place.

Anterior cruciate ligament rupture may be caused by either traumatic or non-contact injury, with the latter being most common (Boden *et al.*, 2000). Non-contact injury often occurs during abrupt deceleration maneuvers involving a subsequent change of direction, such as side-step cutting or landing after a jump (Boden *et al.*, 2000; McLean *et al.*, 1999). Although the exact mechanics that lead to ligament rupture vary, it is commonly thought that most injuries occur immediately following heel strike when the knee is close to full extension and the joint is subjected to both rotational and ab-adduction moments (Boden *et al.*, 2000).

2.5.1 The ACL-deficient knee

Rupture of the ACL has been shown to affect knee biomechanics in many respects, not simply anterior-posterior translation for which this ligament is the primary restraint. Some subjects with ACL deficiency demonstrated reduced knee flexion during weight acceptance as a stabilization strategy to prevent further joint damage (Rudolph *et al.*, 2001; Waite *et al.*, 2005). A reduction in knee flexion moment was also shown to coincide with peak knee flexion during stance (Berchuck *et al.*, 1990; Rudolph *et al.*, 2001; von Porat *et al.*, 2006). Berchuck *et al.* (1990) hypothesized that the reduction in knee moment was due to a decrease in net quadriceps force, effectively minimising the anterior tibial translation that would place stress on the deficient ACL; this was termed 'quadriceps avoidance' gait. In a computational model of normal knee mechanics during walking, Shelburne *et al.* (2004) furthermore found that maximum ACL force occurred during early stance due to anterior shear forces at the knee. Large shear forces were predominantly caused by both the magnitude and anterior direction of the patellar tendon force during weight acceptance (Shelburne *et al.*, 2004).

Using electromyography, some researchers have found that quadriceps muscle activity is not actually reduced during early stance (Reed-Jones & Vallis, 2008; Roberts *et al.*, 1999; Rudolph *et al.*, 2001; Waite *et al.*, 2005); alternatively, the reduced external flexion moment was attributed to greater hamstrings co-contraction, which would result in posterior translation of the proximal tibia (Rudolph *et al.*, 2001). Reducing external flexion moment in order to minimise the risk of anterior tibial translation and ACL strain does not seem to be a consistent strategy across all ACL-deficient individuals, however. In a subsequent study by the same research group that first established the idea of quadriceps avoidance gait, mean flexion moment of the ACLD knees was not found to be significantly lower than the contralateral uninjured knees (Andriacchi & Dyrby, 2005); yet, it was noted that flexion moment varied more in the ACLD subjects than within the normal subject group.

Analysis of transverse plane gait in the ACLD knee is far more limited in the literature than flexion-extension in the sagittal plane for two main reasons: obtaining accurate and reliable internal-external rotational knee kinematics is still problematic due to soft-tissue artefact associated with conventional skin marker-based data collection systems (Alexander & Andriacchi, 2001; Benoit *et al.*, 2007) and the contribution of the ACL to rotational restraint was – until recently – largely overlooked (Georgoulis *et al.*, 2003; Zaffagnini *et al.*, 2000).

As with sagittal plane kinematics and kinetics, IE rotation abnormalities do not appear to be consistent across all ACLD subjects. Georgoulis *et al.* (2003) and Andriacchi & Dyrby (2005) both reported greater internal rotation of the ACL-injured knee when compared with the uninjured knee during walking. In both studies the reduced external rotation occurred during the swing phase of gait; possible mechanisms suggested by the authors included increased activity of the rectus femoris (Georgoulis *et al.*, 2003) or a loss of screw-home movement (Andriacchi & Dyrby, 2005). Zhang *et al.* (2003) on the other hand, demonstrated a net increase in external rotation of the injured knee during walking. An increase and decrease in lateral and medial hamstrings activity, respectively, was proposed as a possible protective mechanism that would result in the observed increase in external rotation (Reed-Jones & Vallis, 2008; Zhang *et al.*, 2003).

The position of the centre of IE rotation on the transverse plane has also been investigated to determine whether it is affected by ACL deficiency. Study results during squatting and deep knee bend activities demonstrated that rotation resulted from movement of the lateral femoral condyle on the tibial plateau while the medial contact point remained unchanged (Dennis *et al.*, 2005; Hill *et al.*, 2000; Johal *et al.*, 2005; Yamaguchi *et al.*, 2009). If the medial compartment contact area remained relatively constant, this would imply that the axis of rotation was on the medial side of the joint. Yamaguchi *et al.* (2009) additionally investigated joint kinematics during a pivoting task in the same group of ACLD individuals and found the centre of rotation occurred just *lateral* of the midpoint of the medial and lateral contact points.

The reason for the contrasting ML centre of rotation position between the two activities was given as the difference in activities: squatting and pivoting were described as 'sagittal' and 'non-sagittal plane' activities, respectively (Yamaguchi *et al.*, 2009). This rationale was not supported by the findings of Koo & Andriacchi (2008), however. They found the knee joint centre of rotation to occur on the lateral side during normal walking, which is considered a sagittal plane activity. The conflicting results of this investigation with those of previous studies were attributed to the difference between ambulatory and non-ambulatory activities. Unfortunately, the findings of Tashman *et al.* (2004) during downhill running were not addressed by Koo & Andriacchi (2008); in that study the observed external rotation was again associated with a shift in contact area in the lateral compartment, suggesting a medial centre of rotation during this ambulatory activity (Tashman *et al.*, 2004).

An additional difference between the two ambulatory activity studies was that the subjects of Tashman *et al.* (2004) had had ACL reconstructive surgery, whereas those participating in the investigation of Koo & Andriacchi (2008) had healthy knees. Although it is possible that this may be the reason for the different findings, this seems dubious given the conclusions of comparable studies that determined centre of rotation positions in injured and uninjured knees. Koo & Andriacchi (2008) found matching patterns of AP and IE motion in the group of 23 healthy subjects and the previous group of 18 subjects (27 knees) with ACL injury; from this they extrapolated a laterally located centre of rotation in ACLD knees. Yamaguchi *et al.* (2009) similary found the location of the centre of rotation, whether on the medial or lateral side, to be consistent between the injured and contralateral knees.

Therefore, although the interpretation of the medial versus lateral position of the centre of rotation during different activities may not be adequately understood, researchers appear to agree that the ML centre of rotation position is not altered by ACL deficiency. Without further information it would be unreasonable to assume that the ACL-reconstructed knee examined by Tashman *et al.* (2004) would differ from both the normal and ACLD knee.

Whereas anterior-posterior laxity is relatively simply to measure with an arthrometer, measurement during dynamic gait is more complex due to the limitations of gait models and data collection systems. Some models, such as that used by the Vicon Clinical Manager, define the knee as a ball-and-socket joint and do not consider tibiofemoral translations at all (Kadaba *et al.*, 1990; Roren, 2005; Vaughan *et al.*, 1999). With side-to-side differences in AP translation being less than 10 mm during passive loading, the accuracy of traditional skin-based marker systems would not necessarily be great enough to differentiate between ACLD and uninjured knees during gait.

More accurate methods of measuring AP translation during dynamic tasks have been devised, however, demonstrating laxity in ACL-injured knees. Beard *et al.* (2001), for example, used optical markers placed on the lateral malleolus, tibial tuberosity, and patella to define the patellar tendon angle, from which anterior tibial translation could be calculated. This study actually measured greater anterior tibial translation *after* ACL reconstruction, whereas no difference had been found between injured and contralateral limb prior to surgery (Beard *et al.*, 2001). Using a point-cluster technique, Andriacchi & Dyrby (2005) found a significant decrease in anterior translation in the ACLD (unreconstructed) knee just prior to heel strike during walking, contrary to the findings of Zhang *et al.* (2003) which showed greater anterior translation of the ACLD knee throughout most of the swing phase of gait. Although differences were found to be statistically significant, no reliability data on their 6 DOF goniometer measurement device were included, nor were quantitative values of AP translation stated in the study by Zhang *et al.* (2003). Waite *et al.* (2005) found no difference in AP laxity between ACLD and contralateral limbs using a method similar to that of Andriacchi & Dyrby (2005).

The counterintuitive findings in AP laxity were interpreted by Andriacchi & Dyrby (2005) as linked to the simultaneously occurring reduction in external rotation; this may be better appreciated by considering the anterior-posterior positions of the medial and lateral tibiofemoral contact points with respect to the location from which AP translation was measured. With the advent of imaging techniques used during dynamic motion analysis, studies have been able to track the positions of the tibiofemoral contact points throughout the activity cycle. Figure 2.6 from Dennis *et al.* (2005) depicts the medial and lateral contact points on the tibial plateau during a deep knee bend. It is clear that rotation caused by anterior-posterior translation of only one side of the femur (i.e. internal-external rotation) corresponds to anterior or posterior translation of the midpoint of the trans-epicondylar femoral axis with respect to the centre of the tibia. In a subsequent publication, Koo & Andriacchi (2008) described a lateral side centre of rotation; if this was also the case with the ACLD subjects in their previous study (Andriacchi & Dyrby, 2005), then it would follow that the observed decrease in external rotation would be accompanied by a decrease in anterior translation.

Since the issue of the medial-lateral position of the axis of rotation during various dynamic activities is still uncertain, and given the interdependence of segment coordinate systems and coupled AP and rotation movements, it may be misguided to base conclusions on measured AP laxity without also considering tibiofemoral IE rotation.

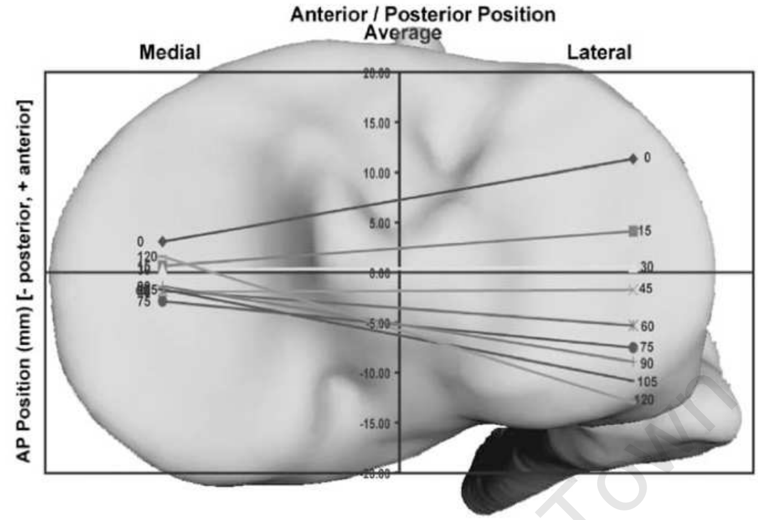


Figure 2.6: Medial and lateral tibiofemoral contact positions plotted during a deep knee bend activity (Dennis *et al.*, 2005).

2.5.2 Does ACL reconstruction restore knee function under physiological loading conditions?

While ACL reconstruction has proven to be an effective treatment for some patients, others have been incapable of returning to pre-injury activity levels. Studies have demonstrated improvement in joint restraint following reconstruction; however, this was often still inferior with respect to the healthy knee (Georgoulis *et al.*, 2003, 2007; Kowalk *et al.*, 1997; Ristanis *et al.*, 2005; Tashman *et al.*, 2004). Gait analysis has been useful in demonstrating complications with specific surgical reconstructive techniques; for example, a study examining biomechanical parameters in subjects who underwent bone-patellar tendon-bone reconstruction during stair ascent found that post-operatively sagittal plane knee joint moment and power was reduced in the injured knee, while moment and power and were increased in the contralateral ankle (Kowalk *et al.*, 1997). The authors suggested that donor site morbidity associated with the BPTB surgical technique resulted in a compensation mechanism that was not present pre-operatively.

These findings were supported by Webster *et al.* (2005), who compared kinematics and kinetics between patients who had received BPTB grafts against those who had received hamstrings tendon grafts. The results also showed a significant reduction in joint flexion moment at mid-stance in the knees that had undergone BPTB reconstruction, in addition to a reduction in extension moment in the hamstrings-reconstructed knees at terminal stance when compared with the control group.

Hamstrings and BPTB reconstruction techniques have been further evaluated in the transverse plane by Chouliaras *et al.* (2007). No differences in IE rotation were found between surgical techniques during a stair descent and pivoting activity; however, both ACL-reconstructed groups were still found to have significantly greater internal rotation than the healthy control group. Although rotational restraint had not been restored in either of these subject groups, previous studies had demonstrated improvement in IE rotational laxity with respect to the pre-operative ACLD state during walking (Georgoulis *et al.*, 2003).

Studies using RSA methods similarly presented unfavourable outcome in transverse plane rotation following ACL reconstruction (Brandsson *et al.*, 2002; Tashman *et al.*, 2004). The subjects evaluated by Brandsson *et al.* (2002) both pre- and post-operatively showed no difference in transverse plane rotation following BPTB reconstruction. Tashman *et al.* (2004), on the other hand, demonstrated an external rotation shift in their ACL-reconstructed subjects when compared with their contralateral uninjured knee during the stance phase of running gait. (The surgical technique used for reconstruction was not specified.)

Despite the growing number of publications comparing passive (i.e. non-weightbearing) laxity outcome between SB and DB surgical techniques, no study could be found examining both types of surgical techniques during dynamic physiological loading activities. Results from passive laxity studies furthermore present conflicting outcomes: some researchers have concluded that the DB technique is superior, several studies have found no statistical differences between clinical function in patients receiving either the SB or DB reconstruction, while still other studies have demonstrated possible complications associated with the DB technique.

A recent meta-analysis of randomised control trials found no significant differences in KT-1000 measured anterior-posterior or pivot shift dynamic joint laxities between SB and DB reconstructions, however, only four studies met the inclusion criteria for their primary analysis (Meredick *et al.*, 2008). Due to the complexity involved in quantitatively measuring *in vivo* rotational kinematics, the results in the literature have not yet been successful in presenting a complete understanding of the biomechanical differences in surgical techniques and the rolls of the AM and PL bundles in constraining torsional loads.

In light of this account of the literature, it is evident that the benefits of the double-bundle surgical technique from a biomechanical standpoint are debatable. Another practical consideration is the added time and cost of this procedure, which would potentially require a 24% reduction in ACL revision surgery to offset its expense (Brophy *et al.*, 2009). A better understanding of not just the ACL, but all structures involved in rotational restraint of the knee, is still required to allow a surgeon to recommend the best treatment for his or her patient.

In summary, the literature on knee rotation and the role played by the ACL (both the native and reconstructed ligament) were presented in this chapter. Cadaver studies have permitted a more precise description of the contribution of various anatomical structures to restraint in the three planes of motion. While *in vivo* studies are more representative of the actual function of the joint, it has been more difficult for investigators to apply precise loads within known and isolated directions of translation and rotation. For this reason, passive laxity studies regarding rotational outcome of single versus double-bundle surgeries are ambiguous: the majority have made conclusions on rotational joint restraint while in fact measuring motion under combined loading situations. Since no weight-bearing studies have assessed these two types of surgery, there is no physiological data with which to compare these passive clinical results. Fundamental research to determine the contribution of these reconstructive techniques is, therefore, still required before conclusions can be drawn on the best procedure to use for each patient.

Chapter 3

Measuring three-dimensional knee kinematics under torsional loading

3.1 Introduction

Studies investigating pathological knee kinematics are focusing increasingly on joint motion in all three planes, rather than simply the primary (i.e. sagittal) plane of motion. It has long been recognized that significant rotation in the transverse plane occurs throughout the range of flexion. Rotational laxity of the knee is now one aspect by which to diagnose knee pathology and evaluate surgical treatment, such as anterior cruciate ligament (ACL) reconstruction (Georgoulis *et al.*, 2003, 2005; Koh *et al.*, 2005; Logan *et al.*, 2004; Mannel *et al.*, 2004; Scopp *et al.*, 2004; Tashman *et al.*, 2004; Yagi *et al.*, 2002; Yamamoto *et al.*, 2004; Zaffagnini *et al.*, 2000).

Prior to medical imaging devices such as magnetic resonance imaging (MRI) that allow the investigator to observe the position and orientation of the underlying bone, non-invasive *in vivo* methods of measuring knee rotation were limited to external devices and skin markers prone to soft tissue artefact. With a much smaller range of knee motion in the transverse plane than in the sagittal plane, it was difficult to acquire results that could be considered reliable with these methods (Koh *et al.*, 2005). Techniques used in cadaveric studies, although more accurate than external devices due to the ability to insert markers directly on the bone, are too invasive to be used *in vivo* on large subject groups. Furthermore, cadaver studies are often confined to older knee specimens, which may not reflect the knee kinematics of a younger population, nor the mechanical properties of *in vivo* tissue.

Several methods for measuring *in vivo* knee and ankle joint kinematics in three dimensions (3D) have now been developed in which the relative positions of the bones were measured using MRI, computed tomography, fluoroscopy and biplane radiography (Bingham & Li, 2006; Fellows *et al.*, 2005b; Küpper *et al.*, 2007; Siegler *et al.*, 2005; Tashman & Anderst, 2003; Udupa *et al.*, 1998; Van Sint Jan *et al.*, 2006). Static or dynamic images of the bones at the joint were registered to their 3D segment models and associated coordinate systems to determine their positions and orientations in 3D space. Results from these studies were found to be more accurate than previous *in vivo* methods of measurement.

In this study, tibiofemoral knee kinematics were measured in 3D while the knee was subjected to torsional loading, thereby simulating a clinical examination. Furthermore, we wished to develop a method by which knee rotations and translations about and along all three axes could be measured *in vivo* with the ultimate intent being its application in the assessment of treatment of knee pathology. Our first objective, therefore, was to design and build a device that would apply a known torsional load to a subject's knee while being scanned using MRI; the images of the knee in the torqued position could then be used to measure six degree-of-freedom motion of the joint accurately and non-invasively. The next objective was to determine the feasibility and repeatability of this methodology with torques applied in internal and external rotation and with the knee in full extension and 30° of flexion. The within-subject variability associated with tibial rotation under different loading conditions would then demonstrate the potential of the system to provide results that are clinically relevant.

3.2 Methods

In order to measure 3D knee laxity objectively under torsional loading, a method for applying a precise load about a fixed axis with respect to the joint was required

(Küpper *et al.*, 2007). Once a specified load could be achieved, position and orientation of the femur and tibia were measured using MRI.

3.2.1 Torsional loading apparatus

In order to simulate clinical examination, the loading apparatus was designed to accommodate the greatest range of knee angles, with limitations governed only by the open-MRI magnet and patient bed. Figure 3.1 shows the computer model of the knee loading device designed around the MRI patient table for imaging of the right knee. Preliminary technical drawings from which the main components were manufactured are included in Appendix A. The aluminum slide rails permitted adjustment of flexion-extension and abduction-adduction angles, which were measured using a goniometer. The subject was positioned semi-supine with the knee joint (within the coil) at the centre of radius of the flexion-extension and ab-adduction tracks, so that only rotation of the shank was required. A plastic boot was connected to the rotation base via extension channels that permitted foot positioning toward the knee coil for shorter subjects as shown in Figure 3.2(a). The knee loading device was rotated about the patient table to permit imaging of the contralateral knee.

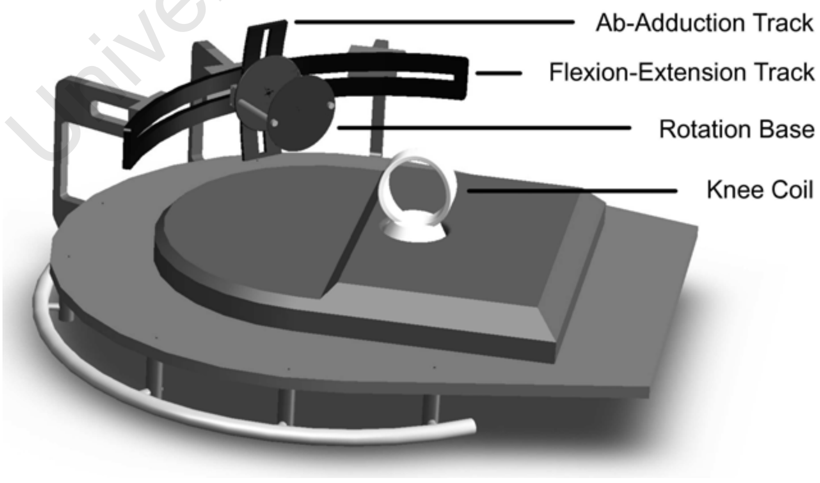


Figure 3.1: Model of knee torsional loading device mounted to MRI patient table.

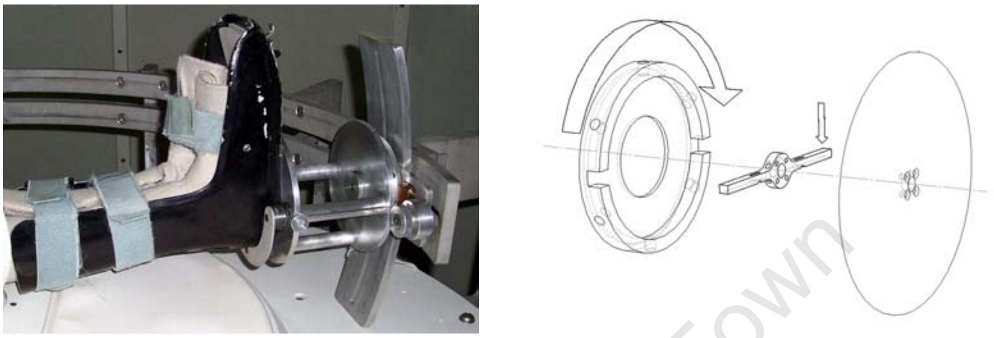
Since the focus of this study was on rotational loading in the transverse plane of the knee, the apparatus was designed to permit torque about only the long axis of the tibia while the other five degrees of freedom were fixed at the distal end of the shank. However, the thigh (and subsequently, the knee joint) was theoretically allowed six degree-of-freedom motion. Virtually full rotational freedom was achievable at the hip and translation was limited by ligament stiffness and joint mechanics, as well as body weight at the pelvis only. While not entirely unconstrained (as would have been the case had the proximal end of the femur been free to move without any restrictions), both rotations and translations at the knee itself were possible, albeit to a limited degree (Zavatsky, 1997).

In order to compare results between individuals and subject groups, a set torque was applied to each knee being examined. However, in order to account for the subject's mass which regularly affects the loads experienced by the knee, the applied torque was normalized to body mass according to the following equation:

$$T = 0.05 \left[\frac{Nm}{kg}\right] M + 1.25 \; [Nm] \tag{3.1}$$

where T is the applied torque in Newton-meters and M refers to the subject's mass in kilograms. This equation was based on data from the literature in addition to pilot data in which various torque values were recorded from minimum to maximum values corresponding to perceptions of comfort level (Blankevoort & Huiskes, 1996; Kanamori *et al.*, 2000; Mannel *et al.*, 2004; Yagi *et al.*, 2002). For several subjects of varying mass, the above equation represented the median of this range of values.

Two methods were used to measure the magnitude of the applied torque. The first method used an electronic load cell as shown in Figure 3.2(b). An internal torque was manually applied to the torque disc, which transmitted a linear force to one end of the load cell. The load cell was rigidly fixed to the rotation base and, in turn, the foot which resisted the applied torque. The resulting strain deformation in the load cell was measured by a strain gauge with data collected using LabVIEW^{TM} (National Instruments); circuit and calibration details are given in Appendix B. External torque was measured by a second strain gauge mounted onto the opposite side of the load cell. Calibration of each side of the load cell was achieved by hanging weights off the end of a lever arm of known length extending from the torque disc while the rotation base was rigidly fixed in place.



(a) Device in the extended position for shorter subjects.

(b) Load cell with torque disc and foot plate.

Figure 3.2: Torsional loading device components.

Since the strain gauge was continually measuring load, the investigators were able to observe a reduction in torque over time with the boot clamped at the angle corresponding to the specified torque. Presumably, this was due to relaxation of the soft tissues, both at the knee and the hip joints. Details of the supplementary investigation of the effect of relaxation on measured torque can be found in Appendix C. It was therefore decided that the torque would be reapplied following a sufficient period in which the rate of change of torque was less than 0.4 Nm/min. After reapplying the correct torque and ensuring a negligible drop in load for the secured position, the strain-gauge was disconnected to prevent image distortion during MRI scanning.

The second method of measuring the applied torque was a simplified approach involving the investigator (AH) pulling a commercial spring scale connected to the perimeter of the rotation disc of the boot via a thin cord. The load measured by the scale was then converted to a torque value based on the distance from the centre of rotation to the point of application of load (i.e. the radius of the rotation disc). The advantage of this method was that it was simpler and more robust, resulting in reduced set-up time and elimination of malfunctions caused by electrical disturbances. However, the protocol for the spring scale system was modified based on knowledge gained from the electronics method to attain greater accuracy when applying the torque: specifically, a two-minute ‘relaxation’ period was permitted as described previously before applying the final torque for the scan.

Using either method described above, it is possible that the applied torque would have decreased once the boot had been clamped at the appropriate position. Figure C.1 demonstrated that the drop in load following the designated two-minute relaxation period would not have been more than approximately 0.25 Nm over the entire three minute scan sequence. This quantity of change in torque did not enable sufficient movement during imaging to cause motion artefact and was, therefore, considered acceptable.

3.2.2 Data collection

Six volunteers with no history of knee injury were recruited for this study, the protocol for which was approved by the Human Ethics Committee of the University of Cape Town (Appendix D). Informed consent was given by each subject prior to data collection.

Since this method of measuring knee laxity was intended for use with patients having knee pathology such as ACL injury, it was necessary that the protocol minimize the time that the patient had to endure knee loading. Prolonged stress on an injured knee could not only cause discomfort for the patient, but could also cause muscle tensioning which would affect the contribution of the ligaments to joint constraint. However, a longer MRI scan sequence would generate higher resolution, and consequently more accurate images over the same field of view. Therefore, 3D models of the femur and tibia were generated from high resolution images scanned in a neutral (unloaded) position and shape-matched to models created from low resolution image volumes of the knee scanned under load. By matching the high and low resolution model of each segment, its position and orientation could be accurately determined without requiring a long MRI scan in a torqued position.

Magnetic resonance images were acquired using the 0.2 Tesla dedicated open-MR system (E-Scan XQ, Esaote, Italy) shown in Figure 3.3. Three-dimensional T1-weighted sequences with a 256 x 256 matrix were used for both high and low resolution transverse images (Figure 3.4). Experts including radiologists and medical imaging physicists alike agreed that the quality of these images was comparable to that of scanners with higher field strength (1.5 Tesla or greater) typically used in research studies; the exceptional images could be attained despite the low field strength due to the compact coil that fit closely around the joint.

Figure 3.3: Torsional loading device with subject's knee at 30° of flexion.

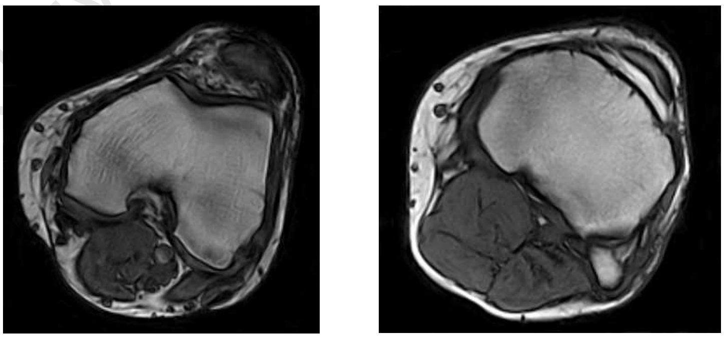


Figure 3.4: Low resolution magnetic resonance images of the femur (left) and tibia (right) scanned while an internal torque was applied to the knee.

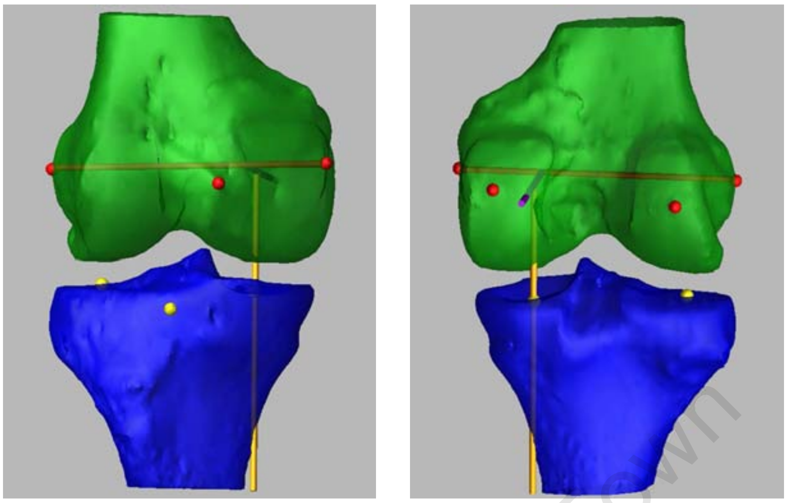
The axial plane was chosen for this sequence as it was the one with the greatest degree of knee motion under torsional loading, and found to have the greatest accuracy when measuring rotation in this plane (Fellows *et al.*, 2005b). The high resolution scan in a neutral knee position generated 90 contiguous slices of 1.56 mm thickness for a 14 cm field-of-view. This 3D image volume was acquired in just over 10 minutes. Four low resolutions scans (22 slices of 6.25 mm thickness) requiring only 2 minutes 50 seconds were taken with the subject's knee under load: internally and externally torqued with the knee in full extension, as well as internally and externally torqued with the knee at 30° of flexion.

3.2.3 Data analysis

Three-dimensional models of the knee were generated from the MR images scanned in the neutral position and for each of the torqued positions using a commercial segmentation software package (Mimics^{TM}, Materialise, Belgium). Point cloud models of each segment were exported to Matlab^{TM} in which the shape-matching procedure was completed. An iterative closest points algorithm based on the method of Fellows *et al.* (2005a) was used to register the points of the high resolution model segment to those of each associated low resolution model. A transformation matrix representing the rotations and translations from the high to low resolution models was recorded and subsequently used in the final description of kinematic position.

Local coordinate systems (LCS) were defined by identifying several anatomical landmarks on the high resolution 3D models of the distal and proximal ends of the femur and tibia, respectively. These 3D position data were then exported into Matlab^{TM} to calculate the LCS. Clinical descriptions of rotation and translation followed the convention developed by Grood & Suntay (1983). The flexion-extension axis was defined as the medial-lateral axis of the femoral coordinate system, the internal-external rotation axis was defined as the long axis of the tibia, and abduction-adduction occurred about the floating axis which was perpendicular to the preceding two axes (Grood & Suntay, 1983).

Figure 3.5 shows the 3D models of the femur and tibia with the anatomical landmarks used to define the LCS for each segment. The y-axis of the right femur extended from the lateral to the medial femoral epicondyle with the origin at its midpoint. (For the left knee the direction was reversed.) A temporary z-axis



(a) Anterior view. (b) Posterior view.

Figure 3.5: Anterior and posterior views of the 3D models of the right femur and tibia with anatomical landmarks. Flexion-extension, internal-external rotation, and abduction-adduction (floating) axes are shown in red, yellow, and purple, respectively.

was normal to the plane defined by the most anterior and posterior points of the medial femoral condyle and the most posterior point on the lateral condyle. The femoral x-axis in the posterior-anterior direction was defined as perpendicular to the y-axis and a temporary z-axis.

The origin of the tibial coordinate system was located in the middle of the medial plateau, since the axis of rotation extends through this position for the flexion range of 10° - 80° (McPherson *et al.*, 2005). The tibial y-axis extended from the lateral to medial tibial plateau midpoints for the right knee and was reversed for the left. The midpoints of the medial and lateral plateaus were defined as the most distal point in the central area of each plateau and could easily be identified on the 3D model. The z-axis was defined as normal to the plane of contact of the femoral condyles, i.e. the tibial plateau. The plane was defined as having the previously described points on the medial and lateral tibial plateaus, as well as the most anterior point on the most proximal slice of the tibial medial condyle. The x-axis of the tibia in the posterior-anterior direction was calculated as the cross-product of the y- and z-axes.

The clinical rotations and translations relating the tibial coordinate system to the femoral coordinate system were calculated before and after loading based on the previously determined transformation matrices derived from the shape-matching algorithm. The position of the tibia under the four conditions of torsional loading was always calculated with respect to the femur in the unloaded neutral position.

3.2.4 Feasibility study

A representative knee model, composed of two cylindrical MRI phantoms designating the femur and tibia respectively, was used to measure the accuracy of the segmentation and shape-matching analysis. Each phantom knee segment was manually positioned on specially designed cradles: one simulating 0° of knee flexion and the other simulating 30° of knee flexion (Figure 3.6). For each value of flexion, the tibial phantom was rotated externally by 20° and internally by 30° and scanned in each position using the low resolution scanning sequence described above. An additional scan simulating 0° of flexion and 0° of rotation was conducted using the high resolution scanning sequence. Local coordinate systems for each segment were aligned with the geometry of the high resolution phantom model rather than theoretical knee landmarks. Measurement of the position and orientation of the tibial with respect to the femoral component in the simulated torqued positions was carried out according to the protocol described in section 3.2.3.

Figure 3.6: Phantom model simulating a left knee at 30° of flexion.

3.2.5 *In vivo* repeatability study

A repeatability study was undertaken to measure the variability of knee joint kinematics under torsional load with each subject's knee at two angles: 30° of flexion and full extension. The protocol outlined in sections 3.2.2 and 3.2.3 was repeated five times by a single investigator (AH) on one knee of each of six subjects, using only one high resolution scan to build a 3D model and segment coordinate systems for each of the five trials. It was presumed that the greatest variation in knee kinematics would be associated with knee morphology under the specified load during data collection rather than the segmentation or shape-matching protocols with which associated errors had already been measured by the phantom knee model.

However, to verify this hypothesis and to limit any inaccuracy associated with the investigator's chosen anatomical landmarks, high resolution MR images were scanned for each trial for one of the six subjects. From each high resolution knee scan, 3D models were created and landmarks were identified to build the LCS for each segment. For these five trials, the repeatability of the identification of the knee landmarks was measured and the effects on the overall knee kinematics in the torqued positions were determined.

One female and five male subjects (age 29.3 $\pm$ 3.6 years, height 178.0 $\pm$ 8.6 cm, mass 72.0 $\pm$ 13.0 kg) were recruited for this study. Subjects had no history of injury for the knee joint of interest. Two left knees and four right knees were examined. A minimum of one day was given between trials for each subject, except for Subject 2 whose five trials were conducted over two days due to time constraints. Intraclass correlation coefficients (ICC) and standard error of measurement (SEM) were calculated for range of rotation data in both extended and flexed knee positions.

3.3 Results

Rotations calculated from the position of the phantom knee model using the segmentation and shape-matching protocol were compared with the actual rotations about the three clinical axes (Table 3.1). In both the simulated extended and flexed positions, the measured degree of internal and external rotation was within 1.6° of the actual rotated position.

Table 3.1: Actual and measured three-dimensional rotation angles (degrees) for the phantom knee model used for validation of the segmentation and shape-matching protocol.

Simulated Torque	**Knee Angle**	**Knee Extended** Measured	Actual	**Knee Flexed** 30° Measured	Actual
External	flexion	-0.2	0.0	27.7	30.0
	adduction	0.0	0.0	0.3	0.0
	external rotation	18.4	20.0	18.4	20.0
Internal	flexion	-0.6	0.0	28.0	30.0
	adduction	-0.3	0.0	0.8	0.0
	external rotation	-30.3	-30.0	-29.3	-30.0

Standard deviations in all three planes for all landmarks identified on the five neutral scans for Subject 1 were found to be less than 2 mm, except for the landmark on the anterior surface of the medial tibial plateau where the standard deviation in the medial-lateral direction was 2.2 mm (Table 3.2). The effect of the landmark position variability on the overall values of tibial rotation was minimal as demonstrated by Figure 3.7.

Table 3.2: Standard deviations (mm) of global x, y, and z positions of knee landmarks on one subject over 5 trials.

Bone	**Knee Landmarks**	Standard Deviation of Position **x**	**y**	**z**
Femur	medial epicondyle	0.1	0.3	1.2
	lateral epicondyle	0.7	0.3	0.9
	posterior surface, medial condyle	0.3	1.7	0.7
	anterior surface, medial condyle	0.3	0.9	1.3
	posterior surface, lateral condyle	0.3	1.2	1.1
Tibia	medial plateau, centre	1.3	1.5	0.7
	lateral plateau, centre	0.4	0.6	0.5
	anterior surface, medial plateau	0.4	2.2	0.8

Mean ranges of tibial rotation for the six subjects varied between 11.6° and 32.2° for the extended position and 17.2° and 28.8° for the flexed position; stan-

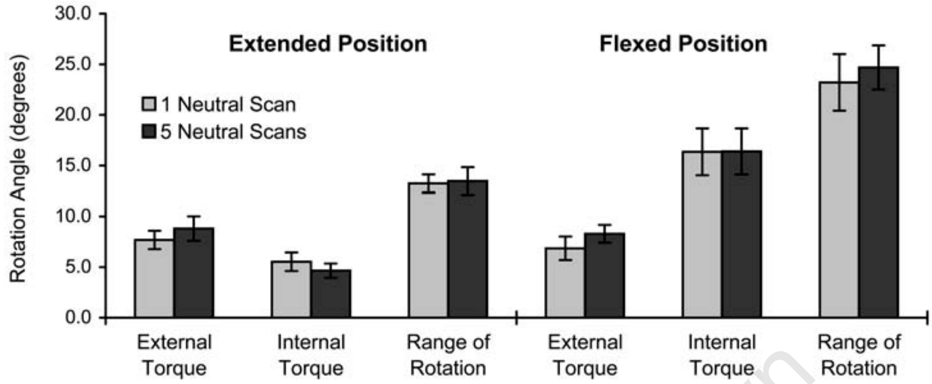


Figure 3.7: Mean and standard deviation of absolute tibial rotation angle (degrees) under external and internal torque loading for Subject 1 using one versus five neutral scans.

dard deviations over the five trials for external and internal rotations were consistently less than 2.5° (Table 3.3). ICC-values for the range of rotation were 0.99 and 0.93 in the extended and flexed knee positions, respectively. The standard error of measure was less than 0.75° for all subjects in both flexion and extension.

Table 3.3: Mean and standard deviation of knee rotation angles under torsional loading (convention: external = positive; internal = negative) and range of rotation for 6 subjects in extended and flexed positions based on 5 sets of data. Rotation angles are in degrees and applied torques are in Nm.

Sbj	Appl Torq	**Knee Extended**						**Knee Flexed** 30°					
		Ext Torq		Int Torq		Range		Ext Torq		Int Torq		Range	
		mean	SD	mean	SD	mean	SD	mean	SD	mean	SD	mean	SD
1	3.9	7.7	0.9	-5.5	0.9	13.2	0.9	6.9	1.2	-16.3	2.3	23.2	2.8
2	5.3	26.7	1.5	-5.5	0.8	32.2	1.7	19.9	1.9	-8.8	1.2	28.8	2.5
3	4.5	6.6	0.7	-5.0	1.6	11.6	1.7	9.0	1.4	-8.2	2.2	17.2	2.4
4	4.7	9.2	0.6	-12.8	2.4	22.0	2.8	8.7	1.2	-14.7	1.5	23.4	2.6
5	5.0	4.9	1.3	-8.6	1.3	13.5	2.2	2.8	2.0	-21.4	1.8	24.2	1.2
6	5.8	7.8	1.8	-7.1	1.6	14.9	2.1	5.7	0.9	-17.8	0.8	23.4	1.4

3.4 Discussion

3.4.1 Phantom knee model

The greatest discrepancy between actual and measured knee phantom position was in the degree of knee flexion in the simulated flexed position. This supported the findings of Fellows *et al.* (2005b) in which it was shown that greater accuracy could be obtained with MR images taken in the plane of motion (i.e. transverse images for rotation in the transverse plane). Since the primary focus of this technique was to measure knee rotation in the transverse plane, with measures of flexion in the sagittal plane being only a secondary objective, these results were considered acceptable.

This investigation gave an indication as to the accuracy of using the Matlab^{TM} registration procedure to match the 3D models produced from the low and high resolution scans. While the shape and features of the phantom did not correspond well to the tibia or femur, the number of registration points generated from the cylinders was similar to that of the *in vivo* bone segments and, therefore, adequately represented the knee joint for the purpose of this sub-study.

3.4.2 Anatomical landmark position

The variation in calculated landmark position over the five trials collected for Subject 1 could be attributed to inconsistencies in the MRI scans or inaccuracies in data processing, such as segmentation of the images and identification of the landmarks on the 3D segment models. The tibial and femoral landmarks for which the greatest standard deviations were measured – the anterior surface of the medial tibial plateau and the posterior surface of the medial condyle – were both used to define the transverse planes of their respective segments and the corresponding normal axes. The identification of the anterior surface of the tibial medial plateau in particular was, therefore, more crucial along the z-axis, rather than the y-axis, as it would be a discrepancy in the distal-proximal direction that would change the orientation of the transverse plane and corresponding axis of rotation.

All observed differences in the overall knee kinematics due to the use of only one versus the complete set of five neutral scans were less than the data processing errors associated with the knee phantom model in Table 3.1. In general, greater variation in the identification of positions of the tibial landmarks was observed. This was because the anatomical features on the proximal tibia chosen to define clinical segment axes were not as prominent as on the distal femur. Suitable landmarks at the distal end of the tibia were not within the limited field of view of the scanner available to us, and could therefore not be used. However, since the variation in knee kinematics was great enough to show differences under specific loading conditions as shown in Table 3.3, it was concluded that using only one high resolution neutral scan to analyse all five trials would be acceptable for each of the remaining subjects.

3.4.3 Measures of clinical rotation under torsional load for six subjects

In this study, an MRI-compatible torsional loading device, as well as data collection and image analysis protocol were developed to measure rotational knee laxity; its feasibility was tested using a phantom knee model. Results showed clinically relevant differences in the degree of knee rotation under four rotational loading conditions. All subjects demonstrated an increase in internal rotation with the knee flexed (Table 3.3), which agreed with the findings of Kanamori *et al.* (2000) and Musahl *et al.* (2007). Although standard deviations for each subject were greater than those reported by Musahl *et al.* (2007), their study used invasive bicortical pins on cadavers for a best case scenario. The large disparity in tibial rotation values in this study, in addition to smaller individual standard deviations versus those reported by Kanamori *et al.* (2000) for 12 cadaveric knees, indicated that variation across a subject group was more substantial than within repeated trials of an individual.

The accuracy of this methodology is furthermore superior to other non-invasive systems that have been used to measure *in vivo* knee rotation. External skin-mounted tracking sensors were used by both Shultz *et al.* (2007) and Tsai *et al.* (2008) with ICC-values between different testing sessions reported as 0.91 and

0.81 for the two investigations, respectively. The advantage of using MRI to prevent soft tissue artefact was also demonstrated by Okazaki *et al.* (2007) who demonstrated ICC of 0.96 and 0.98 when measuring the anterior tibial translation of the medial and lateral compartments at 10° of flexion; these values were comparable to those of 0.99 and 0.93 calculated from our data at full extension and 30° of flexion, respectively. The benefit of our methodology is that the 'matching' of unloaded and torqued knee models is determined mathematically from two sets of MRI data, rather than a comparison of invasive fluoroscopic images with MRI scans as required by the technology used by Okazaki *et al.* (2007).

An advantage of our methodology was the level of accuracy that was maintained despite the decreased MRI scan time required for patients having pain associated with knee pathology. This could be attributed to the individualized bone segment matching protocol, in which the low resolution 3D image volumes were matched to high resolution models developed from the subject's own knee, rather than bone segments from a database. This was reflected in the low SEM and high ICC-values calculated for the range of rotation, which suggest excellent agreement of the data over the different testing days. The non-invasive MRI technique permitted accurate measurement of the underlying bone, thereby avoiding skin motion artefact. Furthermore, it allowed the visualization of soft tissues around the joint; injury to these tissues may best be seen under load. The MRI-compatible torsional loading device and image analysis methodology developed in this study has been demonstrated to provide useful information for further investigation into normal and pathological knee laxity.

Chapter 4

In vivo joint laxity under torsional loading in the healthy knee

4.1 Introduction

In order to characterise pathological changes in joint stability, one must first have an understanding of the biomechanics of the healthy knee joint. The predominant motion of the knee is flexion in the sagittal plane; however, it has long been shown that physiological rotations and translations occur in all three planes of motion. The screw-home mechanism characteristic of the healthy knee is the coupled internal rotation of the tibia with respect to the femur as it flexes; at full extension, the coupled external rotation provides joint restraint (Benoit *et al.*, 2007; Chen *et al.*, 2001; Crawford *et al.*, 2007; Koh *et al.*, 2005; Moglo & Shirazi-adl, 2005; Piazza & Cavanagh, 2000; Shefelbine *et al.*, 2006; Wilson *et al.*, 2000). Furthermore, variable degrees of varus-valgus, anterior-posterior, medial-lateral, and distal-proximal laxities have been measured in healthy subjects under dynamic and passive loading conditions (Benoit *et al.*, 2007; Dennis *et al.*, 2005; Georgoulis *et al.*, 2003; Küpper *et al.*, 2007; Li *et al.*, 2006; Zhang *et al.*, 2003).

The geometries, configurations, and properties of various anatomical structures that comprise the knee, provide this complex joint with the stability required to withstand most loading situations accompanying daily tasks. The articular surfaces of the tibial plateau and femoral condyles, the cruciate and collateral ligaments, iliotibial tract, posterior oblique ligament, arcuate ligament, and menisci are among the main structures indicated to maintain passive restraint of the knee (Amirault *et al.*, 1988; Amis *et al.*, 2005; Blankevoort & Huiskes, 1996; Defrate *et al.*, 2004; Meyer & Haut, 2008; Nordt *et al.*, 1999). Their contributions to overall restraint is dependent on the direction and magnitude of the applied loads.

Considerable research has been dedicated to the mechanism by which the anterior cruciate ligament (ACL) contributes to joint constraint; in most recent years the focus has been specifically on rotational restraint in the transverse plane. Although the exact mechanism of ACL rupture is unknown, it is thought that non-contact injury generally occurs with concomitant valgus bending and external rotation of the joint (Meyer & Haut, 2008). However, knee kinematics have more often been measured under a combined internal and valgus rotatory load simulating the pivot shift phenomenon, which has been shown to correlate to laxity symptoms associated with ACL injury (Amis *et al.*, 2005). Due to the difficulty in quantifying the pivot shift motion, outcome measures *in vivo* have largely been limited to subjective grading systems (Amirault *et al.*, 1988; Järvelä, 2007; Meredick *et al.*, 2008; Streich *et al.*, 2008; Yasuda *et al.*, 2006). Kubo *et al.* (2007) and Yagi *et al.* (2007) presented methods of measuring velocity and acceleration of the tibiofemoral motions as a means by which to quantify the pivot shift; however, actual applied varus-valgus and rotational loads were not measured.

Most clinical trials reporting quantitative laxity measured under a known load used an anterior-posterior (AP) laxity arthrometer, the most accessible validated measurement tool, and thus were limited to AP laxity in a single plane (Meredick *et al.*, 2008). Intra-operative navigation systems have been used to get three-dimensional (3D) quantitative kinematic data; again however, the precise loads applied by the surgeon were generally not recorded (Ferretti *et al.*, 2008; Martelli *et al.*, 2007; Zaffagnini *et al.*, 2007). Several *in vitro* studies have applied precise torques, either independently or combined with varus-valgus or AP loads, and measured resulting internal-external rotations (Amis & Scammell, 1993; Diermann *et al.*, 2008; Gabriel *et al.*, 2004; Kanamori *et al.*, 2000; Kaneda *et al.*, 1997;

Mannel *et al.*, 2004; Meyer & Haut, 2008; Scopp *et al.*, 2004; Yamamoto *et al.*, 2004). Far fewer studies were found in which tibiofemoral transverse plane rotation was measured under known torsional loading *in vivo* (Almquist *et al.*, 2002; Nordt *et al.*, 1999; Schmitz *et al.*, 2008; Shultz *et al.*, 2007; Tsai *et al.*, 2008); due to measurement methods, however, it may not have been feasible to evaluate joint motion in the other anatomical planes.

This chapter investigates the six degree-of-freedom kinematics resulting from internal and external torsional loads applied to the healthy knee at two positions of flexion: full extension at which the knee is locked and rotation is thought to be restricted (Benoit *et al.*, 2007; Crawford *et al.*, 2007; Koh *et al.*, 2005) and 30° of flexion at which non-contact ACL injury commonly arises (Boden *et al.*, 2000). The data gathered was used to establish the normal variability of knee motion in a healthy population under these loading conditions. By testing both left and right knees of each subject, symmetry could be verified in order to support the use of patients' contralateral limbs as controls in future studies. Understanding this data and the mechanisms by which the knee is able to restrain rotational loads is an essential baseline to determine the effects of ACL pathology such as rupture or reconstruction using various surgical techniques.

4.2 Methods

Fifteen subjects (4 female, 11 male) with no history of knee injury were recruited for this study. Informed consent was given by each subject, as required by the protocol approved by the University of Cape Town Ethics Committee (Appendix D). Subjects ranged in age from 22 to 43 years of age and were all moderately to very physically active, representing a normal population in which ACL rupture may occur as a result of sporting injuries. Demographic data is included in Table E.1.

The data collection and analysis protocol followed for this study was described in detail in Chapter 3. External and internal torques were applied to both left and right knees while each knee was in full extension and then repositioned to 30° of flexion. Applied torques were normalized to each subject's body mass. Low resolution 3D T1-weighted images (6.25 mm slice thickness) were generated by the 0.2 Tesla magnetic resonance imaging (MRI) scanner in less than 3 minutes while the joint was under load. The 3D image volume was then shape-matched to a high resolution image volume (1.56 mm slice thickness) scanned in a no-load position. Three-dimensional rotations and translations of the tibia with respect to the femur were calculated by comparing the transformation matrices before and after torque was applied.

Statistical analysis was performed using SPSS 15.0 (SPSS Inc) software. Paired t-tests were used to detect differences in left and right knee measures.

4.3 Results

4.3.1 Torque applied versus rotation measured

Absolute correlation coefficients for applied torsional load versus range of rotation were both less than 0.37, signifying linear independence of these variables in both extended and flexed knee positions (Figure 4.1).

4.3.2 Six degree-of-freedom knee kinematics

Overall subject means and standard deviations for the translations and rotations in the three anatomical planes indicate that the greatest tibiofemoral movement under torsional loading was in internal-external rotation (Figure 4.2) and anterior-posterior translation (Figure 4.3). The increase in range of rotation from 15.4° and 14.3° (left and right, respectively) in extension to 25.6° and 23.5° in the flexed knee position was primarily due to an increase in internal tibial rotation; external rotation values remained similar in the two positions of flexion (Figure 4.2). Measured values of tibiofemoral flexion were approximately 3° - 5° higher under external torque versus internal torque in both positions of flexion. Variation in knee flexion-extension was greater than in ab-adduction, and in general, standard deviations tended to be greater in the flexed knee position. Subject-specific data and individual ranges of rotation are listed in Table E.1.

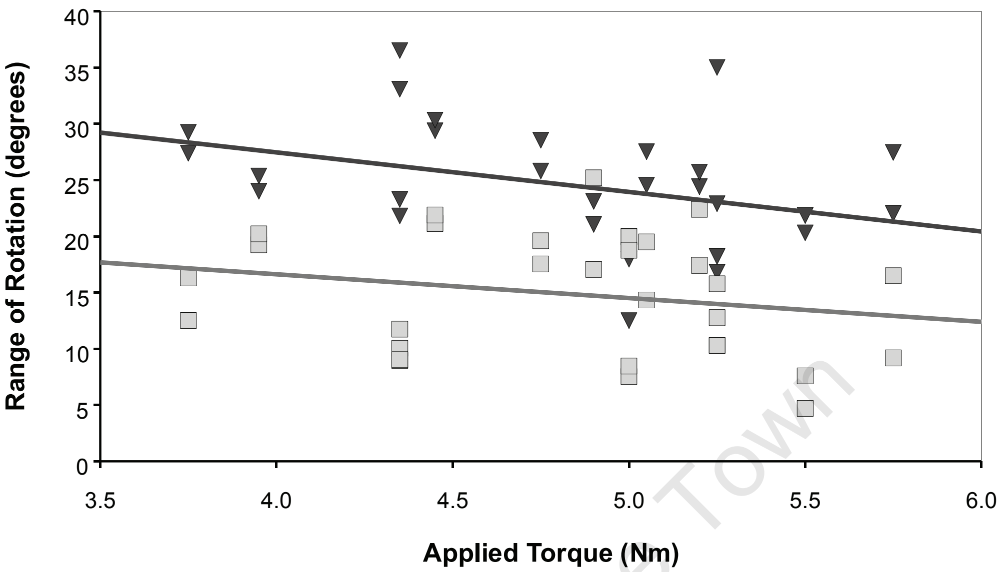


Figure 4.1: Applied torque versus measured rotation for left and right knees. Extended position (■) and flexed position (▼) data are shown with regression lines. R-squared values for linear regressions in extended and flexed positions are 0.05 and 0.13, respectively.

4.3.3 Rotation coupled with anterior-posterior translation

Figures 4.4 and 4.5 show that there was a correlation between internal-external rotation and anterior-posterior translation under torsional loading in both the extended and flexed positions. In the flexed position with an internal torque, a smaller translation was coupled with rotation as compared to the other three loading conditions; this is demonstrated by the smaller slope of the linear regression curve.

4.3.4 Rotation: Left–right symmetry

Significant differences in external rotation were found between left and right knees in both extended and flexed knee positions with the left knee showing greater rotation (Table 4.1). In the internally rotated positions, however, the right knees tended to have increased laxities.

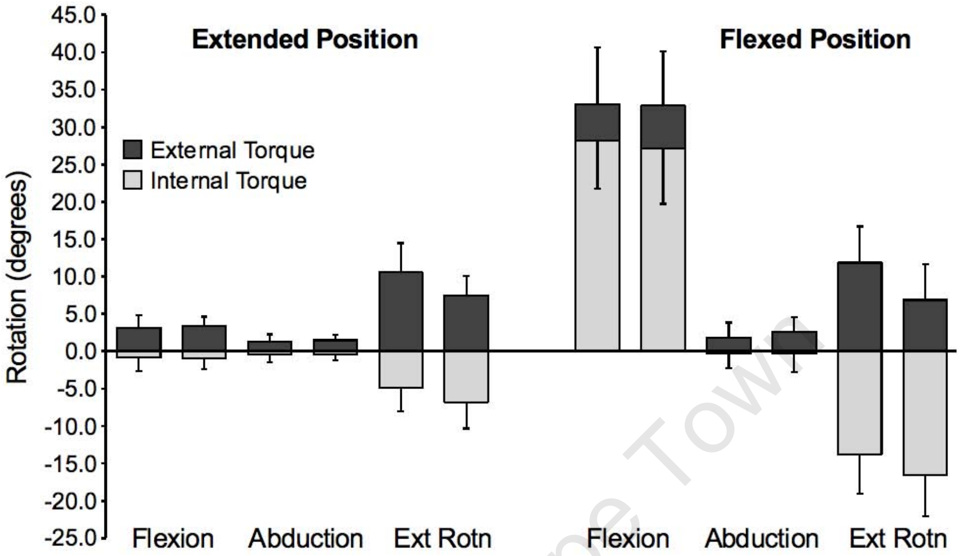


Figure 4.2: Rotation in three planes under torsional loading in extended and flexed knee positions. All values start at 0. Left and right knee data are presented side-by-side for each rotation with positive directions as marked: Flexion, Abduction, and External Rotation.

Table 4.1: Mean left-right differences in absolute internal and external rotations with levels of significance.

Applied Torque	**Knee Extended** Mean difference	p-value	**Knee Flexed 30°** Mean difference	p-value
External	3.0	0.009	5.0	0.001
Internal	-1.9	0.052	-2.8	0.064

4.4 Discussion

The purpose of this study was to determine the 3D kinematics resulting from internal and external torsional loads in the healthy knee joint in order to establish a baseline against which data from ensuing studies involving ACL patients can be compared. Unlike other studies in which kinematics were measured under externally applied loads, this investigation normalized the torque applied according to each subject's body mass. The normalization equation (equation 3.1) assumes a

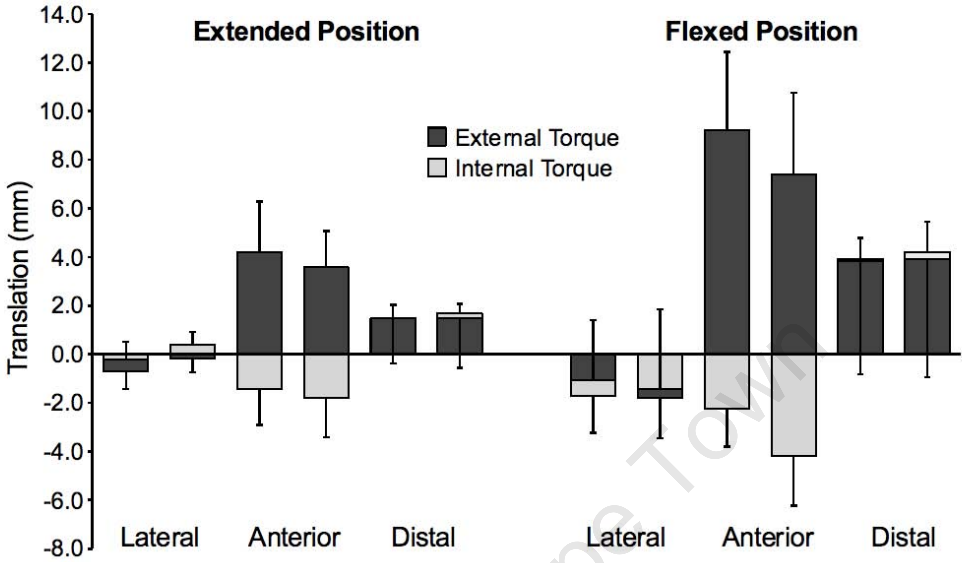


Figure 4.3: Translation in three planes under torsional loading in extended and flexed knee positions. All values start at 0. Left and right knee data are presented side-by-side for each translation with positive directions as marked: Lateral, Anterior, and Distal.

direct relation between subject mass and the torque that can be tolerated, based on observations made in our pilot study. This equation permitted standardization of the applied load, while allowing the greatest possible load to be used without causing discomfort to the subject. Almquist *et al.* (2002) measured rotation using torques of 3 Nm, 6 Nm, and 9 Nm and showed that there is a direct relationship between torque and range of rotation when applied to the same knee at the same flexion angle. Since range of rotation was shown to be independent of applied torque as demonstrated in Figure 4.1, the normalization used was not only valid, but essential when comparing knee laxities of subjects with varying mass.

Four other investigations that applied rotational loads to the knee *in vivo* all used a distinct torque values of either 5 Nm (Nordt *et al.*, 1999; Schmitz *et al.*, 2008; Shultz *et al.*, 2007) or 6 Nm (Tsai *et al.*, 2008) for every subject tested, despite subject mass standard deviations of up to 11.4 kg. One of our subjects weighing only 54 kg expressed mild discomfort with the 4 Nm external torque applied in the flexed position; a 5 Nm torque would be expected to cause this

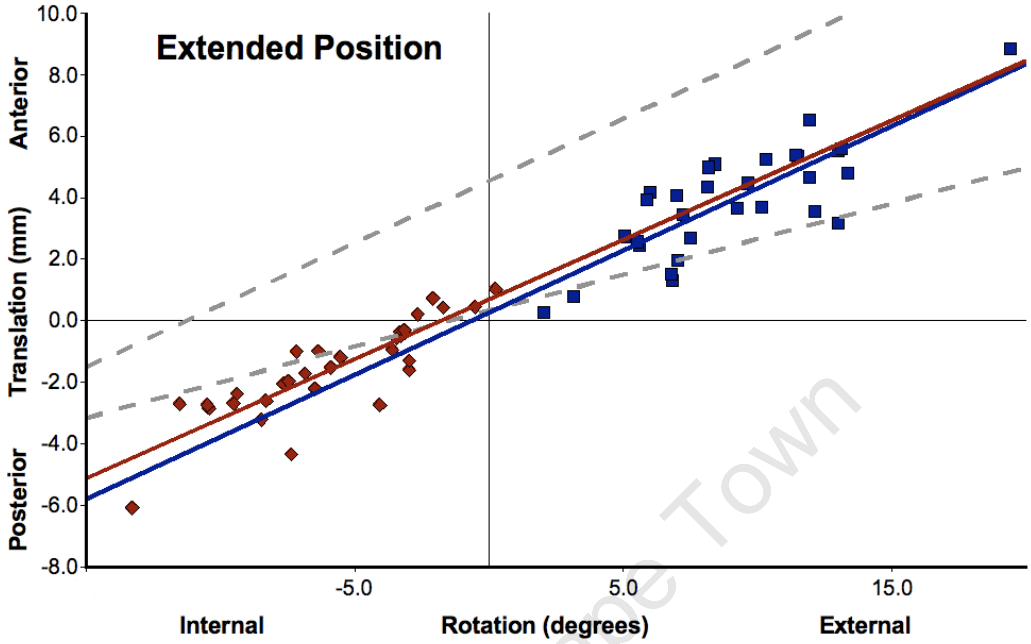


Figure 4.4: Internal-external rotation versus anterior-posterior translation for left and right knees of 15 subjects in the extended knee position. External torque (■) and internal torque (◆) are shown with regression lines. (External and internal torque regressions in flexed position from Figure 4.5 are shown as grey dashed lines for comparison.)

subject to contract the muscles surrounding the joint so as to resist the load, thereby affecting the measured passive laxity. Since subject-specific data were not presented by other authors, it is not known whether their results using a distinct load exhibited a relationship between subject mass and measured laxity. In this study, the mean torque applied was 4.8 Nm, which is very close to the single value used by other researchers; therefore, it is reasonable to compare mean outcomes from the different studies given the lack of similarly derived data in the literature.

The larger variations in knee kinematics observed in the flexed position may be attributed to the imprecise positioning of the knee by the investigator using the manual goniometer. The standard deviations of between 6.4° and 7.6° corresponding to flexion angle readings in the flexed knee position are within normal limits of accuracy using this equipment (Jagodzinski *et al.*, 2000) and are the reason exact measures of knee flexion were recorded using the 3D models developed from the MR images. The low variability associated with the ab-adduction values, and the fact that they were close to 0° in all positions of loading, are a good indicator that there was minimal kinematic crosstalk in the measurements (Charlton *et al.*, 2004; Piazza & Cavanagh, 2000).

The magnitudes of rotational laxity measured in the extended and flexed positions agree well with data from six separate subjects used to validate the methodology (Chapter 3), as well as results from published studies. In cadaveric studies, increases in rotational laxity from an isolated internal torque ranged from just under 8° to approximately 12° (Blankevoort *et al.*, 1988; Kanamori *et al.*, 2000; Musahl *et al.*, 2007), comparable to our findings of a left-right mean increase of 9.5° of internal rotation. Absolute magnitudes of rotation in the extended position closely matched data presented by Blankevoort *et al.* (1988). However, their results in the flexed position and those of Kanamori *et al.* (2000) and Musahl *et al.* (2007) were about 7° to 8° larger in both external and internal

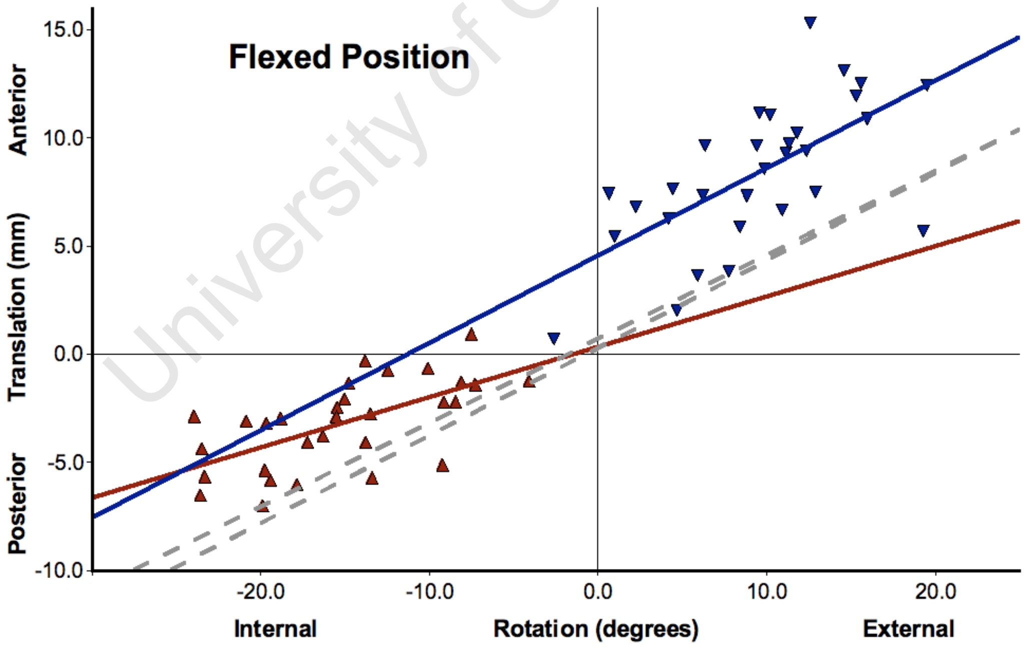


Figure 4.5: Internal-external rotation versus anterior-posterior translation for left and right knees of 15 subjects in the flexed knee position. External torque (▼) and internal torque (▲) are shown with regression lines. (External and internal torque regressions in extended position from Figure 4.4 are shown as grey dashed lines for comparison.)

rotation than the findings of the present study (i.e. about 15° larger in the overall range of rotation). Nonetheless, these rotations were exceeded by at least one subject in each position in our study.

Magnitudes of rotation from our study more closely match results given by Shultz *et al.* (2007) and Nordt *et al.* (1999) measured at 20° of knee flexion, and Tsai *et al.* (2008) at 30° of flexion, indicating that unconscious muscle tensioning may have contributed to joint stiffness *in vivo*. Interestingly, under external torsional loading, our results showed similar rotational laxities in both positions of flexion, unlike the findings of Musahl *et al.* (2007). Since no other study with comparable measurements of external rotation at 0° and 30° of flexion could be found, it cannot be concluded that the contrasts can only be attributed to differences in study design (e.g. *in vivo* versus *in vitro* models).

Interestingly, significant differences of up to 5° in transverse plane rotation were found between left and right knees in the four different loading conditions. Only two other studies could be found in which bilateral knee rotation was measured in healthy subjects (Shultz *et al.*, 2007; Tsai *et al.*, 2008). Neither of these studies found significant side-to-side differences; however, methods of measurement involved skin markers prone to soft tissue artefact, resulting in measurement errors of 5° or more.

The standard error of measurement (SEM) using our methodology was less than 1°, as shown in Chapter 3; therefore, the standard deviations of up to 5.5° in our subject group reflect true inter-subject variation, rather than measurement error. This variation across subjects indicates that knee rotation may vary substantially in a healthy population; the observed side-to-side differences, although statistically significant, may not be clinically relevant. The difference in range of rotation was less than 2.3°, however, which may be evidence that this is a more meaningful measure when using the contralateral limb as a control in studies involving knee pathology.

The asymmetry of internal and external rotation due to torsional loading may be explained in part by the viscoelastic behavior of ligaments. With the knee in flexion, a substantial degree of ‘primary’ rotation – easily up to or beyond 10° in each direction – occurs with relatively small (1 to 2 Nm) initial torque values (Musahl *et al.*, 2007; Wang & Walker, 1974). Rotation in excess of this initial laxity requires disproportionately more torque due to the non-linear stiffness of the ligaments (Woo *et al.*, 2006).

The neutral resting position of the subject in which the high resolution MRI scan was performed with the knee in full extension was *not* assumed to be at 0° of rotation in our study. Instead, the degree of neutral position rotation was subtracted from the torqued measure of rotation to calculate the net rotation under load. At full extension, the degree of primary laxity is likely less than 10° in each direction. However, an imbalance in an individual's neutral knee position may consistently fatigue the rotational restraints in one direction to a greater extent than in the other, resulting in an imbalance in ligament laxity. This would not only account for the differences of up to 5° in one direction, but would also account for the *smaller* differences in total range of knee rotation. If the bilateral differences in internal and external rotational laxity may be attributed to variations occurring within the initial range of primary laxity, this quantity may not be relevant when diagnosing knee injuries under passive loading conditions.

One topic of interest when investigating kinematics of the knee joint is its axis of rotation; the relationship between internal-external rotation and AP translation shown in Figures 4.4 and 4.5 give a good indication as to its position in the four loading conditions investigated in this study.

Although the helical axis could have been calculated to determine the precise location of the centre of rotation, its physical interpretation is clinically meaningless unless no other rotation or translation aside from internal-external rotation and distal-proximal translation were to result from the applied torque. This could not be assumed; nor could it be taken for granted that the helical axis of rotation would be similar for both internal and external rotational loading. Furthermore, to make the data comparable between subjects after establishing an average helical axis and to 'convert' the results to clinically comprehensible rotations and translations, an anatomical reference frame would still be required from which sagittal, coronal, and transverse plane motions could subsequently be calculated (Dennis *et al.*, 2005).

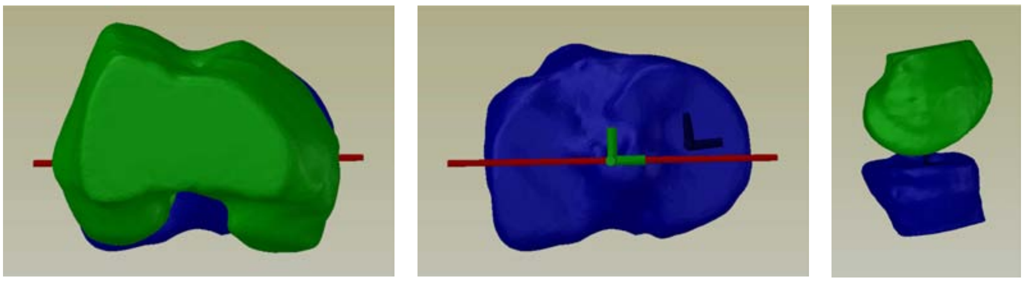
The kinematic model, based on the Grood & Suntay (1983) joint coordinate system, defined rotation about the long axis of the tibia. The position of the rotation axis in this model was at the origin of the tibial coordinate system placed at the centre of the surface of the medial plateau. In the extended position, the coupling of anterior translation with external rotation and posterior translation with internal rotation implies an actual axis of rotation located lateral to the chosen axis position, i.e. closer to the midpoint of the medial and lateral tibial plateaus. Rotation about a more central axis would cause the observed anterior or posterior translation of the chosen origin with respect to the femoral origin located at the midpoint of the epicondyles.

The axial view of the tibial plateau and origin with respect to the position of the femoral origin clearly illustrates this coupled movement (Figure 4.6). This central approximation of the actual location of the rotation axis is supported by Kaneda *et al.* (1997), in which the location of the mean helical axis under 3 Nm of external torque was located at the medial tibial spine in 15 cadaveric specimens. Furthermore, the regression lines calculated from our *in vivo* data for both the external torque (Figure 4.4, blue) and internal torque positions (Figure 4.4, red) show y-intercepts of approximately 0, signifying an absence of translation without rotation; in other words, the translation measured was not real, but simply an artefact of a misplaced tibial origin.

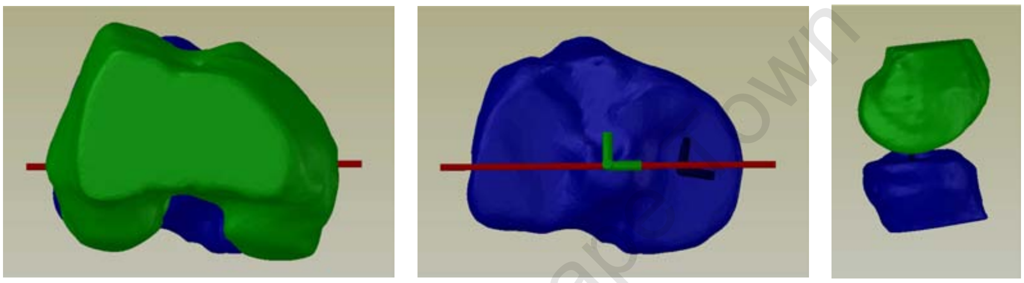
In fact, the coupling of rotation and translation is likely sinusoidal, rather than linear. Figure 4.7 shows the measured AP translation as a function of the *sin* of the degree of rotation and the distance between the chosen origin and the actual centre of rotation; this is described by the following equation:

$$t = d\, sin\theta \qquad (4.1)$$

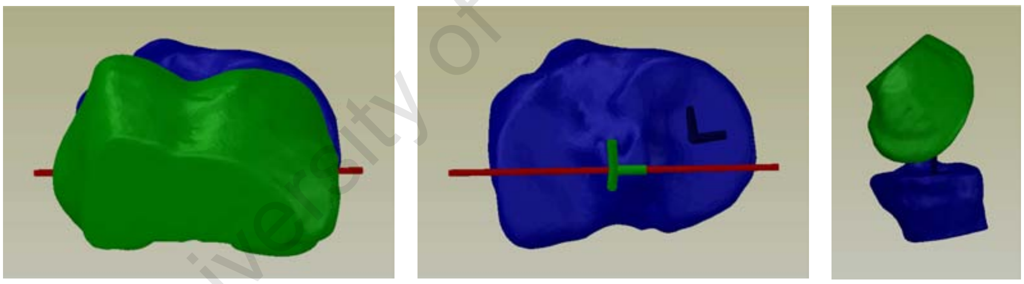
where t is the anterior-posterior translation, d is the distance between actual and chosen origins, and θ is the angle of rotation. (A similar relation was described by Roos *et al.* (2006) with the tibial origin in flexion-extension.) Figures 4.4 and 4.5, however, displays data from 15 different subjects, rather than 15 rotation angles of the same knee, making the linear relationship an acceptable assumption with this limited amount of data. The roughly equivalent slopes in the internal and external rotation positions indicate a similar distance between actual and chosen origins in each condition.



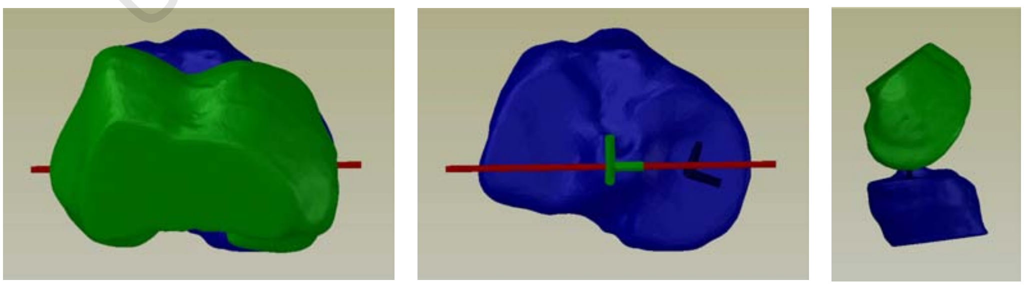
(a) Extended external rotation.



(b) Extended internal rotation.



(c) Flexed external rotation.



(d) Flexed internal rotation.

Figure 4.6: A tibiofemoral model viewed in the transverse and sagittal planes. The middle column shows only the femoral coordinate system and flexion-extension axis to enable relative tibial and femoral origin positions to be seen.

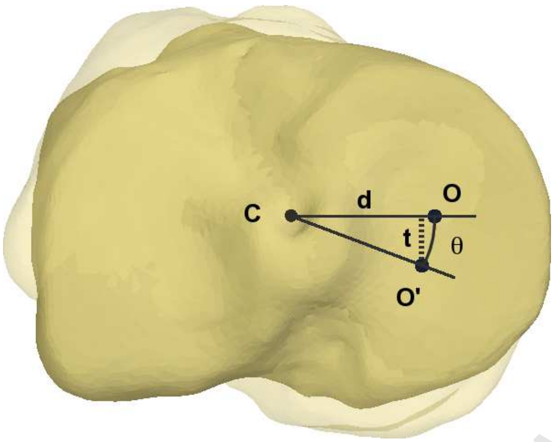
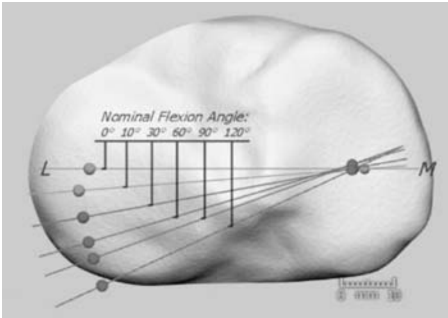


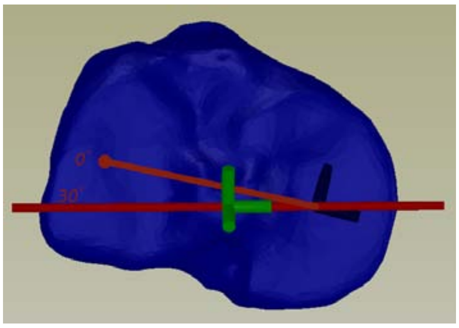
Figure 4.7: Anterior-posterior translation resulting from internal-external rotation of the tibia about an axis located at a different position from the tibial origin. O is the original position of the chosen origin, O' is its position following rotation about the centre C, d is the distance between actual and chosen origins, θ is the angle of internal rotation, and t is the magnitude of posterior translation represented by the dotted line.

In the flexed position with an external torque, the slope of the regression line is about the same as for the extended positions; however, it has been translated anteriorly (Figure 4.5, blue). As with the extended position, a lateral shift of the axis of rotation has likely occurred as illustrated in Figures 4.6(c) and 4.6(d). Equal slopes in this position and the extended positions correspond to an equivalent extent of displacement.

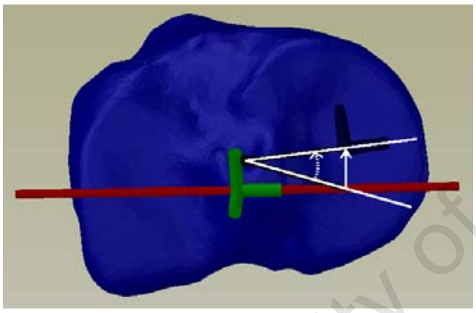
The anterior translation of the regression line can be explained by the screw-home motion that occurs with knee flexion (Figure 4.8). From extension to 30° of knee flexion we know that tibiofemoral roll-back occurs on the lateral, but not the medial side of the knee, resulting in coupled internal tibial rotation about a medially oriented rotation axis (Crawford *et al.*, 2007; Dennis *et al.*, 2005; Iwaki *et al.*, 2000; Johal *et al.*, 2005; Koh *et al.*, 2005; McPherson *et al.*, 2005; Pinskerova *et al.*, 2000; Wilson *et al.*, 2000). Figure 4.8(a) illustrates the posterior translation on the tibial plateau of the lateral femoral condyle. A femoral origin located midway between the epicondyles would consequently also move posteriorly. Since the axis of rotation in our model is on the medial plateau and AP translation is measured along the floating axis perpendicular to the epicondylar axis, this rotation alone would not account for AP translation (Figure 4.8(b)).



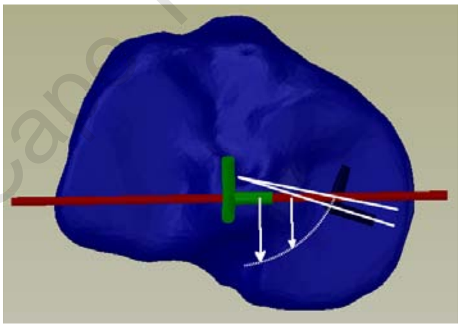
(a) A femoral origin located at the midpoint of the epicondyles would move posteriorly on the tibial plateau with 30° of flexion (McPherson *et al.*, 2005).



(b) The tibia and femoral epicondylar axis have been rotated to show the same internal rotation resulting from flexion, but with the epicondylar axis remaining horizontal.



(c) With an external torque, the tibial origin moves anteriorly relative to the femoral origin.



(d) With an internal torque, the rotation about a central tibial axis causes the tibial origin to translate posteriorly relative to the femoral origin, but with a smaller relative displacement than in external rotation.

Figure 4.8: Coupled rotation and anterior-posterior translation resulting from torsional loads.

The addition of an axial external rotation torque counteracted this internal rotation in every knee, as deduced from exclusively positive values of external rotation. With rotation about a more centrally located axis, the coupled translation measurements indicate that the medial condyle followed the lateral condyle and moved posteriorly on the medial plateau, i.e. the chosen tibial origin moved in an anterior direction with respect to the femoral origin (Figure 4.8(c)). Depending on the laxity of the knee, the external torque may have simply balanced the internal rotation resulting in a net rotation of 0° and anterior translation of about 5 mm. Alternatively, in other subjects, it resulted in external rotation coupled with anterior translation – with the overall amount of translation greater than the amount in the extended position at the same degree of rotation (Figure 4.5).

The predetermined internal rotation of the tibia that occurred with flexion was magnified with the addition of an internal torque and generated the larger increase in overall rotation in the flexed internal torque position compared to the flexed external torque position (Figure 4.2). With an effective axis of rotation anterior to the femoral origin, internal rotation results in a smaller posterior translation than the anterior translation in the external rotation position, since the AP distance between femoral origin and axis of rotation must be taken into account (Figure 4.8(d)). The slope of the regression line illustrating the rotation-translation correlation is less, in the flexed internal torque position (Figure 4.5, red), than the slopes in the other three loading conditions, showing that on average there is less posterior translation of the chosen origin than in the extended position with the same amount of internal rotation.

Joint stability may be compromised with the deficiency of any structure that provides support. Pathological laxity is a result of the joint following the path of least resistance under a specific external loading condition (Nordt *et al.*, 1999). The path of least resistance is the pathway of motion that occurs when the overall force (i.e. resistance torque) of all contributing structures is minimized; the greater the contribution of a specific structure to the net restraint, the greater the displacement will be towards that structure in order to minimize the total resistant force.

The contribution of each structure to the net rotational restraint depends on its material stiffness properties (tissue elasticity) and its perpendicular distance from the location of the applied torque. This can be summarized by the following cross-product equation:

$$\Sigma T = \Sigma (F \times r) \tag{4.2}$$

where T is the restraining torque, F is the force in a specific joint structure (which varies with tissue elasticity), and r is the distance between the point of application of the force and the torque axis.

The reduction in anterior translation due to anterior loading that occurs with fixed internal or external rotation is an example of the lateral and medial collateral ligaments contributing more to joint restraint as they became taught (Amis *et al.*, 2005). In the extended position, in both internal and external rotational loading conditions, our results showed that the net axis of restraint was located in approximately the same position as that of the applied torque axis. Therefore, the combined force-distance contribution of all structures that provided stability were balanced. In extension, there is an increase in tension of all ligamentous structures posterior to the femoral epicondyles, including the posteromedial capsule, the posterior capsule, and the arcuate ligament complex Amis *et al.* (2005). Similar to the reduction in anterior translation noted by Amis *et al.* (2005), this tightening of collateral and posterior structures may have resulted in a stress-shielding effect of the ACL in the extended position (Amis *et al.*, 2005; Csintalan *et al.*, 2006; Nordt *et al.*, 1999).

In the flexed position, conversely, the extra-articular ligaments relax (Amis *et al.*, 2005), resulting in smaller force contributions to the overall joint restraint. Lateral capsular laxity, combined with a more convex lateral tibial plateau, results in a more mobile lateral tibial compartment (Amis *et al.*, 2005; Nordt *et al.*, 1999); this likely led to the increase in internal tibial rotation in the flexed position while the degree of external rotation remained about the same in extended and flexed positions. Although the ACL is protected by the MCL under external tibial torsional loading, in internal rotation it plays a greater role in overall joint restraint (Amis *et al.*, 2005; Csintalan *et al.*, 2006; Harfe *et al.*, 1998; Meyer & Haut, 2008; Nordt *et al.*, 1999). This is due to its oblique orientation relative to the axis of rotation (Blankevoort & Huiskes, 1996); as the tibia rotates internally, the distance between ligament insertions increases, with further tensioning of the ligament taking place as it twists around the PCL.

Given the proximity of the ACL insertion to the actual axis of rotation when compared to the positions of the collateral and posterior ligaments, we would expect the overall contribution of the ACL to rotational restraint to be minimal in both extended and flexed knee positions.

A general trend of increasing knee flexion was observed when subjected to external versus internal torque (Figure 4.2); these connected motions are contrary to the normally observed coupling of internal rotation with flexion in the sagittal plane. This may be explained by the difference in loading conditions in this study, in which pure torsional loading without any additional constraints would cause a simple sliding motion of the femoral condyles on the tibial plateau. However, rotational restraint of the knee is provided by the contact surfaces of the joint in addition to the ligaments of the knee (Blankevoort & Huiskes, 1996). The more concave shape of the medial plateau, together with the increased stiffness of the medial meniscus, may have limited the sliding of the medial condyle and forced the condyle to roll in order to accommodate the applied torque; roll-back in external rotation is converted into an increased flexion angle, whereas roll-forward in internal rotation becomes a decreased flexion angle. Furthermore, the heightened strain on the medial collateral ligament in external rotation may have been offset by an increase in flexion angle between 0° and 30° of flexion. The more convex shape of the lateral tibial plateau and the general increased laxity of this side of the joint may have permitted the sliding motion that resulted in the observed net flexion in external rotation and net extension in internal rotation. In order to confirm this theory, the tibiofemoral contact points and positions of the menisci should be measured under these loading conditions, which was beyond the scope of this study.

In conclusion, the 3D knee kinematics measured under a normalized torque showed a large variation in transverse plane rotation, the primary motion resulting from torsional loading, in a group of healthy individuals. Our rotation data agreed well with that of the literature; a mean increase in range of rotation of about 10° was measured from full extension to 30° of flexion, which could be attributed to an increase in the internal direction. However, significant left-to-right differences in external and internal loading conditions confirm that caution should be taken when comparing knee rotations to their contralateral limb. Coupled anterior-posterior translation with internal-external rotation revealed that the effective axis of rotation is located near the centre of the tibial plateau, lateral to the tibial rotation axis in flexion-extension. It is important to bear in mind that these results are for specific passive loading conditions *in vivo*. While all reasonable measures were taken to avoid active stabilisation strategies used by the subjects, the possibility of muscle recruitment cannot be entirely ruled out.

Chapter 5

Passive rotational laxity of the ACL-deficient and reconstructed knee: Single vs double-bundle surgery

5.1 Introduction

The anterior cruciate ligament (ACL), in addition to its primary role restraining anterior tibial translation, has been shown to contribute to rotational constraint of the knee. Conventional surgical techniques adequately limit anterior-posterior (AP) laxity; however, subjective 'giving-way' symptoms and positive pivot shift reveal that rotational instability often remains. In order to improve rotational laxity outcome, surgical techniques have been modified to reconstruct not just the anteromedial (AM), but also the posterolateral (PL) bundle of the ACL. Although the single-bundle (SB) technique has been shown to improve knee restraint with respect to the injured knee, several biomechanical studies have shown significant reductions in knee laxity and superior functional outcome under anterior and pivot shift loading, when comparing the outcome of the double-bundle (DB) to the SB reconstruction (Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Siebold *et al.*, 2008; Yagi *et al.*, 2002, 2007; Yasuda *et al.*, 2006; Zantop *et al.*, 2006).

At the time of inception of this study in 2006, however, little *in vivo* clinical evidence was available comparing the outcome of the SB and DB techniques. In fact, a meta-analysis published in 2008 found only four randomised control trials (Level I evidence) and an additional five prospective and retrospective comparative studies (Levels II and III) to assess differences in outcome of SB and DB reconstructions; their findings showed that there were no clinically significant differences in KT-1000 arthrometer and pivot shift results between surgical techniques (Meredick *et al.*, 2008). Other reviews have also identified this lack of *in vivo* clinical evidence to support the more complicated DB technique (Amis *et al.*, 2005; Crawford *et al.*, 2007; Lewis *et al.*, 2008; Longo *et al.*, 2008; Steckel *et al.*, 2007b).

Since 2006, the publication of clinical trials investigating SB and DB outcome has accumulated substantially; although much of the evidence supports the use of the DB technique (Järvelä, 2007; Kondo *et al.*, 2008; Muneta *et al.*, 2007; Seon *et al.*, 2007; Siebold *et al.*, 2008; Yasuda *et al.*, 2006), some trials have not found significant differences between clinical outcomes in the patient groups (Asagumo *et al.*, 2007; Streich *et al.*, 2008). Furthermore, there is a proliferation of research describing improvements in both SB and DB surgical technique that also reduce rotational laxity. These include adjusting tunnel placements to accommodate a more horizontal graft and modifying initial graft tensions and specific knee angles at which tensioning occurs (Jepsen *et al.*, 2007; Kondo & Yasuda, 2007; Loh *et al.*, 2003; Markolf *et al.*, 2009; Musahl *et al.*, 2005; Scopp *et al.*, 2004; Yamamoto *et al.*, 2004; Yasuda *et al.*, 2008; Zaffagnini *et al.*, 2008).

The problem with much of the clinical evidence in the literature is that assessments use either a quantitative outcome that does not measure rotation (e.g. AP instability assessed with an arthrometer) or a subjective test that provides only a gross clinical measure of laxity (e.g. pivot shift). Moreover, neither of these tests is able to establish the role of the ACL specifically in rotational restraint, since it is not an isolated torque that is applied. (The pivot shift is a combined internal and valgus torsional load.) The differences in measured outcome could, therefore, be attributed to the contribution of the ACL to anterior or valgus restraint, rather than rotational restraint. Those studies that have examined the effect of ACL surgical technique, graft tension, or tunnel placement on a quantifiable measure of rotational laxity by applying an isolated torque, have done so without measuring the magnitude of the applied load, making comparisons within and between studies difficult due to this subjective component of the study. As no *in vivo* post-operative studies could be found, these also could not be reasonably compared with the existing clinical evidence.

The purpose of this study was, therefore, to determine differences in rotational laxity outcome in SB and DB reconstructions under known isolated torsional loading. The study was designed as a prospective double-blinded randomised control trial (Level I evidence).

5.2 Methods

5.2.1 Participants and interventions

Thirty-two subjects were recruited for this trial from the patient list of the Sports Science Orthopaedic Clinic in Cape Town between November 2006 and March 2008. Testing was generally completed outside of regular clinic hours (i.e. evenings and weekends). A transportation allowance was provided for those patients who did not have their own means of transport. Eligibility criteria included the following:

- Age: 18 - 49 years.
- Injury: complete isolated ACL rupture with minimal injury to other structures (e.g. patients with concomitant meniscal, medial, or lateral structure injury were excluded).
- Previous lower limb pathology: no previous injuries to either affected or contralateral limb.
- Clinical status: ability to walk with no or negligible pain.

Knee laxity tests were performed by a trained investigator prior to and following ACL surgery. Time between surgery and post-operative testing ranged between 2.5 and 9 months (mean 5.2 months) with the final testing completed in July 2008.

Patients underwent one of two surgical procedures to reconstruct the ACL: a single-bundle or a double-bundle reconstruction. All procedures were performed at the Vincent Pallotti Hospital in Cape Town by the same orthopaedic surgeon (Dr. Willem van der Merwe) who had over ten years' experience with both surgical techniques.

The surgical procedure for the double-bundle technique is described in detail by Bellier *et al.* (2004) and a brief description of each procedure is given here. In both procedures, the semitendinosus and gracilis tendons were harvested from the affected limb through the anteromedial incision. Standard arthroscopic evaluation and site preparation were conducted to enable a clear visualization of the anatomical femoral and tibial footprints. At the end of each procedure, the knee was put through a range of motion to confirm an absence of graft impingement and to ensure stability of the graft.

5.2.1.1 Single-bundle surgical procedure

Both semitendinosus and gracilis tendons were folded in half to produce a four-stranded graft. With the knee flexed to 120°, a guidewire was placed at the 10:30 o'clock position (1:30 for the left knee) and a single 7-10 mm femoral tunnel was drilled at the midpoint of the AM and PL attachments. Next, the knee was flexed to 30° and a single 7-10 mm tunnel was drilled though the proximal end of the tibia. In each case, the diameter of the prepared graft was measured, and the tunnel was drilled accordingly. The graft was passed through the tibial and femoral tunnels and an Endobutton (Smith & Nephew Inc) was used for femoral side graft fixation. Once the graft was tensioned to approximately 50 N with the knee flexed to 90°, the graft was secured at the tibial side using a bioabsorbable interference screw (Smith & Nephew Inc).

5.2.1.2 Double-bundle surgical procedure

The double-bundle graft technique used the folded semitendinosus to produce the AM bundle, while the PL bundle was fashioned from the doubled gracilis tendon.

After they were each passed through an Endobutton (Smith & Nephew Inc), the two ends of each graft bundle were sutured together separately. The knee was flexed to 120° to drill the first of the femoral tunnels for the AM bundle with the guidewire placed at the 11 o'clock position (1 o'clock for the left knee). The PL bundle tunnel was drilled next with the guidewire at the 9:30 o'clock position (2:30 for the left knee). The AM tunnel diameter was 6-8 mm, while the PL bundle was slightly smaller at 5-7 mm.

The two tibial tunnels were then created, beginning with the PL tunnel. The anterolateral tibial spine was used as a guide for this tunnel, while the AM tunnel was located between the two tibial spines and anterior to the PL tunnel. Again, the AM tunnel was drilled between 6-8 mm and the PL tunnel was only 5-7 mm in diameter. The PL bundle (gracilis) and AM bundle (semitendinosus) grafts were then passed through the tunnels. Bioabsorbable interference screws were used for fixation of both graft bundles when tensioning to approximately 50 N had been achieved. The PL bundle was fixed first with the knee flexed to 15°. The knee was then flexed to 90° for securing the AM bundle.

5.2.1.3 Testing protocol

Each patient who met the inclusion criteria as determined through a physical exam and MRI scan performed by Dr. van der Merwe was given details of and asked to participate in the study. Those who agreed, signed an informed consent document approved by the Human Ethics Committee of the University of Cape Town (Appendix D).

Details of the data collection and analysis methods are given in Chapter 3. Subjects were tested up to four weeks before their surgeries with scans of both injured and contralateral limbs taken at that time. In some circumstances, it was not possible test both knees pre-operatively, so the contralateral knee was scanned at the time of post-operative testing. Low resolution T1-weighted transverse plane MR images were taken while normalized internal and external torsional loads were applied to the knee in full extension and at 30° of flexion. A high resolution image in a neutral (no-load) position was recorded for shape-matching purposes during data analysis. Complete six degree-of-freedom tibiofemoral kinematics were calculated for the contralateral knee, as well as for the injured knee pre- and post-operatively in the four loading conditions: internal and external torque in extended and flexed positions. All MR imaging was completed at the Sports Science Orthopaedic Clinic; image processing and Matlab^{TM} analysis were conducted off-site on a separate laptop computer.

5.2.2 Objectives and outcome

The primary objective of this study was to compare the magnitude of change in rotational laxity pre- to post-operatively in the single and double-bundle ACL reconstructions with the knee positioned in full extension and at 30° of flexion. Secondary objectives were to determine whether there were differences in laxity in the ACL deficient and reconstructed knee with respect to subjects' contralateral knees. The hypothesis that the mean rotational laxity of the patients' contralateral knees was not different to that of a group of healthy age- and gender-matched control subjects (whose data were presented in Chapter 4) was furthermore tested.

5.2.3 Randomisation and blinding

A random allocation sequence was generated using Matlab ^{TM} to ensure equal numbers in single and double-bundle groups for the first 30 subjects and blocks of 10 subjects thereafter. Participants were enrolled by Dr. van der Merwe; intervention group was assigned at time of surgery by Dr. van der Merwe's administrative assistant who kept the random allocation list. The participants and primary investigator conducting data collection and analysis (AH) were blinded to group assignment; however, the intervention group could be discerned from the post-operative MRI scans during image segmentation.

5.2.4 Statistical analysis

A linear mixed model for repeated measures was applied to detect intervention group differences pre- to post-operatively using SPSS 15.0 (SPSS Inc). Post-hoc analysis was conducted with a two-tailed paired samples t-test. Secondary outcomes – specifically comparisons of contralateral with control group, contralateral with ACL-deficient group, and contralateral with ACL-reconstructed group knee laxity – were also measured using the linear mixed model for repeated measures. Differences with p-values less than or equal to 0.05 were considered to be statistically significant.

5.3 Results

Baseline demographic and clinical subject data are presented in Table 5.1. Of the 32 participants enrolled in this study, three patients allocated to the double-bundle intervention were lost to follow-up (Figure 5.1). Of those, two subjects had completed testing of the contralateral limb during the pre-operative session. An additional two subjects (one in each intervention group) did not have testing completed on the contralateral limb due to patients' personal constraints and consequent withdrawal from the study. The linear mixed model permitted the use of all available data in each analysis; the number of subjects included in each analysis was therefore dependent on the knee, test time (pre- or post-operative), and loading condition examined.

Table 5.1: Baseline demographic and clinical subject data (mean ± SD) for control and patient groups. ACL all includes subjects from both single-bundle (SB) and double-bundle (DB) groups.

Variable	**Control**	**ACL all**	**SB**	**DB**
Sex (F:M)	4:11	8:24	7:10	1:14
Age (yrs)	30.3 ±5.9	30.2 ±6.2	31.5 ± 5.7	26.8 ± 6.0
Height (cm)	174.5 ±9.3	174.5 ±8.8	171.5 ± 6.8	177.9 ± 9.7
Mass (kg)	71.7 ±11.3	79.3 ±14.5	76.3 ± 13.8	82.7 ± 14.9
Applied Torque (Nm)	4.8 ±0.6	5.2 ±0.7	5.1 ± 0.7	5.4 ± 0.7
Time Injury-PreOp (mos)	n/a	5.7 ±8.9	5.5 ± 11.3	5.9 ± 5.5
Time Surgery-PostOp (mos)	n/a	5.2 ±2.0	4.6 ± 1.8	6.1 ± 1.8

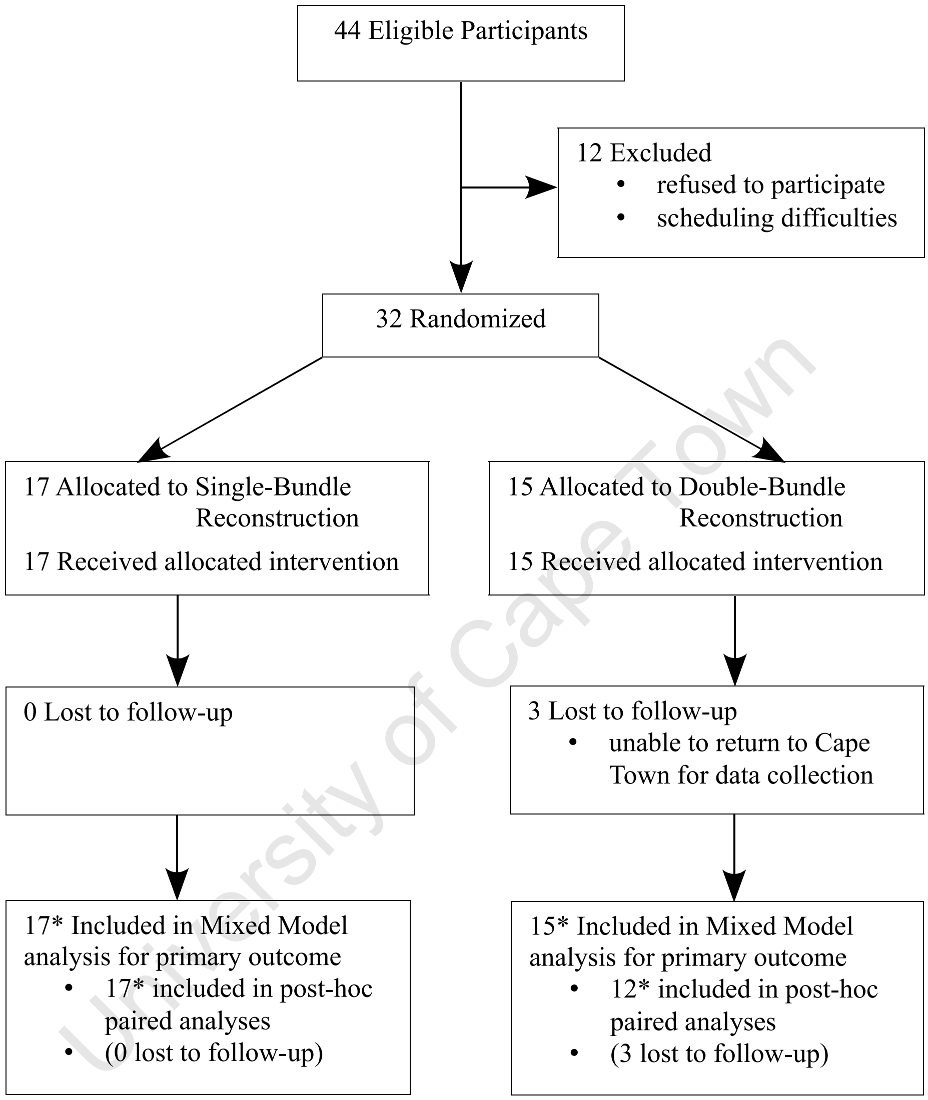


Figure 5.1: Flow diagram of participants through each stage of the randomised control trial for the primary outcome comparing single and double-bundle ACL reconstruction. * Some data were excluded from analyses if adequate torque could not be applied. (See explanation in section 5.3.1.)

5.3.1 Protocol deviations

While all subjects were able to tolerate the pre-calculated normalized torque applied to their contralateral knee, several experienced discomfort with the load applied to the injured knee. In those cases, the knee was scanned under the maximum tolerated load with this value recorded for the specific condition. Since the mixed model is able to evaluate unmatched data, it was important to include all valid data in the 'intention to treat' analysis. However, it has been shown that rotation is directly related to torsional load (Almquist *et al.*, 2002). It was therefore necessary to determine the percentage of the total torque at which the results would be affected by the smaller load. A Kruskal-Wallis one-way ANOVA was used to compare the mean rotation in each of the four loading conditions with the data divided into 3 groups according to the level of torque achieved: Group I = 100% torque, Group II = 75-99% torque, and Group III = 50-74% torque.

Table 5.2: Differences in measured mean rotation according to group. Group I = 100% torque, Group II = 75-99% torque, Group III = 50-74% torque. * post-hoc analysis indicates significant difference between Groups I and III.

Loading	Group I		Group II		Group III		Level of
Condition	N	mean	N	mean	N	mean	Significance
Extended Ext T	84	9.5	2	10.4	2	3.5	0.098*
Extended Int T	84	-7.4	3	-4.0	3	-3.3	0.027*
Flexed Ext T	82	9.6	3	10.3	4	7.0	0.300
Flexed Int T	85	-13.7	3	-10.3	1	-11.5	0.273

Results in Table 5.2 show that significant differences were detected in the extended position with internal torsion and significance was approached in the extended position with external torsion. A Mann-Whitney post-hoc analysis showed significant differences between Groups I and III in each of these loading conditions.

Since no significant differences in measured rotation were detected between Groups I and II, it was decided to exclude only those data in which the achieved level of torque was less than 75% of the pre-calculated normalized value. This was then used as a standard for all subjects across all loading conditions. In total, 13 of the possible 68 data sets (19%) were excluded from the SB group and 5 of the possible 60 data sets (8%) were excluded from the DB group.

5.3.2 Adverse events

Only two adverse reactions were reported at follow-up. One subject complained of swelling and instability of the knee at the time of follow-up, $3\frac{1}{2}$ months post-operatively and could not tolerate the applied torque in all loading conditions. Another subject developed more serious pre-tibial soft tissue swelling due to problems with the Calaxo bioabsorbable screw implant (Smith & Nephew); this product was subsequently recalled. The patient required local debridement and removal of the remaining screw fragments; however, graft-to-bone healing had already occurred by this point. Following an additional 3 months of recovery, this subject agreed to return for post-operative testing (a total of 7 months following the original surgery). Both of these subjects were in the single-bundle reconstruction group.

5.3.3 Outcomes

The only significant interaction between SB and DB surgical techniques when comparing the pre- to post-operative results was in the flexed internally torqued loading condition in which the DB group demonstrated a reduction in transverse plane rotation following ACL reconstruction (Figure 5.2 and Table F.1). No significant differences were found between single and double-bundle groups, however.

In general, ACL reconstruction was shown to reduce rotational laxity in the extended position under internal torsional loading, restoring rotation to that of the contralateral knee (Figure 5.2, Figure 5.3 and Table F.2). No surgical group interaction was observed, however (Table F.1); i.e. this difference was not dependent on the type of reconstruction (SB or DB).

No differences in rotational laxity were found between the left-right averaged knees of the control group and the contralateral knees of the patients in any of the four loading conditions (Figure 5.3 and Table F.2).

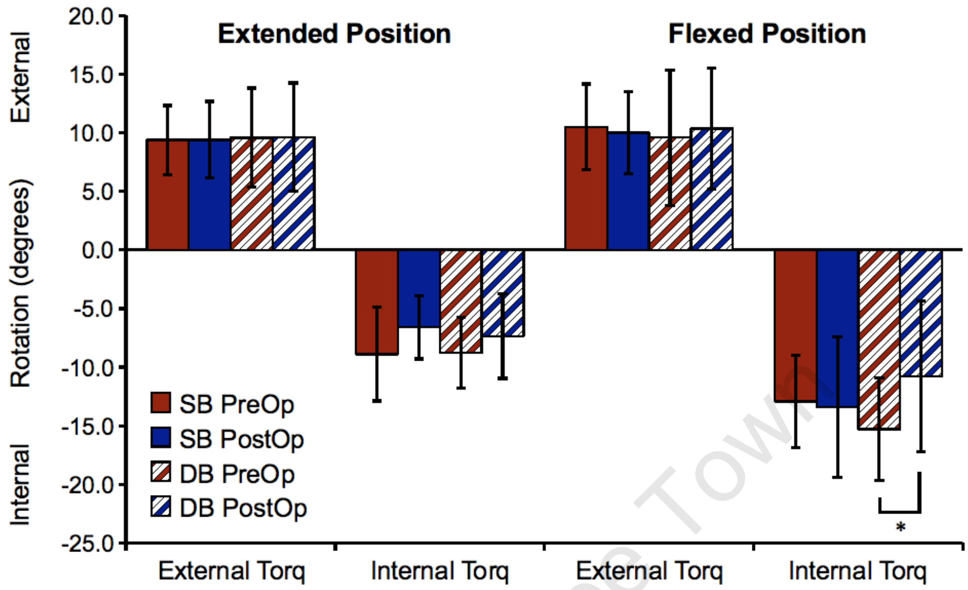


Figure 5.2: Mean internal and external rotation measured pre- and post-operatively in the single and double-bundle groups in the four loading conditions. * indicates significant difference.

5.4 Discussion

In this study, single and double-bundle surgical techniques were compared to determine differences in rotational laxity in patients with isolated rupture of the ACL before and after reconstructive surgery. The findings showed that in only the flexed knee position under internal torsional loading did the DB reconstruction reduce rotational laxity more than the SB technique (with this reduction being statistically significant); however, when compared with rotation of the contralateral knee which demonstrated a mean laxity similar to that of the injured knee, this may have resulted in excessive restraint. Although significant differences were not found between SB, DB, or contralateral knee groups in the flexed, internally torqued knee condition, the mean rotation in the reconstructed SB knee more closely matched that of the contralateral uninjured knee than did the mean DB knee rotation.

Our findings also demonstrated a significant increase in internal rotation of the injured knee with respect to the contralateral and reconstructed knees in

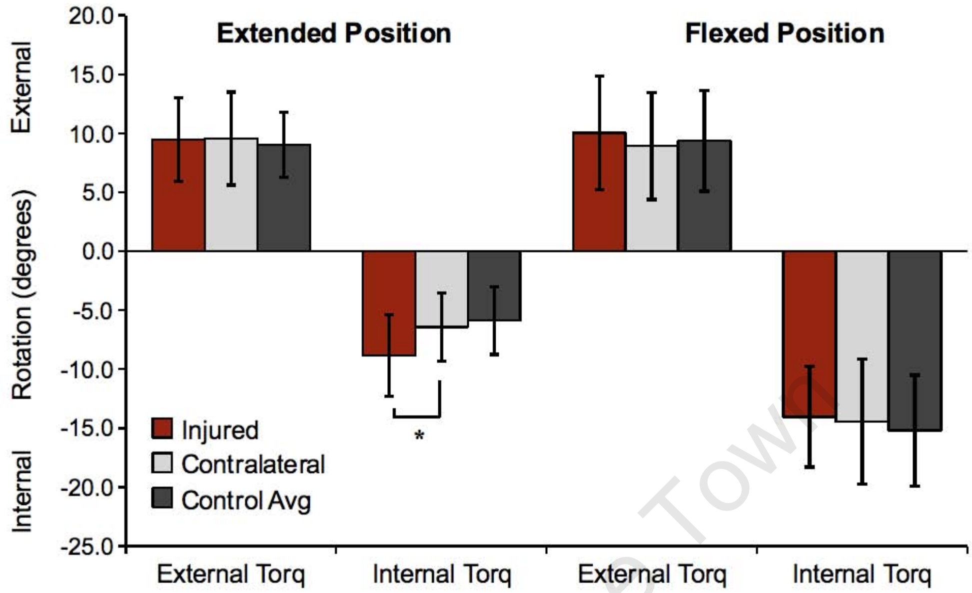


Figure 5.3: Mean internal and external rotation measured in Control (averaged from left and right knee data), Patient contralateral, and Patient injured (pre-operative) knee groups in the four loading conditions. *Post-hoc analysis revealed significant difference. (Note: Injured knee data was not compared to Control Avg data.)

the extended position. In other words, in this loading condition, the reconstruction of the ACL returned the knee kinematics to normal; however, there was no significant effect of surgical technique.

Since no significant differences were found between the contralateral knees of the patients and a healthy age- and gender-matched control group, it was valid to use the data from patients' contralateral knees as reliable controls (Kozanek *et al.*, 2008). Furthermore, demonstrating statistically equivalent results between healthy controls and patients' contralateral knees indicates that there was no preexisting laxity and that the contralateral knees were not affected by ACL injury in this group of patients (Kozanek *et al.*, 2008).

Our study supports the evidence that the ACL contributes to rotational restraint under internal torsional loading, but that it is not the primary restraint to rotational loading. Rotational forces are first constrained by the extraarticular ligaments, which have a mechanical advantage in rotation and thereby shield the ACL from stress under torsional loading (Amis *et al.*, 2005; Csintalan *et al.*, 2006;

Nordt *et al.*, 1999). The position of the ACL insertion and its resulting orientation allows it to provide restraint in internal rotation: the distance between the anteromedial position of the tibial insertion and the posterior position on the medial side of the lateral femoral condyle increases under internal rotation, thereby increasing the tension of the ligament and subsequent restraint of the joint (Amis *et al.*, 2005; Blankevoort & Huiskes, 1996). No differences in rotation were observed in the injured, reconstructed, or contralateral knees under external torsional loading, while differences *were* observed with internal torque, verifying its effect in only the one direction of rotation.

An increased rotational laxity of the ACL-deficient knee with respect to the contralateral knee was observed in only the extended position, while no difference in laxity was observed in the injured knee at 30° of flexion. This distinction between extension and flexion may be attributed to the general laxity of the ACL and other major rotational restraints in these knee positions. It has been shown that with no externally applied load, the tension of the ACL is greater in the extended position than at 30° of flexion (Amis & Dawkins, 1991; Blankevoort *et al.*, 1991; Markolf *et al.*, 2008a; O'Connor & Zavatsky, 1993). The recruitment patterns of various ligaments (and their respective bundles) were illustrated as functions of flexion by Blankevoort *et al.* (1991); with no external loading, the posterolateral bundle of the ACL exhibited near maximum strain with the knee in full extension.

Knowing that the distance between ACL insertions increases with internal rotation, we can deduce that the tension under an additional internal torque would only increase. This hypothesis is supported by Markolf *et al.* (2009) who showed that the *in situ* ACL force increased in both full extension and 30° of flexion with the addition of a 5 Nm load (with the force magnitude greater in the extended than in the flexed position).

The collateral ligament forces that would have permitted a prediction of the relative contribution of the various ligaments under these loading conditions were not presented in the study by Markolf *et al.* (2009). However, Blankevoort *et al.* (1991) assumed the recruitment of the ACL to be less than that of the collateral ligaments in full extension and to decrease even further throughout the first 30° of flexion. This assumption was based on the relative length changes of the cruciate and collateral ligaments through 90° of flexion. While the length (and so the tension) of the ACL decreased, the overall lengths of the collateral ligaments generally decreased to a lesser extent between 0° and 30° of flexion. Even with the addition of a 3 Nm internal torque at 20° of flexion, the overall recruitment of the ACL was assumed to be minimal when compared to the MCL (Blankevoort *et al.*, 1991). (No data for recruitment patterns were presented at 0° and 30° of flexion under torsional loading conditions.)

Amis & Dawkins (1991) also demonstrated that, despite an increase in length of the ACL under 1 Nm of applied torque when compared to the no-load condition, its length under a fixed torsional load still decreased (i.e. the ligament tension decreased) between 0° and 30° of flexion. It is therefore reasonable to assume that the overall contribution of the ACL to rotational restraint is less at 30° of flexion than in the fully extended position.

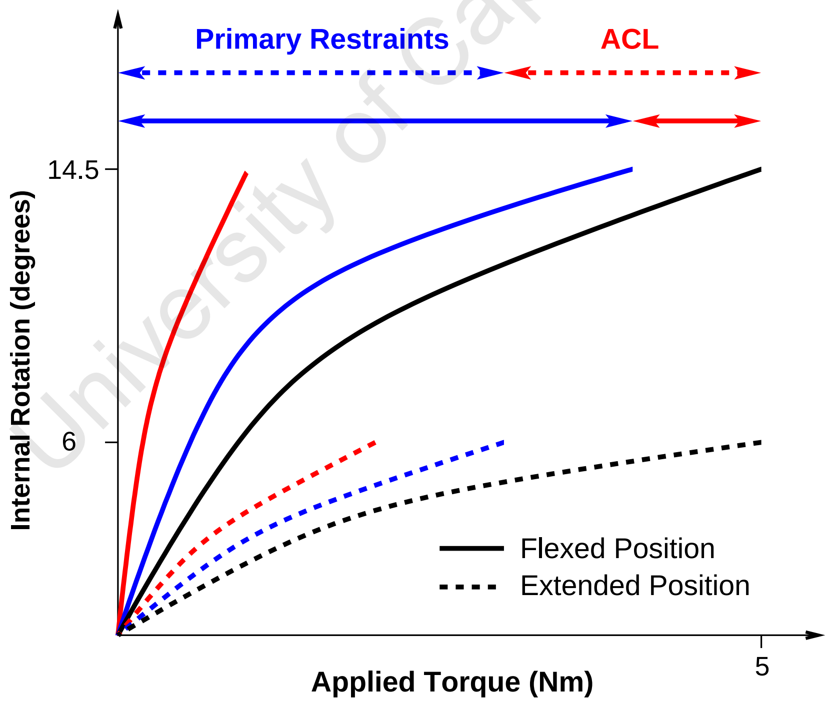


Figure 5.4: Contribution of the primary restraints and ACL to internal rotational restraint in the extended and flexed positions. (Black lines indicate the combined contribution of the primary restraints with the intact ACL.)

This concept is illustrated in Figure 5.4 which shows the proposed recruitment of joint structures based on a typical ligament load-deformation curve during physiological loading conditions (Musahl *et al.*, 2007). In the extended position under internal torsional loading, the primary restraints make the major contribution to the overall restraint required to resist the applied torque; although, the ACL also provides substantial constraint. In the flexed position, all ligaments tend to relax, increasing the slopes of the torque-rotation curves; however, the ACL slackens to a greater extent than the primary restraints (Blankevoort *et al.*, 1991). Its contribution to the overall joint restraint is, therefore smaller at 30° of flexion than in the extended position. Consequently, the rotation resulting from an applied torque in the ACL-deficient knee at 30° of flexion is the same as that of the contralateral knee, while in the fully extended position, there is an increase in rotation in the injured knee (Figure 5.5 and Figure 5.3).

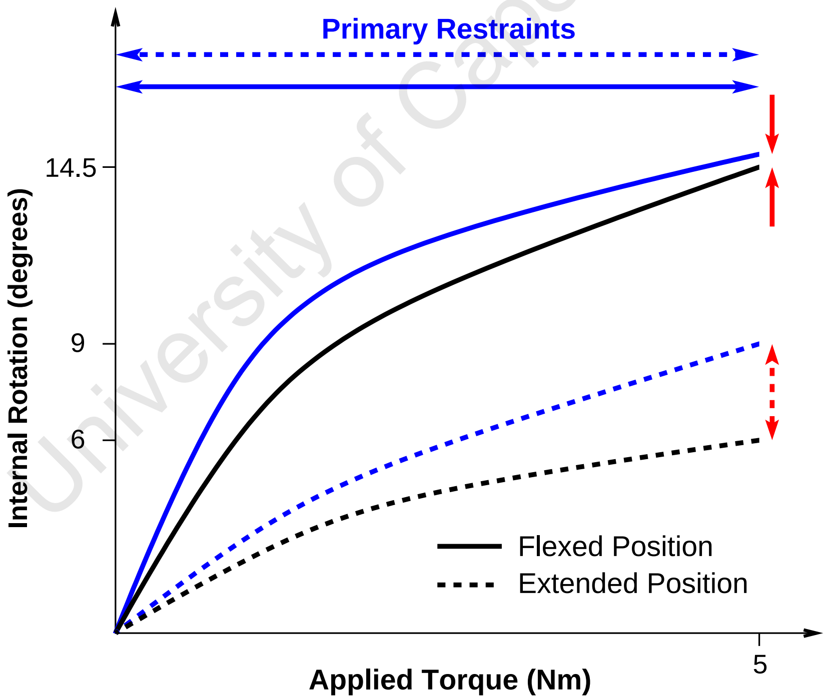


Figure 5.5: In the ACL-deficient knee, the primary restraints restrict the entire load. The change in internal rotation from the ACL-intact to ACL-deficient knee is indicated by the red arrows. (Black lines indicate the combined contribution of the primary restraints with the intact ACL.)

The specific surgical technique, i.e. single or double-bundle, affected rotational restraint in only one loading condition: internal torsion at 30° of flexion. The DB technique limited rotation with respect to the injured knee; however, since the injured knee laxity was actually closer to that of the contralateral knee, this may be considered an overconstraint of rotation. Extending the previous theory to the results obtained in the flexed position, the excessive restraint provided by the DB graft induces less internal rotation with the same applied torque (Figure 5.6).

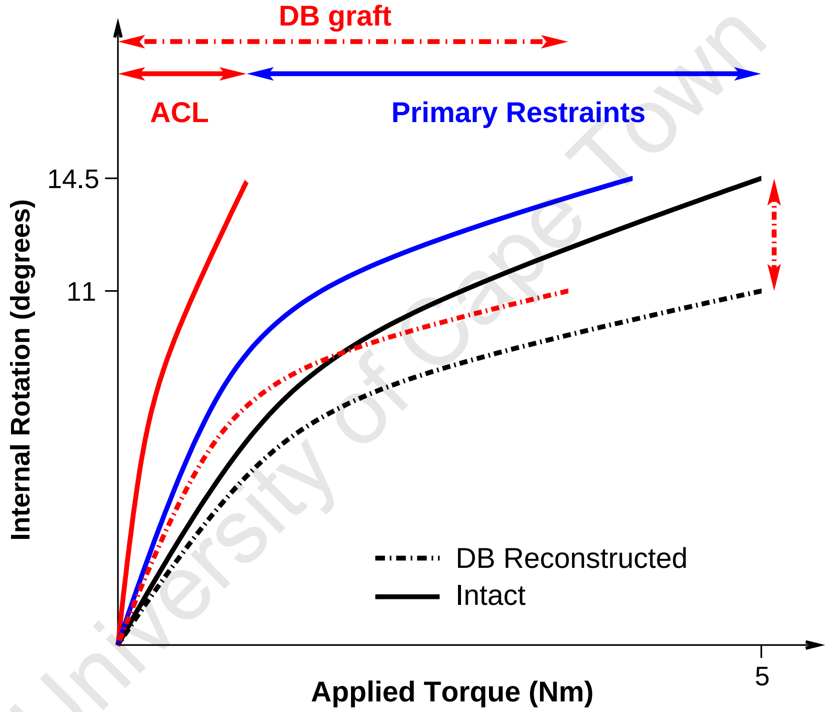


Figure 5.6: In the DB-reconstructed knee, the DB graft restricts more load than the native ACL. The change in internal rotation from the ACL-intact to DB-reconstructed knee is indicated. (Black lines indicate the combined contribution of the primary restraints with the intact or reconstructed ACL.)

This finding was substantiated by several cadaveric studies that examined both rotational laxity and graft tension under torsional loading (Markolf *et al.*, 2008b, 2009; Steckel *et al.*, 2007a). Steckel *et al.* (2007a) found that the DB technique overcorrected internal-external rotation with respect to the intact knee at 15°, 60°, 75°, and 90° of flexion, while rotation following SB reconstruction was not significantly different from the intact knee at all angles except 60° of flexion (where it also overcorrected knee laxity). In a study that compared rotation and graft tension under 5 Nm of internal torque in the ACL intact, sectioned, SB, and DB-reconstructed knee (using four different graft-tensioning protocols), Markolf *et al.* (2009) found that two of the DB techniques overconstrained rotation at higher flexion angles, specifically 50° to 120° of flexion while no differences were found with the SB technique. The resultant force in the posterolateral graft was markedly higher than both the SB graft and the intact ACL, although, similar to our results, no significant differences in rotation were shown between surgical techniques in full extension. In another study conducted by the same group, significant decreases in rotation resulted with the clinical pivot shift test in three of the four DB techniques when compared to the intact knee, while the SB technique restored rotational stability to normal (Markolf *et al.*, 2008b).

Whereas the DB technique reduced internal tibial rotation in the flexed position to a greater extent than the SB technique when comparing pre- and post-operative laxity, no significant differences in knee rotation were found between the two techniques (Figure 5.2). Results in the literature comparing the two techniques vary: while some studies have demonstrated differences in joint laxity between the SB and DB reconstruction (Adachi *et al.*, 2004; Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Lopomo *et al.*, 2009; Markolf *et al.*, 2009; Petersen *et al.*, 2006; Seon *et al.*, 2007; Steckel *et al.*, 2007a; Yagi *et al.*, 2002, 2007; Yamamoto *et al.*, 2004; Yasuda *et al.*, 2006; Zantop *et al.*, 2006), others have shown similar results with both techniques (Ferretti *et al.*, 2008; Steckel *et al.*, 2007a; Streich *et al.*, 2008). Of those that have shown an improvement in rotational restraint with respect to the uninjured knee using the DB technique (rather than an overcorrection), the majority have done so under anterior or pivot shift (i.e. combined valgus and torsional) loading (Adachi *et al.*, 2004; Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Lopomo *et al.*, 2009; Petersen *et al.*, 2006; Steckel *et al.*, 2007a; Yagi *et al.*, 2002, 2007; Yasuda *et al.*, 2006; Zantop *et al.*, 2006).

It is not possible to directly compare our results with these studies since the loading conditions differ from our study in which laxity was examined under an isolated tibial torque. It has been well-established that the ACL is the primary restraint to anterior loading; therefore, an anterior force of 134 N (typically used in the aforementioned studies) would recruit the ACL to a greater extent than torsional loading conditions in which the primary restraints (such as the collateral ligaments or menisci) shield the ACL. Similarly in the pivot shift test, the addition of a valgus moment to a torsional load has been shown to significantly increase ACL forces and strain (Kanamori *et al.*, 2000; Shin *et al.*, 2005), thereby likely recruiting the ACL to a greater extent than with a simple isolated torque.

By increasing the tension of the ACL its contribution to the torsional restraint of the joint will theoretically also increase according to equation 4.2, since the torque provided by the other structures would either decrease to constrain the same overall load or the total restraint provided by the joint structures would increase, thereby decreasing the resulting rotation. This is demonstrated by the following equations (the first of which was derived from equation 4.2):

$$\Sigma T_{restraint} = (F_{ACL} \times r_{ACL}) + \Sigma (F_{Si} \times r_{Si}) \tag{5.1}$$

where T is the restraining torque, F is the force of a specific joint structure, r is the distance between the point of application of the force and the torque axis, and the subscripts refer to the ACL or specific joint structure Si contributing to rotational restraint. As the force of the ACL increases, the forces of the other contributing structures decrease to provide the equivalent overall torque as long as the radius of rotation remains the same. (If the location of the axis of rotation changes, this distance could change.)

This theory is supported by the computational study conducted by Suggs *et al.* (2003) in which an increase in simulated graft tension decreased rotation resulting from an anterior load. With the ACL providing a greater contribution to overall joint laxity in the anterior and pivot shift loading conditions, it is reasonable that differences between SB and DB grafts could be detected.

Despite the possibility of greater overall contribution of the ACL to rotational restraint, two *in vivo* studies demonstrated no differences between graft types with anterior or pivot shift loading (Ferretti *et al.*, 2008; Streich *et al.*, 2008). Ferretti *et al.* (2008) measured anterior and rotational laxity at 30° of flexion under isolated maximum anterior and torsional loading intra-operatively in 20 patients who received either the SB or the DB reconstruction. By applying a subjective measure of manual maximum force, it is possible that the variation in applied load may have resulted in high standard deviations in measured rotation making it difficult to compare the surgical techniques. Since our protocol normalized load according to subject mass, objective comparability across subjects was established and it is more likely that standard deviations of our rotation results were reflective of actual individual subject variation.

Similar to Ferretti *et al.* (2008) in a prospective randomised control trial, Streich *et al.* (2008) found no significant differences in anterior or pivot shift laxity between SB and DB groups. They attributed this inconsistency with the literature to the subjective (and, possibly inaccurate) assessment of the pivot shift and to the placement of the femoral tunnel which permitted a more horizontal orientation of the SB graft.

In fact, several studies have shown that changing tunnel placement will affect rotational laxity in both SB and DB techniques; in general, a more anatomical tunnel placement which allows grafts to attain a more horizontal rather than vertical orientation in the joint, has been shown to improve rotational constraint (Musahl *et al.*, 2005; Scopp *et al.*, 2004; Yamamoto *et al.*, 2004; Zantop *et al.*, 2008). In our study, the tunnels for the SB graft were positioned midway between the native AM and PL bundle insertions, in other words in the SB anatomical position, which may have resulted in similar rotational laxity to the DB technique. Both postoperative graft quality (defined by its thickness and apparent tension) and tensioning during initial fixation of the graft have also been shown to affect joint laxity (Kondo & Yasuda, 2007; Suggs *et al.*, 2003). With similar tensioning protocols used for both surgical techniques and a relatively brief follow-up period of only 5 months in which minimal graft relaxation may have occurred, it is foreseeable that both of our patient groups would have similar graft tension at follow-up, resulting in similar joint laxity.

Our results do not support findings from some studies in which rotational laxity examined at 30° of flexion under isolated torsional loads in cadavers and intraoperatively showed significant differences in rotation between the ACL-reconstructed with the intact or deficient knee. These studies all applied higher loads (6.5 Nm, 10 Nm and maximum manual force) than those used in our *in vivo* study (Kanamori *et al.*, 2000; Martelli *et al.*, 2007; Monaco *et al.*, 2007; Scopp *et al.*, 2004). Furthermore, our study normalized the applied torque to individual subject mass, ensuring that the correlation between the amount of torque and measured rotation would not affect the inter-subject comparison (Almquist *et al.*, 2002).

With a smaller magnitude of only 5 Nm of internal torque, Markolf *et al.* (2008a) showed that cutting the posterolateral bundle did not affect knee rotation. (The effect of cutting both bundles was not examined in that study.) Diermann *et al.* (2008) also found no significant differences of internal rotation in ACL-deficient, intact, and reconstructed knee when a combined 10 Nm valgus and 4 Nm rotational load was applied. Conceivably, in these studies as with ours, in which the applied load was comparatively smaller, the primary restraints were able to control rotation and the torsional load did not stress the intact or reconstructed ACL enough to warrant its contribution to overall joint restraint. In addition, in one of the intraoperative studies that displayed differences between injured and reconstructed knees, 16 of the 30 subjects presented with associated injuries to collateral ligaments (Martelli *et al.*, 2007). In these patients, the capacity of the primary restraints may have been exceeded, with the ACL consequently providing secondary support under high torque conditions, thereby demonstrating rotational differences in ACL-deficient, reconstructed, and intact knees.

To ensure that the uneven distribution of male and female patients between groups (with seven of the eight females allocated an SB reconstruction) did not affect the laxity outcome, further statistical analyses were performed. Specifically, significant difference in rotation between gender subgroups was assessed using an independent samples t-test in each of the four loading conditions (extended knee with external torque, extended knee with internal torque, flexed knee with external torque, and flexed knee with internal torque). The patients were furthermore divided into the following categories: all patients contralateral knees, SB patients injured knees pre-operatively, and SB patients injured knees post-operatively. There were no significant differences in rotational laxity between genders in any subject category or in any loading condition examined. (All p-values were greater than 0.158.) Therefore, the imbalance in gender distribution did not account for differences in laxity.

Another possible limitation of our study was that the post-operative rehabilitation regime was not regulated. Although, it has been shown that rehabilitation has an effect on clinical outcome, several studies have shown that there has been no significant effect on knee laxity, specifically (Beynnon *et al.*, 2005; Grant *et al.*, 2005; Shelbourne & Davis, 1999). Furthermore, due to the brief follow-up time following surgery, differences in rehabilitation protocol would likely not have had a great effect on measured knee rotation.

This limited mean follow-up period of only 5 months may alternatively be viewed as a limitation of the study in that certain structures of the ACL-injured joint require two years to fully recover (Risberg *et al.*, 2004). All patients, however, suffered isolated ACL injury; without concomitant damage to surrounding structures, recovery time should be reduced (Anderson *et al.*, 1992). Moreover, all patients were able to walk pain-free at follow-up; general observation by the primary investigator (AH) found that the variability in patients' perceived comfort during the pre-operative testing session was greater than post-operatively. (No knee scores or other functional tests were available to confirm this observation.) An additional study with a longer follow-up period in which the same rehabilitation protocol is followed by all subjects would be beneficial to confirm these findings following complete ACL recovery.

Important clinical implications for surgeons performing anterior cruciate ligament reconstruction are identified by the results of this study. In determining whether the DB technique would benefit or, in fact, hinder a particular patient, the extent of the injury should be considered. In this study, for isolated ACL rupture with negligible damage to the surrounding soft tissues, a DB reconstruction had no advantage over the SB technique and may even have overconstrained the knee in some cases. However, if the primary restraints to rotation are debilitated and cannot be reconstructed surgically, a DB technique may provide the additional restraint that could prevent or minimize further injury.

This study provides insight into differences in surgical technique under a specific loading condition. However, comparisons of SB and DB reconstructions under different loading conditions must also be considered before determining the best treatment for a particular patient. Further research is required to evaluate these techniques in functional weight-bearing tasks, at higher flexion angles, and in patients with concomitant injury. For this reason, we have conducted another investigation into the outcome of single and double-bundle reconstruction under physiological loading conditions in Chapter 6.

Minimal difference in outcome of SB and DB reconstruction was demonstrated under torsional loading conditions in a group of patients with isolated rupture of the ACL. Since subjects had negligible concomitant injury to structures that have been shown to provide rotational restraint such as the collateral ligaments and menisci, sufficient constraint was likely provided by these structures. The overall evidence presented by this study suggests that the intact ACL does not restrict external rotation, but provides internal rotational restraint when knee conditions generate greater tension and substantial recruitment of the ACL. The rotational laxity that results from isolated ACL injury is restored by both SB and DB surgical techniques.

Chapter 6

Kinematics of the ACL-deficient and reconstructed knee during dynamic activities

6.1 Introduction

The anterior cruciate ligament is the most commonly injured ligament of the knee (Widuchowski *et al.*, 2007). Persistent laxity of the joint following reconstruction of the anterior cruciate ligament (ACL) is thought to lead to osteoarthritis (Chaudhari *et al.*, 2008; DeFrate *et al.*, 2006; Georgoulis *et al.*, 2003); therefore, research has recently concentrated on improving reconstruction techniques to restrict laxity, primarily in the transverse plane (Stergiou *et al.*, 2007). The double-bundle (DB) surgical technique, which reconstructs both anteromedial and posterolateral bundles of the ACL, has been shown to limit rotational laxity to a greater extent than the single-bundle (SB) technique (Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Yagi *et al.*, 2007).

However, little data is available evaluating transverse plane restraint following SB versus DB reconstruction techniques during physiological loading conditions *in vivo*. Jordan *et al.* (2007) demonstrated that the behaviour of the anteromedial and posterolateral bundles of the ACL throughout the range of flexion during weightbearing was inconsistent with observations made in cadaveric studies. They found that both bundles were longest near extension and decreased in length with increasing flexion, while previous non-weightbearing investigations had shown a reciprocal functioning of the length of the two bundles (Amis & Dawkins, 1991).

Conflicting results with respect to the position of the knee joint centre of rotation in the transverse plane were also illustrated by Koo & Andriacchi (2008) when comparing walking with non-ambulatory activities. Whereas passive flexion-extension and squatting activities exhibited a centre of rotation on the medial side (Dennis *et al.*, 2005; Hill *et al.*, 2000; Iwaki *et al.*, 2000), the average centre of rotation was found to be on the lateral side in all subjects during normal walking (Koo & Andriacchi, 2008).

ACL deficiency has been shown to alter three-dimensional (3D) knee kinematics during gait (Andriacchi & Dyrby, 2005; Georgoulis *et al.*, 2003; Ristanis *et al.*, 2003; Tashman *et al.*, 2004; Zhang *et al.*, 2003). Traditional methods of ACL reconstruction using the single-bundle approach are not able to restore normal kinematics (Brandsson *et al.*, 2002; Georgoulis *et al.*, 2007; Ristanis *et al.*, 2005). The objective of this study, therefore, was to confirm whether the findings of our passive loading study (Chapter 5) applied under physiological loading conditions. A randomised control trial would determine whether a double-bundle ACL reconstruction is better able to restore 3D knee kinematics with respect to those of the healthy knee than the single-bundle surgical technique during dynamic, weightbearing activities.

6.2 Methods

6.2.1 Participants and interventions

Thirty-three subjects from either the patient or healthy control groups of the passive knee laxity studies described in Chapters 4 and 5 agreed to participate in this additional study to test dynamic knee laxity. The twenty-two patients were randomly allocated either a single or double-bundle surgical reconstruction (with 11 subjects in each group); 11 age- and gender-matched Control subjects were also selected to continue in the additional trial.

Testing was completed at the Sports Science Institute of South Africa in Cape Town between April 2007 and July 2008; testing of patients was conducted prior to and following ACL reconstruction, while Control subjects were tested once only. Eligibility criteria, surgical procedures, randomisation and blinding are described in detail in Chapter 5 in sections 5.2.1, 5.2.1.1, 5.2.1.2, and 5.2.3 respectively.

6.2.2 Data collection protocol

Subjects' gait during low- and high-demand activities was recorded at 250 Hz in six degrees of freedom using an eight-camera motion analysis system (Vicon Motion Systems, Oxford, UK). Anthropometric data were recorded and fifteen retro-reflective markers were secured to anatomic landmarks based on the modified Helen Hayes marker set (Vaughan *et al.*, 1999). A minimum of five trials were collected for each of the following activities:

Walk – A standard walking trial (low-demand activity) was used as baseline data with which to compare results to other studies. Subjects were instructed to walk along a 10 m walkway at their self-selected pace.

Ninety-degree cut – A cutting activity was designed to actuate a 90° change in direction to simulate a typical game situation in which an offensive player tries to get open from a defender. Three cones were used to mark the start, cut point, and end locations of the cut in order to guide the subject (Figure 6.1). Approximately 3 m of space was available on either side of the cut point cone for approach and termination, which limited the speed and intensity at which this activity could be performed. A demonstration of the activity was performed; however, subjects were allowed to execute the task in the manner most natural to them (i.e. they were not required to follow a specific step-sequence). The cutting activity was repeated on both sides, with subjects instructed to cut first to their right for an acceptable number of trials and then to their left to ensure that both injured and contralateral limbs would be on the inside and outside of the body during the activity.

Jump – Subjects were asked to perform a maximum distance two-foot jump from which a full recovery could be made. A practice trial in which subjects jumped as far as they felt comfortable was used to mark take-off and landing positions so that the same distance would be covered for each recorded trial. If a subject was not able to return to an upright position without losing their balance, the landing marker was brought closer to the take-off position in increments of 5 cm until a two-foot jump and full recovery landing could be achieved. For the patient group, this distance was recorded so that the same distance could be used in the follow-up test session.

6.2.3 Data analysis

The first phase of data processing was completed using Vicon's Workstation software (Oxford Metrics, England). Angles were defined according to the Joint Coordinate System (Grood & Suntay, 1983). The optimised lower-limb gait analysis (OLGA) method as described by Charlton *et al.* (2004) was used to improve the quality of the kinematic output; specifically, this method which uses a Kalman filter, has been shown to reduce variability across trials by minimising artefact due to soft tissue movement and kinematic cross-talk (Charlton *et al.*, 2004; DeGroote *et al.*, 2008). Anthropometric measurements, together with the marker positions recorded during the static trial were used to improve calculations of bone lengths. In combination with the walking trial used for the dynamic calibration, a better estimate of joint centres and segment orientations could be determined, thereby improving the reliability of the joint angle output (Charlton *et al.*, 2004; Roren, 2005). The default settings in Vicon were used for all activities.

Up to five trials from each activity were selected for further processing based on minimum kinematic fit residuals calculated in OLGA. The following gait cycle events for a minimum of three good trials were marked manually for export with the kinematic data: left and right foot strike over 1.5 strides (i.e. 3 steps = 4 foot strikes) during the walking and cutting trials, as well as heel off, toe off, and foot strike for the jump task. Foot strike was defined as the frame at which the first of either the heel or toe marker reached a local minimum during the stance or landing phase.

The cutting activity was split into two components. The first component included the initial step leading up to the change of direction and the step following the 90° cut (e.g. for a right-cut, this included the right, left, and right heel strikes). The second component included the step following the 90° cut and the step concluding the change in direction (e.g. for a right cut this included the left, right, and left step sequence). In other words, the step immediately following the initiation of the change in direction was included in both the first and last components of the activity. This was done to ensure the complete change of direction was captured in the gait cycles; while some subjects were able to finish the rotation in the second step of the three-step analysis, others performed a more rounded cut in which two steps were required to complete the 90° change of direction.

Each component of the cutting task and one stride of the walk activity was normalized to a 100% cycle. The trial time for the jump activity was calculated from first heel off to initial foot contact on landing times; in order to include the recovery period on landing, the whole jump cycle was defined as 65% longer than the trial time from initial heel off. The airborne (i.e. swing) phase of the jump, defined as final toe off to first contact landing, was also analysed as a separate 100% cycle.

The three-dimensional kinematic data and marked gait cycle events were exported from Workstation into Matlab where gait cycle normalisation was accomplished. Individual flexion-extension, add-abduction, and internal-external rotation curves were plotted for each trial, from which outliers were excluded and the remaining trials were used to calculate subject mean curves. Maximum, minimum, and range of rotation data were determined for rotations in each of the three anatomical planes from each subject mean curve for further statistical analysis. Range midpoint, defined as the mean of the maximum and minimum values, was also analysed. Activity mean curves were generated from individual subject mean curves.

Since the cutting task requires asymmetric behaviour of left and right legs, the activity was further divided to examine inside or outside (based on the turning direction) limbs. The inside limbs (i.e. right leg for the right-cut and left leg for the left-cut trials) were subsequently combined, with corresponding groupings carried out for the outside limbs. The activities were consequently categorised and described as follows:

Walk – left heel strike to heel strike,

Cut123Inside – inside limb during initial three foot strike events (i.e. limb that is in stance, then swing),

Cut123Outside – outside limb during initial three foot strike events (i.e. limb that is in swing, then stance),

Cut234Inside – inside limb during final three foot strike events,

Cut234Outside – outside limb during final three foot strike events,

JumpFull – the complete jump task from heel off to standing,

JumpSwing – the airborne/swing phase of the jump task.

6.2.4 Objectives and outcome

The primary objective of this study was to determine differences in transverse plane knee rotation from pre- to post-operative testing sessions in patients who had undergone either single or double-bundle ACL reconstruction during weight-bearing activities. The hypothesis with respect to the primary outcome was that both surgical procedures would reduce the overall range of rotation; however, the DB reconstruction would reduce it to a greater extent than the SB surgical technique. The secondary objectives were to compare the mean control and contralateral knee data with those of the injured knees pre- and post-operatively.

6.2.5 Statistical analysis

Interactions between the single and double-bundle groups from the pre- to post-operative test sessions were determined using a linear mixed model for repeated measures in SPSS 15.0 (SPSS Inc). Post-hoc analysis was conducted using a two-tailed paired samples t-test (or Wilcoxon signed-rank test if data were not normally distributed). The linear mixed model was also used to compare secondary outcomes including contralateral and injured knees of the patients in their subgroups (SB and DB tested pre- and post-operatively) and differences between

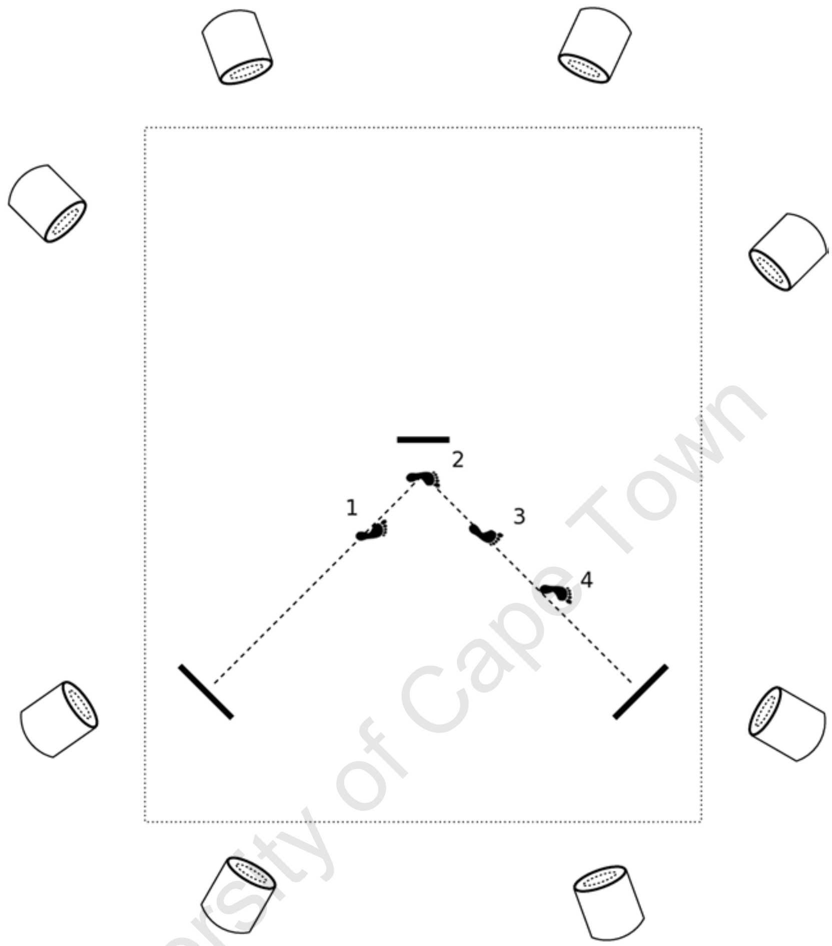


Figure 6.1: Gait lab setup showing typical foot strike positions for the Cut-Right activity and the numbering system used to define the first and second parts of the activity, Cut123 and Cut234, respectively. (Figure is not to scale.)

healthy left and right knees in the Control group. Control left-right averaged and ACL-deficient (ACLD) knee data were compared using independent samples t-tests (or Wilcoxon sum-rank test for non-normally distributed data). P-values less than or equal to 0.05 were considered statistically significant.

6.3 Results

The baseline data presented in Table 6.1 show that all four female patients were randomly allocated a single-bundle reconstruction. The SB group was also slightly older, and mean height and mass were slightly less than the DB group. Furthermore, although the differences were not statistically significant, there was a tendency towards a lower cutting activity cadence post-operatively in the SB group, while the mean cadence in the DB group remained approximately equal between testing sessions. The cadence of the ACLD patients was also significantly higher than that of the Control group for the Cut-Right activity; the difference between Controls and ACLD patients for the Cut-Left activity approached significance.

Table 6.1: Baseline demographic and clinical subject data (mean ± SD) for control and patient groups. ACL all includes subjects from both single-bundle (SB) and double-bundle (DB) groups.

Variable	**Session**	**Control**	**ACL all**	**SB**	**DB**
Sex (F:M)		3:8	4:18	4:7	0:11
Age (yrs)		29.5 ± 5.4	29.0 ± 5.7	32.1 ± 4.9	25.9 ± 4.9
Height (cm)		176.3 ± 9.3	174.3 ± 8.0	170.5 ± 7.5	178.0 ± 6.8
Mass (kg)		73.0 ± 12.0	81.8 ± 13.9	79.1 ± 14.4	84.5 ± 13.5
Time Injury-Pre (mos)		n/a	6.9 ± 10.4	6.7 ± 13.8	7.0 ± 6.0
Time Surg-Post (mos)		n/a	4.6 ± 1.6	3.6 ± 0.7	5.8 ± 1.6
Walk cadence	Pre	114.3 ± 7.9	110.0 ± 7.0	110.4 ± 7.0	109.7 ± 7.4
(steps/min)	Post	n/a	112.8 ± 6.4	114.4 ± 7.4	110.4 ± 3.8
Cut-Right cadence	Pre	149.1 ± 21.9	169.2 ± 28.1	166.9 ± 32.8	171.1 ± 25.1
(steps/min)	Post	n/a	162.9 ± 23.9	156.0 ± 11.2	171.6 ± 32.8
Cut-Left cadence	Pre	150.9 ± 25.4	169.9 ± 27.3	168.6 ± 32.8	170.6 ± 23.4
(steps/min)	Post	n/a	161.8 ± 21.3	156.5 ± 14.7	168.5 ± 27.1

6.3.1 Protocol deviations

Of the 22 patients in this randomised control trial, three allocated to the double-bundle reconstruction group were lost to follow-up (Figure 6.2). Two subjects in the single-bundle group were designated outliers for at least one of the activities. This definition was based on maximum and minimum flexion and rotation angles:

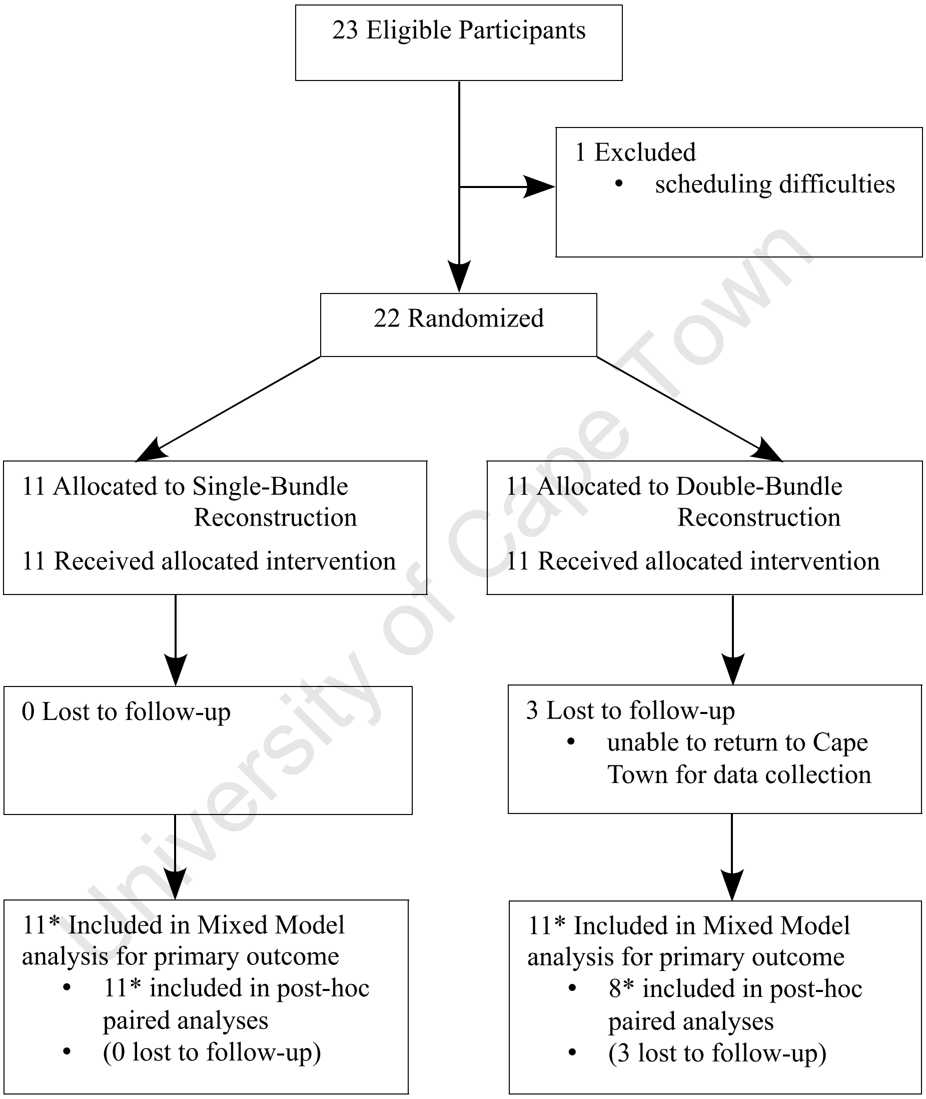


Figure 6.2: Flow diagram of participants through each stage of the randomised control trial for the primary outcome comparing single and double-bundle ACL reconstruction. * Some data were excluded from analyses if subjects were classified as outliers for a particular activity. (See explanation in section 6.3.1.)

if, for a particular activity, both flexion and rotation angles were greater than two standard deviations away from the mean of the overall group including all subjects, the data collection notes were examined for an explanation. The two patients for whom this was the case both experienced discomfort during weight-bearing activities due to their knee injuries. Their data was subsequently omitted from the group analyses for those particular activities in the pre-operative session.

6.3.2 Adverse events

Adverse events were described in detail in section 5.3.2. The two subjects who experienced problems following surgery (one requiring an additional surgery and one with swelling and discomfort) were both able to participate in the follow-up session of the gait study without any problems or physical limitations. Both of these subjects were allocated a SB reconstruction.

6.3.3 Gait activity kinematics in three planes

Maximum knee flexion angles for both Control and ACLD subjects performing the high-demand activities (cutting and jumping) exceeded maximum angles during walking (Table 6.2, Figures 6.3 to 6.5). In general, the inside knee demonstrated greater flexion than the outside knee while performing the 90° cut, with Control subjects consistently demonstrating greater maximum flexion than ACLD subjects for this and the JumpFull activities.

Similar to flexion, maximum adduction angle tended to increase for the more demanding activities; abduction angles (i.e. minimum negative adduction) varied to a lesser extent across activities for the ACLD knees than for the healthy control knees and compared to the maximum adduction angles.

Greater maximum and smaller minimum flexion and adduction angles were achieved throughout the full jump activity (JumpFull) when compared to just the airborne swing phase (JumpSW); however, rotation angles were only different for the maximum values. The minimum internal rotation angles, equivalent to maximum external rotation, were actually slightly less during the swing phase of the activity due to a more consistent alignment of the subjects' individual curves owing to more distinct activity start and end points.

No significant differences were demonstrated in maximum or minimum transverse plane rotation between the ACLD and healthy control knees, although ACL-deficient knees tended to have lower maximum and minimum internal rotation values (Table 6.2).

Since the second part of the cutting activity (Cut234) generally showed similar trends to the first part (Cut123), these data are presented in Appendix G. The following discussion will focus on the first part of the activity.

6.3.4 Outcomes: Transverse plane rotation

No significant interaction was found in range of rotation when comparing SB and DB injured knees during the pre- and post-operative test sessions. Significant interactions *were*, however, found in all activities except walking when analysing the rotation midpoint defined as the mean of the maximum and minimum rotations (Table 6.3). The rotation midpoint indicated the shift in the range of rotation (or rotational alignment) between test sessions shown in Figures 6.7 and 6.13. During both cut and jump activities, the transverse plane rotational alignment of the double-bundle knees was significantly closer to that of the healthy control group than that of the single-bundle reconstructions.

ACLD knees tended to have a more external shift in the range of rotation than did the Control subjects' knees during cutting; this shift, however, was not statistically significant. No significant differences were found in rotation midpoint between left and right knees of the Control subjects in any activities.

No differences in the shift of range of rotation of the patients' contralateral and injured knees were found from the pre- to post-operative testing session (Figures 6.14 and 6.15, Appendix G), i.e. the contralateral knees followed the same pattern as the injured knees from one test session to the next.

6.4 Discussion

This study measured 3D knee kinematics during dynamic weightbearing activities in healthy subjects and in ACL-injured patients prior to and following surgical reconstruction. With patients randomly allocated either a single or double-bundle

Table 6.2: Maximum and minimum joint angles (degrees) in three planes for all activities for the Control and ACL-deficient (pre-operative) knees. (Mean $\pm$ SD)

	Control		ACL-deficient	
Activity	**max**	**min**	**max**	**min**
Flexion				
Walk	62.7 $\pm$ 4.0	5.3 $\pm$ 4.1	59.9 $\pm$ 6.5	5.7 $\pm$ 7.6
Cut123Inside	101.2 $\pm$ 10.6	27.4 $\pm$ 7.2	87.7 $\pm$ 9.1	22.9 $\pm$ 8.8
Cut123Outside	81.8 $\pm$ 8.8	18.3 $\pm$ 6.7	72.2 $\pm$ 15.1	17.2 $\pm$ 7.5
Cut234Inside	101.7 $\pm$ 10.8	18.9 $\pm$ 5.9	87.9 $\pm$ 9.2	19.0 $\pm$ 6.9
Cut234Outside	81.5 $\pm$ 7.5	14.4 $\pm$ 5.0	77.7 $\pm$ 8.1	14.4 $\pm$ 8.0
JumpFull	82.6 $\pm$ 14.7	9.8 $\pm$ 6.5	72.8 $\pm$ 11.0	10.4 $\pm$ 7.8
JumpSW	49.9 $\pm$ 5.1	13.0 $\pm$ 7.5	50.4 $\pm$ 11.6	14.4 $\pm$ 6.5
Adduction				
Walk	14.4 $\pm$ 7.2	-2.1 $\pm$ 4.8	16.1 $\pm$ 9.2	-1.6 $\pm$ 9.2
Cut123Inside	19.3 $\pm$ 9.1	-1.8 $\pm$ 8.6	24.2 $\pm$ 12.6	0.5 $\pm$ 8.5
Cut123Outside	15.2 $\pm$ 8.9	-5.1 $\pm$ 7.9	19.6 $\pm$ 8.9	-1.1 $\pm$ 8.3
Cut234Inside	19.7 $\pm$ 9.2	-3.2 $\pm$ 7.7	24.2 $\pm$ 12.7	-0.8 $\pm$ 7.7
Cut234Outside	17.7 $\pm$ 7.8	-8.2 $\pm$ 10.0	20.4 $\pm$ 7.9	-1.2 $\pm$ 7.6
JumpFull	13.6 $\pm$ 6.8	-6.8 $\pm$ 9.7	16.6 $\pm$ 9.3	-3.8 $\pm$ 9.6
JumpSW	9.8 $\pm$ 5.6	-3.0 $\pm$ 5.4	14.0 $\pm$ 8.9	0.3 $\pm$ 6.7
Internal Rotation				
Walk	2.6 $\pm$ 7.9	-18.9 $\pm$ 7.9	3.2 $\pm$ 8.6	-18.9 $\pm$ 10.2
Cut123Inside	11.8 $\pm$ 8.1	-14.8 $\pm$ 7.6	7.4 $\pm$ 6.8	-18.9 $\pm$ 8.4
Cut123Outside	11.3 $\pm$ 5.5	-15.3 $\pm$ 6.4	8.9 $\pm$ 8.5	-16.8 $\pm$ 8.5
Cut234Inside	12.6 $\pm$ 8.3	-18.4 $\pm$ 6.4	7.5 $\pm$ 7.6	-19.8 $\pm$ 7.6
Cut234Outside	13.2 $\pm$ 6.3	-18.8 $\pm$ 8.0	10.7 $\pm$ 8.4	-19.4 $\pm$ 9.6
JumpFull	12.3 $\pm$ 6.2	-14.4 $\pm$ 6.3	10.7 $\pm$ 8.2	-13.4 $\pm$ 7.0
JumpSW	6.8 $\pm$ 7.1	-15.5 $\pm$ 6.3	4.2 $\pm$ 7.3	-14.3 $\pm$ 7.4

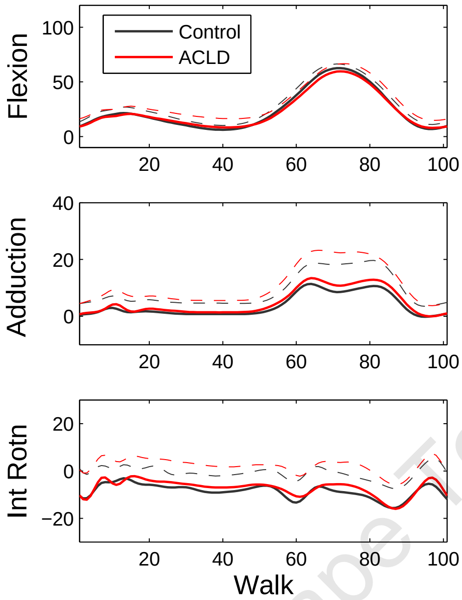


Figure 6.3: Walk three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

reconstruction, an objective comparison of these two treatment methods could be made under physiological loading conditions.

6.4.1 Three-dimensional knee kinematics during low and high demand activities

The walking activity showed comparable knee kinematics between the Control subjects and previously published data (Kadaba *et al.*, 1990), thereby verifying the methods used for this study and providing an acceptable baseline data set with which to compare the results of the other activities and patient outcomes.

No previously published data could be found in which identical methods of cutting or jumping activities were performed with which to compare the kinematic data; however, reasonable comparisons could be made with studies investigating similar tasks. McLean *et al.* (1999) and Sigward & Powers (2006) investigated kinematics during the stance phase of side-step cutting in healthy

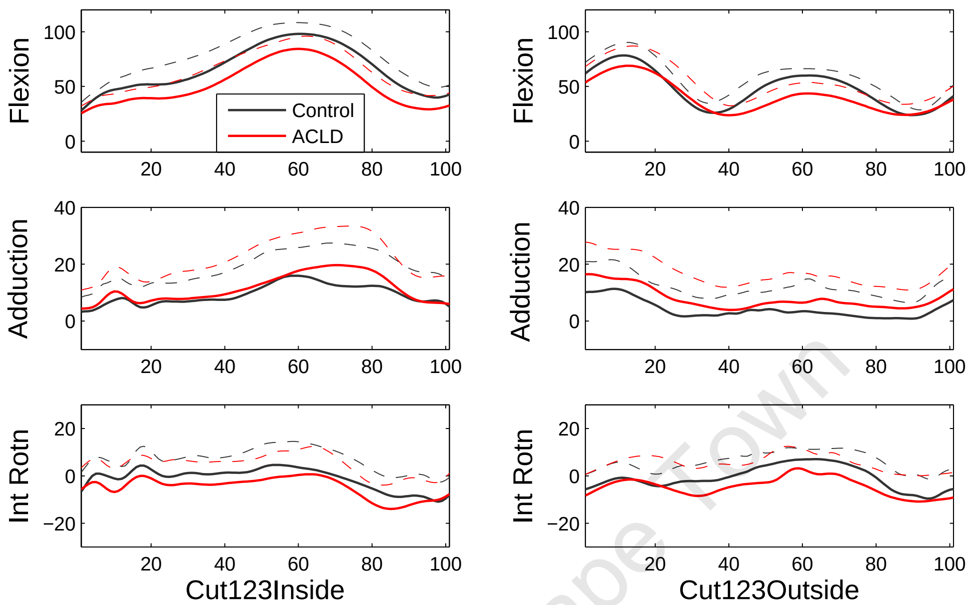


Figure 6.4: Cut123Inside and Cut123Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

subjects. Despite a less pronounced cut angle (45° rather than 90° used in this study), maximum flexion angles reported during the stance phases were slightly lower (approximately 46°) and greater (approximately 55°), respectively (McLean *et al.*, 1999; Sigward & Powers, 2006) than our results of just over 50° during the Cut123Inside stance phase (Figure 6.4).

In the frontal and transverse planes, results from the Control subjects in this study more closely matched those of Sigward & Powers (2006) and Nagano *et al.* (2009) in which subjects demonstrated between 0° and 10° of adduction, as well as initial external rotation followed by approximately 6° of internal rotation. The results of McLean *et al.* (1999) on the other hand, displayed joint angle curves in the abduction and external rotation domains (rather than primarily adduction and internal rotation); these reciprocal findings may be attributed to a simple shift in the curves as a result of differences in the anatomical landmarks chosen and segment coordinate system definitions. A change in the orientation of the flexion-extension axis can alter the calculated ab-adduction and internal-external

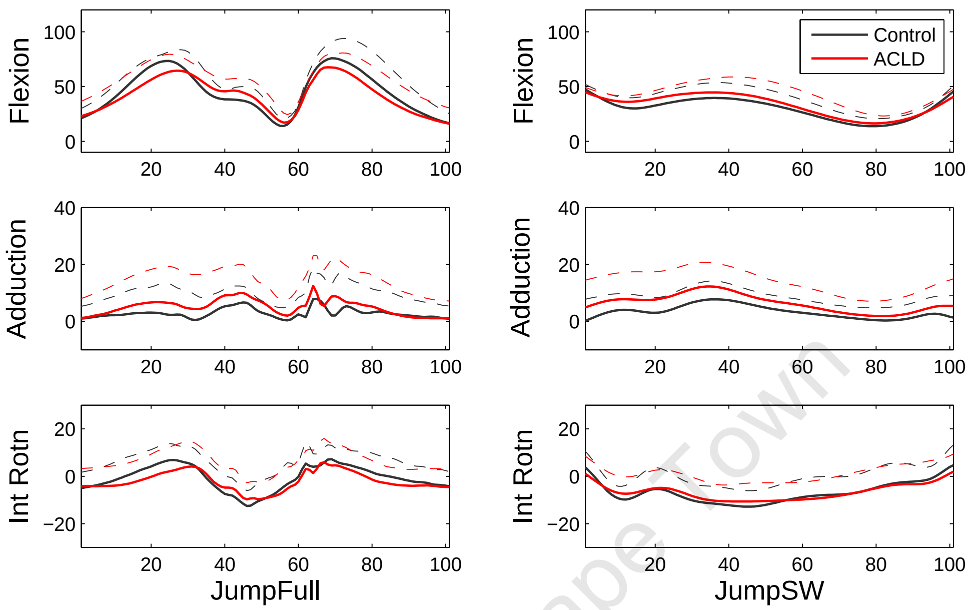


Figure 6.5: JumpFull and JumpSW three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

rotation angles by as much as 15° (Kadaba *et al.*, 1990; Piazza & Cavanagh, 2000). In both this study and that of Sigward & Powers (2006), the Vicon model was used as a basis from which to calculate joint angles.

Range of internal-external rotation over the 100% Cut123Inside cycle was greater than those presented by previous studies for similar activities (McLean *et al.*, 1999; Nagano *et al.*, 2009; Sigward & Powers, 2006); however, the greatest change between maximum and minimum values occurred between 55% and 100% of the cycle (i.e. the swing phase), which was not analysed in these other studies. As changes in kinematics with ACL injury have been observed during the swing phase of gait (Andriacchi & Dyrby, 2005; Georgoulis *et al.*, 2003), we considered it important to include this data in the analysis.

The jump activity cycle demonstrated an increase and then decrease in knee flexion prior to toe-off (discerned by comparing the JumpFull and JumpSW curves). Maximum extension was reached just before landing at which point the knee flexed to absorb the impact from the landing. Adduction and rotation

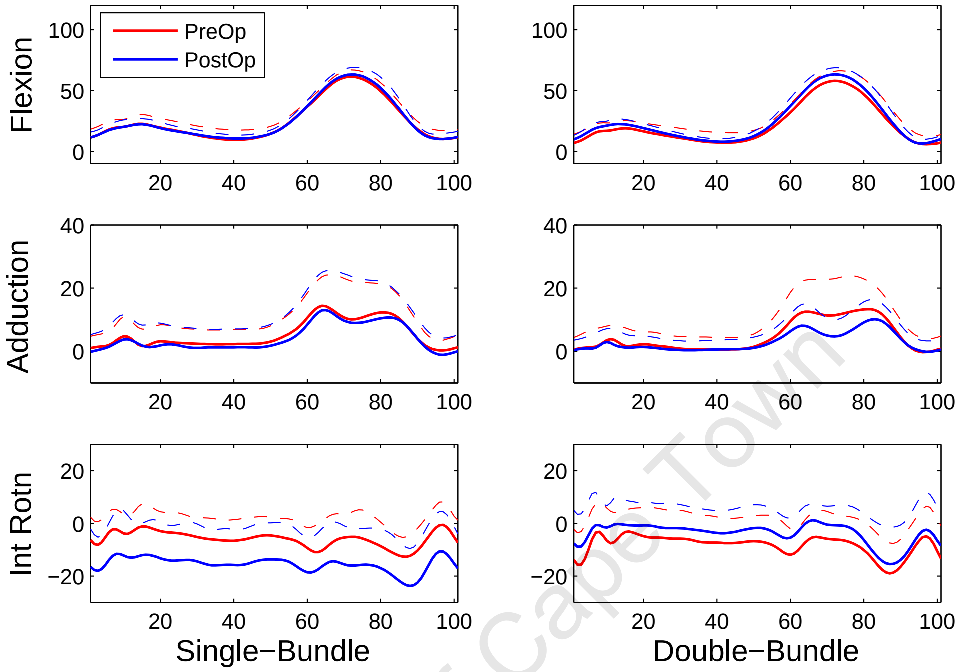


Figure 6.6: Walk three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

curves displayed relatively high frequency changes upon landing, which can most likely be attributed to the reverberations of the wand markers during this high impact stage of landing. Maximum adduction and internal rotation values must, therefore, be interpreted with caution.

The general pattern of the internal-external rotation curve during the take-off and landing (i.e. stance) phases of the JumpFull activity conformed to those of similar activities in other studies, including two-foot vertical jump landing (Nagano *et al.*, 2009) and squatting (Hemmerich *et al.*, 2006; Yamaguchi *et al.*, 2009). The tibia rotated internally with respect to the femur with increasing knee flexion, demonstrating coupled screw-home motion (Benoit *et al.*, 2007; Hill *et al.*, 2000). External rotation accompanied knee extension during the swing phase of this activity (Figure 6.5).

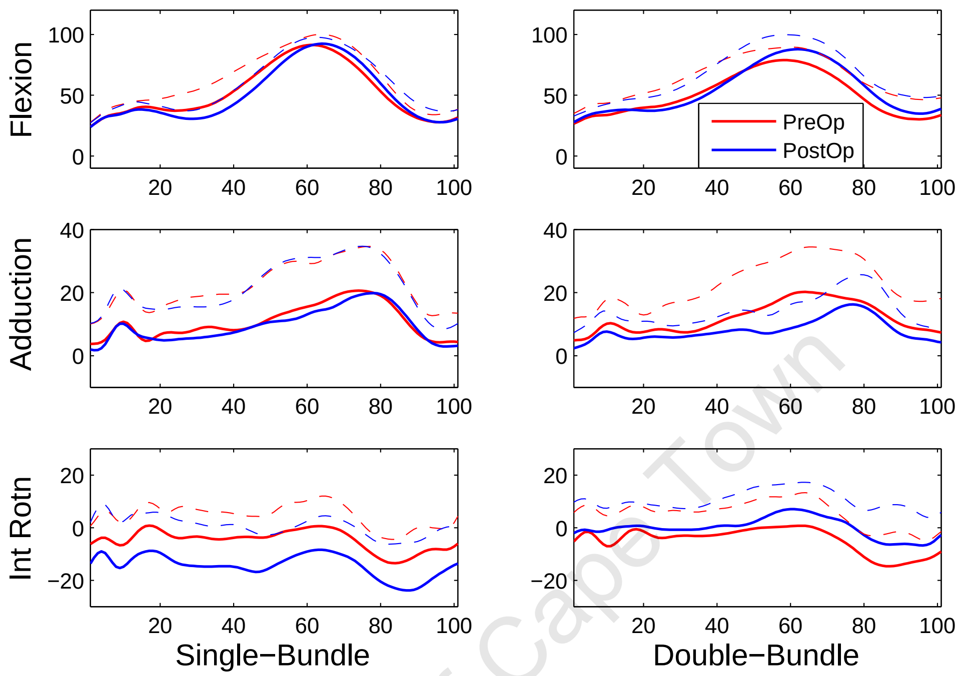


Figure 6.7: Cut123Inside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

6.4.2 Differences in kinematics in healthy Control and ACL-ruptured knees

Joint kinematic curves in all three planes demonstrated similar patterns between the injured and healthy knees. However, the slower cadence of the self-selected walking pace and a decrease in maximum flexion angles during all three activities indicated a possible protection mechanism adopted by the ACLD subjects (Table 6.2). Similar reductions in flexion were observed during the stance phase of walking gait in ACLD subjects when compared with healthy control subjects by several other investigators (Chmielewski *et al.*, 2005; Georgoulis *et al.*, 2003; Rudolph *et al.*, 2001). By increasing knee stiffness and thereby limiting degrees of freedom, a subject with deficient afferent feedback due to injury is able to stabilise the joint (Chmielewski *et al.*, 2005; Georgoulis *et al.*, 2003).

This decrease in flexion angle was more pronounced during cutting and jump-

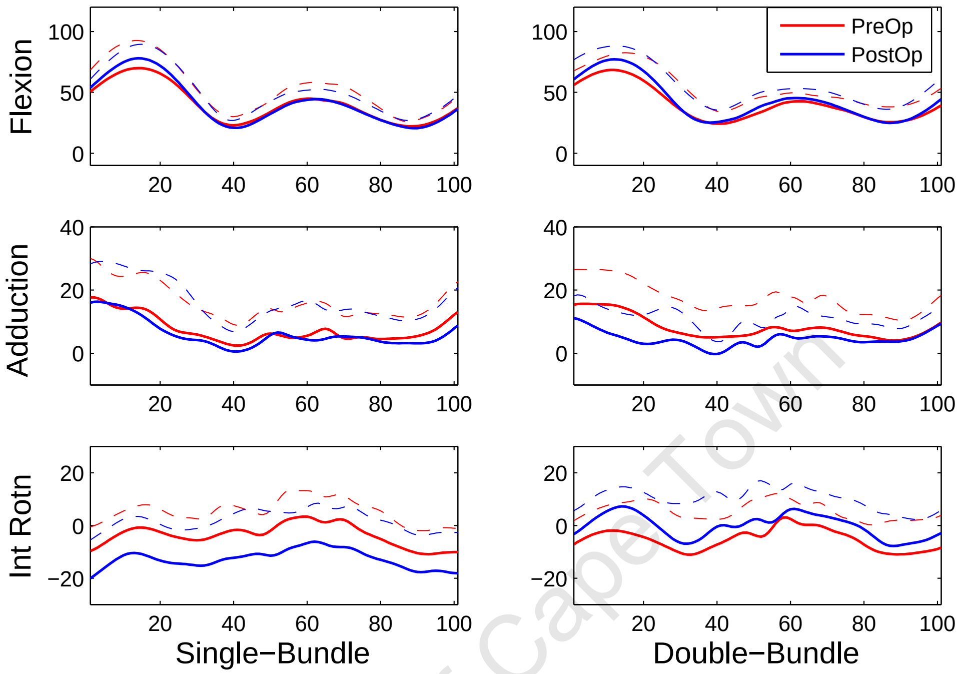


Figure 6.8: Cut123Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

ing than walking with a difference of up to 13.8° between mean maximum flexion between the two groups during the Cut234Inside activity (Table 6.2). The reduced flexion may not be attributed to a more cautious technique by which this activity was performed by the patient group as evidenced by a slight increase in cadence when compared to the the Controls. This shows that, whereas the Control group adopted a relaxed, moderate intensity level, the patient group made greater effort to perform this activity at the maximum intensity level at which they felt comfortable in their injured state.

No significant differences in range of internal-external rotation were found between the ACLD and the healthy Control knees. The ACLD subjects, however, demonstrated a general tendency toward greater adduction and less internal rotation than the Controls during the high demand activities (Table 6.2). Again, this supports the theory that the patients adopted a knee protection strategy.

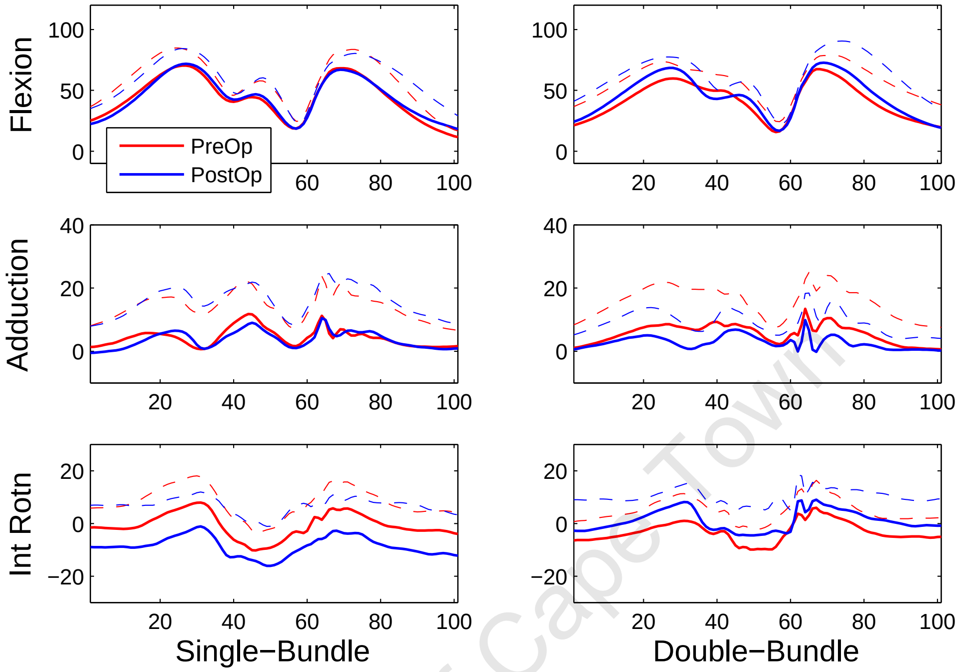


Figure 6.9: JumpFull three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

Passive abduction (valgus moment) combined with internal rotation manifests the pivot shift phenomenon in the ACLD knee and correlates to a patient's sense of instability during gait (Amis *et al.*, 2005; Yamaguchi *et al.*, 2009). In order to prevent a possible 'giving way' occurrence, someone with an injured ACL may actively adduct and externally rotate the tibia through muscle activation. Since the ACL is a secondary restraint to internal rotation, co-contraction of the medial hamstrings would not only stiffen the joint, but would rotate the tibia externally with respect to the femur, reducing the risk of damage to other structures in the absence of restraint of the ACL.

Similar shifts in rotation and adduction have been observed in ACL-deficient subjects during walking and running (Tashman *et al.*, 2004; Zhang *et al.*, 2003). When compared with the normal knees in the healthy control group (Zhang *et al.*, 2003) or contralateral uninjured knees (Tashman *et al.*, 2004), the ALCD knees

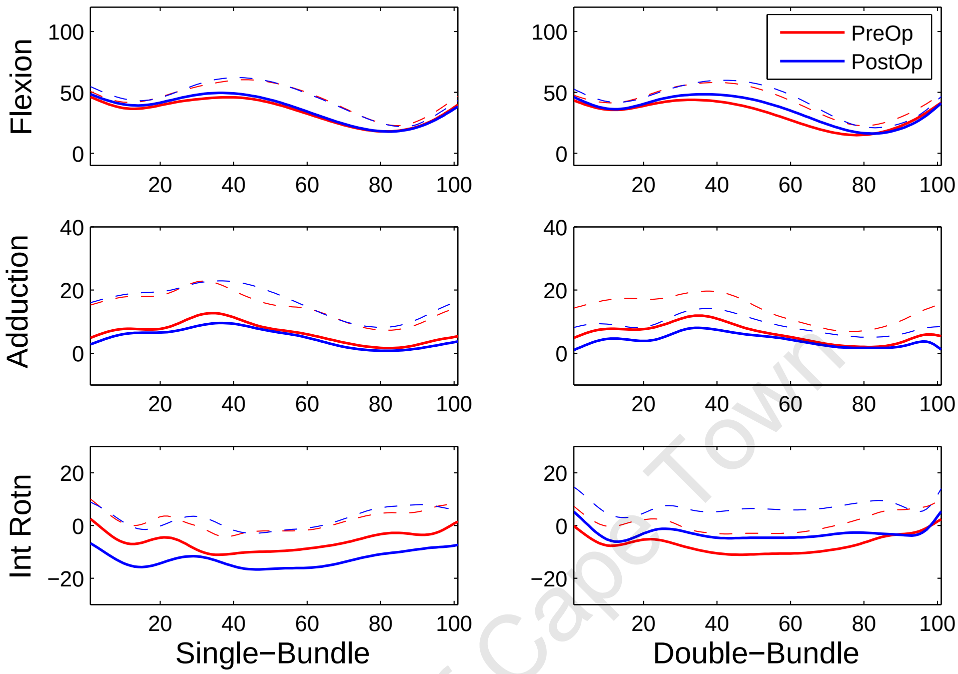


Figure 6.10: JumpSW three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

showed a similar ranges and patterns of rotation, but with a roughly 4° increase in external rotation and 3° increase in adduction.

Opposite trends were observed by Georgoulis *et al.* (2003) and Andriacchi & Dyrby (2005); their ACLD subjects demonstrated less external rotation with differences reaching significance during the swing phase of gait. Interestingly, Andriacchi & Dyrby (2005) observed a simultaneous decrease in anterior translation just prior to heel strike and found that the shift in rotation was correlated with the magnitude of the flexion moment during the initial stance period. In fact, our data also indicates a minimal shift towards internal rotation (although not statistically significant) of the ACLD group during the walking activity (Figure 6.3) with a greater difference occurring at the peak just before heel strike. It is possible that these subjects had developed a different compensation strategy with priority placed on minimizing anterior displacement of the tibia by increasing the

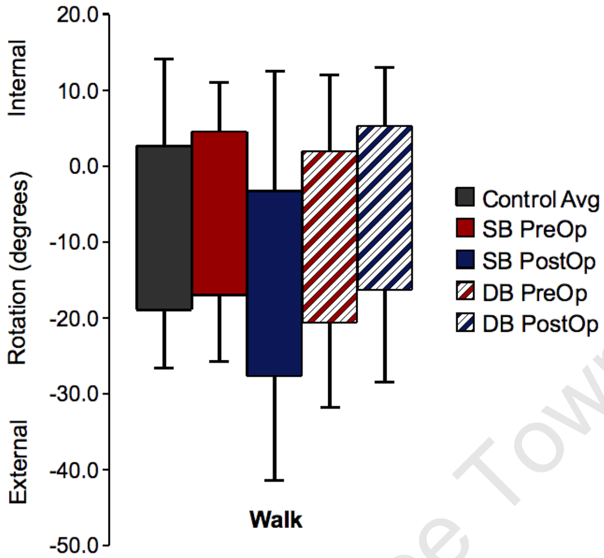


Figure 6.11: Walk rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

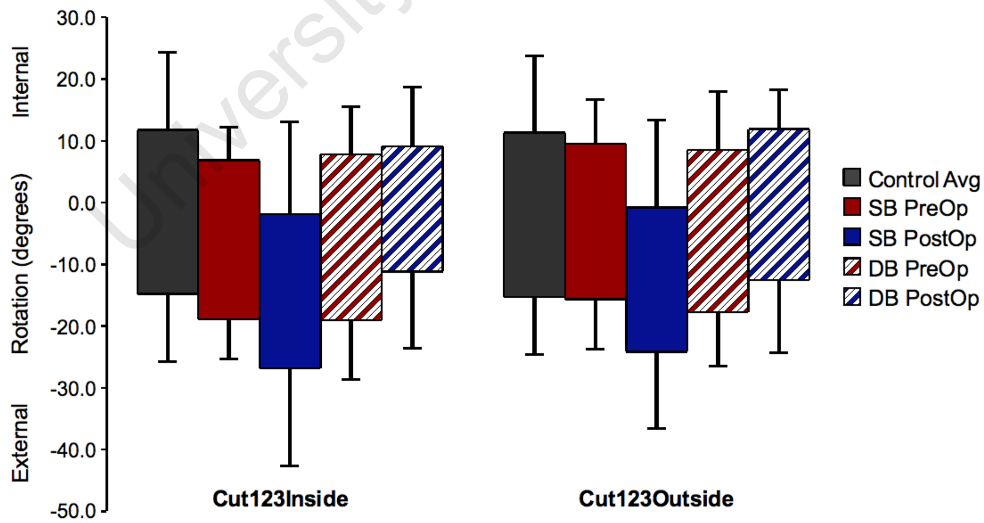


Figure 6.12: Cut123Inside and Cut123Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

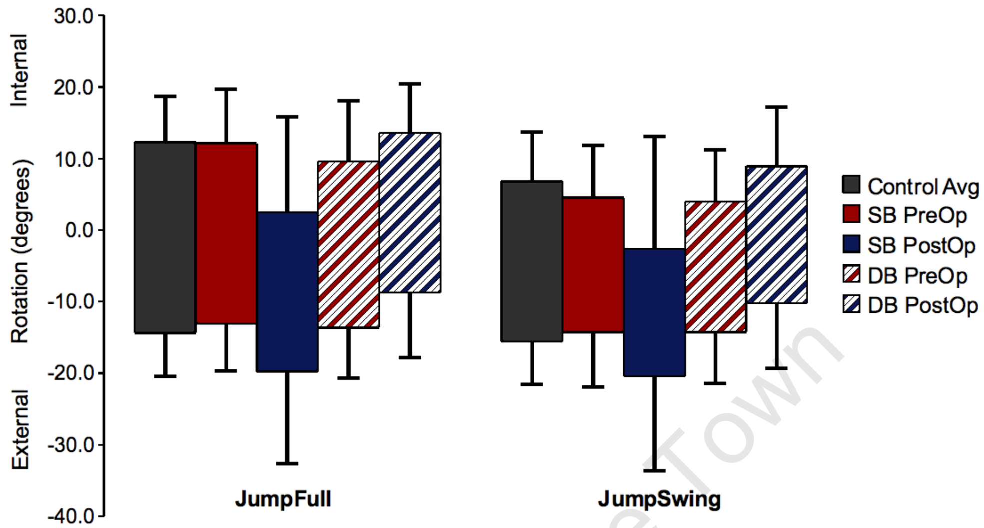


Figure 6.13: JumpFull and JumpSW rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

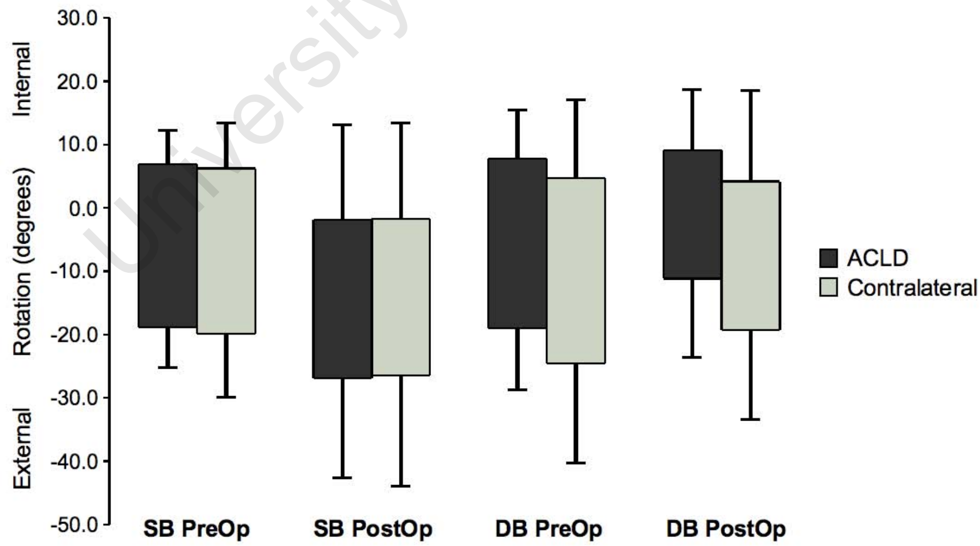


Figure 6.14: Cut123Inside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

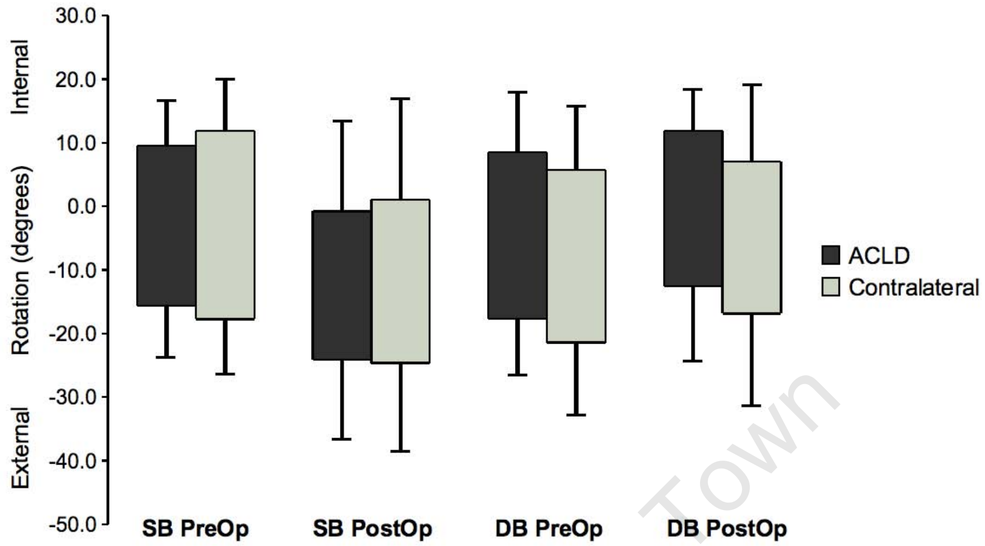


Figure 6.15: Cut123Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

flexion moment during the less demanding activity of walking.

It has been shown that co-contraction of the hamstrings is more effective at reducing the strain on the ACL between 15° and 60° of flexion and reducing internal rotation of the tibia at flexion angles greater than or equal to 30° (Li *et al.*, 1999). Since maximum flexion angles during the weightbearing phase of walking are smaller than during cutting or jumping, a co-contraction strategy would have less effect in reducing the strain on the ACL during this activity. With conflicting reports in the literature, however, further investigation is required to draw accurate conclusions regarding causes for differences in transverse plane rotation between ACLD and healthy knees during walking and high-demand activities.

Table 6.3: Means and standard deviations (in degrees) of rotation midpoint for all subject groups during each dynamic activity. P-values for surgery by test-time interaction are listed for the injured knee groups. (Statistically significant interactions are highlighted.)

Activity	**Test**	**ControlAVG**		**SB Contralat**		**DB Contralat**		**SB Injured**		**DB Injured**		**Srg x Time**
	Time	mean	SD	mean	SD	mean	SD	mean	SD	mean	SD	**p-value**
Walk	PreOp	-8.2	7.8	-7.9	6.7	-12.0	12.3	-6.2	7.2	-9.3	10.4	0.070
	PostOp			-16.9	13.7	-11.0	14.4	-15.5	14.7	-5.5	9.9	
Cut123	PreOp	-1.5	6.9	-6.8	8.6	-9.9	13.6	-6.0	5.7	-5.6	7.1	**0.041**
Inside	PostOp			-14.1	16.1	-7.5	14.2	-14.4	15.4	-1.0	10.0	
Cut123	PreOp	-2.0	5.7	-2.9	7.1	-7.8	10.7	-3.1	7.3	-4.6	8.7	**0.019**
Outside	PostOp			-11.8	14.6	-4.9	13.3	-12.5	13.2	-0.3	8.6	
Cut234	PreOp	-2.9	6.0	-7.5	8.4	-10.3	13.5	-5.5	5.8	-6.8	7.2	**0.023**
Inside	PostOp			-14.9	16.6	-8.3	13.5	-15.1	14.4	-1.9	8.8	
Cut234	PreOp	-2.8	6.6	-5.0	5.8	-8.5	11.0	-4.0	8.0	-4.7	8.8	0.082
Outside	PostOp			-13.0	14.5	-6.4	13.1	-13.4	15.4	-2.1	10.4	
JumpFull	PreOp	-1.1	5.9	-1.3	8.5	-4.8	11.5	-0.4	6.9	-2.0	7.1	**0.018**
	PostOp			-6.8	14.5	-2.0	12.6	-8.6	13.2	2.4	7.9	
JumpSW	PreOp	-4.4	6.2	-3.5	7.1	-7.7	12.0	-4.9	6.9	-5.1	7.2	**0.033**
	PostOp			-10.1	14.3	-4.2	12.9	-11.5	14.6	-0.6	8.7	

6.4.3 Differences in rotational laxity in single versus double-bundle reconstructed knees

Although maximum and minimum rotation values varied between ACLD and Control group knees and between the SB and DB injured knees from pre- to post-operative testing sessions, the overall range of rotation remained relatively constant between groups for all activities; similar results were found by Zhang *et al.* (2003) and Tashman *et al.* (2004) when investigating ACL-deficient knee kinematics. Because maximum and minimum values shifted in the same direction when comparing groups or test sessions (Figures 6.11 to 6.13), the rotation range midpoint gave a better indication of differences concerning transverse plane rotations between groups. The rotational midpoint gives an indication of the rotational alignment of the tibia with respect to the femur throughout the activity. By using the mean of maximum external and internal rotations, the centre of the envelope of rotation for each activity was established. The clinical importance of this quantity is associated with the mechanics of the joint; in this case the potential for joint degeneration caused by increased normal or shear forces in areas where they normally do not occur.

The greater maximum flexion angles and higher cadences of the 90° cut for both the patient and Control groups are clear indications that this activity was more demanding than walking; it was, therefore, not surprising that the trends observed following reconstruction with regard to internal-external rotation midpoint during walking (Figure 6.11) became statistically significant during cutting (Figure 6.12).

Similarity in range of rotation between the SB and DB groups is evidence that the the different surgical reconstruction techniques provided little difference in knee transverse plane laxity; however, the shift in range midpoint in opposing directions from pre- to post-operative testing sessions when comparing groups is a clear indicator that there is a difference in outcome between surgical techniques. The fact that the contralateral (uninjured) knees demonstrated the same shift of range midpoint between test sessions as their injured counterpart (Figure 6.14) suggests that the contrast between SB and DB outcomes may be attributed to more than simply differences in the mechanics of the surgical procedures. Berchuck *et al.* (1990) observed comparable homogeneity between injured and contralateral knees in their group of unilateral ACL-deficient patients when measuring knee flexion angles and moments. They attributed the similarities in injured and uninjured knees to their reprogramming theory, in which the locomotor process adapts to the deficient ACL by avoiding excessive motion of the tibia in order to protect the joint. The fact that the shift in range of rotation occurred over the entire cycle rather than during a specific phase (e.g. stance or swing) supports the concept that this is a neuromuscular adaptation due to poor stability rather than an instantaneous response that would likely show a variation in rotation at the point in the stride cycle following displacement of the tibia (Berchuck *et al.*, 1990).

What is unusual about the interaction between SB and DB groups with respect to the shift in rotation midpoint is that post-operatively, the SB group demonstrated an even greater disparity from the Control group than pre-operatively. In other words, the injured knees tended to exhibit slightly more external rotation than the healthy knees. The DB reconstructed knees' range of rotation then shifted internally with respect to the pre-operative state back to that of the normal knees, while the SB reconstructed knees demonstrated a *further* increase in external rotation range shift of approximately 10° for the cutting activity (Figure 6.12). This response was more pronounced for the high-demand activities than for walking.

Both of these observations indicate the involvement of the muscles around the knee used to stabilise the joint during dynamic gait, specifically, those involved in the ACL surgery. Several studies, including this one, have shown an increase in knee stiffness with ACL injury, described as an 'immature stabilization strategy' as observed by a decrease in maximum knee flexion angle and an increase in co-contraction of the muscles around the joint (Chmielewski *et al.*, 2005; Rudolph *et al.*, 2000, 2001). Co-contraction has been shown to unload the anterior cruciate ligament at higher flexion angles, thereby protecting the injured knee (Li *et al.*, 1999; O'Connor, 1993). Since the semitendinosus and gracilis tendons are in part responsible for internal rotation of the tibia, weakness of these muscles following harvesting of the tendons for the ACL graft may have resulted in an insufficiency in their ability to oppose the external torque produced by the biceps femoris during co-contraction (Viola *et al.*, 2000).

The return to normal of the DB range of rotation midpoint implies that this group did not require the same co-contraction strategy for stabilisation used by the SB group. A tendency to return to normal magnitudes of flexion and adduction in the DB group, while these rotations in the sagittal and frontal planes remained similar to pre-opeative values in the SB group (Figures 6.6 to 6.10), further supports this hypothesis.

Additionally, it is possible that subjects experiencing instability due to injury implemented a reprogramming strategy similar to that suggested by Berchuck *et al.* (1990). Since the ACL is secondarily responsible for restraint of internal torsional loads at the knee (Amis *et al.*, 2005; Blankevoort & Huiskes, 1996), a protection mechanism providing greater overall external rotation of the tibia may have been used to reduce the possibility of further injury that could occur with extreme internal rotation resulting from a ruptured or inadequate ACL ligament or graft. The significant shift towards external rotation observed in the SB group following ACL reconstruction may be a combined effect of the locomotor system adapting to an unstable joint and the inability of the compromised medial hamstring muscles to oppose the external torque produced by the biceps femoris during co-contraction.

6.4.4 Study limitations

One limitation of this study was that all (4 out of 22) female patients were randomly selected to receive the single-bundle reconstruction. It has been found that females suffer a higher incidence of ACL ruptures when compared to their male counterparts; however, knee kinematics were not found to be significantly different between men and women during side-step cutting tasks (McLean *et al.*, 1999; Sigward & Powers, 2006).

Nonetheless, to ensure gender did not influence our findings, additional analyses were carried out to compare test session by rotation range midpoint interaction between the female and male subjects for each activity. Female subjects displayed the same movement toward greater external rotation following ACL reconstruction as male subjects and no statistical differences were found between gender subgroups. (All p-values were greater than 0.31.) We therefore, concluded that this did not influence our overall results.

The mean follow-up period for the DB group was approximately two months longer than the SB group, which may have permitted further healing of the graft in these subjects and differences in the outcome. However, two months is a negligible duration when compared with the overall time of several years required for recovery from this injury with some patients *never* returning to a pre-injury sense of joint stability (Risberg *et al.*, 2004).

Furthermore, if the findings may, in part, be explained by muscle co-contraction and weakness at the donor site, one must take into account the time required for healing of the harvested tendons. In an intra-operative investigation, Ferretti *et al.* (2002) found distinct differences in the collagen fibre bundles of the regenerated semitendinosus tendon at 6 versus 24 months post-reconstruction, indicating that two years or longer are required for complete regeneration. Viola *et al.* (2000) moreover observed significant reductions in internal tibial rotation strength over four years post-operatively. The two-month difference in the follow-up testing period between groups is, therefore, considered clinically negligible. In order to verify this deduction and to additionally determine long-term differences in outcome between these two surgical techniques, further follow-up testing should be conducted at least two to five years post-operatively.

Soft tissue artefact is a common concern when interpreting results in gait analysis. In a recent study in which kinematics from skin markers were compared with those calculated from bone-pin markers, Benoit *et al.* (2006) illustrated up to 4.4° and 13.1° differences for walking and cutting activities, respectively. In this study, an optimization algorithm (OLGA) was used to minimize the effects of soft tissue motion especially in the determination of transverse plane rotation (Charlton *et al.*, 2004; DeGroote *et al.*, 2008; Roren, 2005). Most importantly, the same methods were implemented for all subjects (those receiving single and double-bundle reconstructions) at both pre- and post-operative testing sessions. The statistically significant interaction established between intervention groups and sessions, therefore, could not have been a result of soft tissue artefact.

6.4.5 Conclusions

In this study, the outcome of single and double-bundle reconstruction of an isolated ACL rupture was investigated during dynamic activities. The 3D knee kinematics measured during walking, cutting, and jumping compared well with those of similar activities reported in the literature; high-demand activities revealed greater differences between the healthy Control and ACLD knees pre-operatively, as well as the SB and DB groups post-operatively. Although no significant differences were found between Control and ACLD transverse plane rotations, findings in all three planes suggested the use of a protection mechanism by the injured patient group. The observation that contralateral knee kinematics followed the same directional shift in rotation following ACL reconstruction as the injured knees supported the theory by Berchuck *et al.* (1990) that the locomotor process is reprogrammed to adapt to the deficient joint.

The hypothesis that range of rotation would be affected by ACL reconstruction was not confirmed by this study. However, the interaction of the rotation range midpoint between the SB and DB groups from pre- to post-operative testing sessions indicated persistent laxity in the SB group, while the DB three-dimensional kinematics returned closer to those of the healthy control subjects.

The possibility that a co-contraction stabilization strategy was used with subsequent graft harvest site weakness contributing to changes in transverse plane kinematics may be an important consideration for post-operative rehabilitation. The external rotation shift demonstrated by the SB group following reconstruction (and to a smaller extent by all ACLD subjects) could have significant implications for long-term joint degeneration as different structures within the joint will experience compressive or shear forces that previously were unloaded; concomitant unloading of other tissues will occur in other areas of the joint.

Additional long-term follow-up studies are required to determine the effects on joint laxity, donor site morbidity, and joint degeneration following single and double-bundle ACL reconstruction for a more comprehensive comparison of these surgical techniques; however, these preliminary *in vivo* results under physiological loading conditions indicate improved joint constraint following double-bundle reconstruction.

Chapter 7

Conclusions and recommendations

7.1 Rotational laxity outcome: Does the double-bundle ACL reconstruction provide better restraint than the single-bundle technique?

In this thesis I have addressed the question of whether the double-bundle (DB) reconstruction of a deficient anterior cruciate ligament (ACL) is more advantageous in providing rotational restraint at the knee than single-bundle (SB) surgery. The term '*rotational* laxity' used in the literature is somewhat ambiguous. It has been interpreted as the subjective degree of instability as assessed by a clinician, resulting from a load that incorporates a torsional element, such as the pivot shift test. The instability measured by this test is not confined to transverse plane rotation, however. Rotational laxity has also been assessed in terms of tibiofemoral internal-external rotation. While the kinematics ensuing from daily physiological activities *may* be comprised of a certain degree of axial rotation, the loading conditions may not necessarily incorporate a torsional component at all.

In evaluating the outcome of the two surgical techniques, our measure of laxity focussed on the transverse plane rotation occurring at the tibiofemoral joint; the applied loading conditions varied, however, in the two studies involving ACL-deficient patients. In the first study (Chapter 5), we wished to compare the proficiencies of the single and double-bundle reconstruction against the ACL-deficient and healthy knee to restrain a known isolated torsional load. Previously existing methods of assessing rotational laxity at the knee were judged to be inadequate to accurately and objectively measure knee kinematics under these specific loading conditions. An innovative device was therefore designed to apply a precise static torque about the long axis of the tibia, incorporating a 'relaxation' period to ensure minimal depreciation of applied load once the foot position was fixed. With the custom-built apparatus keeping the knee position stationary under load, the knee was imaged using a magnetic resonance imaging (MRI) scanner, thereby avoiding soft tissue artefact and providing reliable kinematic data (Chapter 3). Knowing that the complex structure of the joint may yield motion in more than one degree-of-freedom, the image analysis methodology was furthermore developed to measure rotations and translations in all three anatomical planes that could result from the applied torque.

The second study that assessed rotational knee laxity in ACL patients (Chapter 6), did so under dynamic weightbearing conditions. Although the specific loads applied at the knee via the ground reaction forces at the foot were not known or precisely controlled as with the first investigation, this study provided a means by which to compare the two surgical techniques under realistic loading conditions. The effects of not just the passive restraints of the knee were taken into account, but also the influence of compressive loads and muscle activation on the measured outcome.

Given the difference in loading conditions between the two clinical studies, it is not surprising that the findings were comparatively at odds: under passive torsional loading at 30° of flexion, the patients who had been allocated the double-bundle reconstruction demonstrated a trend toward less internal rotation than the contralateral uninjured knee, while the rotation measured in the single-bundle group was closer to normal. Alternately under dynamic conditions, while the range of rotation did not differ between the two groups, the transverse plane rotational alignment of the double-bundle reconstructed knees was significantly closer to that of the healthy control group than the transverse-plane position of the single-bundle reconstructions during the cut and jump activities.

In addition to the difference in loading conditions, the other important distinction between the two studies was the flexion angles at which rotation was being measured: under passive loading, tibiofemoral rotation was measured at 0° and 30° of flexion, while under dynamic loading during the high-demand activities flexion angles ranged from 10° to over 100°.

The results from these two studies are not necessarily contradictory and illustrate the importance of the different methods of assessing *in vivo* rotational laxity. To reconcile the outcomes, we must examine how the variations in study conditions could affect the results.

The intact ACL was better able to control the applied torsional load in only the extended position under passive isolated torque; at 30° of flexion there was no difference in either internal or external rotational laxity between the ACL-deficient and the contralateral knees under isolated passive loading conditions. The effect of surgical technique was only observed in the flexed position, however, where the DB reconstruction restricted rotation to a greater extent than both the SB reconstruction and the intact knee. Therefore, the addition of the second graft bundle was able to restrain (or perhaps overconstrain) a torsional load more than both the single-bundle graft and native ACL at a higher angle of flexion.

On the other hand, the divergent rotation shift displayed by the dynamic kinematics in the two patient groups following reconstruction was reasonably consistent over the entire activity cycle and was not restricted to a specific flexion angle. The shift in the knees of the SB group away from the normal rotational alignment following reconstruction, which suggested inferior kinematic constraint when compared to the pre-operative state, was evidence that the graft donor site – compromised with surgery – was involved in providing joint restraint. This, therefore, gave an indication (albeit speculative) that an active stabilisation strategy involving the hamstrings was used by the patients in the SB, but not the DB group under dynamic loading conditions over the range of flexion. A greater sense of security in the DB patients would make any additional co-contraction of the hamstrings unnecessary, resulting in the observed rotational shift towards the normal control group. Further investigation via electromyography could objectively assess this hypothesis.

Since the passive loading investigation demonstrated that there was no difference in rotational laxity with ACL rupture compared with the intact knee at the greater flexion angle, the sense of stability perceived by the DB patients under physiological loading conditions may be attributed to the capability of the double-bundle graft to restrain those forces and moments accompanying torsional loading during dynamic tasks, such as compressive axial force, varus-valgus moments, and anterior-posterior forces. This theory is again supported by the findings of the passive loading study at full extension in which the ACL and other rotational restraints are in greater tension (Amis & Dawkins, 1991; Blankevoort *et al.*, 1991). In this extended position, the ruptured ACL *did* cause an increase in laxity and the reconstruction improved transverse plane restraint. Similarly, with additional loads at the knee that would strain the supporting structures including the ACL during dynamic tasks, the effect of an inferior graft would become apparent.

Axial compression in the absence of any additional external forces at the knee joint has been shown to cause both anterior tibial translation and transverse plane rotation (Liu-Barba *et al.*, 2007; Meyer & Haut, 2008). The augmented ACL strain and further increase in anterior and rotational laxity that has been observed in the ACL-deficient knee under joint compression (Fleming *et al.*, 2001; Liu-Barba *et al.*, 2007; Meyer & Haut, 2008) is consequently logical.

The magnitude of rotation that resulted from isolated compression was less than 10°, however, even at the point of catastrophic failure in the cadaver specimens (Liu-Barba *et al.*, 2007; Meyer & Haut, 2008). This degree of transverse plane rotation, still within normal limits of knee motion, would not cause damage to the ACL or other structures without coupled motion in another degree of freedom. Mean anterior translation at failure under compressive loading was found to be 27 ± 15 mm (Meyer & Haut, 2008) and likely provided a greater contribution to ACL strain than did rotation under axial loading condtions. This anterior tibial displacement is ascribed to the anterior component of the compressive force resulting from the posterior slope of the tibial plateau (Blankevoort & Huiskes, 1996; Liu-Barba *et al.*, 2007; Meyer & Haut, 2008).

With concomitant loads at the knee such as internal or external torques, however, joint compression has actually resulted in decreased rotational laxity when compared with isolated torsional loading conditions; the geometry of the tibiofemoral contact surfaces were thought to provide this additional restraint to the joint (Blankevoort & Huiskes, 1996; Wang & Walker, 1974). Activation of the muscles around the joint provide further stability, not only through the additional stiffness supplied by the muscle fibers, but also by taking advantage of the frictional and normal force contribution of the joint contact surfaces with joint compression (Wang & Walker, 1974). Li *et al.* (1999) demonstrated this experimentally in 10 knee specimens; a decrease in ACL force with co-contraction of the quadriceps and hamstrings was observed when compared with the isolated quadriceps force during a simulated isometric extension of the knee.

During dynamic activities such as the cutting task performed by our subjects, substantial joint moments occur in all three anatomical planes of motion (Sigward & Powers, 2006) and AP forces have been demonstrated during even low-demand activities such as walking (Andriacchi & Dyrby, 2005). While the combined forces at the knee would have undoubtedly added stress to the intact ACL or reconstructed graft in the absence of axial compressive loading, under weightbearing conditions with muscle co-contraction it would not be unreasonable for the measured range of rotation to be equal to or less than the range under isolated loading conditions.

These distinctions have been described as the 'operating point' and 'limits of passive stability' (Tashman *et al.*, 2004); in our subject groups the range of passive transverse plane rotation at 30° of flexion was approximately 25° (Figures 4.2, 5.2, and 5.3), while the range of dynamic rotation over the cutting activity cycle was generally between 20° and 30° with the maximum and minimum rotation peaks typically occurring at or above 30° of flexion (Table 6.2 and Figures 6.4, 6.7, 6.8, 6.12, and 6.14). Given the consistency in range of rotation across subject groups and test times during the dynamic activities, the perceived instability in the SB as compared to the DB group was conceivably due to motion other than axial rotation. The observed external rotation shift in the SB group was therefore not due to the inefficacy of this surgical procedure to restrain rotation, but rather an indirect consequence of joint laxity caused by ACL graft deficiency and the co-contraction stabilisation strategy employed to control it.

The regularity of the operating range of rotation under dynamic loading is highlighted again when comparing it with the asymmetrical limits of passive laxity in left and right knees of the healthy control group under isolated internal and external torque (Chapter 4). With the knee in flexion, a substantial degree of 'secondary' rotation - easily up to or beyond 10° in each direction – occurs with relatively small (1 to 2 Nm) initial torque values (Musahl *et al.*, 2007; Wang & Walker, 1974). Rotation in excess of this initial laxity requires disproportionately more torque due to the non-linear stiffness of the ligaments (Woo *et al.*, 2006). While the difference in left and right passive rotation was statistically significant at 3° and 5°, this quantity may not be clinically relevant under passive loading conditions.

The neutral resting position of the subject in which the high resolution MRI scan was performed with the knee in full extension was *not* assumed to be at 0° of rotation in our study. Instead, the degree of neutral position rotation was subtracted from the torqued measure of rotation to calculate the net rotation under load. At full extension, the degree of secondary laxity is probably less than 10° in each direction, however, it is possible that the 5° left-right difference in total range of rotation could nonetheless be attributed to an imbalance in the neutral knee position, which is not clinically relevant within this range of secondary laxity.

The divergence in the rotation shift between the SB and DB groups during the post-operative testing session of the cutting activity was approximately 14° (Figure 6.12), well beyond the asymmetry observed in the healthy control subjects under passive loading. Combined with the relatively consistent range of rotation under dynamic loading conditions, this rotation shift becomes even more clinically prominent.

The hypothesis that the difference in joint restraint following SB or DB reconstruction is in their capacities to constrain combined loads, rather than simply rotational laxity, is also supported by the studies that have evaluated the two procedures using the pivot shift test. The evidence in the literature review was ambiguous when all studies with conclusions reporting measures of 'rotational' laxity were assessed. However, when segregating only those studies that used a pivot shift to evaluate joint laxity, the evidence was clearly in favour of the double-bundle procedure with Järvelä (2007); Kondo *et al.* (2008); Muneta *et al.* (2007); Siebold *et al.* (2008); Yagi *et al.* (2007) finding better outcome in the DB group, while only Streich *et al.* (2008) found no significant difference between the two techniques and Markolf *et al.* (2008b) demonstrated the potential for overcorrection of joint laxity with the DB reconstruction.

Due to the difficulty in quantitatively evaluating the pivot shift test, no *in vivo* study has quantitatively compared SB and DB reconstructions by the associated degree of rotation. It has moreover been demonstrated that the combined effects of valgus and internal rotational moments on ACL strain are greater than either load in isolation (Shin *et al.*, 2005). Therefore, the conclusion that the DB reconstructive technique provides superior 'rotational' stability due to better pivot shift outcome is inaccurate.

The illustration by Pearle *et al.* (2008) of the AM and PL bundle obliquity throughout the range of flexion (Figure 7.1) may shed some light on both the reason for the apparent improved constraint of the DB technique under combined loading situations, as well as the outcome of the isolated torsional load investigation. The bundles are parallel throughout the first 30° of flexion (Jordan *et al.*, 2007; Pearle *et al.*, 2008). Their contributions to the restraint of isolated torsional loading would, therefore, theoretically be comparable and may wholly account for the equal reduction of internal rotation in both the SB and DB reconstructions in the extended knee position (Chapter 5).

With greater knee flexion, the angles of the two bundles were found to change non-uniformly in the three anatomical planes (Jordan *et al.*, 2007; Pearle *et al.*, 2008). The more oblique orientation of the PL bundle in the transverse plane is used to explain its enhanced ability to restrain rotation (Blankevoort & Huiskes, 1996; Pearle *et al.*, 2008). However, the difference in AM-PL angle obliquity in the sagittal plane is more substantial than in the axial plane (Figure 7.1) and would theoretically favour an AM bundle graft reconstruction – rather than the PL bundle – in rotation restraint depending on the location of the axis of rotation. The more vertical orientation of the PL bundle in the coronal plane would furthermore better resist joint distraction or varus-valgus rotation.

The overconstraint of rotation that was observed in the DB group when a passive internal torque was applied to the knee in the flexed position would have

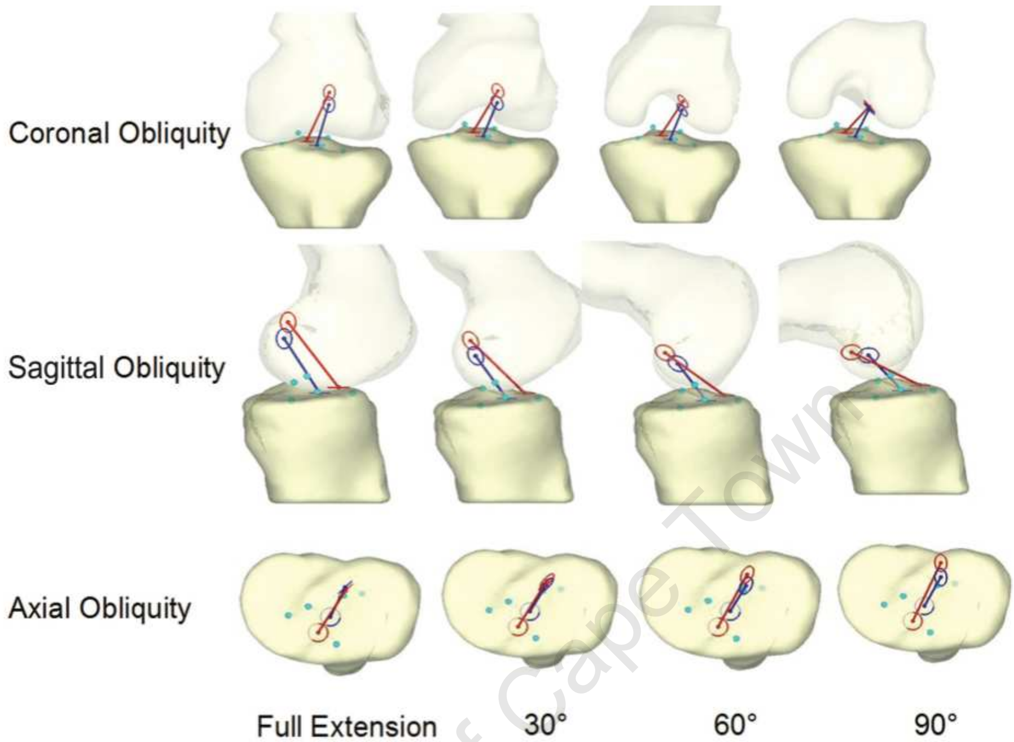


Figure 7.1: Anteromedial and posterolateral bundle obliquity in the three anatomical planes at four angles of knee flexion (Pearle *et al.*, 2008).

occurred at the point at which the AM-PL bundle orientation changed from parallel to oblique with respect to one another. Although the PL bundle loosens to a greater degree than the AM bundle throughout the first 30° of flexion, internal rotation of the tibia counteracts this slackening to a certain extent (Amis & Dawkins, 1991). The additional tension that would be provided by the PL bundle during internal torsional loading would actually have been in a more vertical direction when viewed in the sagittal or coronal planes. In other words, the addition of the PL bundle would essentially pull the tibia and femur closer together than the SB reconstruction could. The distal-proximal direction of force may then have permitted the congruency of the joint contact surfaces to further contribute to joint constraint at the higher angles of flexion experienced throughout the dynamic activities.

If the proposed theory that the improved overall joint constraint is a result of supplemental PL bundle tension normal to the tibial surface is in fact correct, it is still uncertain whether the tension in each graft bundle is similar to that of the native ACL bundles or moreover, whether the graft strain is at a healthy level. It is believed that the contribution of the native PL bundle to joint constraint is greatest at low flexion angles because this bundle slackens to a greater extent than the AM bundle with knee flexion (Markolf *et al.*, 2009; Yagi *et al.*, 2002). If reconstructed grafts are tensioned differently from their native counterparts, the effects on joint stability may appear to be similar under certain loading conditions, but actually have undesirable consequences. It is possible, for example, that excessive tensioning of the *anteromedial*, rather than the additional posterolateral bundle, may account for the observed trend towards overconstraint in the DB group under passive internal torque in the flexed position (Chapter 5), while the lack of vertically directed (distal-proximal) tension in the SB group could simultaneously account for the laxity under dynamic loading conditions (Chapter 6).

Most studies to date have measured ligament strain as an indicator of tension *in vivo* without knowing the initial recruitment length or actual ligament tension. Although equivalent force was employed in both SB and DB surgical techniques to pull the grafts taut, the direct effect of the order of bundle fixation and angles at which the tension was applied on overall graft tension is unknown. Furthermore, after several months of healing and physiological strain, tension in the individual bundles may have changed. In the absence of the ability to measure the force of each native and reconstructed graft bundle directly, it can only be theorized that extreme tensioning of one or both bundles during the DB technique brought about the reduced rotation relative to the contralateral knee under isolated torque in the flexed knee position.

7.2 Research implications and recommendations for the future

The findings presented in this thesis specifically concern the capacity of two techniques of ACL reconstruction to restore *rotational* restraint to the joint *in vivo*. The broader questions are *'Does the double-bundle technique restrict joint laxity* *to a greater extent than the single-bundle technique? If so, how?'* The results from our dynamic study suggest that the DB technique does, in fact, constrain the joint more adequately than the SB reconstruction, allowing patients to apply normal knee kinematics during physiological weightbearing tasks. However, the outcome from the passive torsional loading study indicates that the superior restraint provided by the DB technique is not primarily in opposition to axial rotation, but to another direction of rotation or translation. We have therefore begun to answer the question of *'how?'* by finding that it is *not* as simple as stating that the DB reconstruction provides superior rotational restraint.

Fortunately, we have the means to go back to the drawing board to find the answer to the question of how the DB technique may improve patient outcome after ACL reconstruction. The MRI-compatible loading device has been designed to permit relatively simple modifications in order to apply static loading about or along another axis of rotation/translation. The methodology by which six degree-of-freedom motion can be analysed will facilitate investigations into motion in any anatomical plane under varying loading conditions.

The three-dimensional analysis under torsional loading in our healthy subject group has already provided valuable information with respect to coupled joint motion (Chapter 4); for example, the increase in flexion accompanying external rotation demonstrated a contrasting paired movement under torsional loading from the typical screw-home motion observed with flexion-extension. Therefore, there are either independent or additional structures that control knee kinematics under these different loading conditions, or those structures that contribute to motion restraint behave differently with changes in loading direction. Once these differences are correctly interpreted, the changes in three-dimensional joint kinematics due to knee pathology such as ACL rupture may be more effectively measured. Ultimately, a better understanding of the six degree-of-freedom joint motion will allow surgeons to better predict and assess modifications made to existing surgical reconstructive techniques.

Not only is the measurement of joint motion becoming more accurate with the ability to track tibiofemoral movement in all three anatomical planes, but the number of journal articles describing aspects of surgical techniques, such as the precise three-dimensional location and orientation of the graft bundles, are also continuing to grow (Agneskirchner *et al.*, 2004; Doi *et al.*, 2009; Luites *et al.*, 2007; Zantop *et al.*, 2007). The need for revision surgery has primarily been attributed to incorrect tibial or femoral bone tunnel placement (Crawford *et al.*, 2007; Harner & Poehling, 2004); enhanced accounts of proper surgical technique may, therefore, already improve the outcome of ACL reconstruction without resorting to the latest unsubstantiated surgical trend. For less experienced surgeons, it is probably more valuable to accurately perform a simpler procedure rather than to attempt a more complex technique that may eventually require revision if the initial surgery is poorly executed.

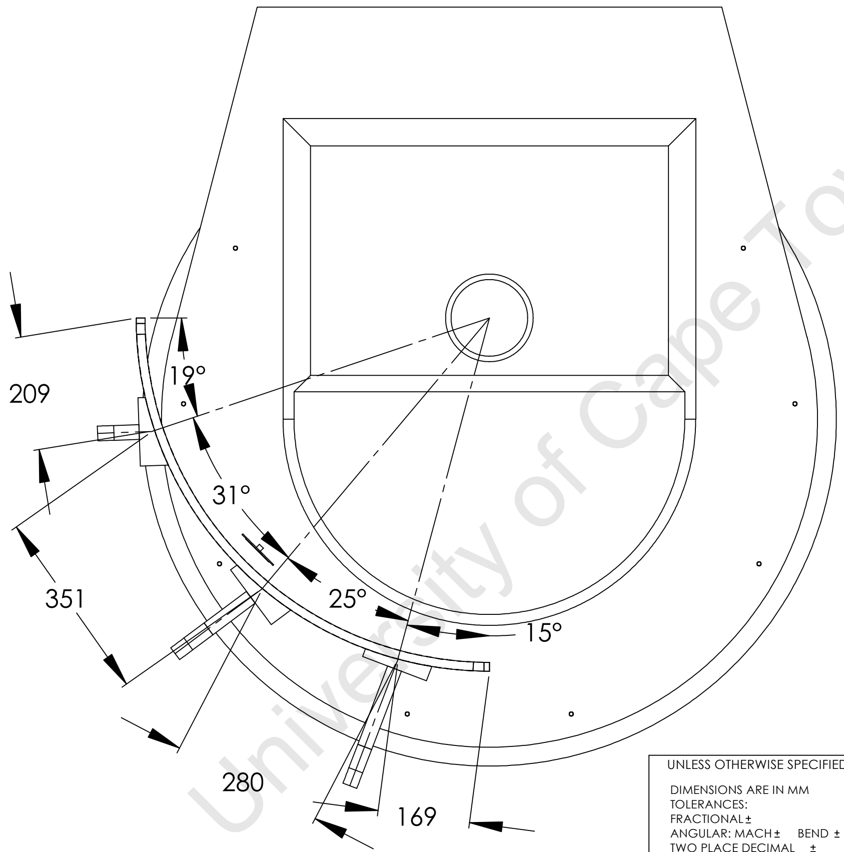
While our dynamic activity randomised control trial did demonstrate kinematics closer to normal in the DB group, the findings examined together with those of our passive torsional loading investigation indicate that the improved constraint is not simply about the tibial axis of rotation. Additionally, the disproportionate reduction in internal rotation in the DB group in the flexed, isolated torque condition may indicate an overconstraint due to excessive graft tensioning. Further research is required into the exact mechanism by which the DB technique may provide superior constraint, as well as the long-term effects this will have at the joint and on the reconstructed graft. Studies using inverse dynamics and electromyography to further investigate the contributions of the muscles on knee function will also add to our understanding of the benefits and detriments that could arise after either type of surgical reconstruction.

It is presumed that kinematics resembling the healthy knee joint, such as those measured in the DB group under dynamic loading conditions, will precipitate fewer long-term complications. However, it has been shown that joint degeneration following ACL reconstruction can be more extensive than in the ACL-deficient knee (Fu *et al.*, 2000; Kessler *et al.*, 2008). It is accordingly possible that a similar counterintuitive finding may be demonstrated with SB and DB reconstructions. Long-term follow-up studies comparing not just the SB and DB techniques, but also the effects of varying tunnel placement, graft tension, concomitant soft tissue injury, and any other factors that influence joint constraint should be undertaken to determine whether a specific technique is associated with a lower incidence of osteoarthritis or disabling joint disease.

This thesis presents new insights regarding the biomechanics of two techniques of ACL reconstruction using either a single or double-bundle graft. The different outcomes of the passive and dynamic loading studies demonstrate that conclusions about the ability of these reconstruction techniques to reduce rotational knee laxity cannot be based on one test alone. While the patients who received the double-bundle reconstruction showed improved performance compared with those who were allocated the single-bundle surgery, further investigation is required to determine whether the evidence of overconstraint of rotation may be detrimental to joint function in the future.

Appendix A

Technical drawings used for manufacturing main components of MRI-compatible torsional loading device



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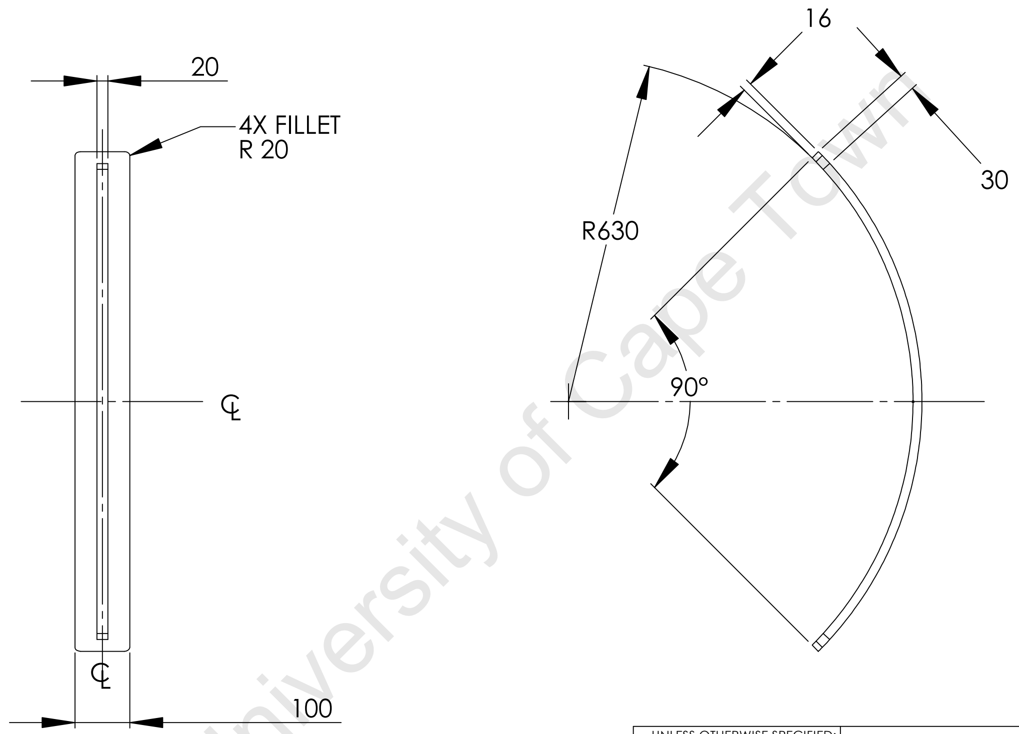

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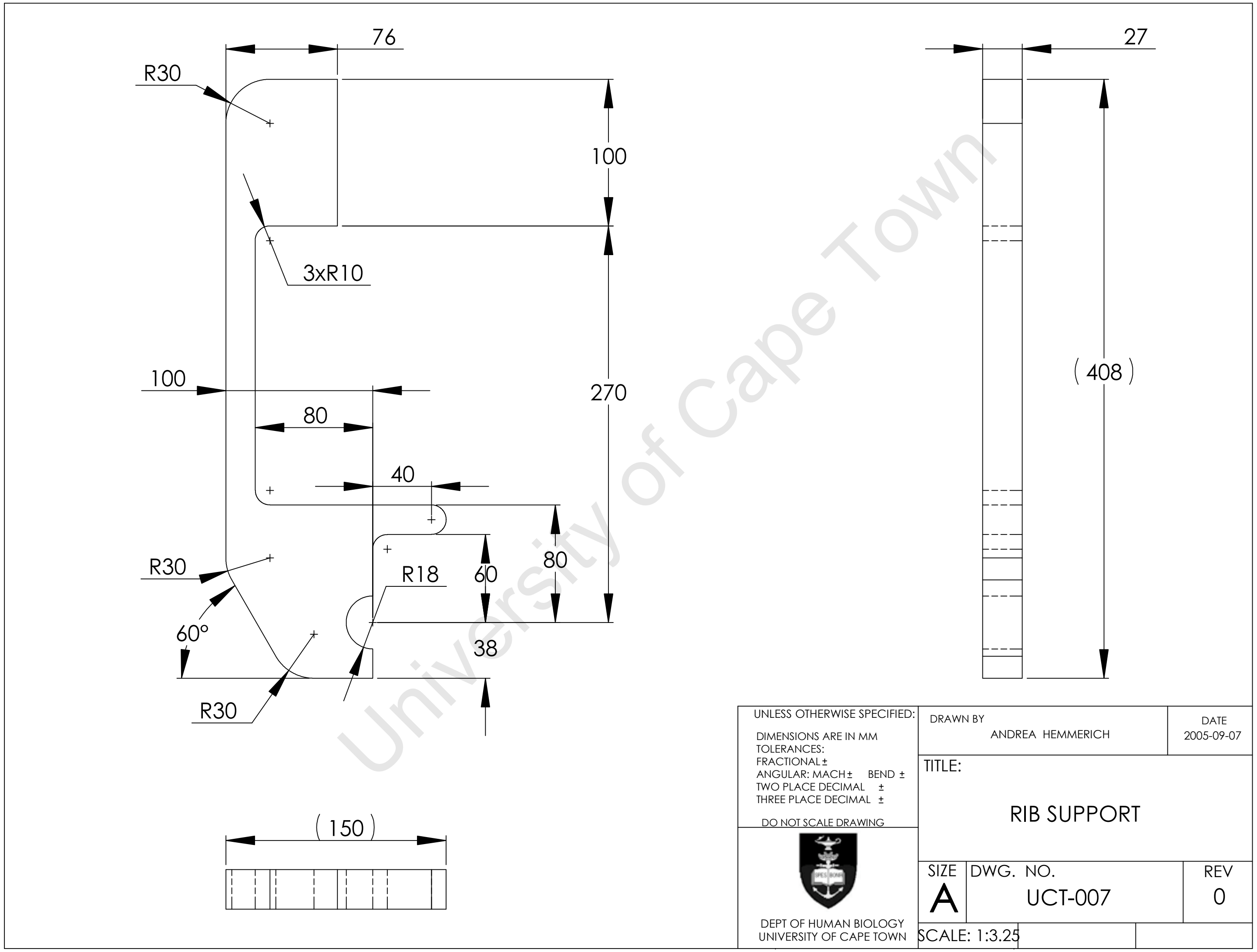
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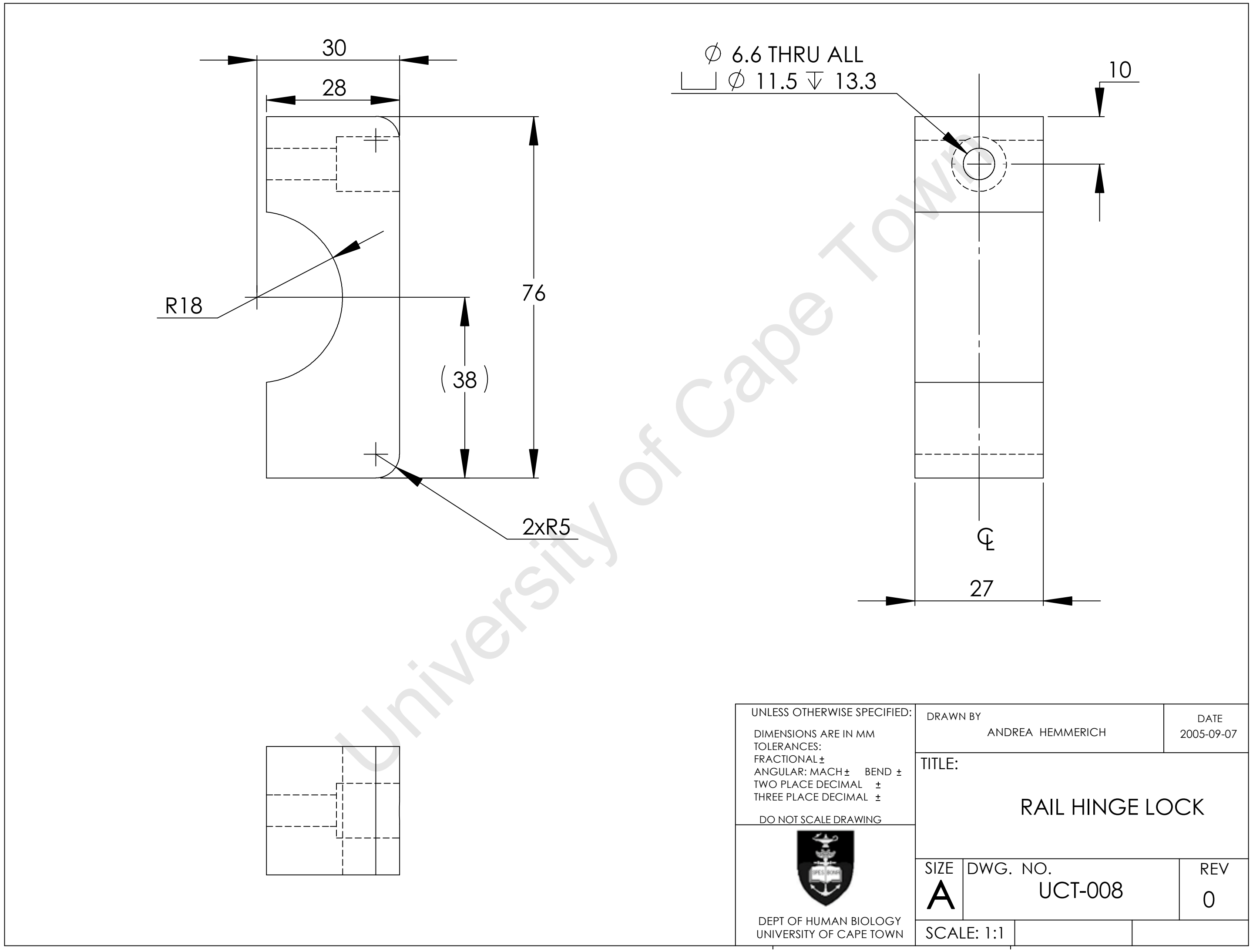
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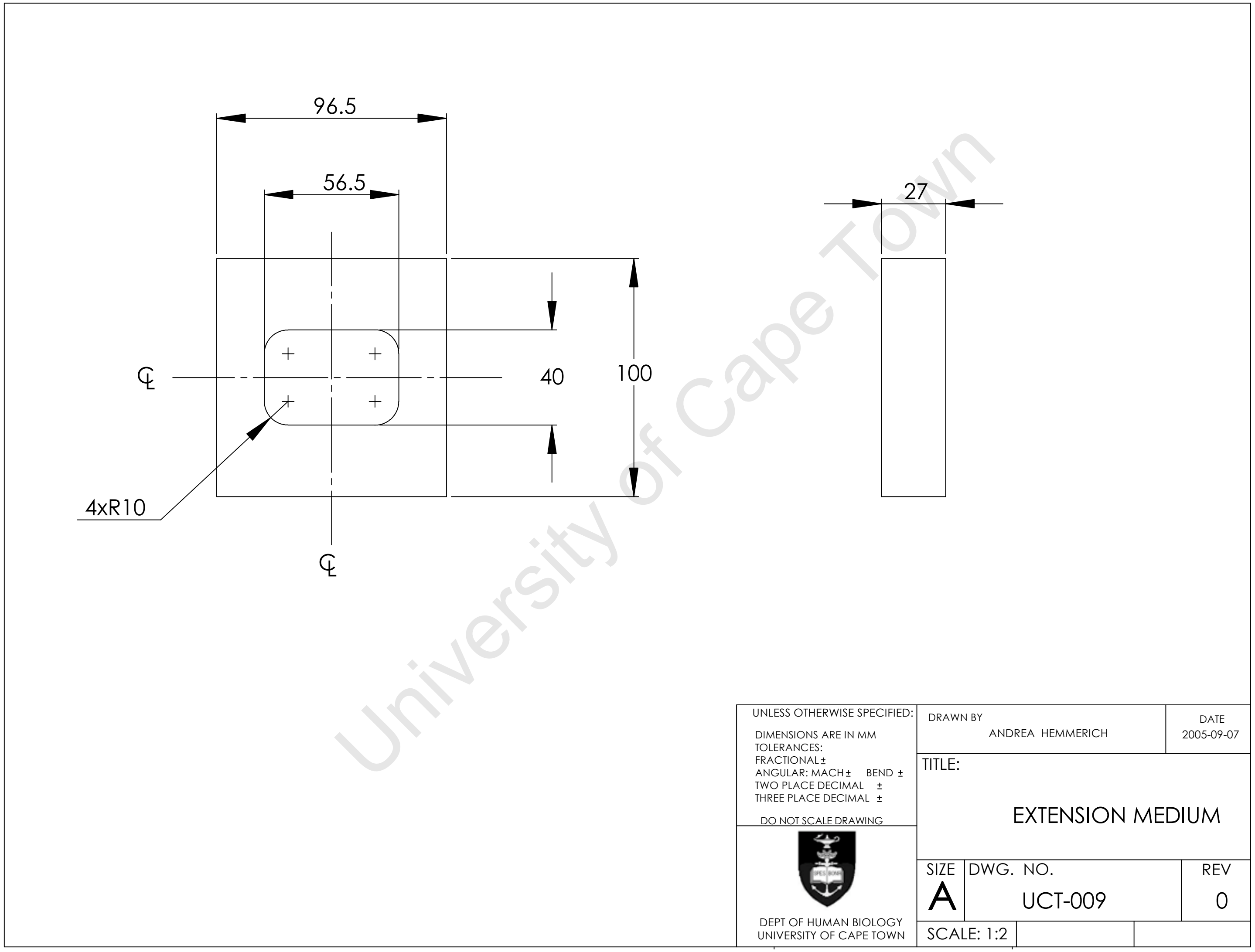
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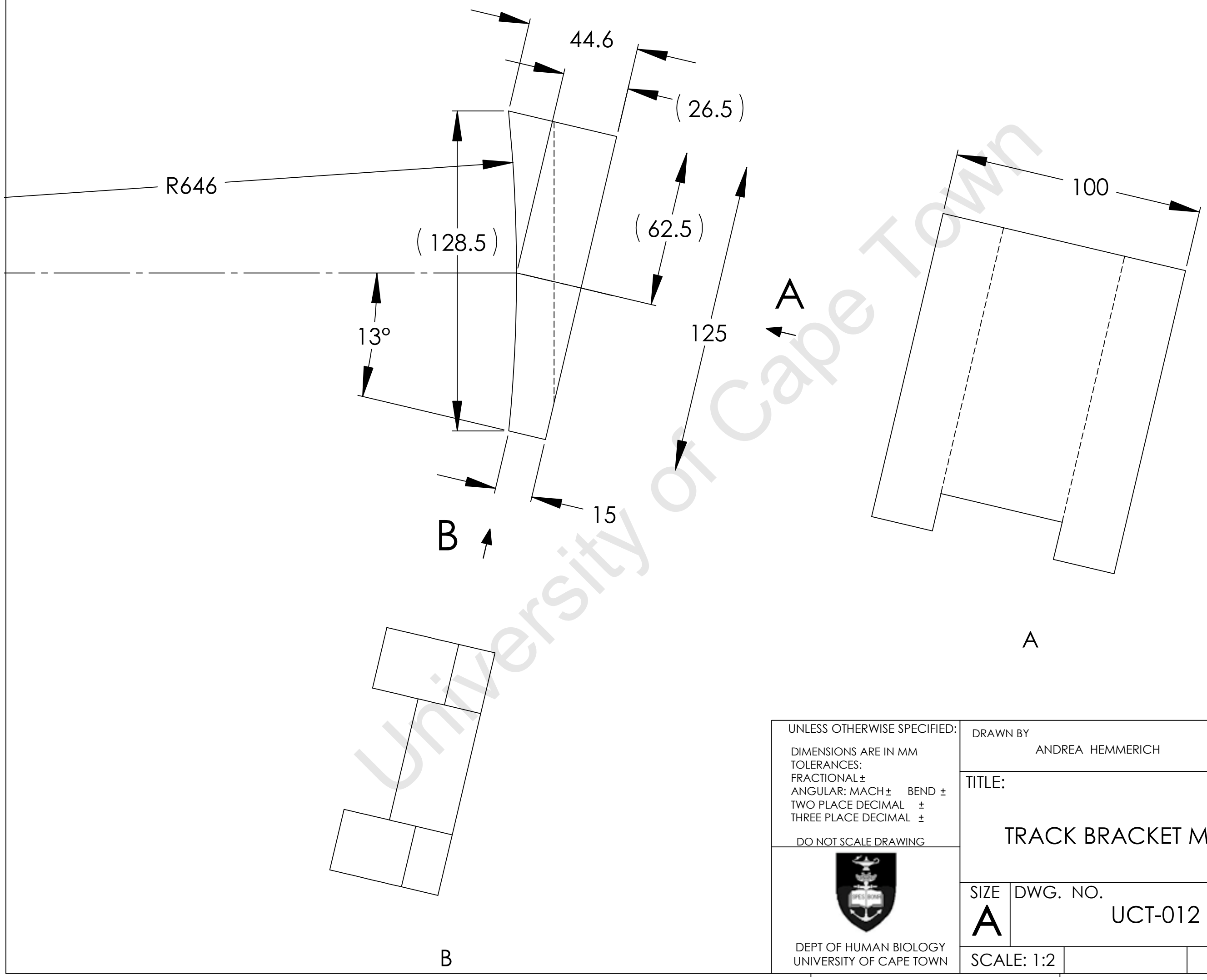

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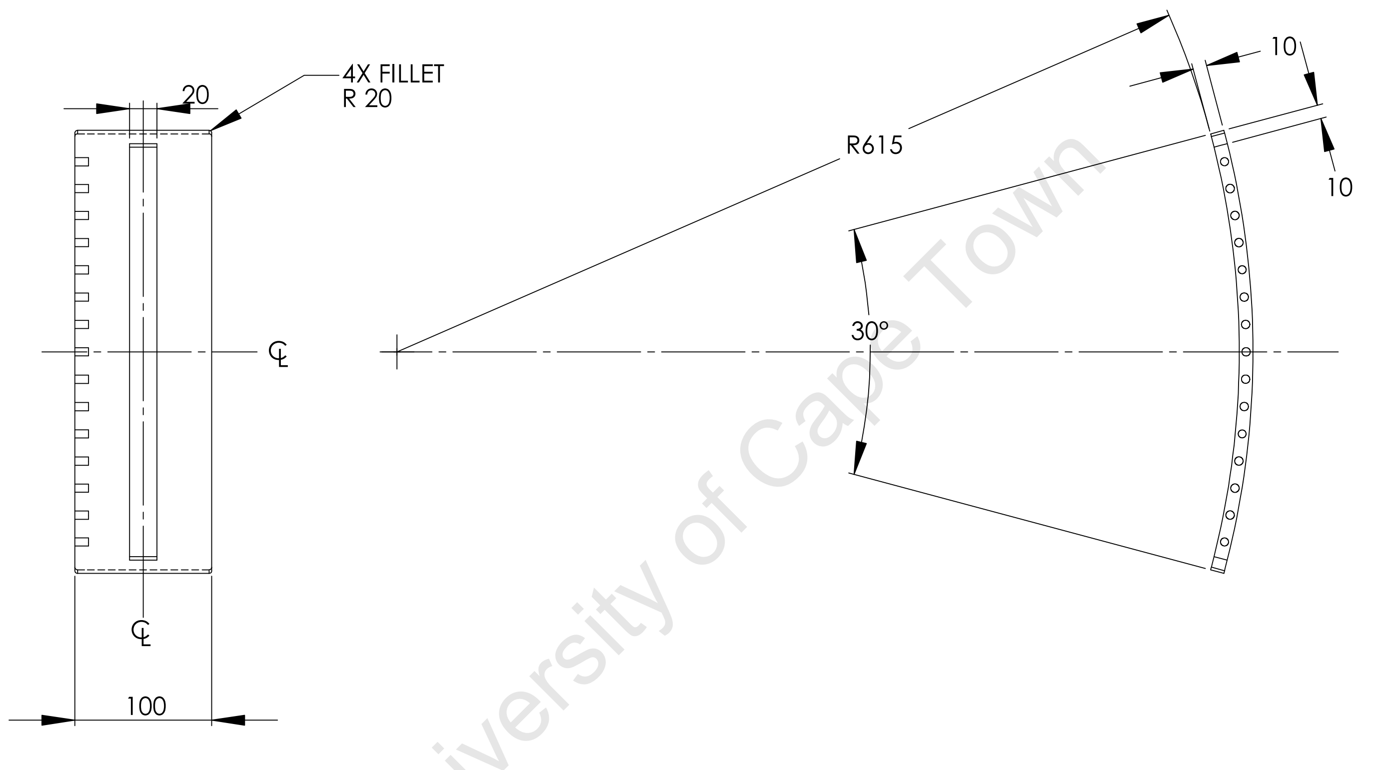

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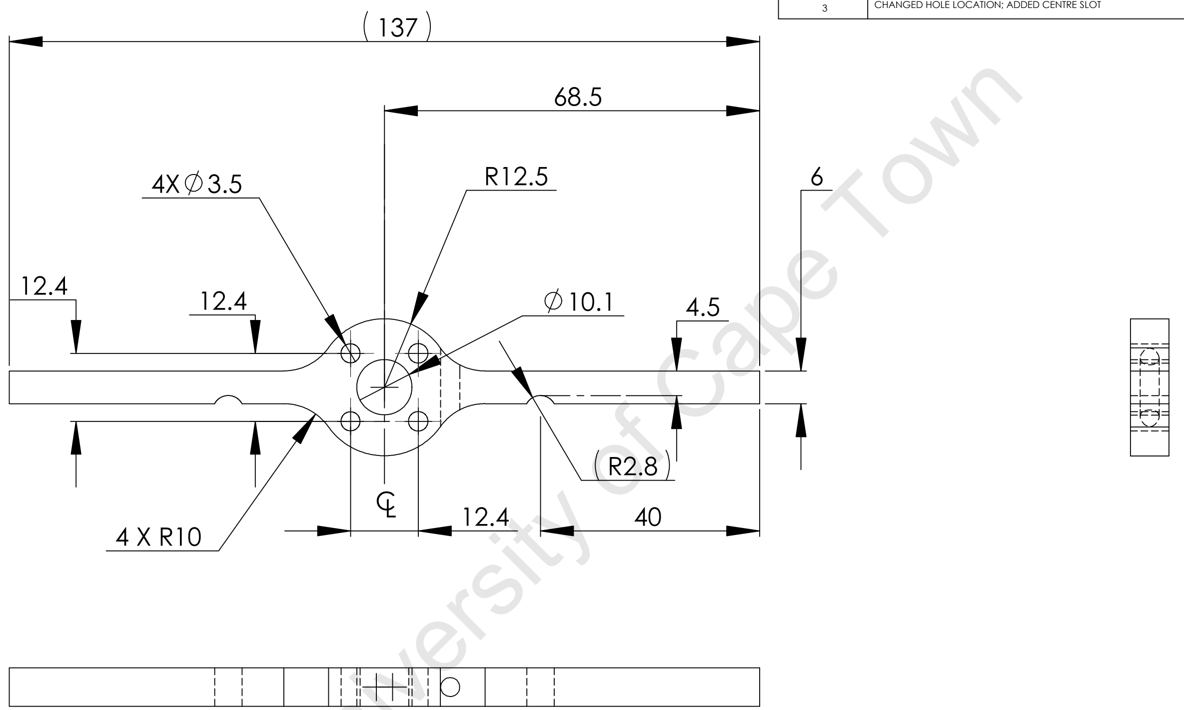

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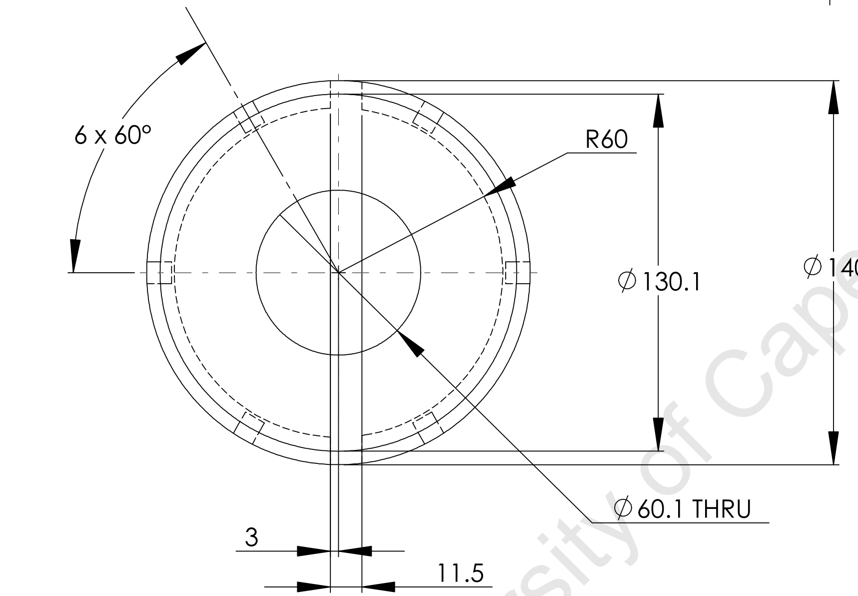

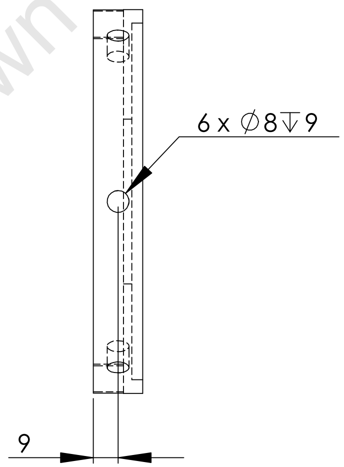

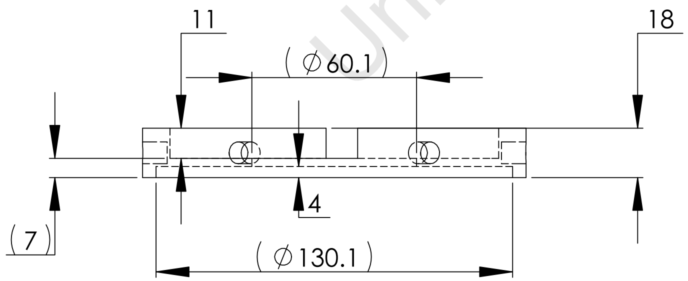

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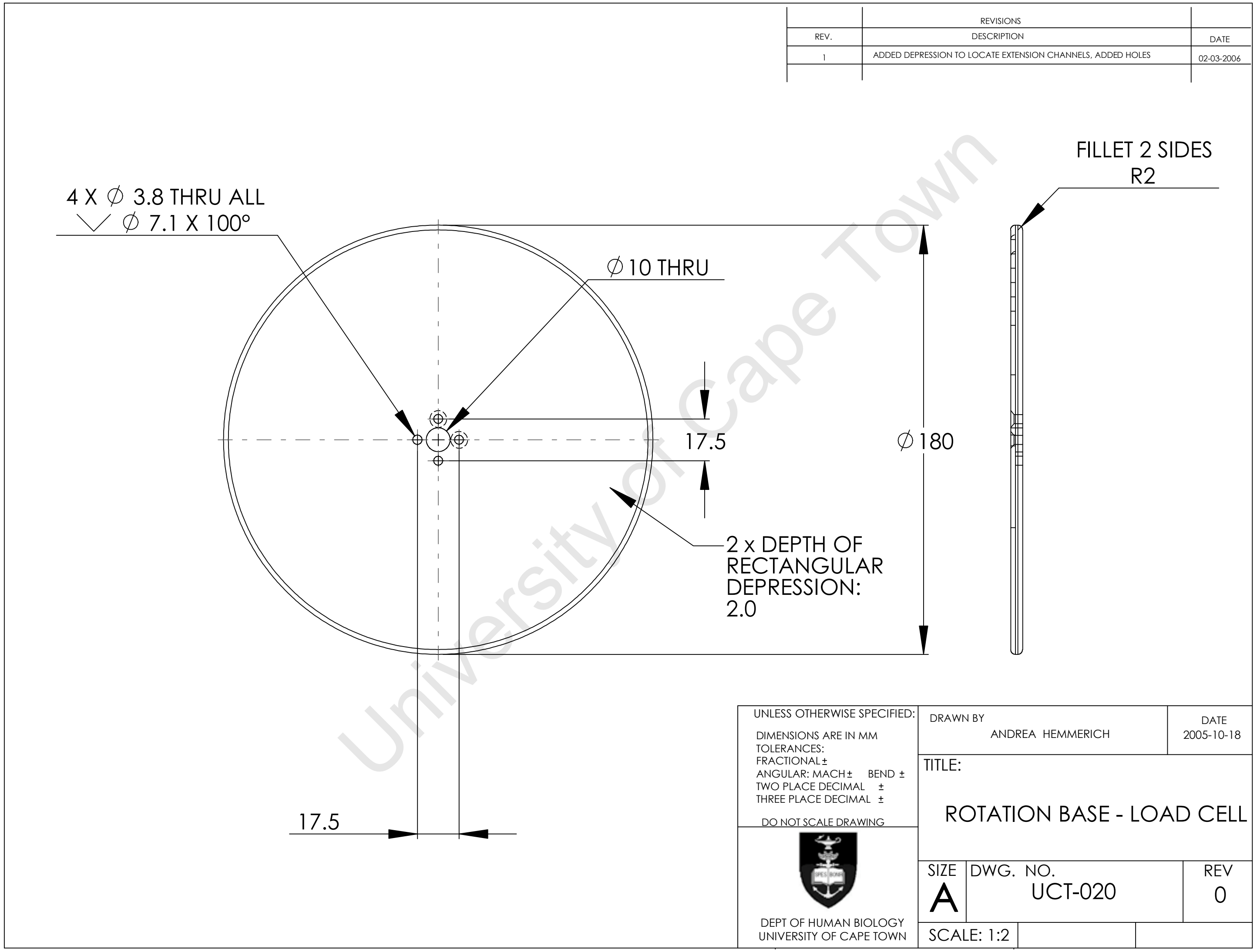
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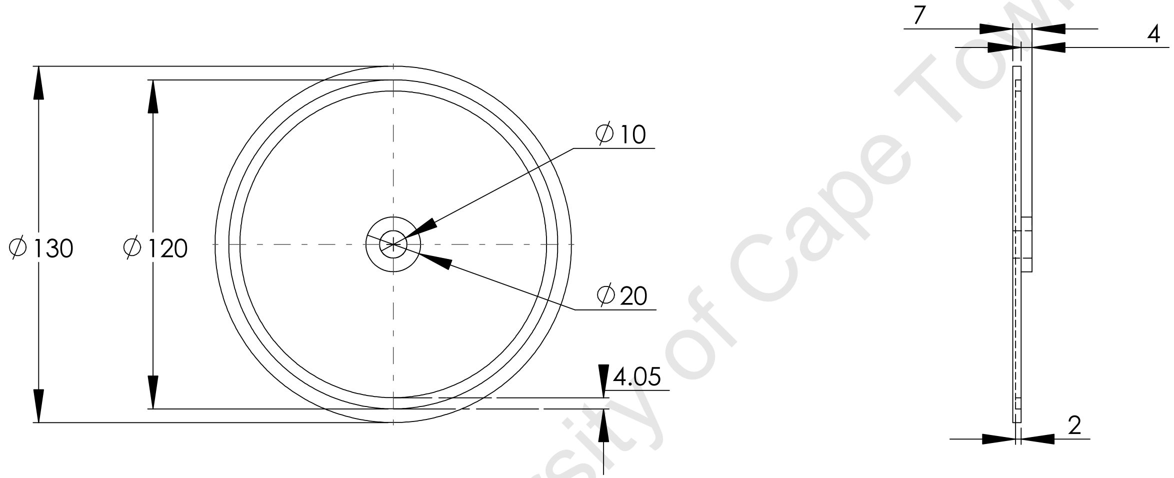

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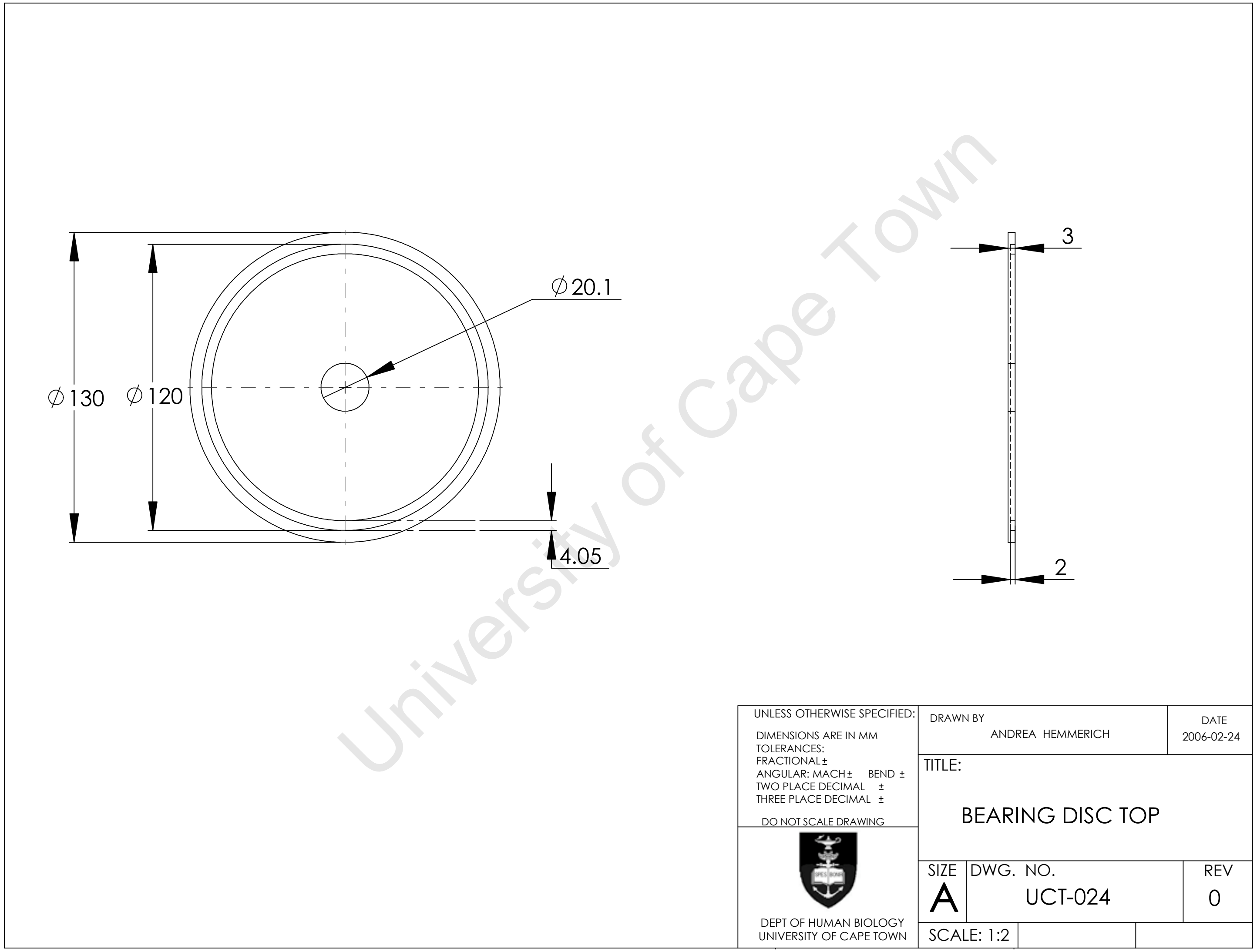

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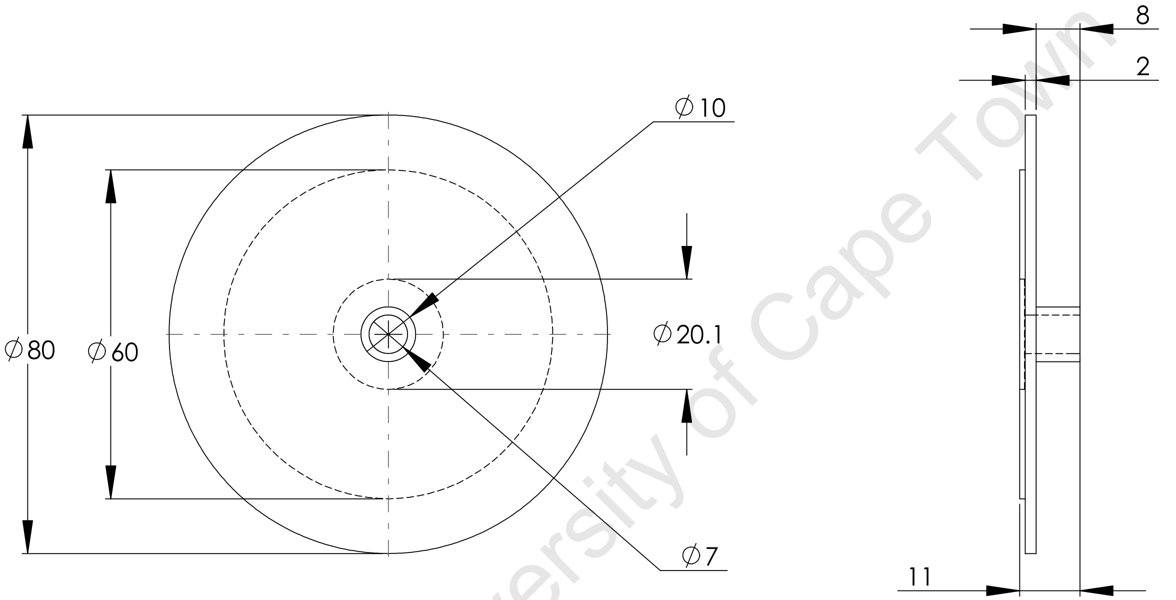

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DEPT OF HUMAN BIOLOGY
UNIVERSITY OF CAPE TOWN

DRAWN BY ANDREA HEMMERICH

DATE 2006-02-24

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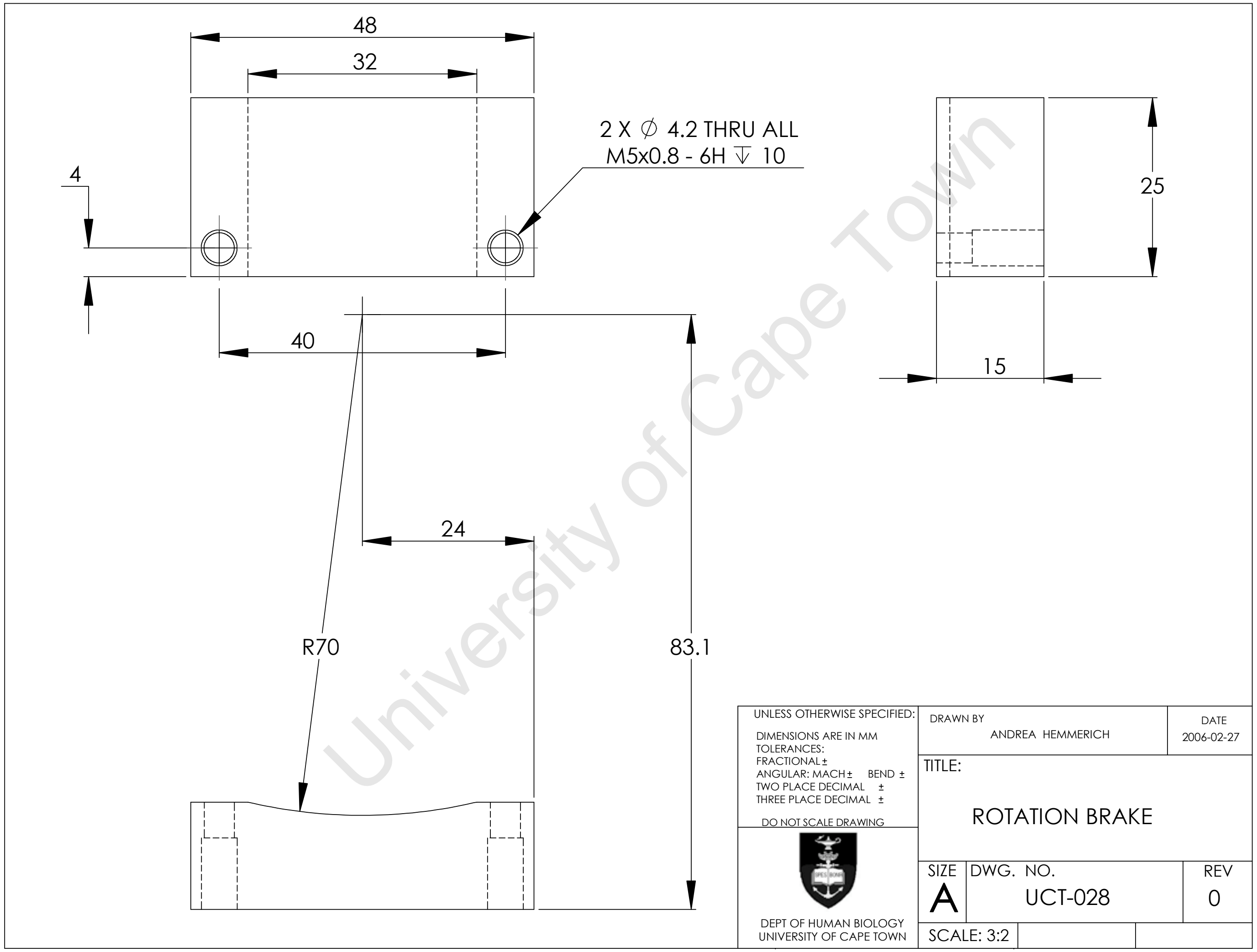
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DRAWN BY
ANDREA HEMMERICH
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Appendix B

Details of the strain gauge circuit used to measure torque

A strain gauge bridge circuit, as shown in Figure B.1, was used to measure the torque applied to the load cell. With rotation of the torque disc, a normal force was exerted at the end of the cantilevered load cell, thereby applying tension to the strain gauge and changing its resistance. The response to the mechanical strain was indicated by the voltmeter. A calibration procedure in which fixed weights applied known loads to the strain gauge, allowed the resulting voltage to be linearly correlated with the torque in LabVIEW^{TM}.

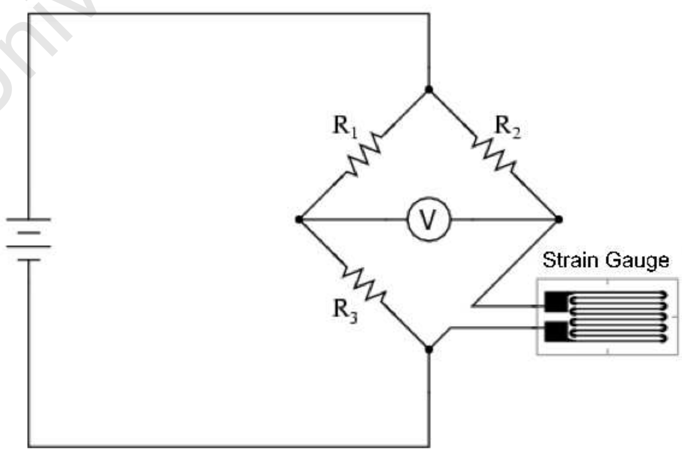

Figure B.1: A quarter-bridge strain gauge circuit was used to measure torque applied to the load cell.

While a half-bridge circuit (using two strain gauges and two resistors) is more typically used in this type of application to compensate for temperature change or other sources of resistance-induced error, a quarter-bridge circuit was considered sufficient for the purposes of this study since the LabVIEW^{TM} software had the capability of 'zeroing' the measured torque before the load was applied. Since the voltage-torque relationship was linear, the initial torque reading that was subtracted by the software did not affect the final measurement.

Appendix C

Torque relaxation measured at fixed rotation angle for one subject

Stress relaxation is used to describe the behaviour of viscoelastic materials such as the ligaments of the knee when subjected to a constant load over time (Woo *et al.*, 2006). In our study, a torque applied manually to the distal end of the shank was kept constant by locking the rotated position of the foot once the specified torque was reached. Figure C.1 shows the change of measured torque as the load was applied and then adjusted to meet the desired magnitude of 4.25 Nm calculated from equation 3.1 for a 60 kg subject. The specified torque was attained at approximately 100 seconds, at which point the rotation brake was applied; a spike in the signal at approximately 120 seconds was likely due to a muscle spasm and the subject adjusting to the applied load. The torsional load measured by the strain gauge was then recorded for an additional 7 minutes before the brake at the boot was released.

The regression curve fit to the data over the 7-minute period of constant loading resembles a typical stress-relaxation curve measured for joint ligaments (Woo *et al.*, 2006). In our experimental setup, the subject's thigh was free to move; consequently, the tibia could rotate with respect to the femur and the femur could move with respect to the pelvis. The observed relaxation, therefore, may reasonably be attributed to the soft tissues at both the knee and hip joints.

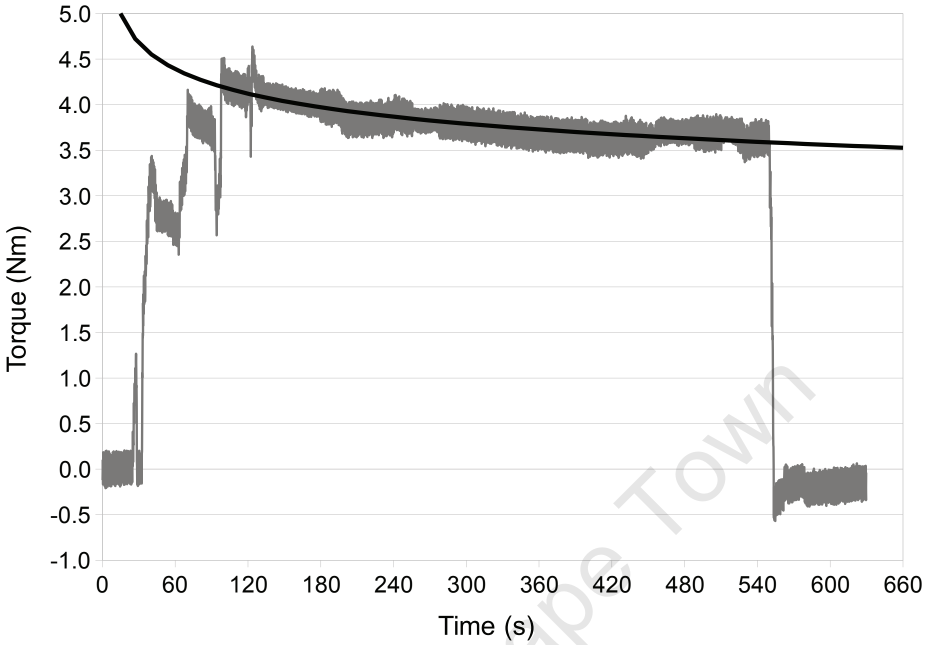

Figure C.1: Torque measured with manual adjustment of initial load and following fixation of rotation brake (shown in grey). Black regression curve displays general trend of time-torque curve, believed to be primarily a result of stress relaxation of the soft tissues and ligaments at the knee and hip joints.

The greatest decrease in measured torque occurred within the first 120 seconds of constant loading. Waiting approximately two minutes after applying the torque not only permitted relaxation of the soft tissues around the joints, but also allowed the subject to become accustomed to the load, thereby minimising the possibility of image artefact resulting from muscle activation and movement. Once sufficient stabilization of the measured torque value was achieved, the correct torque could be reapplied without significant reduction in load because the soft tissues had been 'pre-strained.' For this reason, the two-minute 'relaxation' period was incorporated into the protocol before MR imaging was performed.

Appendix D

Ethics approval letter and subject informed consent form

UNIVERSITY OF CAPE TOWN

Health Sciences Faculty
Research Ethics Committee
Room E53-24 Groote Schuur Hospital Old Main Building
Observatory 7925
Telephone [021] 406 6338 • **Facsimile** [021] 406 6411
e-mail: preaward@curie.uct..ac.za

15 November 2005

REC REF: 392/2005

Prof CL Vaughan
Human Biology

Dear Prof Vaughan

PROJECT TITLE: THE EFFECTS OF THE INTACT DEFICIENT, AND RECONSTRUCTED ANTERIOR CRUCIATE LIGAMENT ON ROTATIONAL STABILITY OF THE KNEE JOINT

Thank you for submitting your study to the Research Ethics Committee for review.

It is a pleasure to inform you that the Ethics Committee has **formally approved** the above-mentioned study on the 11 November 2005.

Please quote the REC. REF in all your correspondence.

Yours sincerely

PROFESSOR T. ZABOW
CHAIRPERSON, HSF HUMAN ETHICS

lemjedi

Informed Consent

I, ________________________________, agree voluntarily to participate in the research project titled "The Effects of the Intact, Deficient, and Reconstructed Anterior Cruciate Ligament on Rotational Stability of the Knee Joint" conducted at the Department of Human Biology of the University of Cape Town. The data for this project will be collected at the Sports Science Institute of South Africa (Cape Town).

The following procedures and concepts have been explained to me in full:

I. Measurement of Knee Instability using Magnetic Resonance Imaging (MRI) Compatible Device

1. The lower limb will be placed in a neutral position (full extension with no rotation). A high resolution knee scan lasting just over ten minutes will be taken in this position.
2. The lower limb will be positioned passively at a specific knee flexion angle within the magnetic resonance imaging (MRI) coil. A set torque (rotational stress) will be applied to the foot with the aid of a specially designed stress-testing device. The torque will be applied gradually by the investigator and can be stopped if any discomfort should occur.
3. The lower limb will be held in place for a period of approximately three minutes by the stress-testing device while an MRI scan is performed.
4. This procedure will be repeated for 4 rotated positions for each limb (i.e. 8 MRI scans in total).

II. Gait Analysis

5. Several retro-reflective markers will be placed on the thigh and shank of each leg.
6. You will walk with your customary gait along a 10 metre walkway.
7. During walking, your gait will be captured by a six-camera motion analysis system and ground reaction force plate.

III. ACL Reconstructive Surgery

There are two commonly performed surgical procedures to repair the ruptured anterior cruciate ligament (ACL), usually referred to as the "single-bundle" and "double-bundle" techniques. At present, no clear evidence exists to suggest that one technique has better

Knee Torque Study – Informed Consent page 1 of 3

results than the other; orthopaedic surgeons worldwide tend to use the procedure with which they are most comfortable and have the most experience. Dr. van der Merwe has over 10 years of experience performing both types of surgery on his patients. For the current study you will be randomly selected to have either one of the surgical techniques performed by Dr. van der Merwe. If he feels that there is a clear indication one way or the other that you should have a specific surgical procedure performed in order to best treat your injury, he will do so and you will no longer be a part of the study.

Procedures I and II will be performed twice over the course of the study: once prior to surgery and once about three months after the surgery (following sufficient time for rehabilitation).

Benefits/Risks

Risks associated with this study do not exceed those associated with normal clinical assessment by the patients' orthopaedic surgeon, Dr. Willem van der Merwe, or that of normal walking.

The information obtained in this study may or may not be of direct use to you. However, it will provide important information concerning the motion of the knee joint and may be of direct importance in the future for the improvement of anterior cruciate ligament reconstruction techniques. You would benefit directly from this information if you required an ACL reconstruction.

Voluntary Participation

Participation in this study is voluntary. You may withdraw at any time without prejudice to you in any way. You may also decline to participate without any negative repercussions.

Confidentiality

All information obtained during this study is confidential. The information obtained will only be available to the investigators involved in the study. The identity of subjects will not be disclosed in any published findings of the study.

Copy

You will be given a copy of this signed Consent Form for your own information.

Contact People

If you have any questions or concerns about the study, you may contact the following people.

Principal Investigator: Prof. Kit Vaughan, Faculty of Health Sciences, University of Cape Town	021 406 6238
Co-investigator: Andrea Hemmerich, Faculty of Health Sciences, University of Cape Town	021 406 6549
Co-investigator: Dr. Willem van der Merwe, Sports Science Orthopaedics Clinic	021 686 1196
Chair, Ethics Committee: Dr. Lesley Henley, Faculty of Health Sciences, University of Cape Town	021 658 5304

I have read the preceding form and understand the testing procedures outlined therein. I understand any accompanying risks and discomforts. Knowing these risks and discomforts and having had the opportunity to pose questions answered to my satisfaction, I hereby consent to participate in this study. I understand that I may withdraw from this study at any time without further question. I have been informed that the individual data derived from my participation in these protocols will remain confidential.

Signature of Subject:________________________ **Date:__________**

Signature of Investigator:____________________ **Date:__________**

Appendix E

Subject level and passive knee rotation data for 15 healthy Control subjects

Table E.1: Subject level data and range of rotation in extended and flexed knee positions for 15 Control subjects.

Subj	Sex	Age (yrs)	Height (cm)	Mass (kg)	Torque (Nm)	Range of Rotation Knee Extended Left	Knee Extended Right	Knee Flexed 30° Left	Knee Flexed 30° Right
1	M	31.8	174	73	4.9	17.1	25.2	23.1	21.0
2	M	29.3	182	75	5.0	20.0	18.8	18.0	20.0
3	F	29.7	157	54	4.0	19.3	20.2	25.4	24.0
4	M	23.8	175	62	4.4	11.7	9.0	36.5	33.1
5	M	25.9	171	75	5.0	7.5	8.5	19.9	12.5
6	F	37.6	170	62	4.4	10.0	9.0	21.8	23.3
7	M	38.0	170	79	5.2	22.4	17.4	24.4	25.7
8	F	29.8	166	64	4.5	21.1	21.9	29.4	30.3
9	F	31.5	158	50	3.8	16.3	12.5	27.4	29.2
10	M	34.0	183	90	5.8	16.5	9.2	27.4	22.0
11	M	25.8	181	76	5.1	19.5	14.3	27.5	24.5
12	M	22.2	181	85	5.5	4.7	7.6	21.8	20.3
13	M	43.0	176	70	4.8	19.6	17.5	28.5	25.8
14	M	24.1	183	80	5.3	15.8	12.8	35.0	22.9
15	M	28.1	190	80	5.3	10.3	10.3	18.2	16.8
mean		30.3	174.5	71.7	4.8	15.8	14.6	26.2	23.9
SD		5.9	9.3	11.3	0.6	5.4	5.6	5.5	5.2

Appendix F

Primary and secondary outcome results for randomised control trial under passive torsional loading

Table F.1: Means, standard deviations, and p-values for measured rotation (in degrees) of subject groups in the primary outcome analyses for the passive rotational laxity study.

						p-value	
Loading Condition	**Analysis**	**Subject Group**	N	mean	SD	Mixed Model	Paired t-test
Extended Ext Torque	Test Session	PreOp	27	9.5	3.5	0.712	
		PostOp	28	9.5	3.8		
	Interaction	SB PreOp	14	9.4	2.9	0.930	
	Test Session	SB PostOp	16	9.4	3.3		
	by	DB PreOp	13	9.6	4.2		
	Surg Technique	DB PostOp	12	9.6	4.6		
Extended Int Torque	Test Session	PreOp	28	-8.8	3.5	**0.028**	
		PostOp	29	-6.9	3.1		
	Interaction	SB PreOp	14	-8.9	4.0	0.924	
	Test Session	SB PostOp	17	-6.6	2.7		
	by	DB PreOp	14	-8.8	3.0		
	Surg Technique	DB PostOp	12	-7.4	3.6		
Flexed Ext Torque	Test Session	PreOp	27	10.0	4.8	0.535	
		PostOp	28	10.2	4.2		
	Interaction	SB PreOp	13	10.5	3.7	0.841	
	Test Session	SB PostOp	16	10.0	3.5		
	by	DB PreOp	14	9.6	5.8		
	Surg Technique	DB PostOp	12	10.4	5.2		
Flexed Int Torque	Test Session	PreOp	30	-14.0	4.3	0.032	
		PostOp	29	-12.3	6.2		
	Interaction	SB PreOp	16	-12.9	3.9	**0.012**	0.793
	Test Session	SB PostOp	17	-13.4	6.0		
	by	DB PreOp	14	-15.3	4.4		**0.002**
	Surg Technique	DB PostOp	12	-10.8	6.4		

Table F.2: Means, standard deviations, and p-values for measured rotation (in degrees) of subject groups in the secondary outcome analyses for the passive rotational laxity study.

Loading Condition	**Analysis**	**Subject Group**	N	mean	SD	**p-value** Mixed Model	Paired t-test
Extended Ext Torque	Healthy-Uninjured	Control Average	15	9.0	2.8	0.635	n/a
		Patient Contralateral	29	9.6	3.9		
	Injured-Uninjured	Patient ACL-deficient	27	9.5	3.5	0.553	
Extended Int Torque	Healthy-Uninjured	Control Average	15	-5.9	2.9	0.551	n/a
		Patient Contralateral	28	-6.4	2.9		
	Injured-Uninjured	Patient ACL-deficient	27	-8.8	3.5	**0.054**	**0.011**
Flexed Ext Torque	Healthy-Uninjured	Control Average	15	9.4	4.3	0.756	n/a
		Patient Contralateral	27	8.9	4.5		
	Injured-Uninjured	Patient ACL-deficient	28	10.0	4.8	0.572	
Flexed Int Torque	Healthy-Uninjured	Control Average	15	-15.2	4.7	0.645	n/a
		Patient Contralateral	29	-14.4	5.3		
	Injured-Uninjured	Patient ACL-deficient	29	-14.0	4.3	0.588	

Appendix G

Additional dynamic activity data

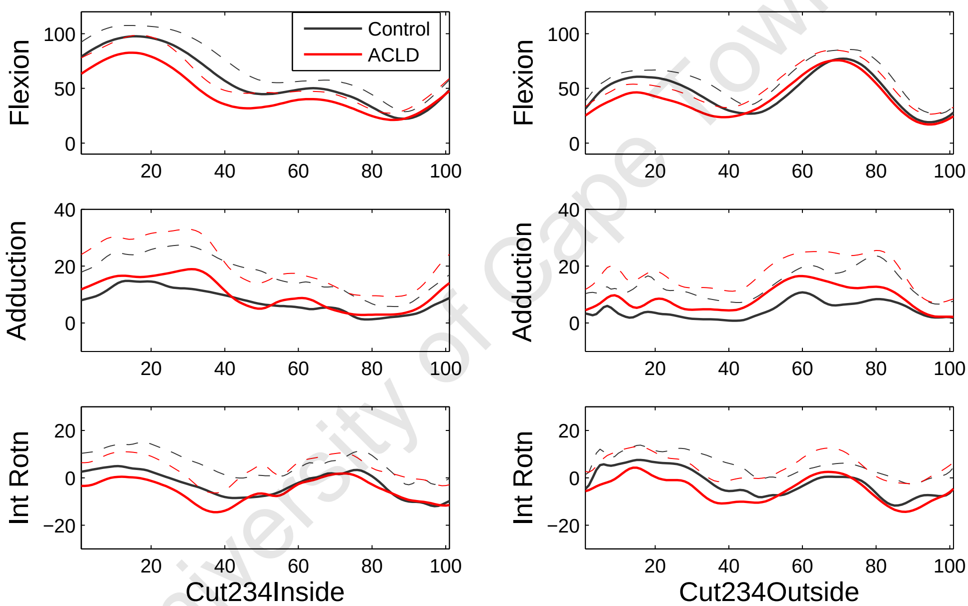

Figure G.1: Cut234Inside and Cut234Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

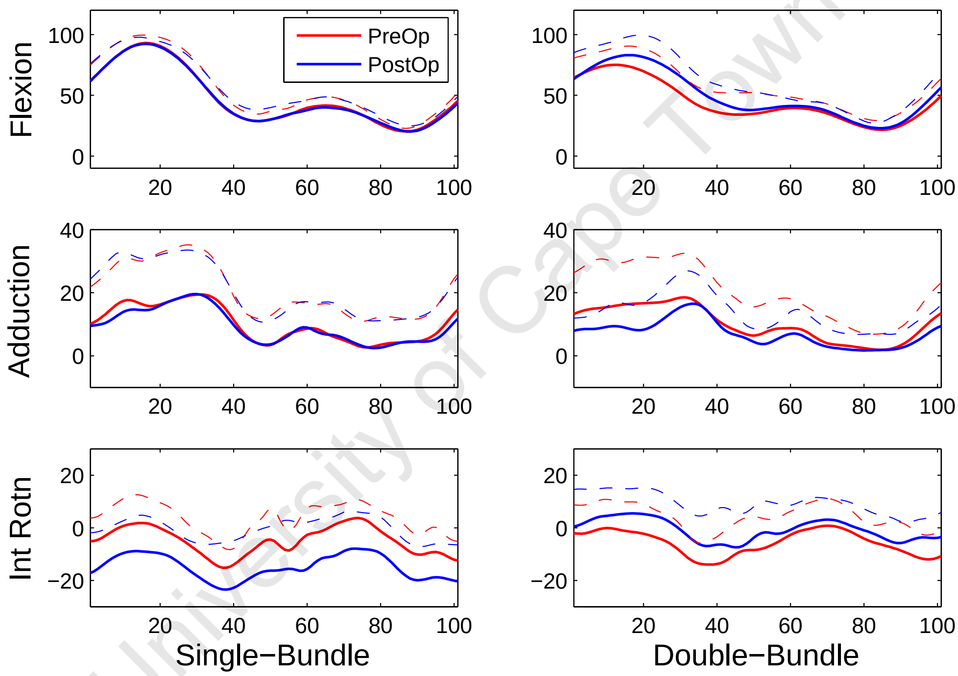

Figure G.2: Cut234Inside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

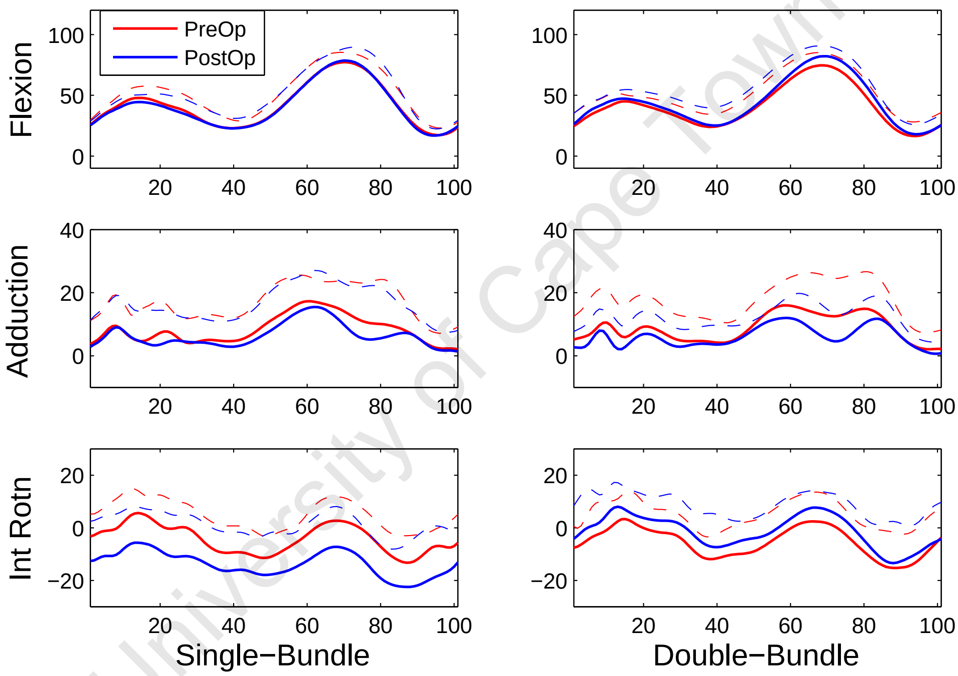

Figure G.3: Cut234Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

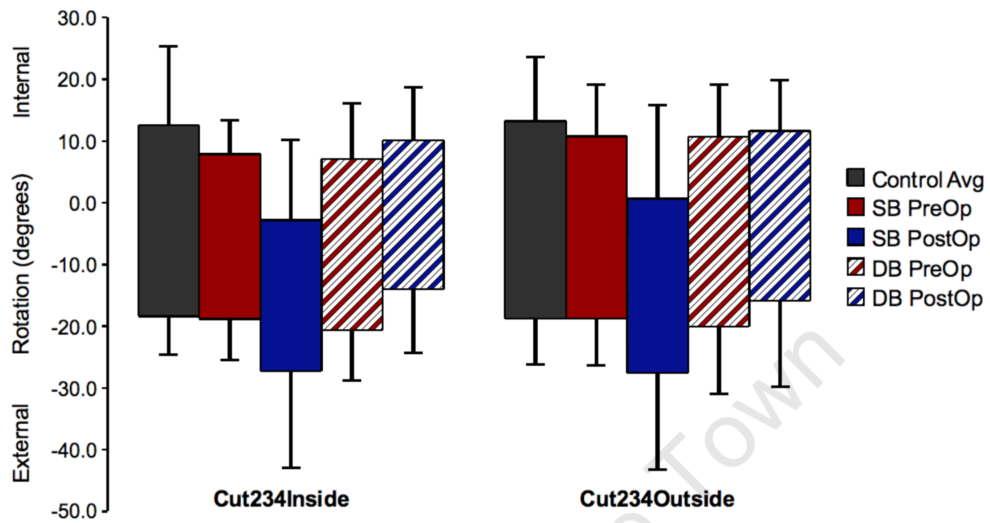

Figure G.4: Cut234Inside and Cut234Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

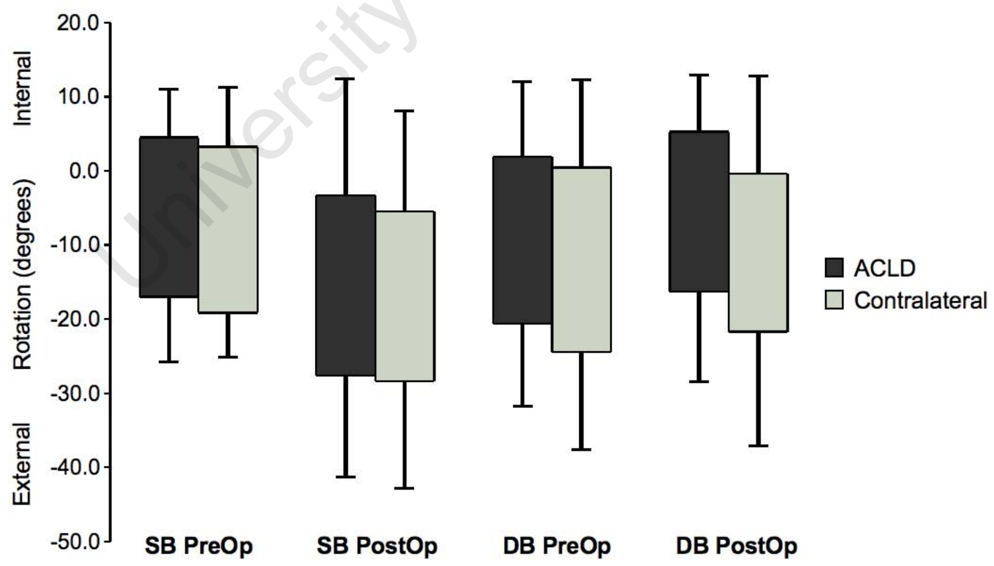

Figure G.5: Walk rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

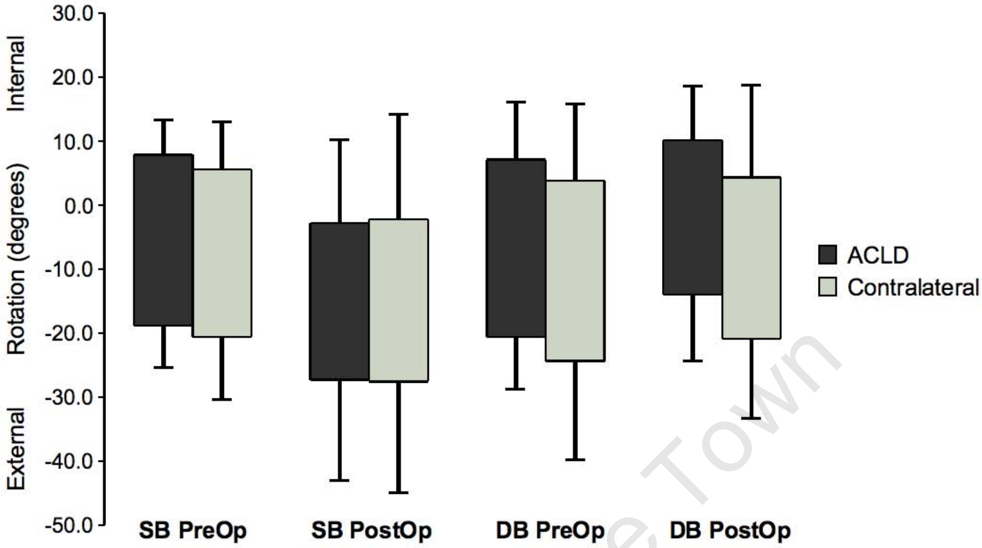

Figure G.6: Cut234Inside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

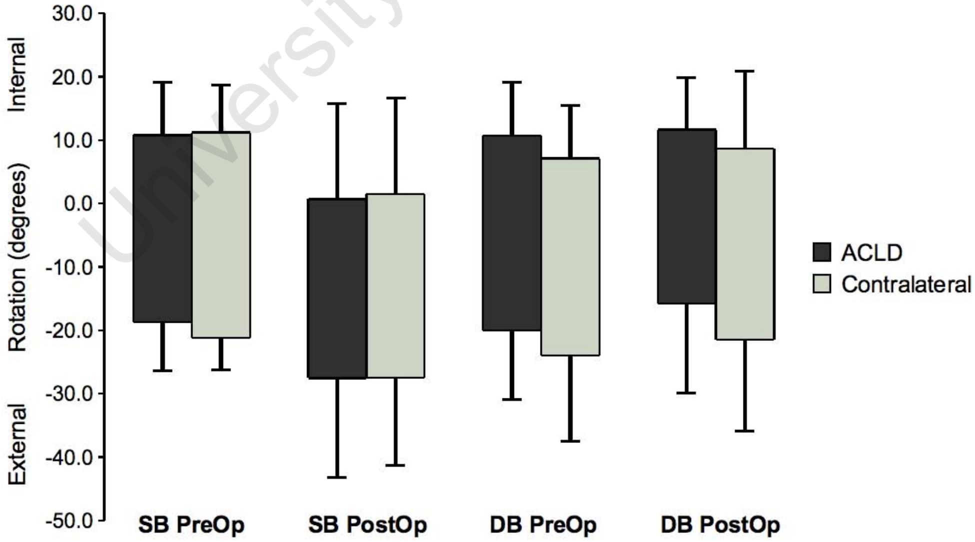

Figure G.7: Cut234Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

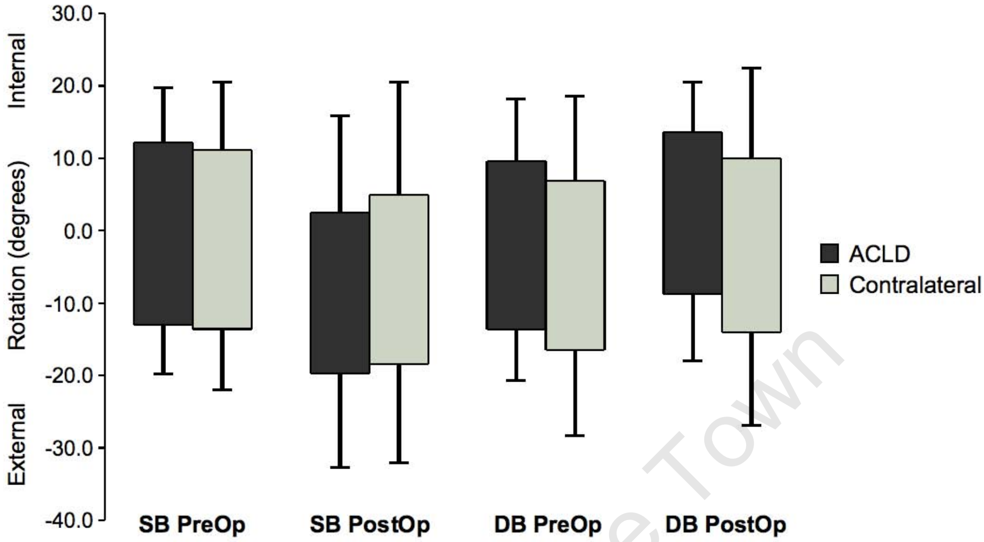

Figure G.8: JumpFull rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

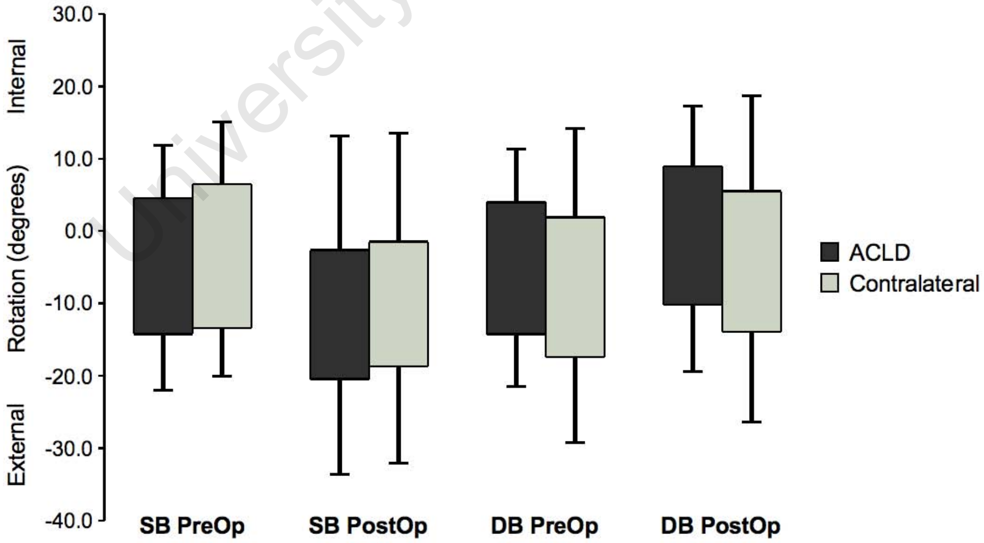

Figure G.9: JumpSW rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

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Rotational Laxity of the Knee following Reconstruction of the Anterior Cruciate Ligament using Single vs Double-Bundle Surgery

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Doctor of Philosophy

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Declaration

Rotational Laxity of the Knee following Reconstruction of the Anterior Cruciate Ligament using Single vs Double-Bundle Surgery

I, ANDREA HEMMERICH, hereby declare that:

1. the above thesis is my own unaided work both in concept and execution, and that apart from the normal guidance from my supervisor, I have received no assistance;
2. neither the substance nor any part of the above thesis has been submitted in the past, or is being, or is to be submitted for a degree at the University of Cape Town or any other university.

The thesis has been presented by me for examination for the degree of Doctor of Philosophy in Biomedical Engineering.

Signature

Date

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Abstract

Knee laxity following anterior cruciate ligament (ACL) injury may lead to long-term joint degeneration causing osteoarthritis. Although traditional surgical techniques are sufficient in providing anterior-posterior knee strength, laxity in remaining degrees of freedom persists. The double-bundle surgical technique, reconstructing both anteromedial and posterolateral bundles of the ACL, is maintained to provide superior rotational restraint; however, transverse plane kinematics have not been accurately assessed and clinical evidence is generally restricted to subjective qualitative measurements of laxity under passive loading conditions.

A magnetic resonance imaging (MRI) compatible device and three-dimensional image processing technique was therefore developed to assess passive knee laxity under known torsional loading *in vivo*, with repeatability results demonstrating a standard error of measurement of less than 0.75° in transverse plane rotational measures. A randomised control trial was conducted with 32 patients exhibiting isolated ACL rupture; subjects were allocated either a single or double-bundle reconstruction and tested prior to and approximately five months following surgery. Passive rotational laxity was quantified using the verified testing apparatus and dynamic kinematics of 22 of those subjects were measured using established gait analysis methods. Three-dimensional kinematics were concurrently assessed in left and right knees of a group of healthy control subjects under both passive and dynamic testing conditions to establish baseline data sets. Linear mixed model statistical analyses enabled a comparison of results across surgical

groups pre- and post-operatively, as well as with the contralateral uninjured knee and healthy control groups.

Passive laxity assessment of the 15 Control subjects demonstrated asymmetry in left and right rotational kinematics when evaluating internal and external rotation independently, thereby suggesting that control comparisons in pathological assessment should not be confined to contralateral knee data. A 2.4° increase in internal rotational laxity observed in the ACL-deficient relative to the normal knee in extension was restored by both single and double-bundle reconstructions. A significant interaction between single and double-bundle surgical techniques pre- to post-operatively was demonstrated when assessing internal rotation at 30° of knee flexion. With single-bundle knee rotation closer to that of the uninjured group mean, the decreased degree of internal rotation observed in the double-bundle knees indicated a propensity to overconstrain motion following reconstruction of isolated ACL tears.

While no difference in overall range of rotation was found under physiological loading conditions, a significant surgery by test-time interaction of the midpoint of the range of movement was observed during the high-demand activities. A greater external rotational shift in the single-bundle group following reconstruction suggested a muscle co-contraction stabilization strategy associated with ineffective internal rotational torque due to hamstrings tendon donor-site morbidity. The kinematics of the double-bundle patient group were closer to those of the control group, suggesting improved joint restraint.

The findings from passive laxity testing indicated that the contribution of the intact and reconstructed ACL to joint restraint is limited under isolated torsional loading. Divergent outcome following single and double-bundle surgical techniques under dynamic loading conditions suggests, however, that ACL deficiency significantly affects the functional capacity of those structures that are primarily responsible for rotational constraint of the knee. While the double-bundle reconstruction provides physiologic control resulting in knee kinematics closer to normal than the single-bundle surgery, it is proposed that the improved stability is due to loading restraint in another degree-of-freedom, rather than specifically axial rotation. Furthermore, it should be cautioned that this restraint may be a consequence of excessive graft tensioning, which could simultaneously account for the outcome of the passive rotational laxity study.

Further improvements to ACL surgical techniques are required to better reproduce passive and weight-bearing kinematics of the uninjured knee and to prevent long-term joint degeneration. While the double-bundle technique demonstrates superior constraint in dynamic loading situations, care must be taken by orthopaedic surgeons to avoid excessive graft tension and overconstraint of joint motion. Consideration should be given to treating the primary restraints of rotation and to high-quality procedures, rather than simply relying on the double-bundle reconstruction to provide sufficient joint stability.

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Nomenclature

2D	two-dimensional
3D	three-dimensional
ACL	anterior cruciate ligament
ACLD	anterior cruciate ligament deficient
AM	anteromedial
AP	anterior-posterior
BPTB	bone-patellar tendon-bone
CAS	computer assisted surgery
CT	computed tomography
DB	double-bundle
DOF	degree-of-freedom
DP	distal-proximal
ICC	intraclass correlation coefficient
IE	internal-external
LCL	lateral collateral ligament
MCL	medial collateral ligament
ML	medial-lateral
MRI	magnetic resonance imaging
PCL	posterior cruciate ligament
PL	posterolateral
RCT	randomised control trial
RSA	roentgen stereogrammetric analysis
SB	single-bundle
SEM	standard error of measurement

Chapter 1

Introduction

1.1 Background

Treatment of the ruptured anterior cruciate ligament (ACL) has improved greatly over the last twenty years with emerging research in this field of orthopaedic medicine. This has resulted in increased success at restoring normal function to the knee joint and allowing patients to return to their activities of daily living in the months and years following injury.

However, long-term complications subsequent to surgical reconstruction have clinicians questioning in greater depth the role of the ACL and, accordingly, how to improve methods of treatment. The function of the ACL in stabilizing the joint in the sagittal plane has been understood for some time since the most predominant direction of tibial laxity following an isolated ACL injury is anteriorly. Due to the emphasis placed on restoring anterior-posterior (AP) translational constraint, the potential supplementary capacity of this ligament was often overlooked; its contribution to rotational constraint is, therefore, still uncertain. Not until patients with seemingly stable knees developed additional knee deficiencies or re-injured themselves, did clinicians begin associating the injured ACL with rotational laxity.

With the aspiration to improve long-term knee biomechanics following ACL reconstruction, surgeons have increasingly endeavored to recreate the native ligament properties. One aspect that has been closely examined in recent years is the significance of the two bundles – anteromedial and posterolateral – of the ACL, with scientific studies supporting the theory that the two bundles function discretely to help constrain the loads experienced at the joint. Research has shown that ligament bundle position, orientation, and tension are not only different from one another, but also vary with knee flexion angle and specific loads applied to the joint. One attempt by which to improve the outcome of ACL surgery, has therefore been to reconstruct both bundles (double-bundle technique), rather than just one bundle (single-bundle technique) of the ligament.

Investigations comparing these two surgical techniques have primarily been conducted on cadaver knees. To eliminate the effects of transformed joint tissue properties on biomechanical outcome, more *in vivo* research is required; however, limitations with existing, non-invasive laxity measurement tools have hindered this field of study.

In order to determine the function of the ACL in the transverse plane, a reliable method of measuring the kinematics is required. Devices used by clinicians to diagnose ACL injury are currently limited to measurement of static AP translation since this is the direction in which the most severe laxity is observed. Measurement of rotational laxity of the joint is therefore simply based on the subjective assessment of the surgeon. Advanced methods of laxity measurement in all three planes of knee motion are required to improve diagnosis and assess treatment.

Advancement in medical imaging has made it a more accessible resource for use in the diagnosis of musculoskeletal injury. Imaging techniques, such as magnetic resonance imaging (MRI), have similarly been applied by biomechanists to develop more accurate methods of measuring joint kinematics *in vivo*. The advantage of MRI is that it is non-invasive while having the ability to determine the precise three-dimensional position and orientation of the underlying bone, thereby avoiding soft tissue artefact associated with skin-based measurements.

While static passive laxity measures are the simplest clinical method of determining knee pathology, gait analysis has been beneficial in demonstrating subtle distinctions between the healthy and compromised limb under dynamic, physiological loading conditions. With improved technology enabling more accurate three-dimensional measurement of tibiofemoral kinematics, this tool is becoming an indispensible means by which to evaluate *in vivo* knee laxity in patient groups.

Both medical imaging and gait analysis provide accurate and objective methods by which to determine changes in knee joint laxity resulting from treatment such as single or double-bundle surgical reconstruction of a ruptured ACL. Scientific research generated from the implementation of these devices promises to be a valuable contribution to the pursuit of improved methods by which to treat ACL deficiency and avoid long-term joint degeneration.

1.2 Objectives

The objectives of this thesis were to determine the role of the native anterior cruciate ligament (ACL) and the effects of ACL reconstruction on *in vivo* rotational laxity of the knee joint. In particular, the outcome of single and double-bundle reconstruction in constraining rotational laxity was of interest.

The first specific goal was thus to design a device to accurately apply a known torsional load to the knee while being scanned using MRI. A procedure by which the MR images could be analysed to determine the precise 3D position and orientation of the tibia with respect to the femur was furthermore required to accurately describe the joint laxity under the specific loading conditions.

The next objective was to use the torsional laxity apparatus to determine the rotational knee laxity of healthy individuals, patients with isolated ACL-rupture, and patients with single-bundle and double-bundle reconstructions. We furthermore wished to compare the laxity of patients' contralateral knees with that of the healthy, uninjured population in order to examine the possibility of inherent knee laxity in people with ACL injury and to assess whether the contralateral limb may be used as a control when testing for rotational laxity. Effectively, our intention was to gain a better understanding of the behaviour of the knee under internal versus external torsional loads at full extension and 30° of flexion in the specified functional status groups.

The purpose of the final study was to evaluate the functional laxity outcome of the single and double-bundle surgical techniques under physiological loading conditions, focussing on the transverse plane rotational knee kinematics. Using established gait analysis methods, the aim was to determine whether the double-bundle procedure demonstrated superior rotational constraint to the single-bundle reconstruction with the knee experiencing realistic forces from everyday activities.

The knowledge gained from this thesis is intended to extend our comprehension of the effects of single and double-bundle surgical reconstruction techniques on *in vivo* knee biomechanics and to improve methods of treatment of ACL injury.

1.3 Document overview

Chapter 2 reviews the relevant literature in the field. The function of the anterior cruciate ligament in transverse plane rotational constraint is addressed, focussing on methods of assessment, in addition to passive and dynamic joint laxity in the ACL-deficient and reconstructed knee. In particular, the investigations comparing single and double-bundle ACL-reconstruction are critiqued.

Chapter 3 describes the MRI-compatible device, data collection and image processing methodology developed to measure three-dimensional kinematics of the knee under torsional loading. Results from feasibility and *in vivo* repeatability studies are presented.

Chapter 4 investigates the three-dimensional knee kinematics of a group of 15 healthy subjects under passive torsional loading using the methodology presented in Chapter 3. Left-right symmetry is analysed and the significance of coupled motion is discussed.

Chapter 5 presents the findings of a randomised control trial in which 32 patients with isolated ACL rupture were allocated either a single or double-bundle surgical reconstruction. Patients were tested pre- and post-operatively under the same passive torsional loading conditions as the healthy subjects in Chapter 4. The interaction of surgical technique by test time (preceeding and following surgery) is investigated.

Chapter 6 presents the outcome of dynamic joint laxity testing in 22 patients allocated either single or double-bundle surgical procedures to reconstruct the injured ACL. Three-dimensional knee kinematic data were collected prior to and following surgery during low and high-demand gait activities. The discussion emphasizes the significance of the contribution of muscles as dynamic stabilisers on rotational kinematics of the knee.

Chapter 7 qualitatively compares the conclusions of the passive and dynamic rotational laxity studies. Final conclusions and recommendations for future work are discussed.

1.4 Publications originating from this PhD thesis research

An expanded edition of the following journal publication is presented as Chapter 3:

- Hemmerich A, van der Merwe W, Vaughan CL. (2009). Measuring three-dimensional knee kinematics under torsional loading. *Journal of Biomechanics* **42**, 183-186.

The following peer-reviewed abstracts are also direct outcomes of this thesis and have been presented as seminars or poster exhibits at international conferences:

- Hemmerich A, van der Merwe W, Batterham M, Vaughan CL. (2009). Rotational laxity in anterior cruciate deficient and reconstructed knees: A prospective randomised control trial comparing single and double-bundle surgical techniques. *22nd Congress for the International Society of Biomechanics*. Cape Town, South Africa
- Hemmerich A, Vaughan CL, van der Merwe W. (2009). Prospective randomised study to compare single-bundle versus double-bundle ACL reconstruction in restoring rotational 3D kinematics of the knee. *7th Biennial Congress of the International Society of Arthoscopy, Knee Surgery & Orthopaedic Sports Medicine*. Osaka, Japan

- Hemmerich A, van der Merwe W, Vaughan CL. (2007). Repeatability of 3D knee joint kinematic measurements in vivo under torsional load. *21st Congress for the International Society of Biomechanics*. Taipei, Taiwan

- Hemmerich A, van der Merwe W, Vaughan CL. (2006). Three-dimensional in vivo knee joint laxity under torsional loading. *5th World Congress of Biomechanics*. Munich, Germany

- Hemmerich A, van der Merwe W, Vaughan CL. (2006). Three-dimensional in vivo motion analysis of knee joint laxity under torsional loading. *9th Symposium on 3D Analysis of Human Movement*. Valenciennes, France.

Chapter 2

Literature review

2.1 Rotational laxity of the knee joint

2.1.1 Knee anatomy and rotational restraint

Several structures have been shown to limit rotational laxity of the knee joint including the joint capsule, the collateral and the cruciate ligaments (Fuss, 1991; Markolf *et al.*, 1976; Wang & Walker, 1974); however conflicting reports as to the degree to which each of these structures, in particular the anterior cruciate ligament (ACL), contribute to knee rotational restraint have been found in the literature. The primary role of the ACL is to restrain anterior displacement of the tibia relative to the femur (O'Connor & Zavatsky, 1993); with research focussing on the ACL and anterior-posterior (AP) translation of the joint, its role in preventing rotational laxity has been largely overlooked (Zaffagnini *et al.*, 2000).

Although some researchers have concluded that the ACL does not play a major role in rotational constraint (Lane *et al.*, 1994), more recent studies have demonstrated a significant increase in rotational laxity following ACL injury (Georgoulis *et al.*, 2003; Tashman *et al.*, 2004; Yagi *et al.*, 2002; Zaffagnini *et al.*, 2000). The capacity of the ACL to restrain rotation in the transverse plane has been attributed to the location of its tibial and femoral insertion sites and its orientation within the joint.

The ACL extends from the anterior part of the tibial plateau to the medial side of the lateral femoral condyle. It is comprised of a multitude of fibers that are commonly considered to be divided into two bundles – anteromedial (AM) and posterolateral (PL) – whose nomenclature is based on their tibial insertions (Figure 2.1). Its oblique orientation causes ligament tensioning and resistance as the tibia pivots about the transverse plane axis of rotation, due to an increase in the distance between tibial and femoral insertion sites. With internal rotation, twisting of the ACL about the posterior cruciate ligament (PCL) precipitates further tensioning of the ligament (Blankevoort & Huiskes, 1996). It has been moreover suggested that the PL bundle has a greater mechanical advantage in restraining rotation with its femoral insertion further from the axis of rotation than that of the AM bundle (Yagi *et al.*, 2002). However, these suggestions are speculative; there is no objective data to support this.

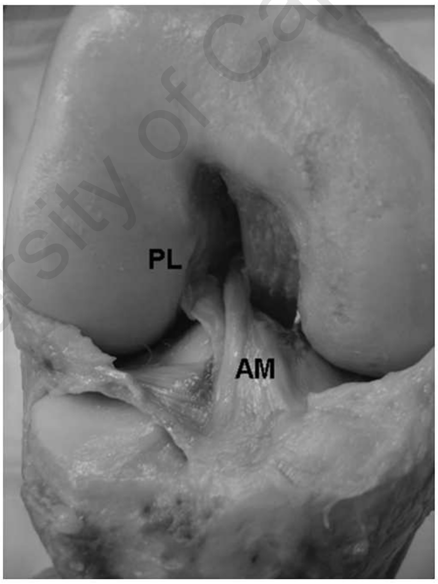

Figure 2.1: Cadaver knee showing AM and PL bundles (Petersen *et al.*, 2006).

Cadaveric and computational studies investigating the mechanics of the ACL bundles during passive flexion-extension have shown that the anteromedial bundle is taut in flexion while the posterolateral bundle is taut in extension, this is accredited to the position of the femoral insertions and resulting orientation of the individual bundles as illustrated in Figure 2.2 (Amis & Dawkins, 1991; Chhabra *et al.*, 2006; O'Connor & Zavatsky, 1993). Due to its relative enhanced tension at lower flexion angles, it has been found that the PL bundle is of greater importance in limiting joint laxity between 0° and 30° (Amis & Dawkins, 1991; Markolf *et al.*, 2009; Yagi *et al.*, 2002).

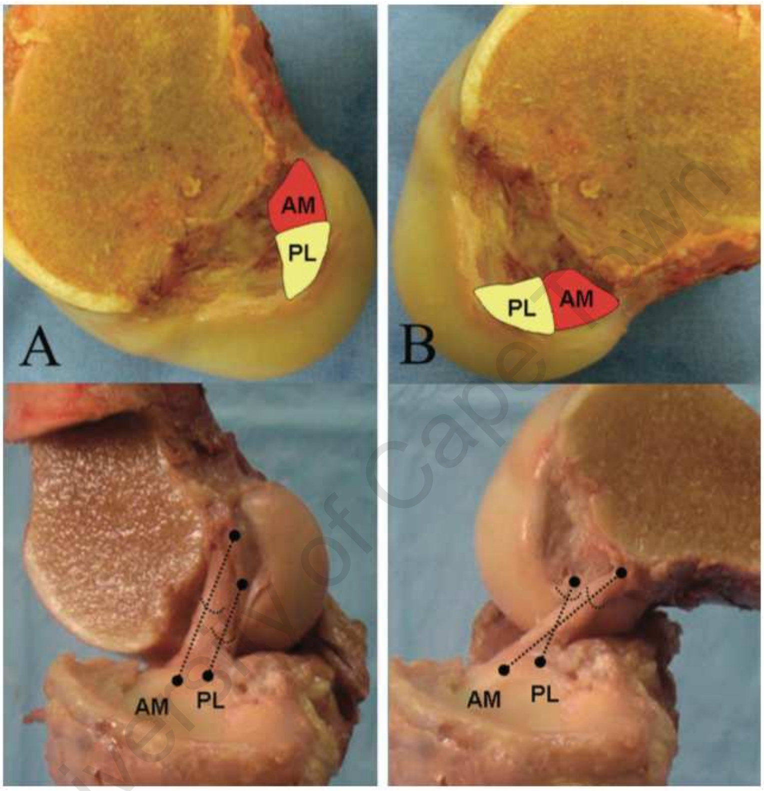

Figure 2.2: Sagittal sectioned view of tibiofemoral joint. A: Anteromedial and posterolateral bundles are parallel with knee in extension. B: Bundles are crossed at 90° of flexion and femoral insertion sites are now horizontal (Chhabra *et al.*, 2006).

The ACL is, nonetheless, only a secondary restraint to axial rotation. It has been extensively maintained that the medial collateral ligament (MCL) is the primary restraint to external rotation (Csintalan *et al.*, 2006; Harfe *et al.*, 1998; Meyer & Haut, 2008; Nordt *et al.*, 1999); however, the lateral collateral ligament (LCL) and posterolateral structures have also been shown to contribute to external rotation restraint (Blankevoort *et al.*, 1991; Kaneda *et al.*, 1997; Markolf *et al.*, 1976). Active joint rotational stability is provided by the hamstrings and iliotibial band, which externally rotate the tibia due to greater influence of the biceps femoris than the medial hamstrings (Kwak *et al.*, 2000). Passive constraint of internal rotation is not only provided by the collateral ligaments, but also by the tibiofemoral contact surfaces (Blankevoort & Huiskes, 1996; Wang & Walker, 1974). The greater posterior slope of the medial tibial surface provides an appreciable contribution to internal rotation restraint at 90° of flexion (Blankevoort & Huiskes, 1996). The menisci are also considered to be secondary joint stabilisers, with the medial maintaining greater stiffness than the lateral meniscus (Masouros *et al.*, 2008; Wang & Walker, 1974).

2.1.2 Importance of maintaining normal knee kinematics

Although it is often possible for a patient with a deficient ligament to perform the majority of activities that he or she conducted on a daily basis prior to injury, treatment to restore normal knee kinematics is important for various reasons. Restoration of joint stability and the elimination of 'giving way' symptoms can allow patients to return to higher intensity level activities within a year of surgery in most cases.

Long-term laxity associated with ACL deficiency is less well understood. Pathological knee kinematics result in changes in positions of tibiofemoral contact points and, consequently, altered stress distributions in the articular cartilage and greater loads on the surrounding joint structures (Chaudhari *et al.*, 2008; Li *et al.*, 2006; Stergiou *et al.*, 2007). Meniscal tears, damage to cartilage, as well as excessive ligament loading may be consequences of chronic ACL deficiency; resulting degeneration of these joint structures may lead to osteoarthritis (Chaudhari *et al.*, 2008; Li *et al.*, 2006; Masouros *et al.*, 2008; Sharma *et al.*, 1999; Shefelbine *et al.*, 2006; Stergiou *et al.*, 2007).

2.2 The ACL-deficient knee: Methods of treatment

An ACL injury may result in a partial or complete rupture of the ligament with possible damage to surrounding structures such as collateral ligaments and the meniscus. Treatment should be based on the extent of the injury in addition to the level of activity to which the patient intends to return and his or her individual characteristics such as age and medical condition (Miller, 2004).

2.2.1 Nonoperative

Nonoperative treatment is usually reserved for those patients who are satisfied to limit their activity level following injury (often older people); it is seldom recommended for those who wish to return to competitive sport, especially activities involving pivoting. Nonoperative treatment is typically limited to a rehabilitation program involving exercise to strengthen muscles. In order to prevent repetitive impact loading or situations that may cause further injury, patients are taught to modify the ways in which activities are conducted. Knee braces can also provide additional stability for the joint (Larson, 1993).

2.2.2 Surgical reconstruction

Surgical reconstruction of the torn ACL was first successfully achieved in the late 19th century (Colombet *et al.*, 1999). The procedure is now common practice with an estimate of over 100,000 reconstructions performed every year (Lewis *et al.*, 2008). The remnants of the native ACL are excised and a replacement graft is extended from the tibial to the femoral insertion site of the native ACL. The graft is fixed to the bone through tunnels extending from the anterior side of the tibia through the tibial plateau and from the medial side of the lateral femoral condyle passing proximally towards the lateral side of the femur.

Fixation devices include the interference screw, endobutton, and staple; choice of hardware is partially dependant on the type of graft used. The most common types of autografts used for ACL reconstruction include the bone-patellar tendon-bone (BPTB) and the four-strand semitendinosus/gracilis tendon autografts (Fu *et al.*, 2000).

The BPTB graft, which harvests the middle third (approximately 10mm in width and 100 mm in length) of the patellar tendon including sections of bone from the insertion sites at the patella and tibia, has been a preferred technique because the bone interference fit within the tibial and femoral tunnels permits bone to bone healing and consequently more rapid recovery. Its high stiffness and ultimate tensile strength, furthermore, limit graft failure (Fu *et al.*, 2000).

The disadvantage with this procedure, however, is that the bone tunnel placement is dependant on the length of the graft: the tunnel must be located so that a sufficient tunnel length is available to adequately tension the graft (Jackson & Lemos, 1993). Furthermore, due to long-term morbidity in the donor knee, such as pain in the patellar region and flexion contracture, as well as decreased joint power, alternative surgical techniques are often preferred (Feller *et al.*, 2001; Kowalk *et al.*, 1997; Marcacci *et al.*, 2003).

Initially, ACL reconstruction was performed using a single-bundle (SB) technique. As the significance of the two separate bundles was not fully understood, a graft simulating only the AM bundle of the ACL was used, which restricted AP translation of the knee. Since then, ACL reconstruction has been improved with graft construction now accounting for both bundles of the native ligament. This double-bundle (DB) approach is another advantage of the hamstrings graft procedure, in which the four strands of the graft are typically composed of the semitendinosus and gracilis tendons with each folded back on itself. Whereas the single-bundle technique has only one fixation site at both the femur and tibia, the double-bundle technique customarily requires two tunnels in both bones as shown in Figure 2.3. Both the AM and PL bundles are reconstructed and fixed independently in the separate tunnels. Although the DB technique was shown to have better post-operative functional results in certain studies, the procedure is more complex, time-consuming, and expensive to perform than the SB technique (Brophy *et al.*, 2009). Additional complications concerning revision surgery due to a second tunnel in each bone have also been intimated (Harner & Poehling, 2004). (A detailed comparison of the functional outcome of the SB and DB techniques is given in section 2.4.1.)

Bone-patellar tendon-bone graft procedures have generally been associated with increased donor-site morbidity than hamstrings grafts (Fu *et al.*, 2000). Not only do patients more often complain of anterior knee pain following BPTB graft reconstruction, in particular during kneeling (Feller *et al.*, 2001), Kowalk *et al.* (1997) also demonstrated reduced flexion moment and power at the injured knee during stair ascent that had not been observed pre-operatively. Simultaneous

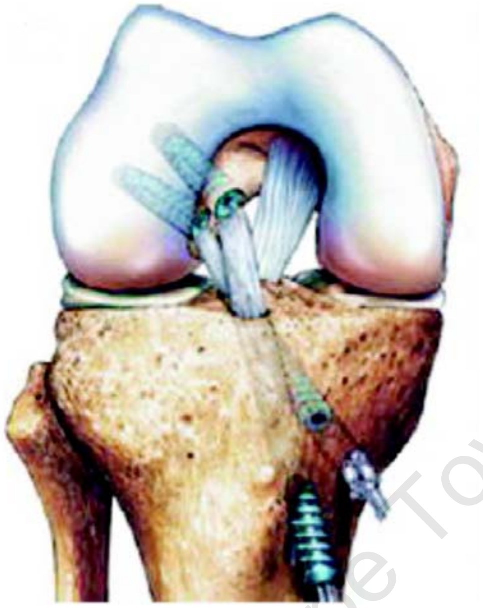

Figure 2.3: Schematic of double-bundle reconstruction (Järvelä, 2007).

increases in joint moment and power of the contralateral ankle suggested a compensation mechanism due to morbidity of the graft harvest site (Kowalk *et al.*, 1997).

Morbidity associated with hamstrings graft reconstruction includes weakening of knee flexion and internal rotation strength (Aune *et al.*, 2001; Viola *et al.*, 2000). Figure 2.4 illustrates the mechanical advantage of the semitendinosus and gracilis muscles in flexion and rotation due to the positions of their distal insertion on the medial aspect of the tibia. Functional deficit is generally minimal with only a 5% decrease in muscle strength reported three years post-operatively (Fu *et al.*, 2000).

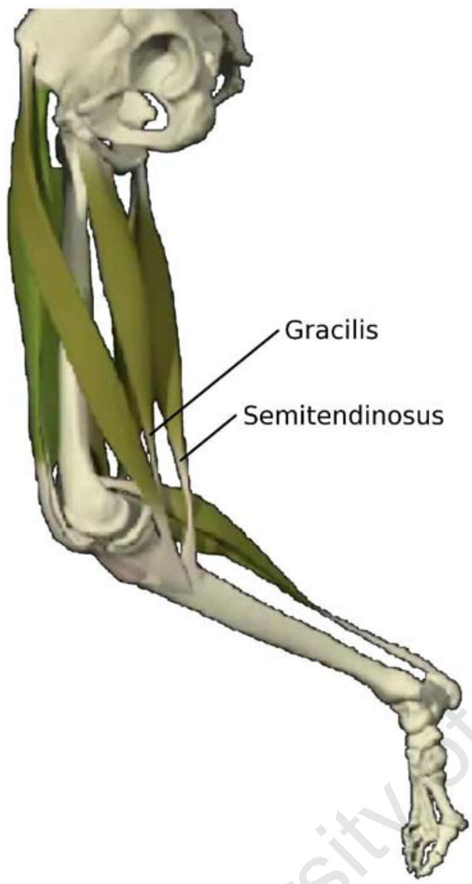

Figure 2.4: Semitendinosus and gracilis tendons shown with knee in flexion (adapted from Moore & Dalley (2005)).

2.3 Measuring joint laxity in the ACL-deficient knee

2.3.1 Concepts of joint motion

Descriptions of joint motion found in the literature are often ambiguous, making research outcomes unintelligible and comparisons between studies difficult. In order to standardize the conventions used in biomechanics research, the International Society of Biomechanics recommended the 'joint coordinate system' (Grood & Suntay, 1983) for the description of tibiofemoral kinematics (Wu & Cavanagh, 1995). This system presents the six degree-of-freedom (DOF) motion of the distal segment (e.g. the tibia) with respect to the proximal segment (e.g. the femur) as rotations about and translations along anatomical axes, making it easily understood by clinicians (Grood & Suntay, 1983). Segment axes are defined based on anatomical landmarks. Figure 2.5 illustrates the definition used by Hoshino *et al.* (2007) where the flexion-extension axis was embedded in the distal femur, internal-external (IE) rotation occurred about the longitudinal axis of the tibia, and ab-adduction took place about the floating axis which was defined by Grood & Suntay (1983) as perpendicular to the previous two axes.

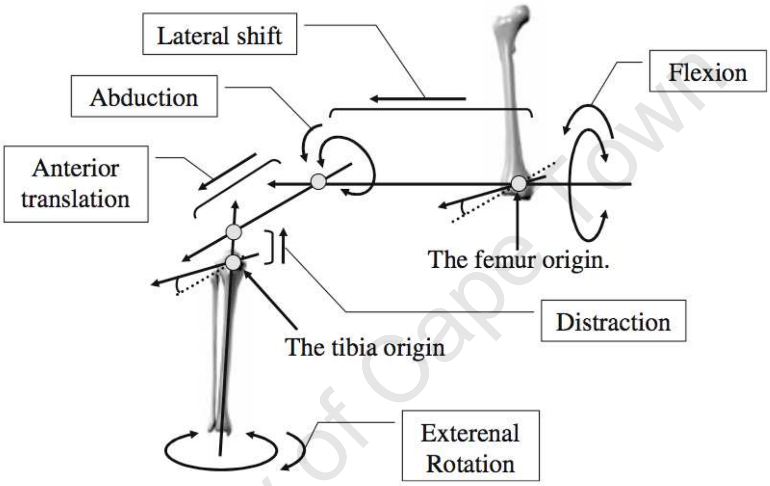

Figure 2.5: Grood & Suntay (1983) developed the joint coordinate system to describe the kinematics of the knee joint. This figure illustrates its application by Hoshino *et al.* (2007).

One limitation of the joint coordinate system convention is its susceptibility to kinematic crosstalk. This can occur when a defined axis of rotation is misaligned with the actual axis of rotation; the result is that rotation in one anatomical plane is misinterpreted for rotation in another (Piazza & Cavanagh, 2000). Crosstalk errors of up to 15° can easily transpire with proportional axis misalignment (Kadaba *et al.*, 1990; Piazza & Cavanagh, 2000).

Crosstalk and the alignment error that can occur from axes defined using anatomical landmarks have been motivation for some biomechanists to measure the 'helical axis' of rotation based on the three-dimensional (3D) tibiofemoral movement (Besier *et al.*, 2003; Dennis *et al.*, 2005; Mannel *et al.*, 2004; Marin *et al.*, 2003). Instead of defining three rotational and translational axes from which tibiofemoral motion is measured, the helical axis method describes segment motion as a rotation about and translation along a single axis which is defined from the tibiofemoral motion (Woltring, 1991).

The helical axis corresponds to the anatomical axis as long as rotation in a single anatomical plane occurs (e.g. flexion-extension); the difficulty arises with the physical interpretation of coupled motions (Woltring, 1991). An example was demonstrated by Dennis *et al.* (2005), who illustrated significant variation in position and orientation of the helical axis during a deep knee bend activity, thereby reflecting the complex motion of the knee. To describe this motion clinically in terms of the combined anterior-posterior translation and internal-external rotation, it was nonetheless necessary for the authors to use an anatomical tibial reference frame in which tibiofemoral contact points were described and subsequent clinical joint motion was measured (Dennis *et al.*, 2005).

Interpretation of anterior-posterior translation of the tibia with respect to the femur may seem simple; however, due to differences in measurement methods and/or segment coordinate systems, results from one study may not be comparable to those of another. Studies using cartesian coordinate systems, have reported AP translation measured along the tibial anterior axis (Robinson *et al.*, 2007; Yamaguchi *et al.*, 2009) as well as the floating axis of the joint coordinate system as illustrated in Figure 2.5 (Benoit *et al.*, 2007; Grood & Suntay, 1983; Hoshino *et al.*, 2007; Reed-Jones & Vallis, 2008; Woo *et al.*, 2006). Depending on the tibial IE rotation angle, this distinction could have significant effects on the translation measured.

The position of the origins of the respective segments can also affect measured translation (Roos *et al.*, 2006). In a study conducted by Beardsley *et al.* (2007), differences in AP translation were calculated using two different coordinate systems under anterior loading conditions. Considerable rotations (e.g. up to 31° in the sagittal plane) accompanied the translation in all three anatomical planes, which contributed to the discrepancies in translation measured between the two coordinate systems. A mean difference in AP translation of 4.2 mm was measured between coordinate systems and it was found that a 3° rotation in each anatomical plane would result in a 2 mm difference in AP translation; these values are routinely considered clinically significant (Beardsley *et al.*, 2007).

This section outlining the complexities involved with the description of joint motion was not intended as a complete review of the literature on this subject, but rather as a brief overview to enable a better understanding of the ensuing sections which critique scientific studies that have investigated joint laxity.

2.3.2 Instrumentation used to measure *in vivo* joint laxity

A clinician can perform numerous tests in order to assess the structural integrity of the knee ligaments. A positive outcome is commonly given by a qualitative description, rather than a quantitative value; for example, motion that varies from the contralateral knee or an audible clunk characteristic of subluxation of the tibial plateau on the femoral condyle (Magee, 1992). These tests examine AP, medial-lateral (ML), and rotational laxity and most commonly include the Lachman, the anterior drawer, and the pivot shift tests. Although these examinations, when performed correctly by an experienced clinician, can verify whether or not there is an ACL deficiency with high reliability, the extent of the injury can be difficult to determine (Magee, 1992).

Instrumented knee laxity measurement devices have been developed to quantify the extent of knee laxity. The most commonly used arthrometers are the KT-1000 and KT-2000 (MEDmetric Corp, San Diego, CA). Only passive AP laxity is tested by these devices; quantitative clinically accessible devices used to measure other degrees of joint laxity are not currently available. However, alternative methods have been developed by several researchers to measure joint motion in one or more planes of motion under various loading conditions (Koh *et al.*, 2005). These can, by and large, be divided into two categories: tracking marker devices and medical imaging techniques.

2.3.2.1 Tracking marker devices

Tracking marker measuring systems have traditionally been used in the gait analysis laboratory and may consist of optoelectric systems (Vaughan *et al.*, 1999) or electromagnetic tracking devices (Hemmerich *et al.*, 2006). In this environment, markers are assumed to be rigidly fixed to the segments of the joint in question. The position and orientation of the proximal and distal segments (e.g. femur and tibia) can then be determined under physiological loading conditions from the information provided by the markers.

Tracking devices have also been used to measure passive joint laxity, such as varus-valgus and IE rotation. Measurement accuracy was reported by Shultz *et al.* (2007) as generally within 2° under a 10 Nm varus-valgus load and 3° to 4° under a 5 Nm internal-external torsional load and by Tsai *et al.* (2008) as within 5° of rotation under 6 Nm of torque. In both of these studies, the error was attributed in part to skin motion artefact.

An even simpler method of tracking IE rotation using a protractor demonstrated limited accuracy; measurement errors determined by comparing the device results with those calculated using roentgen stereogrammetric analysis (RSA) showed a systematic overestimate of 100 % of the actual rotation angle (Almquist *et al.*, 2002). A further disadvantage of this system was the reliance on the precise alignment of the tibia with the external tracking device, i.e. the protractor.

To avoid soft tissue artefact, tracking markers are ideally fixed to the underlying bone. While ethically, this procedure is not authorized under most circumstances due to its invasive nature, several researchers have used these methods intraoperatively where computer-assisted surgical (CAS) instruments are already secured to the tibia and femur (Bull *et al.*, 2002; Ishibashi *et al.*, 2005; Martelli *et al.*, 2007; Robinson *et al.*, 2007). Although this technique has demonstrated greater precision of 3D joint measurements, the load was applied manually in most cases, relying on the investigator to accurately apply consistent quantities of force and/or torque to the joint.

2.3.2.2 Medical imaging techniques

The main advantage of medical imaging techniques used to measure knee joint laxity is the elimination of soft tissue artefact. The Telos stress device was developed to determine the position of the tibia with respect to the femur in the sagittal plane using X-ray under anterior-posterior loading (Schulz *et al.*, 2005). A similar method using X-ray was developed by Sawant *et al.* (2004) to measure valgus laxity associated with combined cruciate and MCL injury. As with arthrometers, motion outside of a single plane cannot be measured and acquiring the transverse plane X-ray images that would be required to measure IE rotation is impractical.

The capability of magnetic resonance imaging (MRI) and computed-tomography (CT) to generate several two-dimensional (2D) slices has enabled investigators to overcome this limitation. In some studies, the three-dimensional nature of these imaging techniques was not fully exploited; rotation was simply determined by tracking the anterior-posterior translation of both the medial and lateral condyles on the tibial plateau in parallel sagittal plane slices (Iwaki *et al.*, 2000; Logan *et al.*, 2004; Okazaki *et al.*, 2007). This technique, although less sophisticated than more recent developments in 3D motion tracking using CT and MRI, was nevertheless able to illustrate the coupled internal rotation with flexion of the knee joint known as screw-home motion that results from medial side sliding and lateral side rollback of the femoral condyles (Hill *et al.*, 2000; Iwaki *et al.*, 2000).

By reconstructing 3D bone segments from several slice medical images, complete 6 DOF motion can be traced. Li *et al.* (2004b) superimposed the segment models generated from 3D fluoroscopy onto 2D X-rays taken at different angles of knee flexion. By correctly orienting the models in the orthogonal planes, 3D position and orientation were determined to accuracies of 0.1 mm and 0.1 degrees (Li *et al.*, 2004b).

This methodology was modified by the same research group for weight-bearing kinematic measurements. Three-dimensional images of the femur and tibia were generated from MR images and two orthogonal images were captured during a lunge activity using the 3D fluoroscope (Li *et al.*, 2004b). Using this technique, this research group was able to identify not only anterior-posterior movement of the ACL-deficient (ACLD) tibia, but also a significant lateral shift of the femur on the tibial surface (DeFrate *et al.*, 2006; Li *et al.*, 2006).

The technique employed by Fellows *et al.* (2005b) was slightly different in that low resolution 3D MRI scans were shape-matched to high resolution images taken at a neutral position. The advantage of these techniques that superimpose low resolution (or 2D) images onto high resolution 3D models is that the low resolution scans can be acquired in a relatively short period of time while the instrumentation used to generate the models may require high scan times for adequate resolution or may limit the position at which scans can be taken (e.g. full knee extension). While the short scan time of the low resolution images permits imaging of the knee in positions that could potentially not be maintained by the subject for the duration of the longer scan time required for the high resolution 3D images, they are nonetheless not instantaneous. These techniques are, therefore, still considered to be 'quasi-static' (Li *et al.*, 2004b).

Roentgen stereogrammetric analysis (RSA), in which small tantalum markers are surgically implanted into the bones, is employed similarly to optoelectric systems, except it uses radiographic images to track the markers. Brandsson *et al.* (2002) demonstrated the value of this technique to track dynamic motion while patients ascended an 8 cm high platform. With this biplane radiographic imaging technique, kinematic data was collected at 2 to 4 exposures per second. RSA was also used by Tashman *et al.* (2004). With a much greater capture rate of 250 Hz, 6 DOF kinematic data could be collected during downhill treadmill running with an accuracy within 1° for IE tibial rotation (Tashman & Anderst, 2003).

2.4 Surgical techniques and rotational laxity outcome

2.4.1 Single versus double-bundle reconstruction

At the commencement of this PhD thesis in early 2005, no journal publications could be found directly comparing SB (one tibial and one femoral tunnel) and DB (two tibial and two femoral tunnels) surgical techniques in a controlled investigation. Joint laxity following SB and 'non-anatomic' DB (one tibial and two femoral tunnels) had been investigated (Adachi *et al.*, 2004; Yagi *et al.*, 2002); however, these studies did not adequately address the issue of rotational laxity as only anterior translation was measured under the specific loading conditions. Nevertheless, in the cadaveric study conducted by Yagi *et al.* (2002), the combined valgus and internal torsional loading conditions did indicate a difference in joint laxity between the two reconstructive techniques. No difference between SB and DB techniques was found with anterior loading by Adachi *et al.* (2004).

Most of the studies comparing SB and anatomic DB reconstruction listed in Table 2.1 suggested superior outcome using the DB technique; however, the specific evidence may not make this general conclusion as elementary as the earlier literature proclaimed. Only six studies actually measured transverse plane rotation (Ferretti *et al.*, 2008; Ishibashi *et al.*, 2005; Markolf *et al.*, 2008b, 2009; Seon *et al.*, 2009; Steckel *et al.*, 2007a), while the remaining studies acquired other measures of restraint, primarily anterior-posterior laxity. Of those that measured IE rotation, only one applied a quantified internal torque to the knee joint and this was in concurrence with a valgus torque to simulate the pivot shift (Markolf *et al.*, 2009). Those that administered isolated torsional loading typically used 'manual [maximum] force' (Ferretti *et al.*, 2008; Ishibashi *et al.*, 2005; Seon *et al.*, 2009).

The other studies in Table 2.1 that formed conclusions regarding rotational laxity following SB or DB reconstruction did so based on subjective measures of the pivot shift test (Asagumo *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Markolf *et al.*, 2008b; Muneta *et al.*, 2007; Siebold *et al.*, 2008; Streich *et al.*, 2008; Yagi *et al.*, 2007). The pivot shift was typically evaluated on a positive-negative or four-point grade indicating the examiner's estimate of the degree of instability. There was no quantitative measure of the kinematics (either tibiofemoral rotation or translation in any anatomical plane) and it has been shown that the clinical assessment varies between examiners (Bull & Amis, 1998). The only distinct association to rotational restraint is the fact that internal torque is one component of the applied load, but even that has not been quantitatively measured.

An attempt to quantify *in vivo* kinematics resulting from the pivot shift has, in fact, been made by several researchers (Bull *et al.*, 2002; Hoshino *et al.*, 2007; Kubo *et al.*, 2007; Lane *et al.*, 2008; Lopomo *et al.*, 2009; Robinson *et al.*, 2007; Yagi *et al.*, 2007). These studies have used electromagnetic or optoelectric systems to track tibiofemoral movement in 3D space. Although measures of tibial translation, rotation, velocity, and acceleration during pivot shift contribute to its objective evaluation and understanding of the 3D kinematic laxity, measuring the magnitude of the applied loads is not possible with these instruments.

Accordingly, it is possible to summarise the methodologies of those studies that have directly compared SB and DB surgical techniques (Table 2.1) into four categories: those that applied transverse plane IE torque and measured rotation in the same plane (Ferretti *et al.*, 2008; Ishibashi *et al.*, 2005; Markolf *et al.*, 2008b, 2009; Seon *et al.*, 2009), those that applied transverse plane torque and measured kinematics other than IE rotation (Asagumo *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Muneta *et al.*, 2007; Siebold *et al.*, 2008; Streich *et al.*, 2008; Yagi *et al.*, 2007), those that applied loads other than transverse plane torque but still measured transverse plane rotation (Steckel *et al.*, 2007a), and those that applied loads other than transverse plane torque and measured kinematics other than rotation (Seon *et al.*, 2007; Yasuda *et al.*, 2006). Of those that have applied transverse plane torsional loads, only four studies applied *isolated* torsional loads, and only one applied a *quantified* torsional load (Markolf *et al.*, 2009); that particular study was conducted on cadaveric specimens and was, therefore, unable to account for the healing process following reconstruction or differences in mechanical properties from living joint tissue. Still, the DB reconstruction is routinely cited as the technique that is able to control *rotational* laxity better than SB surgery following ACL injury.

In actuality, the lack of standardised loading criteria precipitates a more subjective evaluation of joint laxity based on the discretion of the investigator, makes it difficult to compare study outcomes, and may be the reason for some of the conflicting results in the literature; for example, while Ferretti *et al.* (2008) found no difference in either anterior translation or rotation between surgical techniques under maximal manual anterior force and IE torsion, respectively, Seon *et al.* (2009) demonstrated better DB laxity control in both directions and Ishibashi *et al.* (2005) found that only DB anterior laxity (not rotation) was significantly reduced when compared to the SB technique under similar loading conditions. Furthermore, some studies, such as those of Kondo *et al.* (2008); Muneta *et al.* (2007); Seon *et al.* (2009); Siebold *et al.* (2008); Steckel *et al.* (2007a), and Yasuda *et al.* (2006) were in agreement with Ishibashi *et al.* (2005) regarding anterior-posterior constraint under anterior loading conditions, whereas others found no significant difference in measured anterior-posterior translation between SB and DB techniques (Adachi *et al.*, 2004; Asagumo *et al.*, 2007; Ferretti *et al.*, 2008; Järvelä, 2007; Streich *et al.*, 2008; Yagi *et al.*, 2007).

Outcome of the clinical pivot shift test was only slightly less ambiguous with most studies in agreement that the DB reconstruction provided superior constraint (Järvelä, 2007; Kondo *et al.*, 2008; Muneta *et al.*, 2007; Siebold *et al.*, 2008; Yagi *et al.*, 2007), while only Asagumo *et al.* (2007) and Streich *et al.* (2008) showed no difference between surgical techniques. To complicate the matter further, both Markolf *et al.* (2008b) and Steckel *et al.* (2007a) showed that the DB reconstruction actually overcorrected joint laxity in rotation, while the SB technique produced results closest to normal.

Table 2.1: Studies comparing single-bundle and double-bundle surgical techniques under passive loading conditions.

Note: Only anatomic (two-tibial + two-femoral tunnel) double-bundle techniques were included; studies investigating non-anatomic (single-tunnel, double-bundle) techniques have, therefore, been excluded from this list. (* quantity of applied load was not indicated. RCT is abbreviation for randomised control trial.)

Author (Year)	**Loads Applied**	**Data Acquired**	**Study Design and Measurement Methods**	**Findings**
Ishibashi *et al.* (2005)	Manual force*: anterior, internal-external torque	Anterior-posterior translation, internal-external rotation	Intraoperative (32 patients); optoelectric bone markers	DB showed improved anterior-posterior constraint to SB; no difference in rotation.
Yasuda *et al.* (2006)	133 N anterior load	Side-to-side anterior laxity	Prospective study (72 patients); KT-2000	Anatomic DB produced better anterior laxity than SB.
Asagumo *et al.* (2007)	Manual force*: anterior drawer, Lachman, pivot shift	Side-to-side anterior and dynamic joint laxity	Retrospective study (123 patients); KT-1000	No differences between DB and SB outcomes.
Järvelä (2007)	134 N anterior force, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (65 patients); KT-1000	DB showed better pivot shift control than SB; no difference in anterior laxity.
Muneta *et al.* (2007)	Manual force*: anterior drawer, Lachman, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (68 patients); KT-1000	DB produced better anterior laxity and pivot shift results than SB.
Seon *et al.* (2007)	No load (passive flexion)	Anterior translation of medial and lateral tibiofemoral compartments	Retrospective study (20 patients); MR imaging	DB produced better lateral side anterior laxity than SB; no difference in medial side laxity.

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Table 2.1 – continued

Author (Year)	Loads Applied	Data Acquired	Study Design and Measurement Methods	Findings
Steckel *et al.* (2007a)	Manual force*: anterior drawer, Lachman	Anterior translation, internal-external rotation	Cadaver; optoelectric bone markers	DB produced better anterior laxity outcome than SB; DB overconstrained rotation.
Yagi *et al.* (2007)	Manual force*: Lachman, pivot shift	Side-to-side anterior and dynamic joint laxity, tibial acceleration	RCT (60 patients); KT-1000, electromagnetic skin sensors	DB showed better pivot shift control than SB; no difference in anterior laxity.
Ferretti *et al.* (2008)	Manual maximum force*: anterior, internal-external torque	Anterior-posterior translation, internal-external rotation	Intraoperative (20 patients); optoelectric bone markers	No differences between DB and SB in anterior laxity or rotation.
Kondo *et al.* (2008)	133 N anterior force, pivot shift	Side-to-side anterior and dynamic joint laxity	Prospective study (328 patients); KT-2000	DB showed better anterior laxity and pivot shift control than SB.
Markolf *et al.* (2008b)	Valgus torque*, pivot shift	Anterior-posterior translation, varus-valgus, and internal-external rotation; graft forces	Cadaver; robotic testing system	DB overcorrected, while SB adequately constrained pivot shift laxity.
Siebold *et al.* (2008)	Manual anterior force*, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (70 patients); KT-1000	DB showed better anterior and pivot shift control than SB.
Streich *et al.* (2008)	134 N anterior force, pivot shift	Side-to-side anterior and dynamic joint laxity	RCT (50 patients); KT-1000	No difference between DB and SB in any measured outcome.

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Table 2.1 – continued

Author (Year)	**Loads Applied**	**Data Acquired**	**Study Design and Measurement Methods**	**Findings**
Markolf *et al.* (2009)	100 N anterior-posterior force, 5 Nm internal, and 5 Nm valgus torque	Anterior-posterior translation, varus-valgus, and internal-external rotation; graft forces and length change	Cadaver; robotic testing system	SB produced graft forces and knee kinematics closest to normal; DB showed high forces in posterolateral bundle near full extension.
Seon *et al.* (2009)	Manual maximum force*: anterior, internal-external torque	Anterior-posterior translation, internal-external rotation	Intraoperative (40 patients); optoelectric bone markers	DB showed improved anterior-posterior and rotational constraint to SB.

2.4.2 The roles of the anteromedial and posterolateral bundles

Greater insight is gained by reviewing the studies that have addressed the functions of the different bundles of the ACL. It is widely accepted that the AM bundle contributes most to knee constraint at higher angles of flexion, while the PL bundle contributes significantly to joint restraint near full extension (Amis & Dawkins, 1991; Chhabra *et al.*, 2006; Fuss, 1989; Gabriel *et al.*, 2004; Jordan *et al.*, 2007; Mae *et al.*, 2006; Robinson *et al.*, 2007; Yasuda *et al.*, 2008; Zantop *et al.*, 2006). During passive flexion-extension both bundles lose tension from 0° to 30° of flexion; however, the decrease in PL bundle tension is more considerable than that of the AM bundle (Amis & Dawkins, 1991; O'Connor & Zavatsky, 1993; Yasuda *et al.*, 2008). While the PL bundle continues to shorten (i.e. slacken) beyond 30° of flexion, the AM bundle begins to tighten again. The AM bundle, is consequently allocated the greatest proportion of the load of the ACL at flexion angles greater than 30° (Amis & Dawkins, 1991).

With the introduction of an externally applied load, the PL bundle has demonstrated similar properties throughout the range of flexion as in the unloaded condition; the AM bundle, however, did not slacken over the first 30° of flexion when an additional load was applied (Gabriel *et al.*, 2004; Jordan *et al.*, 2007; Li *et al.*, 2004a; Mae *et al.*, 2006; Yagi *et al.*, 2002). Instead, the elongation and tension of the AM bundle remained relatively constant or increased between 0° and 60° of flexion and then decreased through 120° of flexion (Gabriel *et al.*, 2004; Jordan *et al.*, 2007; Vercillo *et al.*, 2007). Despite the drop in tension of the AM bundle at higher flexion in the loaded condition, its overall tension was still significantly greater than that of the PL bundle at respective flexion angles (Gabriel *et al.*, 2004).

Although it does not contribute significantly to anterior-posterior joint restraint at higher angles of flexion, the PL bundle is considered important in maintaining rotational constraint, primarily at lower flexion angles (Yasuda *et al.*, 2008). Robinson *et al.* (2007) found that the mean transverse plane rotation measured intraoperatively during the pivot shift was substantially less with an isolated PL bundle than with only the AM bundle. It has furthermore been shown that the graft forces experienced during a simulated pivot shift test are closer to those of the intact ACL when both AM and PL bundles are reconstructed separately as compared to a single-bundle reconstruction in which all four strands of the graft were fixed in a single femoral and single tibial tunnel (Yagi *et al.*, 2002). The ability of the PL bundle to control rotation has been attributed to its more horizontal orientation (Robinson *et al.*, 2007).

2.4.3 Alternative strategies for surgically constraining rotational laxity

Having considered the structures involved in rotational constraint of the knee joint, it would be naïve to surmise that the number of ACL bundles that are reconstructed would be the only surgical consideration to affect joint laxity. The debate as to which injured structures should be surgically mended has been long-established, with advantages including improved mechanical properties of one particular structure and disadvantages encompassing further trauma and possible damage to other joint tissues. Amirault *et al.* (1988) described a study in which Macintosh's lateral substitution reconstruction was performed on 27 patients with chronic ACL deficiency to reinforce the lateral collateral ligament; 75% of these patients showed subjective improvement in knee constraint.

Nordt *et al.* (1999) and Zaffagnini *et al.* (2007) also recognized that residual joint laxity remains with concomitant medial ligament injuries. In a study comparing 20 patients with combined ACL and MCL injury to 37 patients with isolated ACL injury intraoperatively, greater varus-valgus and AP laxity was found in the combined injury group (Zaffagnini *et al.*, 2007). This study supported the post-operative findings of Nordt *et al.* (1999) in which eight of the 21 knees studied in their acutely injured ACL patients accounted for the greatest difference in measured IE rotation between reconstructed and contralateral uninjured knees.

For this reason, it may be imprudent to simply accept the conclusions of those studies comparing SB and DB surgical techniques (Table 2.1) that did not address the ramifications of participants with concomitant ligament or meniscal injuries on their outcomes (Asagumo *et al.*, 2007; Ishibashi *et al.*, 2005; Järvelä, 2007; Muneta *et al.*, 2007). In fact, only two out of thirteen of these *in vivo* studies cited meniscal injury as an exclusion criterion (Seon *et al.*, 2007; Siebold *et al.*, 2008).

Several studies have also considered the effects of tunnel position and resulting graft orientation on rotational laxity following ACL reconstruction. Not only has it been demonstrated that a more oblique or horizontal femoral tunnel position results in better rotational constraint than vertical graft orientation (Scopp *et al.*, 2004; Stevenson & Johnson, 2007; Zaffagnini *et al.*, 2008); it was also found that there was no significant difference in rotational and AP laxity between DB and laterally oriented SB reconstructions at most flexion angles (Yamamoto *et al.*, 2004). Furthermore, anatomical DB surgical reconstructions also showed significantly superior anterior-posterior and rotational constraint when compared to reconstructions with deeper (i.e. non-anatomical and, effectively, more vertical) posterolateral bundle positioning (Zantop *et al.*, 2008).

This should also be taken into account when considering the results of the studies in Table 2.1 that compared the SB and DB techniques. In some cases the SB reconstructions were performed with the graft placed at either the AM or the PL position in order to facilitate further analysis of the DB reconstruction in the same knee (Ishibashi *et al.*, 2005; Steckel *et al.*, 2007a). This may not be the ideal or typical position for the SB graft and may have generated inaccurate results for the SB reconstruction.

Initial graft tension likewise affects knee laxity following both SB and DB reconstruction. In general, increasing initial graft tension was found to decrease translational and rotational kinematics (Markolf *et al.*, 2008b, 2009; Suggs *et al.*, 2003). When measuring laxity specifically in the DB reconstructed knee, the order in which the AM and PL bundles were tensioned, the amount of force in each bundle, and the flexion angle at which they were tensioned all distinctly influenced AP translation and transverse plane rotation (Cuomo *et al.*, 2007; Hoshino *et al.*, 2007; Markolf *et al.*, 2008b, 2009; Suggs *et al.*, 2003).

All of these studies demonstrated tensioning conditions in which either translation and/or rotation was restricted to less than that of the normal knee; suggestions to avoid overconstraint included tensioning both AM and PL bundles simultaneously at low flexion angles (10° to 20° of flexion (Cuomo *et al.*, 2007; Markolf *et al.*, 2008b) and applying moderate to minimal tension, generally less than 40 N, to both bundles (Hoshino *et al.*, 2007; Markolf *et al.*, 2008b, 2009; Suggs *et al.*, 2003). These studies were conducted on cadaveric or computational models, however, so it is not clear whether similar results would be found *in vivo* following a period of graft healing.

2.5 Dynamic joint constraint

Although loading conditions may not be as precise as certain passive joint laxity assessment techniques, the main advantage of dynamic weightbearing analysis is that both passive restraints and actively generated muscle forces interact throughout physiological movement tasks. Measuring stability of the ACL-deficient knee during gait activities that are performed on a regular basis is not only a means by which to determine abnormal laxity that may lead to long-term joint degeneration, but is also an approach that can help investigators to understand the mechanism which resulted in injury in the first place.

Anterior cruciate ligament rupture may be caused by either traumatic or non-contact injury, with the latter being most common (Boden *et al.*, 2000). Non-contact injury often occurs during abrupt deceleration maneuvers involving a subsequent change of direction, such as side-step cutting or landing after a jump (Boden *et al.*, 2000; McLean *et al.*, 1999). Although the exact mechanics that lead to ligament rupture vary, it is commonly thought that most injuries occur immediately following heel strike when the knee is close to full extension and the joint is subjected to both rotational and ab-adduction moments (Boden *et al.*, 2000).

2.5.1 The ACL-deficient knee

Rupture of the ACL has been shown to affect knee biomechanics in many respects, not simply anterior-posterior translation for which this ligament is the primary restraint. Some subjects with ACL deficiency demonstrated reduced knee flexion during weight acceptance as a stabilization strategy to prevent further joint damage (Rudolph *et al.*, 2001; Waite *et al.*, 2005). A reduction in knee flexion moment was also shown to coincide with peak knee flexion during stance (Berchuck *et al.*, 1990; Rudolph *et al.*, 2001; von Porat *et al.*, 2006). Berchuck *et al.* (1990) hypothesized that the reduction in knee moment was due to a decrease in net quadriceps force, effectively minimising the anterior tibial translation that would place stress on the deficient ACL; this was termed 'quadriceps avoidance' gait. In a computational model of normal knee mechanics during walking, Shelburne *et al.* (2004) furthermore found that maximum ACL force occurred during early stance due to anterior shear forces at the knee. Large shear forces were predominantly caused by both the magnitude and anterior direction of the patellar tendon force during weight acceptance (Shelburne *et al.*, 2004).

Using electromyography, some researchers have found that quadriceps muscle activity is not actually reduced during early stance (Reed-Jones & Vallis, 2008; Roberts *et al.*, 1999; Rudolph *et al.*, 2001; Waite *et al.*, 2005); alternatively, the reduced external flexion moment was attributed to greater hamstrings co-contraction, which would result in posterior translation of the proximal tibia (Rudolph *et al.*, 2001). Reducing external flexion moment in order to minimise the risk of anterior tibial translation and ACL strain does not seem to be a consistent strategy across all ACL-deficient individuals, however. In a subsequent study by the same research group that first established the idea of quadriceps avoidance gait, mean flexion moment of the ACLD knees was not found to be significantly lower than the contralateral uninjured knees (Andriacchi & Dyrby, 2005); yet, it was noted that flexion moment varied more in the ACLD subjects than within the normal subject group.

Analysis of transverse plane gait in the ACLD knee is far more limited in the literature than flexion-extension in the sagittal plane for two main reasons: obtaining accurate and reliable internal-external rotational knee kinematics is still problematic due to soft-tissue artefact associated with conventional skin marker-based data collection systems (Alexander & Andriacchi, 2001; Benoit *et al.*, 2007) and the contribution of the ACL to rotational restraint was – until recently – largely overlooked (Georgoulis *et al.*, 2003; Zaffagnini *et al.*, 2000).

As with sagittal plane kinematics and kinetics, IE rotation abnormalities do not appear to be consistent across all ACLD subjects. Georgoulis *et al.* (2003) and Andriacchi & Dyrby (2005) both reported greater internal rotation of the ACL-injured knee when compared with the uninjured knee during walking. In both studies the reduced external rotation occurred during the swing phase of gait; possible mechanisms suggested by the authors included increased activity of the rectus femoris (Georgoulis *et al.*, 2003) or a loss of screw-home movement (Andriacchi & Dyrby, 2005). Zhang *et al.* (2003) on the other hand, demonstrated a net increase in external rotation of the injured knee during walking. An increase and decrease in lateral and medial hamstrings activity, respectively, was proposed as a possible protective mechanism that would result in the observed increase in external rotation (Reed-Jones & Vallis, 2008; Zhang *et al.*, 2003).

The position of the centre of IE rotation on the transverse plane has also been investigated to determine whether it is affected by ACL deficiency. Study results during squatting and deep knee bend activities demonstrated that rotation resulted from movement of the lateral femoral condyle on the tibial plateau while the medial contact point remained unchanged (Dennis *et al.*, 2005; Hill *et al.*, 2000; Johal *et al.*, 2005; Yamaguchi *et al.*, 2009). If the medial compartment contact area remained relatively constant, this would imply that the axis of rotation was on the medial side of the joint. Yamaguchi *et al.* (2009) additionally investigated joint kinematics during a pivoting task in the same group of ACLD individuals and found the centre of rotation occurred just *lateral* of the midpoint of the medial and lateral contact points.

The reason for the contrasting ML centre of rotation position between the two activities was given as the difference in activities: squatting and pivoting were described as 'sagittal' and 'non-sagittal plane' activities, respectively (Yamaguchi *et al.*, 2009). This rationale was not supported by the findings of Koo & Andriacchi (2008), however. They found the knee joint centre of rotation to occur on the lateral side during normal walking, which is considered a sagittal plane activity. The conflicting results of this investigation with those of previous studies were attributed to the difference between ambulatory and non-ambulatory activities. Unfortunately, the findings of Tashman *et al.* (2004) during downhill running were not addressed by Koo & Andriacchi (2008); in that study the observed external rotation was again associated with a shift in contact area in the lateral compartment, suggesting a medial centre of rotation during this ambulatory activity (Tashman *et al.*, 2004).

An additional difference between the two ambulatory activity studies was that the subjects of Tashman *et al.* (2004) had had ACL reconstructive surgery, whereas those participating in the investigation of Koo & Andriacchi (2008) had healthy knees. Although it is possible that this may be the reason for the different findings, this seems dubious given the conclusions of comparable studies that determined centre of rotation positions in injured and uninjured knees. Koo & Andriacchi (2008) found matching patterns of AP and IE motion in the group of 23 healthy subjects and the previous group of 18 subjects (27 knees) with ACL injury; from this they extrapolated a laterally located centre of rotation in ACLD knees. Yamaguchi *et al.* (2009) similary found the location of the centre of rotation, whether on the medial or lateral side, to be consistent between the injured and contralateral knees.

Therefore, although the interpretation of the medial versus lateral position of the centre of rotation during different activities may not be adequately understood, researchers appear to agree that the ML centre of rotation position is not altered by ACL deficiency. Without further information it would be unreasonable to assume that the ACL-reconstructed knee examined by Tashman *et al.* (2004) would differ from both the normal and ACLD knee.

Whereas anterior-posterior laxity is relatively simply to measure with an arthrometer, measurement during dynamic gait is more complex due to the limitations of gait models and data collection systems. Some models, such as that used by the Vicon Clinical Manager, define the knee as a ball-and-socket joint and do not consider tibiofemoral translations at all (Kadaba *et al.*, 1990; Roren, 2005; Vaughan *et al.*, 1999). With side-to-side differences in AP translation being less than 10 mm during passive loading, the accuracy of traditional skin-based marker systems would not necessarily be great enough to differentiate between ACLD and uninjured knees during gait.

More accurate methods of measuring AP translation during dynamic tasks have been devised, however, demonstrating laxity in ACL-injured knees. Beard *et al.* (2001), for example, used optical markers placed on the lateral malleolus, tibial tuberosity, and patella to define the patellar tendon angle, from which anterior tibial translation could be calculated. This study actually measured greater anterior tibial translation *after* ACL reconstruction, whereas no difference had been found between injured and contralateral limb prior to surgery (Beard *et al.*, 2001). Using a point-cluster technique, Andriacchi & Dyrby (2005) found a significant decrease in anterior translation in the ACLD (unreconstructed) knee just prior to heel strike during walking, contrary to the findings of Zhang *et al.* (2003) which showed greater anterior translation of the ACLD knee throughout most of the swing phase of gait. Although differences were found to be statistically significant, no reliability data on their 6 DOF goniometer measurement device were included, nor were quantitative values of AP translation stated in the study by Zhang *et al.* (2003). Waite *et al.* (2005) found no difference in AP laxity between ACLD and contralateral limbs using a method similar to that of Andriacchi & Dyrby (2005).

The counterintuitive findings in AP laxity were interpreted by Andriacchi & Dyrby (2005) as linked to the simultaneously occurring reduction in external rotation; this may be better appreciated by considering the anterior-posterior positions of the medial and lateral tibiofemoral contact points with respect to the location from which AP translation was measured. With the advent of imaging techniques used during dynamic motion analysis, studies have been able to track the positions of the tibiofemoral contact points throughout the activity cycle. Figure 2.6 from Dennis *et al.* (2005) depicts the medial and lateral contact points on the tibial plateau during a deep knee bend. It is clear that rotation caused by anterior-posterior translation of only one side of the femur (i.e. internal-external rotation) corresponds to anterior or posterior translation of the midpoint of the trans-epicondylar femoral axis with respect to the centre of the tibia. In a subsequent publication, Koo & Andriacchi (2008) described a lateral side centre of rotation; if this was also the case with the ACLD subjects in their previous study (Andriacchi & Dyrby, 2005), then it would follow that the observed decrease in external rotation would be accompanied by a decrease in anterior translation.

Since the issue of the medial-lateral position of the axis of rotation during various dynamic activities is still uncertain, and given the interdependence of segment coordinate systems and coupled AP and rotation movements, it may be misguided to base conclusions on measured AP laxity without also considering tibiofemoral IE rotation.

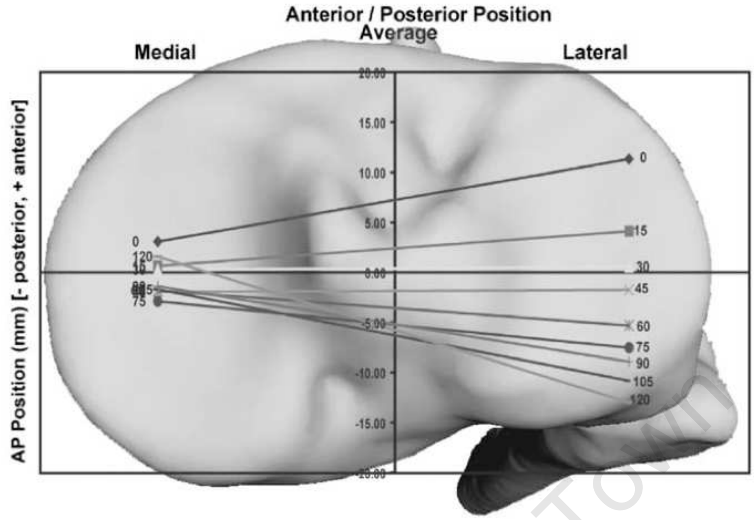


Figure 2.6: Medial and lateral tibiofemoral contact positions plotted during a deep knee bend activity (Dennis *et al.*, 2005).

2.5.2 Does ACL reconstruction restore knee function under physiological loading conditions?

While ACL reconstruction has proven to be an effective treatment for some patients, others have been incapable of returning to pre-injury activity levels. Studies have demonstrated improvement in joint restraint following reconstruction; however, this was often still inferior with respect to the healthy knee (Georgoulis *et al.*, 2003, 2007; Kowalk *et al.*, 1997; Ristanis *et al.*, 2005; Tashman *et al.*, 2004). Gait analysis has been useful in demonstrating complications with specific surgical reconstructive techniques; for example, a study examining biomechanical parameters in subjects who underwent bone-patellar tendon-bone reconstruction during stair ascent found that post-operatively sagittal plane knee joint moment and power was reduced in the injured knee, while moment and power and were increased in the contralateral ankle (Kowalk *et al.*, 1997). The authors suggested that donor site morbidity associated with the BPTB surgical technique resulted in a compensation mechanism that was not present pre-operatively.

These findings were supported by Webster *et al.* (2005), who compared kinematics and kinetics between patients who had received BPTB grafts against those who had received hamstrings tendon grafts. The results also showed a significant reduction in joint flexion moment at mid-stance in the knees that had undergone BPTB reconstruction, in addition to a reduction in extension moment in the hamstrings-reconstructed knees at terminal stance when compared with the control group.

Hamstrings and BPTB reconstruction techniques have been further evaluated in the transverse plane by Chouliaras *et al.* (2007). No differences in IE rotation were found between surgical techniques during a stair descent and pivoting activity; however, both ACL-reconstructed groups were still found to have significantly greater internal rotation than the healthy control group. Although rotational restraint had not been restored in either of these subject groups, previous studies had demonstrated improvement in IE rotational laxity with respect to the pre-operative ACLD state during walking (Georgoulis *et al.*, 2003).

Studies using RSA methods similarly presented unfavourable outcome in transverse plane rotation following ACL reconstruction (Brandsson *et al.*, 2002; Tashman *et al.*, 2004). The subjects evaluated by Brandsson *et al.* (2002) both pre- and post-operatively showed no difference in transverse plane rotation following BPTB reconstruction. Tashman *et al.* (2004), on the other hand, demonstrated an external rotation shift in their ACL-reconstructed subjects when compared with their contralateral uninjured knee during the stance phase of running gait. (The surgical technique used for reconstruction was not specified.)

Despite the growing number of publications comparing passive (i.e. non-weightbearing) laxity outcome between SB and DB surgical techniques, no study could be found examining both types of surgical techniques during dynamic physiological loading activities. Results from passive laxity studies furthermore present conflicting outcomes: some researchers have concluded that the DB technique is superior, several studies have found no statistical differences between clinical function in patients receiving either the SB or DB reconstruction, while still other studies have demonstrated possible complications associated with the DB technique.

A recent meta-analysis of randomised control trials found no significant differences in KT-1000 measured anterior-posterior or pivot shift dynamic joint laxities between SB and DB reconstructions, however, only four studies met the inclusion criteria for their primary analysis (Meredick *et al.*, 2008). Due to the complexity involved in quantitatively measuring *in vivo* rotational kinematics, the results in the literature have not yet been successful in presenting a complete understanding of the biomechanical differences in surgical techniques and the rolls of the AM and PL bundles in constraining torsional loads.

In light of this account of the literature, it is evident that the benefits of the double-bundle surgical technique from a biomechanical standpoint are debatable. Another practical consideration is the added time and cost of this procedure, which would potentially require a 24% reduction in ACL revision surgery to offset its expense (Brophy *et al.*, 2009). A better understanding of not just the ACL, but all structures involved in rotational restraint of the knee, is still required to allow a surgeon to recommend the best treatment for his or her patient.

In summary, the literature on knee rotation and the role played by the ACL (both the native and reconstructed ligament) were presented in this chapter. Cadaver studies have permitted a more precise description of the contribution of various anatomical structures to restraint in the three planes of motion. While *in vivo* studies are more representative of the actual function of the joint, it has been more difficult for investigators to apply precise loads within known and isolated directions of translation and rotation. For this reason, passive laxity studies regarding rotational outcome of single versus double-bundle surgeries are ambiguous: the majority have made conclusions on rotational joint restraint while in fact measuring motion under combined loading situations. Since no weight-bearing studies have assessed these two types of surgery, there is no physiological data with which to compare these passive clinical results. Fundamental research to determine the contribution of these reconstructive techniques is, therefore, still required before conclusions can be drawn on the best procedure to use for each patient.

Chapter 3

Measuring three-dimensional knee kinematics under torsional loading

3.1 Introduction

Studies investigating pathological knee kinematics are focusing increasingly on joint motion in all three planes, rather than simply the primary (i.e. sagittal) plane of motion. It has long been recognized that significant rotation in the transverse plane occurs throughout the range of flexion. Rotational laxity of the knee is now one aspect by which to diagnose knee pathology and evaluate surgical treatment, such as anterior cruciate ligament (ACL) reconstruction (Georgoulis *et al.*, 2003, 2005; Koh *et al.*, 2005; Logan *et al.*, 2004; Mannel *et al.*, 2004; Scopp *et al.*, 2004; Tashman *et al.*, 2004; Yagi *et al.*, 2002; Yamamoto *et al.*, 2004; Zaffagnini *et al.*, 2000).

Prior to medical imaging devices such as magnetic resonance imaging (MRI) that allow the investigator to observe the position and orientation of the underlying bone, non-invasive *in vivo* methods of measuring knee rotation were limited to external devices and skin markers prone to soft tissue artefact. With a much smaller range of knee motion in the transverse plane than in the sagittal plane, it was difficult to acquire results that could be considered reliable with these methods (Koh *et al.*, 2005). Techniques used in cadaveric studies, although more accurate than external devices due to the ability to insert markers directly on the bone, are too invasive to be used *in vivo* on large subject groups. Furthermore, cadaver studies are often confined to older knee specimens, which may not reflect the knee kinematics of a younger population, nor the mechanical properties of *in vivo* tissue.

Several methods for measuring *in vivo* knee and ankle joint kinematics in three dimensions (3D) have now been developed in which the relative positions of the bones were measured using MRI, computed tomography, fluoroscopy and biplane radiography (Bingham & Li, 2006; Fellows *et al.*, 2005b; Küpper *et al.*, 2007; Siegler *et al.*, 2005; Tashman & Anderst, 2003; Udupa *et al.*, 1998; Van Sint Jan *et al.*, 2006). Static or dynamic images of the bones at the joint were registered to their 3D segment models and associated coordinate systems to determine their positions and orientations in 3D space. Results from these studies were found to be more accurate than previous *in vivo* methods of measurement.

In this study, tibiofemoral knee kinematics were measured in 3D while the knee was subjected to torsional loading, thereby simulating a clinical examination. Furthermore, we wished to develop a method by which knee rotations and translations about and along all three axes could be measured *in vivo* with the ultimate intent being its application in the assessment of treatment of knee pathology. Our first objective, therefore, was to design and build a device that would apply a known torsional load to a subject's knee while being scanned using MRI; the images of the knee in the torqued position could then be used to measure six degree-of-freedom motion of the joint accurately and non-invasively. The next objective was to determine the feasibility and repeatability of this methodology with torques applied in internal and external rotation and with the knee in full extension and 30° of flexion. The within-subject variability associated with tibial rotation under different loading conditions would then demonstrate the potential of the system to provide results that are clinically relevant.

3.2 Methods

In order to measure 3D knee laxity objectively under torsional loading, a method for applying a precise load about a fixed axis with respect to the joint was required

(Küpper *et al.*, 2007). Once a specified load could be achieved, position and orientation of the femur and tibia were measured using MRI.

3.2.1 Torsional loading apparatus

In order to simulate clinical examination, the loading apparatus was designed to accommodate the greatest range of knee angles, with limitations governed only by the open-MRI magnet and patient bed. Figure 3.1 shows the computer model of the knee loading device designed around the MRI patient table for imaging of the right knee. Preliminary technical drawings from which the main components were manufactured are included in Appendix A. The aluminum slide rails permitted adjustment of flexion-extension and abduction-adduction angles, which were measured using a goniometer. The subject was positioned semi-supine with the knee joint (within the coil) at the centre of radius of the flexion-extension and ab-adduction tracks, so that only rotation of the shank was required. A plastic boot was connected to the rotation base via extension channels that permitted foot positioning toward the knee coil for shorter subjects as shown in Figure 3.2(a). The knee loading device was rotated about the patient table to permit imaging of the contralateral knee.

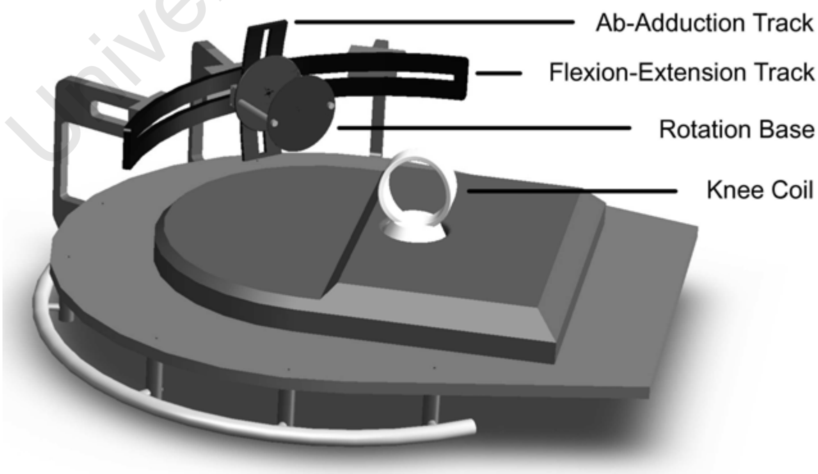


Figure 3.1: Model of knee torsional loading device mounted to MRI patient table.

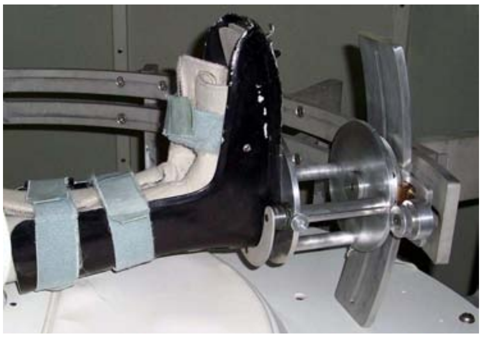
Since the focus of this study was on rotational loading in the transverse plane of the knee, the apparatus was designed to permit torque about only the long axis of the tibia while the other five degrees of freedom were fixed at the distal end of the shank. However, the thigh (and subsequently, the knee joint) was theoretically allowed six degree-of-freedom motion. Virtually full rotational freedom was achievable at the hip and translation was limited by ligament stiffness and joint mechanics, as well as body weight at the pelvis only. While not entirely unconstrained (as would have been the case had the proximal end of the femur been free to move without any restrictions), both rotations and translations at the knee itself were possible, albeit to a limited degree (Zavatsky, 1997).

In order to compare results between individuals and subject groups, a set torque was applied to each knee being examined. However, in order to account for the subject's mass which regularly affects the loads experienced by the knee, the applied torque was normalized to body mass according to the following equation:

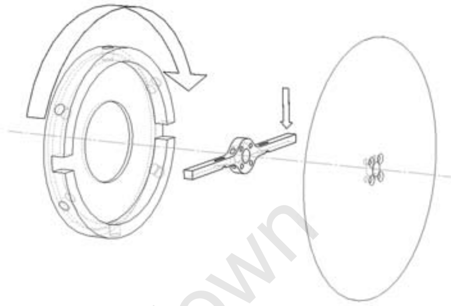
$$T = 0.05 \left[\frac{Nm}{kg}\right] M + 1.25 \; [Nm] \tag{3.1}$$

where T is the applied torque in Newton-meters and M refers to the subject's mass in kilograms. This equation was based on data from the literature in addition to pilot data in which various torque values were recorded from minimum to maximum values corresponding to perceptions of comfort level (Blankevoort & Huiskes, 1996; Kanamori *et al.*, 2000; Mannel *et al.*, 2004; Yagi *et al.*, 2002). For several subjects of varying mass, the above equation represented the median of this range of values.

Two methods were used to measure the magnitude of the applied torque. The first method used an electronic load cell as shown in Figure 3.2(b). An internal torque was manually applied to the torque disc, which transmitted a linear force to one end of the load cell. The load cell was rigidly fixed to the rotation base and, in turn, the foot which resisted the applied torque. The resulting strain deformation in the load cell was measured by a strain gauge with data collected using LabVIEW^{TM} (National Instruments); circuit and calibration details are given in Appendix B. External torque was measured by a second strain gauge mounted onto the opposite side of the load cell. Calibration of each side of the load cell was achieved by hanging weights off the end of a lever arm of known length extending from the torque disc while the rotation base was rigidly fixed in place.



(a) Device in the extended position for shorter subjects.



(b) Load cell with torque disc and foot plate.

Figure 3.2: Torsional loading device components.

Since the strain gauge was continually measuring load, the investigators were able to observe a reduction in torque over time with the boot clamped at the angle corresponding to the specified torque. Presumably, this was due to relaxation of the soft tissues, both at the knee and the hip joints. Details of the supplementary investigation of the effect of relaxation on measured torque can be found in Appendix C. It was therefore decided that the torque would be reapplied following a sufficient period in which the rate of change of torque was less than 0.4 Nm/min. After reapplying the correct torque and ensuring a negligible drop in load for the secured position, the strain-gauge was disconnected to prevent image distortion during MRI scanning.

The second method of measuring the applied torque was a simplified approach involving the investigator (AH) pulling a commercial spring scale connected to the perimeter of the rotation disc of the boot via a thin cord. The load measured by the scale was then converted to a torque value based on the distance from the centre of rotation to the point of application of load (i.e. the radius of the rotation disc). The advantage of this method was that it was simpler and more robust, resulting in reduced set-up time and elimination of malfunctions caused by electrical disturbances. However, the protocol for the spring scale system was modified based on knowledge gained from the electronics method to attain greater accuracy when applying the torque: specifically, a two-minute 'relaxation' period was permitted as described previously before applying the final torque for the scan.

Using either method described above, it is possible that the applied torque would have decreased once the boot had been clamped at the appropriate position. Figure C.1 demonstrated that the drop in load following the designated two-minute relaxation period would not have been more than approximately 0.25 Nm over the entire three minute scan sequence. This quantity of change in torque did not enable sufficient movement during imaging to cause motion artefact and was, therefore, considered acceptable.

3.2.2 Data collection

Six volunteers with no history of knee injury were recruited for this study, the protocol for which was approved by the Human Ethics Committee of the University of Cape Town (Appendix D). Informed consent was given by each subject prior to data collection.

Since this method of measuring knee laxity was intended for use with patients having knee pathology such as ACL injury, it was necessary that the protocol minimize the time that the patient had to endure knee loading. Prolonged stress on an injured knee could not only cause discomfort for the patient, but could also cause muscle tensioning which would affect the contribution of the ligaments to joint constraint. However, a longer MRI scan sequence would generate higher resolution, and consequently more accurate images over the same field of view. Therefore, 3D models of the femur and tibia were generated from high resolution images scanned in a neutral (unloaded) position and shape-matched to models created from low resolution image volumes of the knee scanned under load. By matching the high and low resolution model of each segment, its position and orientation could be accurately determined without requiring a long MRI scan in a torqued position.

Magnetic resonance images were acquired using the 0.2 Tesla dedicated open-MR system (E-Scan XQ, Esaote, Italy) shown in Figure 3.3. Three-dimensional T1-weighted sequences with a 256 x 256 matrix were used for both high and low resolution transverse images (Figure 3.4). Experts including radiologists and medical imaging physicists alike agreed that the quality of these images was comparable to that of scanners with higher field strength (1.5 Tesla or greater) typically used in research studies; the exceptional images could be attained despite the low field strength due to the compact coil that fit closely around the joint.

Figure 3.3: Torsional loading device with subject's knee at 30° of flexion.

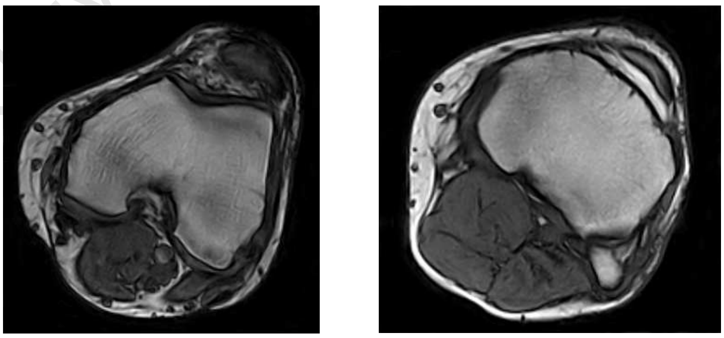


Figure 3.4: Low resolution magnetic resonance images of the femur (left) and tibia (right) scanned while an internal torque was applied to the knee.

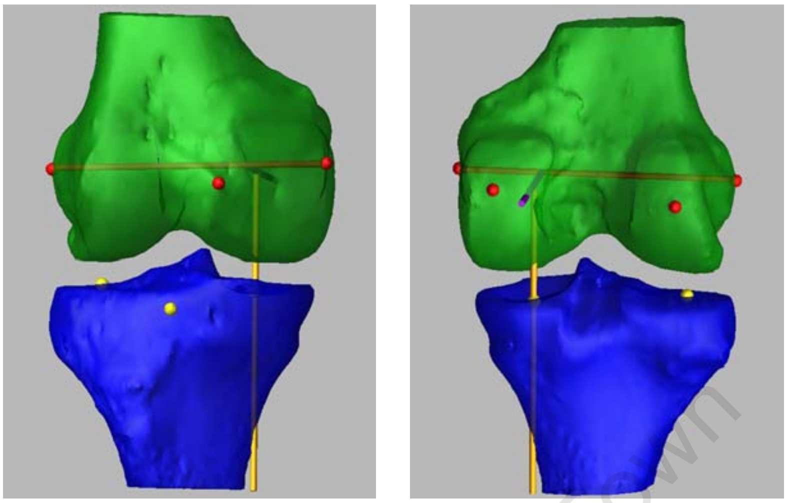
The axial plane was chosen for this sequence as it was the one with the greatest degree of knee motion under torsional loading, and found to have the greatest accuracy when measuring rotation in this plane (Fellows *et al.*, 2005b). The high resolution scan in a neutral knee position generated 90 contiguous slices of 1.56 mm thickness for a 14 cm field-of-view. This 3D image volume was acquired in just over 10 minutes. Four low resolutions scans (22 slices of 6.25 mm thickness) requiring only 2 minutes 50 seconds were taken with the subject's knee under load: internally and externally torqued with the knee in full extension, as well as internally and externally torqued with the knee at 30° of flexion.

3.2.3 Data analysis

Three-dimensional models of the knee were generated from the MR images scanned in the neutral position and for each of the torqued positions using a commercial segmentation software package (Mimics^{TM}, Materialise, Belgium). Point cloud models of each segment were exported to Matlab^{TM} in which the shape-matching procedure was completed. An iterative closest points algorithm based on the method of Fellows *et al.* (2005a) was used to register the points of the high resolution model segment to those of each associated low resolution model. A transformation matrix representing the rotations and translations from the high to low resolution models was recorded and subsequently used in the final description of kinematic position.

Local coordinate systems (LCS) were defined by identifying several anatomical landmarks on the high resolution 3D models of the distal and proximal ends of the femur and tibia, respectively. These 3D position data were then exported into Matlab^{TM} to calculate the LCS. Clinical descriptions of rotation and translation followed the convention developed by Grood & Suntay (1983). The flexion-extension axis was defined as the medial-lateral axis of the femoral coordinate system, the internal-external rotation axis was defined as the long axis of the tibia, and abduction-adduction occurred about the floating axis which was perpendicular to the preceding two axes (Grood & Suntay, 1983).

Figure 3.5 shows the 3D models of the femur and tibia with the anatomical landmarks used to define the LCS for each segment. The y-axis of the right femur extended from the lateral to the medial femoral epicondyle with the origin at its midpoint. (For the left knee the direction was reversed.) A temporary z-axis



(a) Anterior view. (b) Posterior view.

Figure 3.5: Anterior and posterior views of the 3D models of the right femur and tibia with anatomical landmarks. Flexion-extension, internal-external rotation, and abduction-adduction (floating) axes are shown in red, yellow, and purple, respectively.

was normal to the plane defined by the most anterior and posterior points of the medial femoral condyle and the most posterior point on the lateral condyle. The femoral x-axis in the posterior-anterior direction was defined as perpendicular to the y-axis and a temporary z-axis.

The origin of the tibial coordinate system was located in the middle of the medial plateau, since the axis of rotation extends through this position for the flexion range of 10° - 80° (McPherson *et al.*, 2005). The tibial y-axis extended from the lateral to medial tibial plateau midpoints for the right knee and was reversed for the left. The midpoints of the medial and lateral plateaus were defined as the most distal point in the central area of each plateau and could easily be identified on the 3D model. The z-axis was defined as normal to the plane of contact of the femoral condyles, i.e. the tibial plateau. The plane was defined as having the previously described points on the medial and lateral tibial plateaus, as well as the most anterior point on the most proximal slice of the tibial medial condyle. The x-axis of the tibia in the posterior-anterior direction was calculated as the cross-product of the y- and z-axes.

The clinical rotations and translations relating the tibial coordinate system to the femoral coordinate system were calculated before and after loading based on the previously determined transformation matrices derived from the shape-matching algorithm. The position of the tibia under the four conditions of torsional loading was always calculated with respect to the femur in the unloaded neutral position.

3.2.4 Feasibility study

A representative knee model, composed of two cylindrical MRI phantoms designating the femur and tibia respectively, was used to measure the accuracy of the segmentation and shape-matching analysis. Each phantom knee segment was manually positioned on specially designed cradles: one simulating 0° of knee flexion and the other simulating 30° of knee flexion (Figure 3.6). For each value of flexion, the tibial phantom was rotated externally by 20° and internally by 30° and scanned in each position using the low resolution scanning sequence described above. An additional scan simulating 0° of flexion and 0° of rotation was conducted using the high resolution scanning sequence. Local coordinate systems for each segment were aligned with the geometry of the high resolution phantom model rather than theoretical knee landmarks. Measurement of the position and orientation of the tibial with respect to the femoral component in the simulated torqued positions was carried out according to the protocol described in section 3.2.3.

Figure 3.6: Phantom model simulating a left knee at 30° of flexion.

3.2.5 *In vivo* repeatability study

A repeatability study was undertaken to measure the variability of knee joint kinematics under torsional load with each subject's knee at two angles: 30° of flexion and full extension. The protocol outlined in sections 3.2.2 and 3.2.3 was repeated five times by a single investigator (AH) on one knee of each of six subjects, using only one high resolution scan to build a 3D model and segment coordinate systems for each of the five trials. It was presumed that the greatest variation in knee kinematics would be associated with knee morphology under the specified load during data collection rather than the segmentation or shape-matching protocols with which associated errors had already been measured by the phantom knee model.

However, to verify this hypothesis and to limit any inaccuracy associated with the investigator's chosen anatomical landmarks, high resolution MR images were scanned for each trial for one of the six subjects. From each high resolution knee scan, 3D models were created and landmarks were identified to build the LCS for each segment. For these five trials, the repeatability of the identification of the knee landmarks was measured and the effects on the overall knee kinematics in the torqued positions were determined.

One female and five male subjects (age 29.3 $\pm$ 3.6 years, height 178.0 $\pm$ 8.6 cm, mass 72.0 $\pm$ 13.0 kg) were recruited for this study. Subjects had no history of injury for the knee joint of interest. Two left knees and four right knees were examined. A minimum of one day was given between trials for each subject, except for Subject 2 whose five trials were conducted over two days due to time constraints. Intraclass correlation coefficients (ICC) and standard error of measurement (SEM) were calculated for range of rotation data in both extended and flexed knee positions.

3.3 Results

Rotations calculated from the position of the phantom knee model using the segmentation and shape-matching protocol were compared with the actual rotations about the three clinical axes (Table 3.1). In both the simulated extended and flexed positions, the measured degree of internal and external rotation was within 1.6° of the actual rotated position.

Table 3.1: Actual and measured three-dimensional rotation angles (degrees) for the phantom knee model used for validation of the segmentation and shape-matching protocol.

Simulated Torque	**Knee Angle**	**Knee Extended**		**Knee Flexed** 30°	
		Measured	Actual	Measured	Actual
External	flexion	-0.2	0.0	27.7	30.0
	adduction	0.0	0.0	0.3	0.0
	external rotation	18.4	20.0	18.4	20.0
Internal	flexion	-0.6	0.0	28.0	30.0
	adduction	-0.3	0.0	0.8	0.0
	external rotation	-30.3	-30.0	-29.3	-30.0

Standard deviations in all three planes for all landmarks identified on the five neutral scans for Subject 1 were found to be less than 2 mm, except for the landmark on the anterior surface of the medial tibial plateau where the standard deviation in the medial-lateral direction was 2.2 mm (Table 3.2). The effect of the landmark position variability on the overall values of tibial rotation was minimal as demonstrated by Figure 3.7.

Table 3.2: Standard deviations (mm) of global x, y, and z positions of knee landmarks on one subject over 5 trials.

Bone	**Knee Landmarks**	Standard Deviation of Position		
		x	**y**	**z**
Femur	medial epicondyle	0.1	0.3	1.2
	lateral epicondyle	0.7	0.3	0.9
	posterior surface, medial condyle	0.3	1.7	0.7
	anterior surface, medial condyle	0.3	0.9	1.3
	posterior surface, lateral condyle	0.3	1.2	1.1
Tibia	medial plateau, centre	1.3	1.5	0.7
	lateral plateau, centre	0.4	0.6	0.5
	anterior surface, medial plateau	0.4	2.2	0.8

Mean ranges of tibial rotation for the six subjects varied between 11.6° and 32.2° for the extended position and 17.2° and 28.8° for the flexed position; stan-

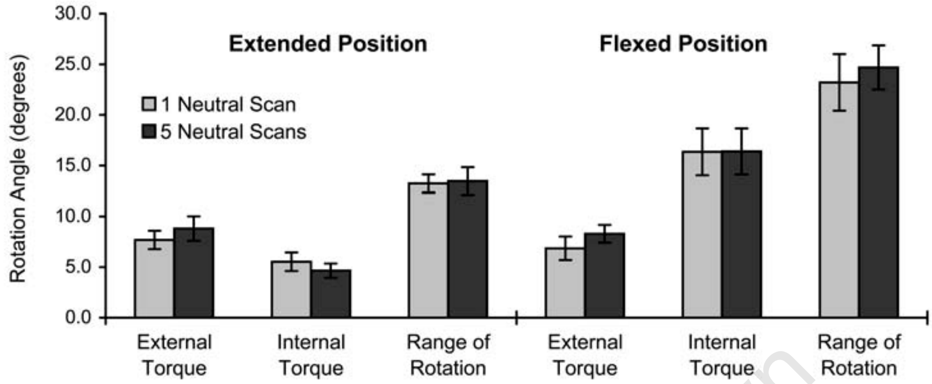


Figure 3.7: Mean and standard deviation of absolute tibial rotation angle (degrees) under external and internal torque loading for Subject 1 using one versus five neutral scans.

dard deviations over the five trials for external and internal rotations were consistently less than 2.5° (Table 3.3). ICC-values for the range of rotation were 0.99 and 0.93 in the extended and flexed knee positions, respectively. The standard error of measure was less than 0.75° for all subjects in both flexion and extension.

Table 3.3: Mean and standard deviation of knee rotation angles under torsional loading (convention: external = positive; internal = negative) and range of rotation for 6 subjects in extended and flexed positions based on 5 sets of data. Rotation angles are in degrees and applied torques are in Nm.

Sbj	Appl Torq	**Knee Extended**						**Knee Flexed 30°**					
		Ext Torq		Int Torq		Range		Ext Torq		Int Torq		Range	
		mean	SD	mean	SD	mean	SD	mean	SD	mean	SD	mean	SD
1	3.9	7.7	0.9	-5.5	0.9	13.2	0.9	6.9	1.2	-16.3	2.3	23.2	2.8
2	5.3	26.7	1.5	-5.5	0.8	32.2	1.7	19.9	1.9	-8.8	1.2	28.8	2.5
3	4.5	6.6	0.7	-5.0	1.6	11.6	1.7	9.0	1.4	-8.2	2.2	17.2	2.4
4	4.7	9.2	0.6	-12.8	2.4	22.0	2.8	8.7	1.2	-14.7	1.5	23.4	2.6
5	5.0	4.9	1.3	-8.6	1.3	13.5	2.2	2.8	2.0	-21.4	1.8	24.2	1.2
6	5.8	7.8	1.8	-7.1	1.6	14.9	2.1	5.7	0.9	-17.8	0.8	23.4	1.4

3.4 Discussion

3.4.1 Phantom knee model

The greatest discrepancy between actual and measured knee phantom position was in the degree of knee flexion in the simulated flexed position. This supported the findings of Fellows *et al.* (2005b) in which it was shown that greater accuracy could be obtained with MR images taken in the plane of motion (i.e. transverse images for rotation in the transverse plane). Since the primary focus of this technique was to measure knee rotation in the transverse plane, with measures of flexion in the sagittal plane being only a secondary objective, these results were considered acceptable.

This investigation gave an indication as to the accuracy of using the Matlab^{TM} registration procedure to match the 3D models produced from the low and high resolution scans. While the shape and features of the phantom did not correspond well to the tibia or femur, the number of registration points generated from the cylinders was similar to that of the *in vivo* bone segments and, therefore, adequately represented the knee joint for the purpose of this sub-study.

3.4.2 Anatomical landmark position

The variation in calculated landmark position over the five trials collected for Subject 1 could be attributed to inconsistencies in the MRI scans or inaccuracies in data processing, such as segmentation of the images and identification of the landmarks on the 3D segment models. The tibial and femoral landmarks for which the greatest standard deviations were measured – the anterior surface of the medial tibial plateau and the posterior surface of the medial condyle – were both used to define the transverse planes of their respective segments and the corresponding normal axes. The identification of the anterior surface of the tibial medial plateau in particular was, therefore, more crucial along the z-axis, rather than the y-axis, as it would be a discrepancy in the distal-proximal direction that would change the orientation of the transverse plane and corresponding axis of rotation.

All observed differences in the overall knee kinematics due to the use of only one versus the complete set of five neutral scans were less than the data processing errors associated with the knee phantom model in Table 3.1. In general, greater variation in the identification of positions of the tibial landmarks was observed. This was because the anatomical features on the proximal tibia chosen to define clinical segment axes were not as prominent as on the distal femur. Suitable landmarks at the distal end of the tibia were not within the limited field of view of the scanner available to us, and could therefore not be used. However, since the variation in knee kinematics was great enough to show differences under specific loading conditions as shown in Table 3.3, it was concluded that using only one high resolution neutral scan to analyse all five trials would be acceptable for each of the remaining subjects.

3.4.3 Measures of clinical rotation under torsional load for six subjects

In this study, an MRI-compatible torsional loading device, as well as data collection and image analysis protocol were developed to measure rotational knee laxity; its feasibility was tested using a phantom knee model. Results showed clinically relevant differences in the degree of knee rotation under four rotational loading conditions. All subjects demonstrated an increase in internal rotation with the knee flexed (Table 3.3), which agreed with the findings of Kanamori *et al.* (2000) and Musahl *et al.* (2007). Although standard deviations for each subject were greater than those reported by Musahl *et al.* (2007), their study used invasive bicortical pins on cadavers for a best case scenario. The large disparity in tibial rotation values in this study, in addition to smaller individual standard deviations versus those reported by Kanamori *et al.* (2000) for 12 cadaveric knees, indicated that variation across a subject group was more substantial than within repeated trials of an individual.

The accuracy of this methodology is furthermore superior to other non-invasive systems that have been used to measure *in vivo* knee rotation. External skin-mounted tracking sensors were used by both Shultz *et al.* (2007) and Tsai *et al.* (2008) with ICC-values between different testing sessions reported as 0.91 and

0.81 for the two investigations, respectively. The advantage of using MRI to prevent soft tissue artefact was also demonstrated by Okazaki *et al.* (2007) who demonstrated ICC of 0.96 and 0.98 when measuring the anterior tibial translation of the medial and lateral compartments at 10° of flexion; these values were comparable to those of 0.99 and 0.93 calculated from our data at full extension and 30° of flexion, respectively. The benefit of our methodology is that the 'matching' of unloaded and torqued knee models is determined mathematically from two sets of MRI data, rather than a comparison of invasive fluoroscopic images with MRI scans as required by the technology used by Okazaki *et al.* (2007).

An advantage of our methodology was the level of accuracy that was maintained despite the decreased MRI scan time required for patients having pain associated with knee pathology. This could be attributed to the individualized bone segment matching protocol, in which the low resolution 3D image volumes were matched to high resolution models developed from the subject's own knee, rather than bone segments from a database. This was reflected in the low SEM and high ICC-values calculated for the range of rotation, which suggest excellent agreement of the data over the different testing days. The non-invasive MRI technique permitted accurate measurement of the underlying bone, thereby avoiding skin motion artefact. Furthermore, it allowed the visualization of soft tissues around the joint; injury to these tissues may best be seen under load. The MRI-compatible torsional loading device and image analysis methodology developed in this study has been demonstrated to provide useful information for further investigation into normal and pathological knee laxity.

Chapter 4

In vivo joint laxity under torsional loading in the healthy knee

4.1 Introduction

In order to characterise pathological changes in joint stability, one must first have an understanding of the biomechanics of the healthy knee joint. The predominant motion of the knee is flexion in the sagittal plane; however, it has long been shown that physiological rotations and translations occur in all three planes of motion. The screw-home mechanism characteristic of the healthy knee is the coupled internal rotation of the tibia with respect to the femur as it flexes; at full extension, the coupled external rotation provides joint restraint (Benoit *et al.*, 2007; Chen *et al.*, 2001; Crawford *et al.*, 2007; Koh *et al.*, 2005; Moglo & Shirazi-adl, 2005; Piazza & Cavanagh, 2000; Shefelbine *et al.*, 2006; Wilson *et al.*, 2000). Furthermore, variable degrees of varus-valgus, anterior-posterior, medial-lateral, and distal-proximal laxities have been measured in healthy subjects under dynamic and passive loading conditions (Benoit *et al.*, 2007; Dennis *et al.*, 2005; Georgoulis *et al.*, 2003; Küpper *et al.*, 2007; Li *et al.*, 2006; Zhang *et al.*, 2003).

The geometries, configurations, and properties of various anatomical structures that comprise the knee, provide this complex joint with the stability required to withstand most loading situations accompanying daily tasks. The articular surfaces of the tibial plateau and femoral condyles, the cruciate and collateral ligaments, iliotibial tract, posterior oblique ligament, arcuate ligament, and menisci are among the main structures indicated to maintain passive restraint of the knee (Amirault *et al.*, 1988; Amis *et al.*, 2005; Blankevoort & Huiskes, 1996; Defrate *et al.*, 2004; Meyer & Haut, 2008; Nordt *et al.*, 1999). Their contributions to overall restraint is dependent on the direction and magnitude of the applied loads.

Considerable research has been dedicated to the mechanism by which the anterior cruciate ligament (ACL) contributes to joint constraint; in most recent years the focus has been specifically on rotational restraint in the transverse plane. Although the exact mechanism of ACL rupture is unknown, it is thought that non-contact injury generally occurs with concomitant valgus bending and external rotation of the joint (Meyer & Haut, 2008). However, knee kinematics have more often been measured under a combined internal and valgus rotatory load simulating the pivot shift phenomenon, which has been shown to correlate to laxity symptoms associated with ACL injury (Amis *et al.*, 2005). Due to the difficulty in quantifying the pivot shift motion, outcome measures *in vivo* have largely been limited to subjective grading systems (Amirault *et al.*, 1988; Järvelä, 2007; Meredick *et al.*, 2008; Streich *et al.*, 2008; Yasuda *et al.*, 2006). Kubo *et al.* (2007) and Yagi *et al.* (2007) presented methods of measuring velocity and acceleration of the tibiofemoral motions as a means by which to quantify the pivot shift; however, actual applied varus-valgus and rotational loads were not measured.

Most clinical trials reporting quantitative laxity measured under a known load used an anterior-posterior (AP) laxity arthrometer, the most accessible validated measurement tool, and thus were limited to AP laxity in a single plane (Meredick *et al.*, 2008). Intra-operative navigation systems have been used to get three-dimensional (3D) quantitative kinematic data; again however, the precise loads applied by the surgeon were generally not recorded (Ferretti *et al.*, 2008; Martelli *et al.*, 2007; Zaffagnini *et al.*, 2007). Several *in vitro* studies have applied precise torques, either independently or combined with varus-valgus or AP loads, and measured resulting internal-external rotations (Amis & Scammell, 1993; Diermann *et al.*, 2008; Gabriel *et al.*, 2004; Kanamori *et al.*, 2000; Kaneda *et al.*, 1997;

Mannel *et al.*, 2004; Meyer & Haut, 2008; Scopp *et al.*, 2004; Yamamoto *et al.*, 2004). Far fewer studies were found in which tibiofemoral transverse plane rotation was measured under known torsional loading *in vivo* (Almquist *et al.*, 2002; Nordt *et al.*, 1999; Schmitz *et al.*, 2008; Shultz *et al.*, 2007; Tsai *et al.*, 2008); due to measurement methods, however, it may not have been feasible to evaluate joint motion in the other anatomical planes.

This chapter investigates the six degree-of-freedom kinematics resulting from internal and external torsional loads applied to the healthy knee at two positions of flexion: full extension at which the knee is locked and rotation is thought to be restricted (Benoit *et al.*, 2007; Crawford *et al.*, 2007; Koh *et al.*, 2005) and 30° of flexion at which non-contact ACL injury commonly arises (Boden *et al.*, 2000). The data gathered was used to establish the normal variability of knee motion in a healthy population under these loading conditions. By testing both left and right knees of each subject, symmetry could be verified in order to support the use of patients' contralateral limbs as controls in future studies. Understanding this data and the mechanisms by which the knee is able to restrain rotational loads is an essential baseline to determine the effects of ACL pathology such as rupture or reconstruction using various surgical techniques.

4.2 Methods

Fifteen subjects (4 female, 11 male) with no history of knee injury were recruited for this study. Informed consent was given by each subject, as required by the protocol approved by the University of Cape Town Ethics Committee (Appendix D). Subjects ranged in age from 22 to 43 years of age and were all moderately to very physically active, representing a normal population in which ACL rupture may occur as a result of sporting injuries. Demographic data is included in Table E.1.

The data collection and analysis protocol followed for this study was described in detail in Chapter 3. External and internal torques were applied to both left and right knees while each knee was in full extension and then repositioned to 30° of flexion. Applied torques were normalized to each subject's body mass. Low resolution 3D T1-weighted images (6.25 mm slice thickness) were generated by the 0.2 Tesla magnetic resonance imaging (MRI) scanner in less than 3 minutes while the joint was under load. The 3D image volume was then shape-matched to a high resolution image volume (1.56 mm slice thickness) scanned in a no-load position. Three-dimensional rotations and translations of the tibia with respect to the femur were calculated by comparing the transformation matrices before and after torque was applied.

Statistical analysis was performed using SPSS 15.0 (SPSS Inc) software. Paired t-tests were used to detect differences in left and right knee measures.

4.3 Results

4.3.1 Torque applied versus rotation measured

Absolute correlation coefficients for applied torsional load versus range of rotation were both less than 0.37, signifying linear independence of these variables in both extended and flexed knee positions (Figure 4.1).

4.3.2 Six degree-of-freedom knee kinematics

Overall subject means and standard deviations for the translations and rotations in the three anatomical planes indicate that the greatest tibiofemoral movement under torsional loading was in internal-external rotation (Figure 4.2) and anterior-posterior translation (Figure 4.3). The increase in range of rotation from 15.4° and 14.3° (left and right, respectively) in extension to 25.6° and 23.5° in the flexed knee position was primarily due to an increase in internal tibial rotation; external rotation values remained similar in the two positions of flexion (Figure 4.2). Measured values of tibiofemoral flexion were approximately 3° - 5° higher under external torque versus internal torque in both positions of flexion. Variation in knee flexion-extension was greater than in ab-adduction, and in general, standard deviations tended to be greater in the flexed knee position. Subject-specific data and individual ranges of rotation are listed in Table E.1.

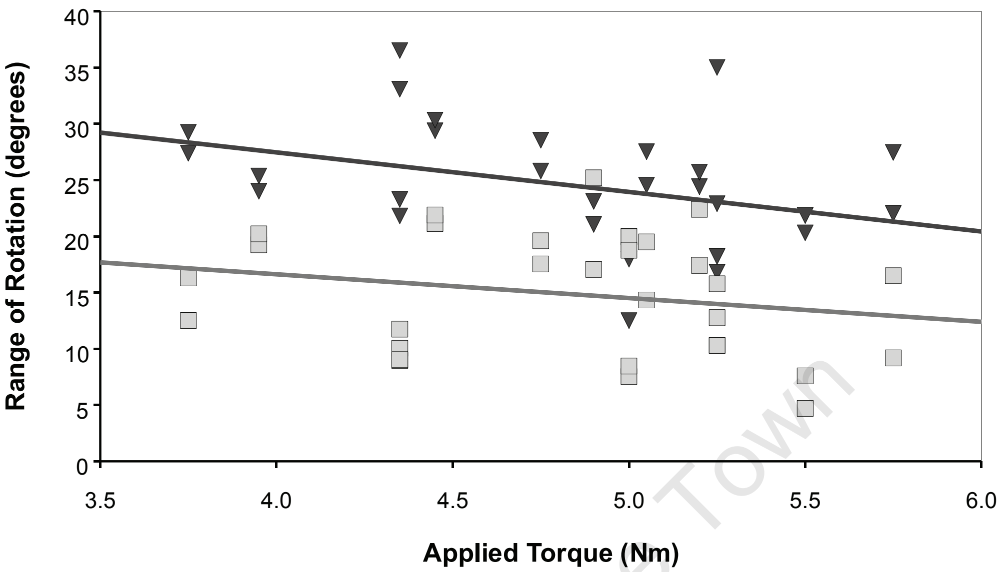


Figure 4.1: Applied torque versus measured rotation for left and right knees. Extended position (■) and flexed position (▼) data are shown with regression lines. R-squared values for linear regressions in extended and flexed positions are 0.05 and 0.13, respectively.

4.3.3 Rotation coupled with anterior-posterior translation

Figures 4.4 and 4.5 show that there was a correlation between internal-external rotation and anterior-posterior translation under torsional loading in both the extended and flexed positions. In the flexed position with an internal torque, a smaller translation was coupled with rotation as compared to the other three loading conditions; this is demonstrated by the smaller slope of the linear regression curve.

4.3.4 Rotation: Left–right symmetry

Significant differences in external rotation were found between left and right knees in both extended and flexed knee positions with the left knee showing greater rotation (Table 4.1). In the internally rotated positions, however, the right knees tended to have increased laxities.

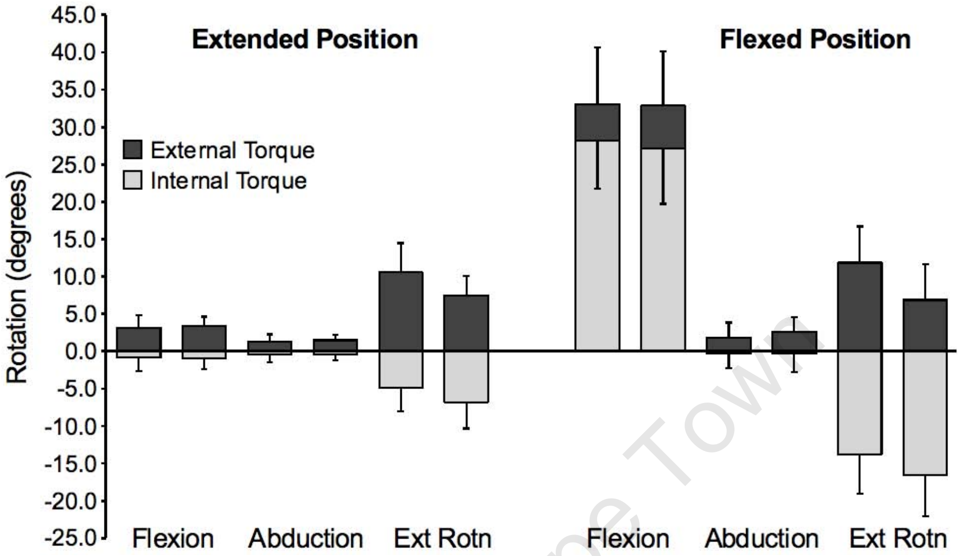


Figure 4.2: Rotation in three planes under torsional loading in extended and flexed knee positions. All values start at 0. Left and right knee data are presented side-by-side for each rotation with positive directions as marked: Flexion, Abduction, and External Rotation.

Table 4.1: Mean left-right differences in absolute internal and external rotations with levels of significance.

Applied Torque	**Knee Extended**		**Knee Flexed 30°**	
	Mean difference	p-value	Mean difference	p-value
External	3.0	0.009	5.0	0.001
Internal	-1.9	0.052	-2.8	0.064

4.4 Discussion

The purpose of this study was to determine the 3D kinematics resulting from internal and external torsional loads in the healthy knee joint in order to establish a baseline against which data from ensuing studies involving ACL patients can be compared. Unlike other studies in which kinematics were measured under externally applied loads, this investigation normalized the torque applied according to each subject's body mass. The normalization equation (equation 3.1) assumes a

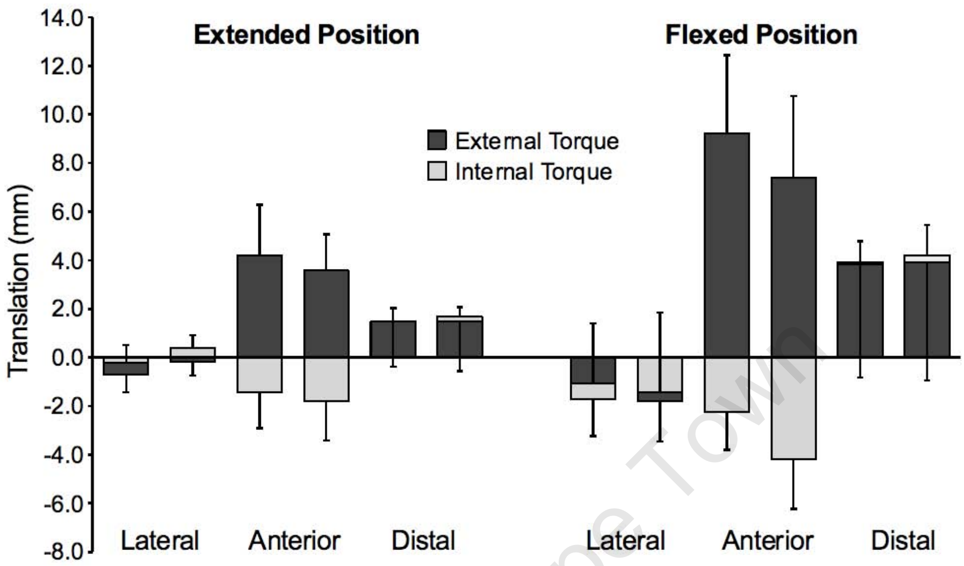


Figure 4.3: Translation in three planes under torsional loading in extended and flexed knee positions. All values start at 0. Left and right knee data are presented side-by-side for each translation with positive directions as marked: Lateral, Anterior, and Distal.

direct relation between subject mass and the torque that can be tolerated, based on observations made in our pilot study. This equation permitted standardization of the applied load, while allowing the greatest possible load to be used without causing discomfort to the subject. Almquist *et al.* (2002) measured rotation using torques of 3 Nm, 6 Nm, and 9 Nm and showed that there is a direct relationship between torque and range of rotation when applied to the same knee at the same flexion angle. Since range of rotation was shown to be independent of applied torque as demonstrated in Figure 4.1, the normalization used was not only valid, but essential when comparing knee laxities of subjects with varying mass.

Four other investigations that applied rotational loads to the knee *in vivo* all used a distinct torque values of either 5 Nm (Nordt *et al.*, 1999; Schmitz *et al.*, 2008; Shultz *et al.*, 2007) or 6 Nm (Tsai *et al.*, 2008) for every subject tested, despite subject mass standard deviations of up to 11.4 kg. One of our subjects weighing only 54 kg expressed mild discomfort with the 4 Nm external torque applied in the flexed position; a 5 Nm torque would be expected to cause this

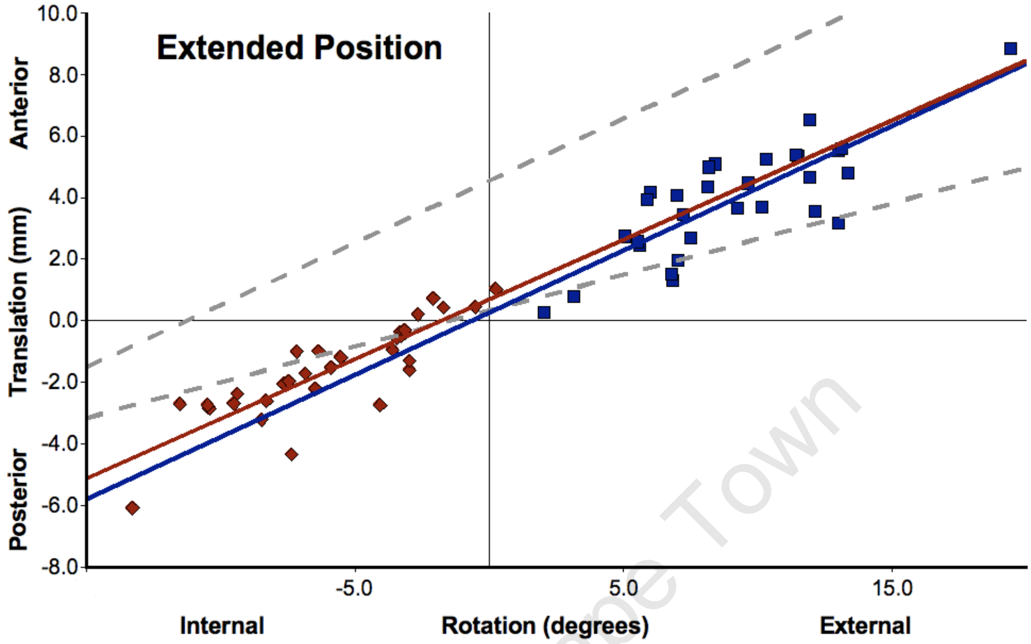


Figure 4.4: Internal-external rotation versus anterior-posterior translation for left and right knees of 15 subjects in the extended knee position. External torque (■) and internal torque (◆) are shown with regression lines. (External and internal torque regressions in flexed position from Figure 4.5 are shown as grey dashed lines for comparison.)

subject to contract the muscles surrounding the joint so as to resist the load, thereby affecting the measured passive laxity. Since subject-specific data were not presented by other authors, it is not known whether their results using a distinct load exhibited a relationship between subject mass and measured laxity. In this study, the mean torque applied was 4.8 Nm, which is very close to the single value used by other researchers; therefore, it is reasonable to compare mean outcomes from the different studies given the lack of similarly derived data in the literature.

The larger variations in knee kinematics observed in the flexed position may be attributed to the imprecise positioning of the knee by the investigator using the manual goniometer. The standard deviations of between 6.4° and 7.6° corresponding to flexion angle readings in the flexed knee position are within normal limits of accuracy using this equipment (Jagodzinski *et al.*, 2000) and are the reason exact measures of knee flexion were recorded using the 3D models developed from the MR images. The low variability associated with the ab-adduction values, and the fact that they were close to 0° in all positions of loading, are a good indicator that there was minimal kinematic crosstalk in the measurements (Charlton *et al.*, 2004; Piazza & Cavanagh, 2000).

The magnitudes of rotational laxity measured in the extended and flexed positions agree well with data from six separate subjects used to validate the methodology (Chapter 3), as well as results from published studies. In cadaveric studies, increases in rotational laxity from an isolated internal torque ranged from just under 8° to approximately 12° (Blankevoort *et al.*, 1988; Kanamori *et al.*, 2000; Musahl *et al.*, 2007), comparable to our findings of a left-right mean increase of 9.5° of internal rotation. Absolute magnitudes of rotation in the extended position closely matched data presented by Blankevoort *et al.* (1988). However, their results in the flexed position and those of Kanamori *et al.* (2000) and Musahl *et al.* (2007) were about 7° to 8° larger in both external and internal

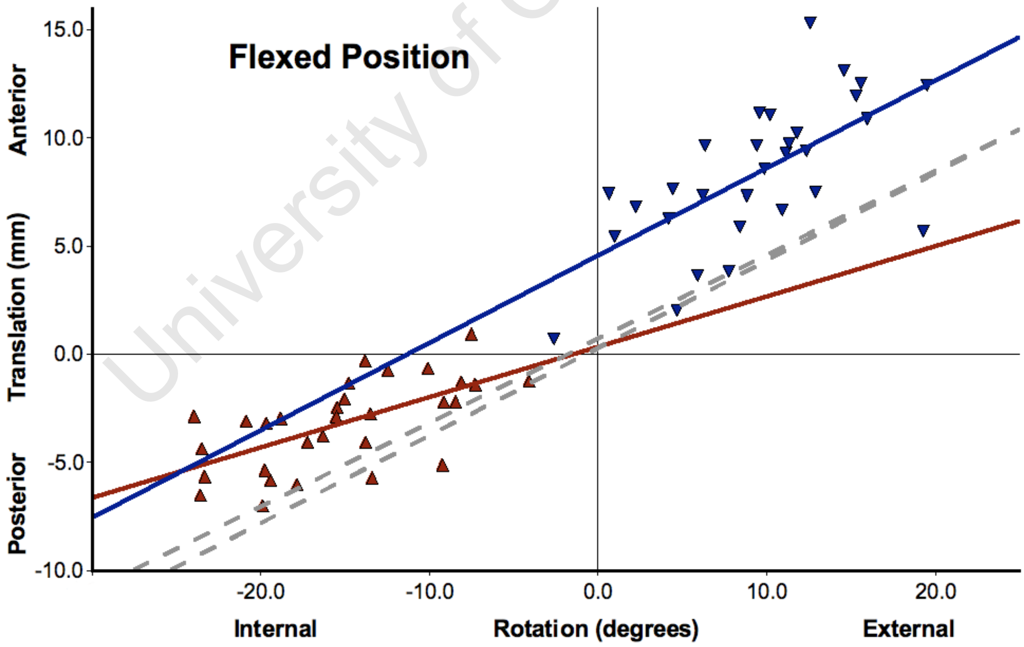


Figure 4.5: Internal-external rotation versus anterior-posterior translation for left and right knees of 15 subjects in the flexed knee position. External torque (▼) and internal torque (▲) are shown with regression lines. (External and internal torque regressions in extended position from Figure 4.4 are shown as grey dashed lines for comparison.)

rotation than the findings of the present study (i.e. about 15° larger in the overall range of rotation). Nonetheless, these rotations were exceeded by at least one subject in each position in our study.

Magnitudes of rotation from our study more closely match results given by Shultz *et al.* (2007) and Nordt *et al.* (1999) measured at 20° of knee flexion, and Tsai *et al.* (2008) at 30° of flexion, indicating that unconscious muscle tensioning may have contributed to joint stiffness *in vivo*. Interestingly, under external torsional loading, our results showed similar rotational laxities in both positions of flexion, unlike the findings of Musahl *et al.* (2007). Since no other study with comparable measurements of external rotation at 0° and 30° of flexion could be found, it cannot be concluded that the contrasts can only be attributed to differences in study design (e.g. *in vivo* versus *in vitro* models).

Interestingly, significant differences of up to 5° in transverse plane rotation were found between left and right knees in the four different loading conditions. Only two other studies could be found in which bilateral knee rotation was measured in healthy subjects (Shultz *et al.*, 2007; Tsai *et al.*, 2008). Neither of these studies found significant side-to-side differences; however, methods of measurement involved skin markers prone to soft tissue artefact, resulting in measurement errors of 5° or more.

The standard error of measurement (SEM) using our methodology was less than 1°, as shown in Chapter 3; therefore, the standard deviations of up to 5.5° in our subject group reflect true inter-subject variation, rather than measurement error. This variation across subjects indicates that knee rotation may vary substantially in a healthy population; the observed side-to-side differences, although statistically significant, may not be clinically relevant. The difference in range of rotation was less than 2.3°, however, which may be evidence that this is a more meaningful measure when using the contralateral limb as a control in studies involving knee pathology.

The asymmetry of internal and external rotation due to torsional loading may be explained in part by the viscoelastic behavior of ligaments. With the knee in flexion, a substantial degree of 'primary' rotation – easily up to or beyond 10° in each direction – occurs with relatively small (1 to 2 Nm) initial torque values (Musahl *et al.*, 2007; Wang & Walker, 1974). Rotation in excess of this initial laxity requires disproportionately more torque due to the non-linear stiffness of the ligaments (Woo *et al.*, 2006).

The neutral resting position of the subject in which the high resolution MRI scan was performed with the knee in full extension was *not* assumed to be at 0° of rotation in our study. Instead, the degree of neutral position rotation was subtracted from the torqued measure of rotation to calculate the net rotation under load. At full extension, the degree of primary laxity is likely less than 10° in each direction. However, an imbalance in an individual's neutral knee position may consistently fatigue the rotational restraints in one direction to a greater extent than in the other, resulting in an imbalance in ligament laxity. This would not only account for the differences of up to 5° in one direction, but would also account for the *smaller* differences in total range of knee rotation. If the bilateral differences in internal and external rotational laxity may be attributed to variations occurring within the initial range of primary laxity, this quantity may not be relevant when diagnosing knee injuries under passive loading conditions.

One topic of interest when investigating kinematics of the knee joint is its axis of rotation; the relationship between internal-external rotation and AP translation shown in Figures 4.4 and 4.5 give a good indication as to its position in the four loading conditions investigated in this study.

Although the helical axis could have been calculated to determine the precise location of the centre of rotation, its physical interpretation is clinically meaningless unless no other rotation or translation aside from internal-external rotation and distal-proximal translation were to result from the applied torque. This could not be assumed; nor could it be taken for granted that the helical axis of rotation would be similar for both internal and external rotational loading. Furthermore, to make the data comparable between subjects after establishing an average helical axis and to 'convert' the results to clinically comprehensible rotations and translations, an anatomical reference frame would still be required from which sagittal, coronal, and transverse plane motions could subsequently be calculated (Dennis *et al.*, 2005).

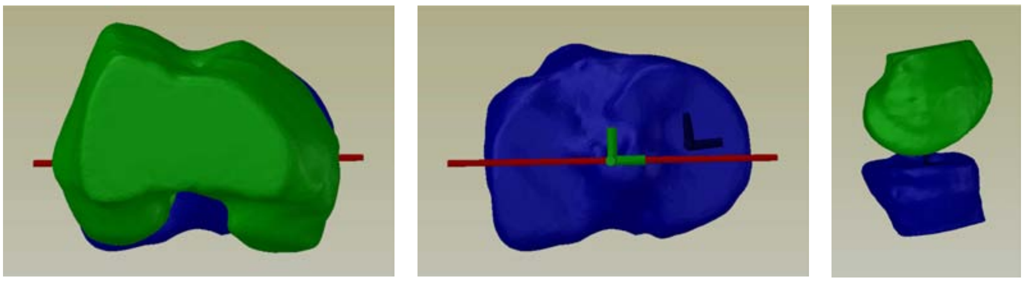
The kinematic model, based on the Grood & Suntay (1983) joint coordinate system, defined rotation about the long axis of the tibia. The position of the rotation axis in this model was at the origin of the tibial coordinate system placed at the centre of the surface of the medial plateau. In the extended position, the coupling of anterior translation with external rotation and posterior translation with internal rotation implies an actual axis of rotation located lateral to the chosen axis position, i.e. closer to the midpoint of the medial and lateral tibial plateaus. Rotation about a more central axis would cause the observed anterior or posterior translation of the chosen origin with respect to the femoral origin located at the midpoint of the epicondyles.

The axial view of the tibial plateau and origin with respect to the position of the femoral origin clearly illustrates this coupled movement (Figure 4.6). This central approximation of the actual location of the rotation axis is supported by Kaneda *et al.* (1997), in which the location of the mean helical axis under 3 Nm of external torque was located at the medial tibial spine in 15 cadaveric specimens. Furthermore, the regression lines calculated from our *in vivo* data for both the external torque (Figure 4.4, blue) and internal torque positions (Figure 4.4, red) show y-intercepts of approximately 0, signifying an absence of translation without rotation; in other words, the translation measured was not real, but simply an artefact of a misplaced tibial origin.

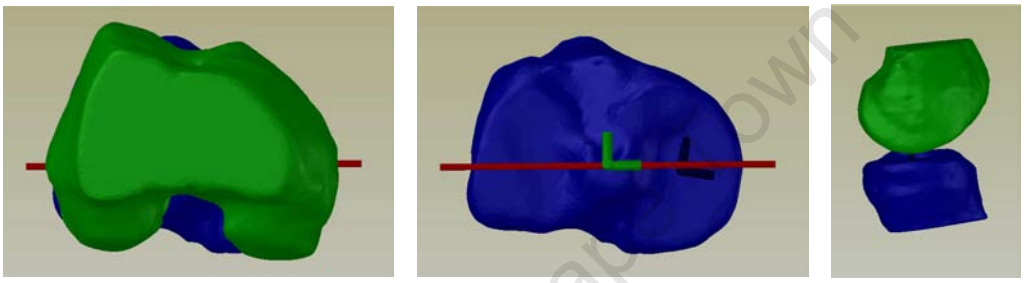
In fact, the coupling of rotation and translation is likely sinusoidal, rather than linear. Figure 4.7 shows the measured AP translation as a function of the *sin* of the degree of rotation and the distance between the chosen origin and the actual centre of rotation; this is described by the following equation:

$$t = d\, sin\theta \tag{4.1}$$

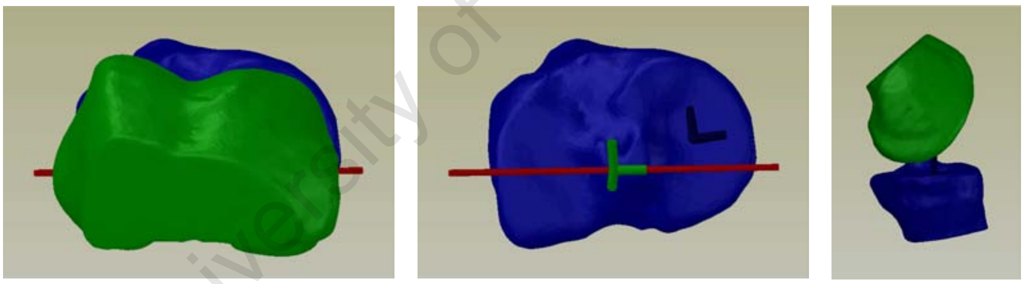
where t is the anterior-posterior translation, d is the distance between actual and chosen origins, and θ is the angle of rotation. (A similar relation was described by Roos *et al.* (2006) with the tibial origin in flexion-extension.) Figures 4.4 and 4.5, however, displays data from 15 different subjects, rather than 15 rotation angles of the same knee, making the linear relationship an acceptable assumption with this limited amount of data. The roughly equivalent slopes in the internal and external rotation positions indicate a similar distance between actual and chosen origins in each condition.



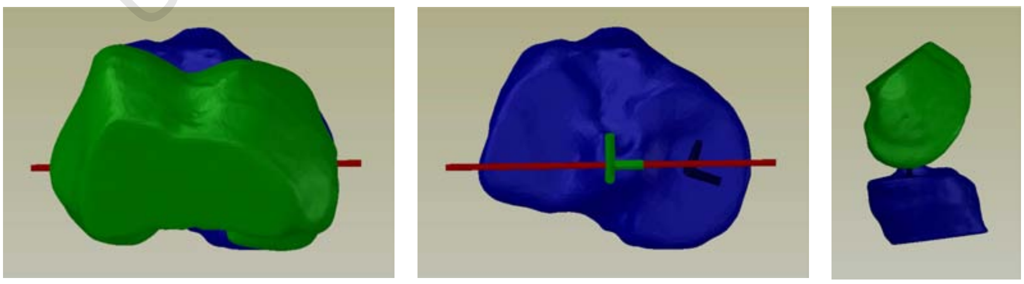
(a) Extended external rotation.



(b) Extended internal rotation.



(c) Flexed external rotation.



(d) Flexed internal rotation.

Figure 4.6: A tibiofemoral model viewed in the transverse and sagittal planes. The middle column shows only the femoral coordinate system and flexion-extension axis to enable relative tibial and femoral origin positions to be seen.

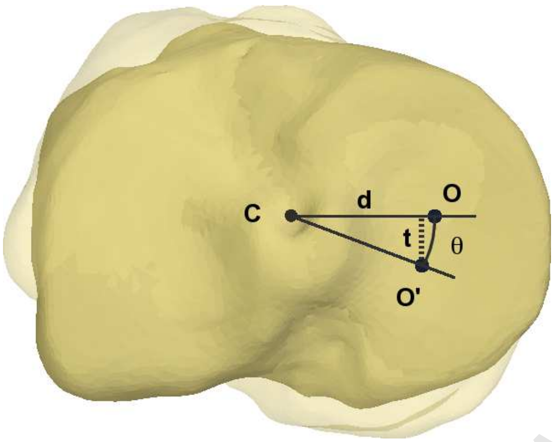
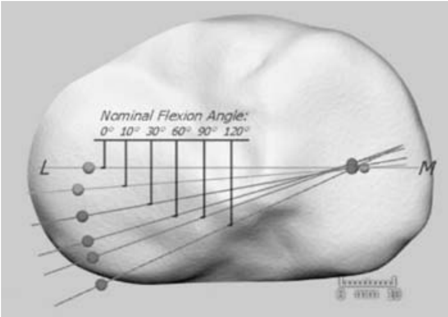


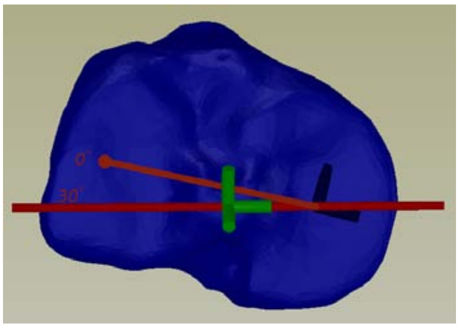
Figure 4.7: Anterior-posterior translation resulting from internal-external rotation of the tibia about an axis located at a different position from the tibial origin. O is the original position of the chosen origin, O' is its position following rotation about the centre C, d is the distance between actual and chosen origins, θ is the angle of internal rotation, and t is the magnitude of posterior translation represented by the dotted line.

In the flexed position with an external torque, the slope of the regression line is about the same as for the extended positions; however, it has been translated anteriorly (Figure 4.5, blue). As with the extended position, a lateral shift of the axis of rotation has likely occurred as illustrated in Figures 4.6(c) and 4.6(d). Equal slopes in this position and the extended positions correspond to an equivalent extent of displacement.

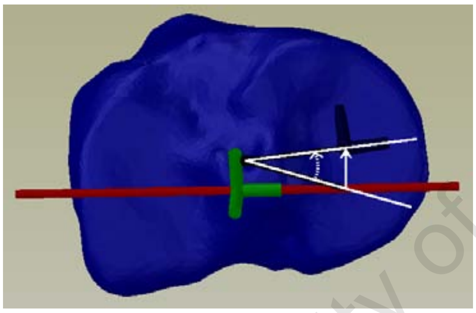
The anterior translation of the regression line can be explained by the screw-home motion that occurs with knee flexion (Figure 4.8). From extension to 30° of knee flexion we know that tibiofemoral roll-back occurs on the lateral, but not the medial side of the knee, resulting in coupled internal tibial rotation about a medially oriented rotation axis (Crawford *et al.*, 2007; Dennis *et al.*, 2005; Iwaki *et al.*, 2000; Johal *et al.*, 2005; Koh *et al.*, 2005; McPherson *et al.*, 2005; Pinskerova *et al.*, 2000; Wilson *et al.*, 2000). Figure 4.8(a) illustrates the posterior translation on the tibial plateau of the lateral femoral condyle. A femoral origin located midway between the epicondyles would consequently also move posteriorly. Since the axis of rotation in our model is on the medial plateau and AP translation is measured along the floating axis perpendicular to the epicondylar axis, this rotation alone would not account for AP translation (Figure 4.8(b)).



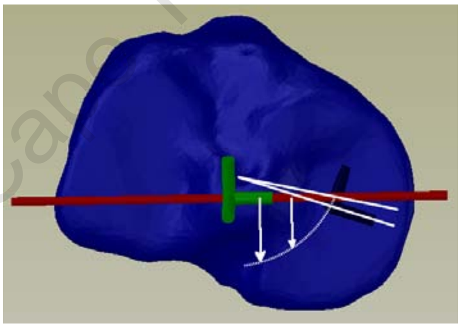
(a) A femoral origin located at the mid-point of the epicondyles would move posteriorly on the tibial plateau with 30° of flexion (McPherson *et al.*, 2005).



(b) The tibia and femoral epicondylar axis have been rotated to show the same internal rotation resulting from flexion, but with the epicondylar axis remaining horizontal.



(c) With an external torque, the tibial origin moves anteriorly relative to the femoral origin.



(d) With an internal torque, the rotation about a central tibial axis causes the tibial origin to translate posteriorly relative to the femoral origin, but with a smaller relative displacement than in external rotation.

Figure 4.8: Coupled rotation and anterior-posterior translation resulting from torsional loads.

The addition of an axial external rotation torque counteracted this internal rotation in every knee, as deduced from exclusively positive values of external rotation. With rotation about a more centrally located axis, the coupled translation measurements indicate that the medial condyle followed the lateral condyle and moved posteriorly on the medial plateau, i.e. the chosen tibial origin moved in an anterior direction with respect to the femoral origin (Figure 4.8(c)). Depending on the laxity of the knee, the external torque may have simply balanced the internal rotation resulting in a net rotation of 0° and anterior translation of about 5 mm. Alternatively, in other subjects, it resulted in external rotation coupled with anterior translation – with the overall amount of translation greater than the amount in the extended position at the same degree of rotation (Figure 4.5).

The predetermined internal rotation of the tibia that occurred with flexion was magnified with the addition of an internal torque and generated the larger increase in overall rotation in the flexed internal torque position compared to the flexed external torque position (Figure 4.2). With an effective axis of rotation anterior to the femoral origin, internal rotation results in a smaller posterior translation than the anterior translation in the external rotation position, since the AP distance between femoral origin and axis of rotation must be taken into account (Figure 4.8(d)). The slope of the regression line illustrating the rotation-translation correlation is less, in the flexed internal torque position (Figure 4.5, red), than the slopes in the other three loading conditions, showing that on average there is less posterior translation of the chosen origin than in the extended position with the same amount of internal rotation.

Joint stability may be compromised with the deficiency of any structure that provides support. Pathological laxity is a result of the joint following the path of least resistance under a specific external loading condition (Nordt *et al.*, 1999). The path of least resistance is the pathway of motion that occurs when the overall force (i.e. resistance torque) of all contributing structures is minimized; the greater the contribution of a specific structure to the net restraint, the greater the displacement will be towards that structure in order to minimize the total resistant force.

The contribution of each structure to the net rotational restraint depends on its material stiffness properties (tissue elasticity) and its perpendicular distance from the location of the applied torque. This can be summarized by the following cross-product equation:

$$\Sigma T = \Sigma (F \times r) \tag{4.2}$$

where T is the restraining torque, F is the force in a specific joint structure (which varies with tissue elasticity), and r is the distance between the point of application of the force and the torque axis.

The reduction in anterior translation due to anterior loading that occurs with fixed internal or external rotation is an example of the lateral and medial collateral ligaments contributing more to joint restraint as they became taught (Amis *et al.*, 2005). In the extended position, in both internal and external rotational loading conditions, our results showed that the net axis of restraint was located in approximately the same position as that of the applied torque axis. Therefore, the combined force-distance contribution of all structures that provided stability were balanced. In extension, there is an increase in tension of all ligamentous structures posterior to the femoral epicondyles, including the posteromedial capsule, the posterior capsule, and the arcuate ligament complex Amis *et al.* (2005). Similar to the reduction in anterior translation noted by Amis *et al.* (2005), this tightening of collateral and posterior structures may have resulted in a stress-shielding effect of the ACL in the extended position (Amis *et al.*, 2005; Csintalan *et al.*, 2006; Nordt *et al.*, 1999).

In the flexed position, conversely, the extra-articular ligaments relax (Amis *et al.*, 2005), resulting in smaller force contributions to the overall joint restraint. Lateral capsular laxity, combined with a more convex lateral tibial plateau, results in a more mobile lateral tibial compartment (Amis *et al.*, 2005; Nordt *et al.*, 1999); this likely led to the increase in internal tibial rotation in the flexed position while the degree of external rotation remained about the same in extended and flexed positions. Although the ACL is protected by the MCL under external tibial torsional loading, in internal rotation it plays a greater role in overall joint restraint (Amis *et al.*, 2005; Csintalan *et al.*, 2006; Harfe *et al.*, 1998; Meyer & Haut, 2008; Nordt *et al.*, 1999). This is due to its oblique orientation relative to the axis of rotation (Blankevoort & Huiskes, 1996); as the tibia rotates internally, the distance between ligament insertions increases, with further tensioning of the ligament taking place as it twists around the PCL.

Given the proximity of the ACL insertion to the actual axis of rotation when compared to the positions of the collateral and posterior ligaments, we would expect the overall contribution of the ACL to rotational restraint to be minimal in both extended and flexed knee positions.

A general trend of increasing knee flexion was observed when subjected to external versus internal torque (Figure 4.2); these connected motions are contrary to the normally observed coupling of internal rotation with flexion in the sagittal plane. This may be explained by the difference in loading conditions in this study, in which pure torsional loading without any additional constraints would cause a simple sliding motion of the femoral condyles on the tibial plateau. However, rotational restraint of the knee is provided by the contact surfaces of the joint in addition to the ligaments of the knee (Blankevoort & Huiskes, 1996). The more concave shape of the medial plateau, together with the increased stiffness of the medial meniscus, may have limited the sliding of the medial condyle and forced the condyle to roll in order to accommodate the applied torque; roll-back in external rotation is converted into an increased flexion angle, whereas roll-forward in internal rotation becomes a decreased flexion angle. Furthermore, the heightened strain on the medial collateral ligament in external rotation may have been offset by an increase in flexion angle between 0° and 30° of flexion. The more convex shape of the lateral tibial plateau and the general increased laxity of this side of the joint may have permitted the sliding motion that resulted in the observed net flexion in external rotation and net extension in internal rotation. In order to confirm this theory, the tibiofemoral contact points and positions of the menisci should be measured under these loading conditions, which was beyond the scope of this study.

In conclusion, the 3D knee kinematics measured under a normalized torque showed a large variation in transverse plane rotation, the primary motion resulting from torsional loading, in a group of healthy individuals. Our rotation data agreed well with that of the literature; a mean increase in range of rotation of about 10° was measured from full extension to 30° of flexion, which could be attributed to an increase in the internal direction. However, significant left-to-right differences in external and internal loading conditions confirm that caution should be taken when comparing knee rotations to their contralateral limb. Coupled anterior-posterior translation with internal-external rotation revealed that the effective axis of rotation is located near the centre of the tibial plateau, lateral to the tibial rotation axis in flexion-extension. It is important to bear in mind that these results are for specific passive loading conditions *in vivo*. While all reasonable measures were taken to avoid active stabilisation strategies used by the subjects, the possibility of muscle recruitment cannot be entirely ruled out.

Chapter 5

Passive rotational laxity of the ACL-deficient and reconstructed knee: Single vs double-bundle surgery

5.1 Introduction

The anterior cruciate ligament (ACL), in addition to its primary role restraining anterior tibial translation, has been shown to contribute to rotational constraint of the knee. Conventional surgical techniques adequately limit anterior-posterior (AP) laxity; however, subjective 'giving-way' symptoms and positive pivot shift reveal that rotational instability often remains. In order to improve rotational laxity outcome, surgical techniques have been modified to reconstruct not just the anteromedial (AM), but also the posterolateral (PL) bundle of the ACL. Although the single-bundle (SB) technique has been shown to improve knee restraint with respect to the injured knee, several biomechanical studies have shown significant reductions in knee laxity and superior functional outcome under anterior and pivot shift loading, when comparing the outcome of the double-bundle (DB) to the SB reconstruction (Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Siebold *et al.*, 2008; Yagi *et al.*, 2002, 2007; Yasuda *et al.*, 2006; Zantop *et al.*, 2006).

At the time of inception of this study in 2006, however, little *in vivo* clinical evidence was available comparing the outcome of the SB and DB techniques. In fact, a meta-analysis published in 2008 found only four randomised control trials (Level I evidence) and an additional five prospective and retrospective comparative studies (Levels II and III) to assess differences in outcome of SB and DB reconstructions; their findings showed that there were no clinically significant differences in KT-1000 arthrometer and pivot shift results between surgical techniques (Meredick *et al.*, 2008). Other reviews have also identified this lack of *in vivo* clinical evidence to support the more complicated DB technique (Amis *et al.*, 2005; Crawford *et al.*, 2007; Lewis *et al.*, 2008; Longo *et al.*, 2008; Steckel *et al.*, 2007b).

Since 2006, the publication of clinical trials investigating SB and DB outcome has accumulated substantially; although much of the evidence supports the use of the DB technique (Järvelä, 2007; Kondo *et al.*, 2008; Muneta *et al.*, 2007; Seon *et al.*, 2007; Siebold *et al.*, 2008; Yasuda *et al.*, 2006), some trials have not found significant differences between clinical outcomes in the patient groups (Asagumo *et al.*, 2007; Streich *et al.*, 2008). Furthermore, there is a proliferation of research describing improvements in both SB and DB surgical technique that also reduce rotational laxity. These include adjusting tunnel placements to accommodate a more horizontal graft and modifying initial graft tensions and specific knee angles at which tensioning occurs (Jepsen *et al.*, 2007; Kondo & Yasuda, 2007; Loh *et al.*, 2003; Markolf *et al.*, 2009; Musahl *et al.*, 2005; Scopp *et al.*, 2004; Yamamoto *et al.*, 2004; Yasuda *et al.*, 2008; Zaffagnini *et al.*, 2008).

The problem with much of the clinical evidence in the literature is that assessments use either a quantitative outcome that does not measure rotation (e.g. AP instability assessed with an arthrometer) or a subjective test that provides only a gross clinical measure of laxity (e.g. pivot shift). Moreover, neither of these tests is able to establish the role of the ACL specifically in rotational restraint, since it is not an isolated torque that is applied. (The pivot shift is a combined internal and valgus torsional load.) The differences in measured outcome could, therefore, be attributed to the contribution of the ACL to anterior or valgus restraint, rather than rotational restraint. Those studies that have examined the effect of ACL surgical technique, graft tension, or tunnel placement on a quantifiable measure of rotational laxity by applying an isolated torque, have done so without measuring the magnitude of the applied load, making comparisons within and between studies difficult due to this subjective component of the study. As no *in vivo* post-operative studies could be found, these also could not be reasonably compared with the existing clinical evidence.

The purpose of this study was, therefore, to determine differences in rotational laxity outcome in SB and DB reconstructions under known isolated torsional loading. The study was designed as a prospective double-blinded randomised control trial (Level I evidence).

5.2 Methods

5.2.1 Participants and interventions

Thirty-two subjects were recruited for this trial from the patient list of the Sports Science Orthopaedic Clinic in Cape Town between November 2006 and March 2008. Testing was generally completed outside of regular clinic hours (i.e. evenings and weekends). A transportation allowance was provided for those patients who did not have their own means of transport. Eligibility criteria included the following:

- Age: 18 - 49 years.
- Injury: complete isolated ACL rupture with minimal injury to other structures (e.g. patients with concomitant meniscal, medial, or lateral structure injury were excluded).
- Previous lower limb pathology: no previous injuries to either affected or contralateral limb.
- Clinical status: ability to walk with no or negligible pain.

Knee laxity tests were performed by a trained investigator prior to and following ACL surgery. Time between surgery and post-operative testing ranged between 2.5 and 9 months (mean 5.2 months) with the final testing completed in July 2008.

Patients underwent one of two surgical procedures to reconstruct the ACL: a single-bundle or a double-bundle reconstruction. All procedures were performed at the Vincent Pallotti Hospital in Cape Town by the same orthopaedic surgeon (Dr. Willem van der Merwe) who had over ten years' experience with both surgical techniques.

The surgical procedure for the double-bundle technique is described in detail by Bellier *et al.* (2004) and a brief description of each procedure is given here. In both procedures, the semitendinosus and gracilis tendons were harvested from the affected limb through the anteromedial incision. Standard arthroscopic evaluation and site preparation were conducted to enable a clear visualization of the anatomical femoral and tibial footprints. At the end of each procedure, the knee was put through a range of motion to confirm an absence of graft impingement and to ensure stability of the graft.

5.2.1.1 Single-bundle surgical procedure

Both semitendinosus and gracilis tendons were folded in half to produce a four-stranded graft. With the knee flexed to 120°, a guidewire was placed at the 10:30 o'clock position (1:30 for the left knee) and a single 7-10 mm femoral tunnel was drilled at the midpoint of the AM and PL attachments. Next, the knee was flexed to 30° and a single 7-10 mm tunnel was drilled though the proximal end of the tibia. In each case, the diameter of the prepared graft was measured, and the tunnel was drilled accordingly. The graft was passed through the tibial and femoral tunnels and an Endobutton (Smith & Nephew Inc) was used for femoral side graft fixation. Once the graft was tensioned to approximately 50 N with the knee flexed to 90°, the graft was secured at the tibial side using a bioabsorbable interference screw (Smith & Nephew Inc).

5.2.1.2 Double-bundle surgical procedure

The double-bundle graft technique used the folded semitendinosus to produce the AM bundle, while the PL bundle was fashioned from the doubled gracilis tendon.

After they were each passed through an Endobutton (Smith & Nephew Inc), the two ends of each graft bundle were sutured together separately. The knee was flexed to 120° to drill the first of the femoral tunnels for the AM bundle with the guidewire placed at the 11 o'clock position (1 o'clock for the left knee). The PL bundle tunnel was drilled next with the guidewire at the 9:30 o'clock position (2:30 for the left knee). The AM tunnel diameter was 6-8 mm, while the PL bundle was slightly smaller at 5-7 mm.

The two tibial tunnels were then created, beginning with the PL tunnel. The anterolateral tibial spine was used as a guide for this tunnel, while the AM tunnel was located between the two tibial spines and anterior to the PL tunnel. Again, the AM tunnel was drilled between 6-8 mm and the PL tunnel was only 5-7 mm in diameter. The PL bundle (gracilis) and AM bundle (semitendinosus) grafts were then passed through the tunnels. Bioabsorbable interference screws were used for fixation of both graft bundles when tensioning to approximately 50 N had been achieved. The PL bundle was fixed first with the knee flexed to 15°. The knee was then flexed to 90° for securing the AM bundle.

5.2.1.3 Testing protocol

Each patient who met the inclusion criteria as determined through a physical exam and MRI scan performed by Dr. van der Merwe was given details of and asked to participate in the study. Those who agreed, signed an informed consent document approved by the Human Ethics Committee of the University of Cape Town (Appendix D).

Details of the data collection and analysis methods are given in Chapter 3. Subjects were tested up to four weeks before their surgeries with scans of both injured and contralateral limbs taken at that time. In some circumstances, it was not possible test both knees pre-operatively, so the contralateral knee was scanned at the time of post-operative testing. Low resolution T1-weighted transverse plane MR images were taken while normalized internal and external torsional loads were applied to the knee in full extension and at 30° of flexion. A high resolution image in a neutral (no-load) position was recorded for shape-matching purposes during data analysis. Complete six degree-of-freedom tibiofemoral kinematics were calculated for the contralateral knee, as well as for the injured knee pre- and post-operatively in the four loading conditions: internal and external torque in extended and flexed positions. All MR imaging was completed at the Sports Science Orthopaedic Clinic; image processing and Matlab^{TM} analysis were conducted off-site on a separate laptop computer.

5.2.2 Objectives and outcome

The primary objective of this study was to compare the magnitude of change in rotational laxity pre- to post-operatively in the single and double-bundle ACL reconstructions with the knee positioned in full extension and at 30° of flexion. Secondary objectives were to determine whether there were differences in laxity in the ACL deficient and reconstructed knee with respect to subjects' contralateral knees. The hypothesis that the mean rotational laxity of the patients' contralateral knees was not different to that of a group of healthy age- and gender-matched control subjects (whose data were presented in Chapter 4) was furthermore tested.

5.2.3 Randomisation and blinding

A random allocation sequence was generated using Matlab ^{TM} to ensure equal numbers in single and double-bundle groups for the first 30 subjects and blocks of 10 subjects thereafter. Participants were enrolled by Dr. van der Merwe; intervention group was assigned at time of surgery by Dr. van der Merwe's administrative assistant who kept the random allocation list. The participants and primary investigator conducting data collection and analysis (AH) were blinded to group assignment; however, the intervention group could be discerned from the post-operative MRI scans during image segmentation.

5.2.4 Statistical analysis

A linear mixed model for repeated measures was applied to detect intervention group differences pre- to post-operatively using SPSS 15.0 (SPSS Inc). Post-hoc analysis was conducted with a two-tailed paired samples t-test. Secondary outcomes – specifically comparisons of contralateral with control group, contralateral with ACL-deficient group, and contralateral with ACL-reconstructed group knee laxity – were also measured using the linear mixed model for repeated measures. Differences with p-values less than or equal to 0.05 were considered to be statistically significant.

5.3 Results

Baseline demographic and clinical subject data are presented in Table 5.1. Of the 32 participants enrolled in this study, three patients allocated to the double-bundle intervention were lost to follow-up (Figure 5.1). Of those, two subjects had completed testing of the contralateral limb during the pre-operative session. An additional two subjects (one in each intervention group) did not have testing completed on the contralateral limb due to patients' personal constraints and consequent withdrawal from the study. The linear mixed model permitted the use of all available data in each analysis; the number of subjects included in each analysis was therefore dependent on the knee, test time (pre- or post-operative), and loading condition examined.

Table 5.1: Baseline demographic and clinical subject data (mean ± SD) for control and patient groups. ACL all includes subjects from both single-bundle (SB) and double-bundle (DB) groups.

Variable	**Control**	**ACL all**	**SB**	**DB**
Sex (F:M)	4:11	8:24	7:10	1:14
Age (yrs)	30.3 ±5.9	30.2 ±6.2	31.5 ± 5.7	26.8 ± 6.0
Height (cm)	174.5 ±9.3	174.5 ±8.8	171.5 ± 6.8	177.9 ± 9.7
Mass (kg)	71.7 ±11.3	79.3 ±14.5	76.3 ± 13.8	82.7 ± 14.9
Applied Torque (Nm)	4.8 ±0.6	5.2 ±0.7	5.1 ± 0.7	5.4 ± 0.7
Time Injury-PreOp (mos)	n/a	5.7 ±8.9	5.5 ± 11.3	5.9 ± 5.5
Time Surgery-PostOp (mos)	n/a	5.2 ±2.0	4.6 ± 1.8	6.1 ± 1.8

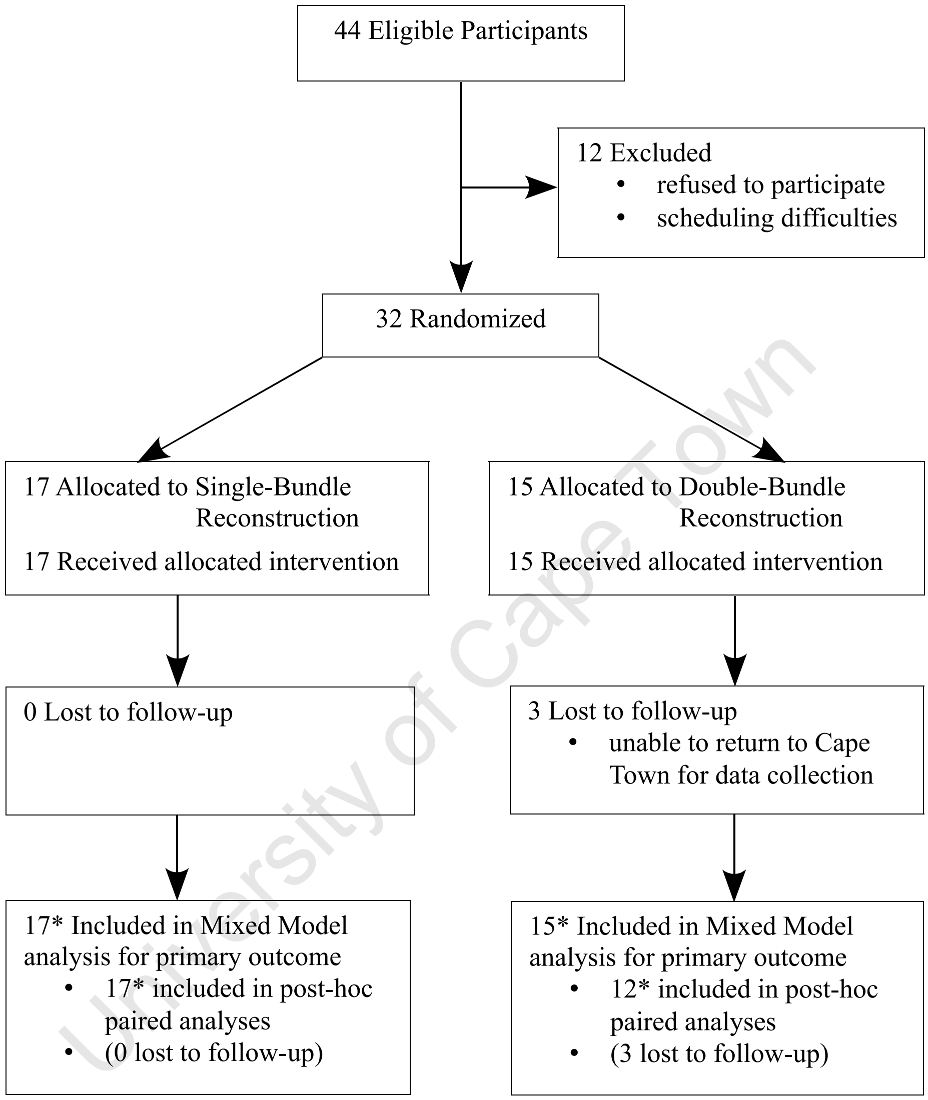


Figure 5.1: Flow diagram of participants through each stage of the randomised control trial for the primary outcome comparing single and double-bundle ACL reconstruction. * Some data were excluded from analyses if adequate torque could not be applied. (See explanation in section 5.3.1.)

5.3.1 Protocol deviations

While all subjects were able to tolerate the pre-calculated normalized torque applied to their contralateral knee, several experienced discomfort with the load applied to the injured knee. In those cases, the knee was scanned under the maximum tolerated load with this value recorded for the specific condition. Since the mixed model is able to evaluate unmatched data, it was important to include all valid data in the 'intention to treat' analysis. However, it has been shown that rotation is directly related to torsional load (Almquist *et al.*, 2002). It was therefore necessary to determine the percentage of the total torque at which the results would be affected by the smaller load. A Kruskal-Wallis one-way ANOVA was used to compare the mean rotation in each of the four loading conditions with the data divided into 3 groups according to the level of torque achieved: Group I = 100% torque, Group II = 75-99% torque, and Group III = 50-74% torque.

Table 5.2: Differences in measured mean rotation according to group. Group I = 100% torque, Group II = 75-99% torque, Group III = 50-74% torque. * post-hoc analysis indicates significant difference between Groups I and III.

Loading	Group I		Group II		Group III		Level of
Condition	N	mean	N	mean	N	mean	Significance
Extended Ext T	84	9.5	2	10.4	2	3.5	0.098*
Extended Int T	84	-7.4	3	-4.0	3	-3.3	0.027*
Flexed Ext T	82	9.6	3	10.3	4	7.0	0.300
Flexed Int T	85	-13.7	3	-10.3	1	-11.5	0.273

Results in Table 5.2 show that significant differences were detected in the extended position with internal torsion and significance was approached in the extended position with external torsion. A Mann-Whitney post-hoc analysis showed significant differences between Groups I and III in each of these loading conditions.

Since no significant differences in measured rotation were detected between Groups I and II, it was decided to exclude only those data in which the achieved level of torque was less than 75% of the pre-calculated normalized value. This was then used as a standard for all subjects across all loading conditions. In total, 13 of the possible 68 data sets (19%) were excluded from the SB group and 5 of the possible 60 data sets (8%) were excluded from the DB group.

5.3.2 Adverse events

Only two adverse reactions were reported at follow-up. One subject complained of swelling and instability of the knee at the time of follow-up, $3\frac{1}{2}$ months post-operatively and could not tolerate the applied torque in all loading conditions. Another subject developed more serious pre-tibial soft tissue swelling due to problems with the Calaxo bioabsorbable screw implant (Smith & Nephew); this product was subsequently recalled. The patient required local debridement and removal of the remaining screw fragments; however, graft-to-bone healing had already occurred by this point. Following an additional 3 months of recovery, this subject agreed to return for post-operative testing (a total of 7 months following the original surgery). Both of these subjects were in the single-bundle reconstruction group.

5.3.3 Outcomes

The only significant interaction between SB and DB surgical techniques when comparing the pre- to post-operative results was in the flexed internally torqued loading condition in which the DB group demonstrated a reduction in transverse plane rotation following ACL reconstruction (Figure 5.2 and Table F.1). No significant differences were found between single and double-bundle groups, however.

In general, ACL reconstruction was shown to reduce rotational laxity in the extended position under internal torsional loading, restoring rotation to that of the contralateral knee (Figure 5.2, Figure 5.3 and Table F.2). No surgical group interaction was observed, however (Table F.1); i.e. this difference was not dependent on the type of reconstruction (SB or DB).

No differences in rotational laxity were found between the left-right averaged knees of the control group and the contralateral knees of the patients in any of the four loading conditions (Figure 5.3 and Table F.2).

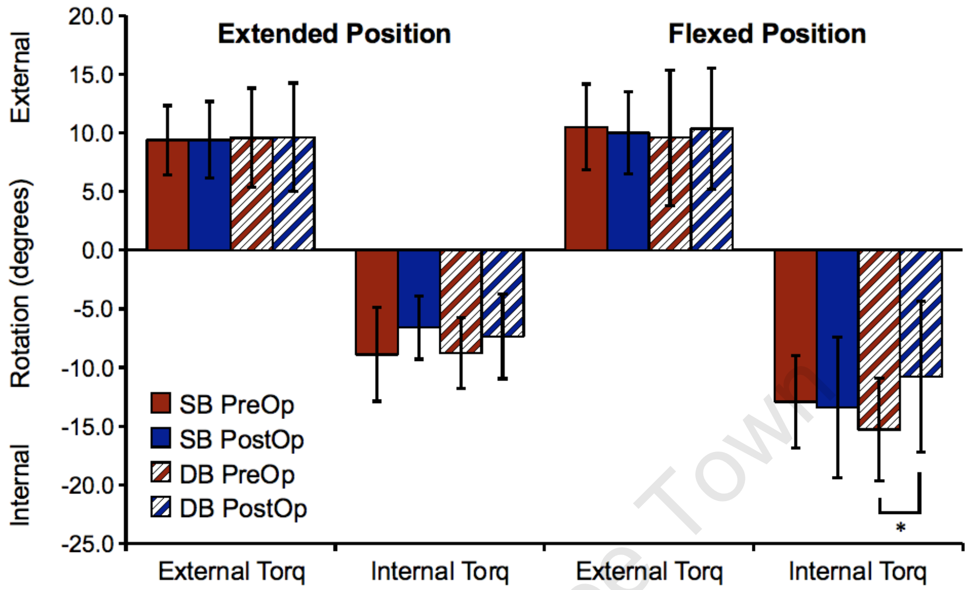


Figure 5.2: Mean internal and external rotation measured pre- and post-operatively in the single and double-bundle groups in the four loading conditions. * indicates significant difference.

5.4 Discussion

In this study, single and double-bundle surgical techniques were compared to determine differences in rotational laxity in patients with isolated rupture of the ACL before and after reconstructive surgery. The findings showed that in only the flexed knee position under internal torsional loading did the DB reconstruction reduce rotational laxity more than the SB technique (with this reduction being statistically significant); however, when compared with rotation of the contralateral knee which demonstrated a mean laxity similar to that of the injured knee, this may have resulted in excessive restraint. Although significant differences were not found between SB, DB, or contralateral knee groups in the flexed, internally torqued knee condition, the mean rotation in the reconstructed SB knee more closely matched that of the contralateral uninjured knee than did the mean DB knee rotation.

Our findings also demonstrated a significant increase in internal rotation of the injured knee with respect to the contralateral and reconstructed knees in

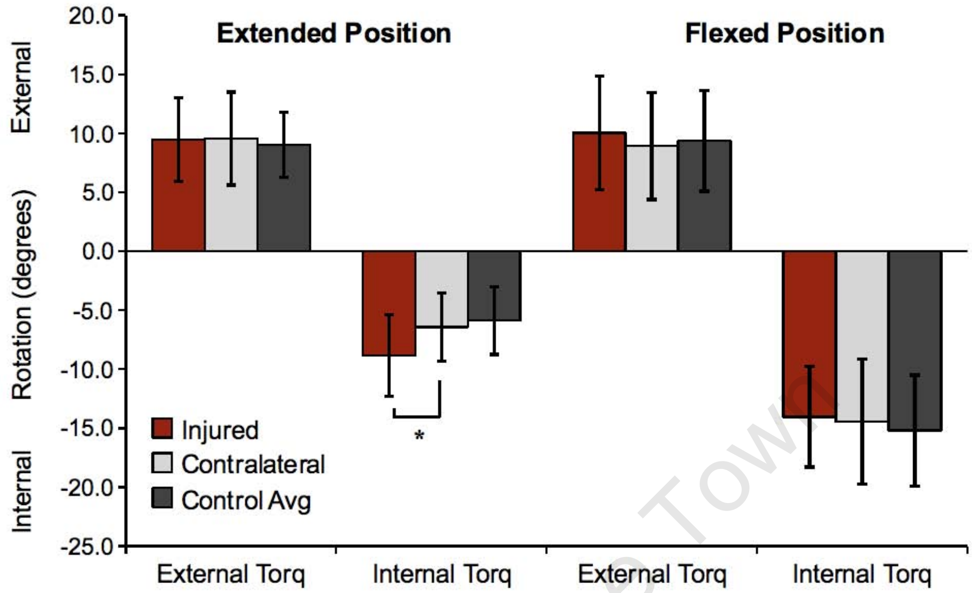


Figure 5.3: Mean internal and external rotation measured in Control (averaged from left and right knee data), Patient contralateral, and Patient injured (pre-operative) knee groups in the four loading conditions. *Post-hoc analysis revealed significant difference. (Note: Injured knee data was not compared to Control Avg data.)

the extended position. In other words, in this loading condition, the reconstruction of the ACL returned the knee kinematics to normal; however, there was no significant effect of surgical technique.

Since no significant differences were found between the contralateral knees of the patients and a healthy age- and gender-matched control group, it was valid to use the data from patients' contralateral knees as reliable controls (Kozanek *et al.*, 2008). Furthermore, demonstrating statistically equivalent results between healthy controls and patients' contralateral knees indicates that there was no preexisting laxity and that the contralateral knees were not affected by ACL injury in this group of patients (Kozanek *et al.*, 2008).

Our study supports the evidence that the ACL contributes to rotational restraint under internal torsional loading, but that it is not the primary restraint to rotational loading. Rotational forces are first constrained by the extraarticular ligaments, which have a mechanical advantage in rotation and thereby shield the ACL from stress under torsional loading (Amis *et al.*, 2005; Csintalan *et al.*, 2006;

Nordt *et al.*, 1999). The position of the ACL insertion and its resulting orientation allows it to provide restraint in internal rotation: the distance between the anteromedial position of the tibial insertion and the posterior position on the medial side of the lateral femoral condyle increases under internal rotation, thereby increasing the tension of the ligament and subsequent restraint of the joint (Amis *et al.*, 2005; Blankevoort & Huiskes, 1996). No differences in rotation were observed in the injured, reconstructed, or contralateral knees under external torsional loading, while differences *were* observed with internal torque, verifying its effect in only the one direction of rotation.

An increased rotational laxity of the ACL-deficient knee with respect to the contralateral knee was observed in only the extended position, while no difference in laxity was observed in the injured knee at 30° of flexion. This distinction between extension and flexion may be attributed to the general laxity of the ACL and other major rotational restraints in these knee positions. It has been shown that with no externally applied load, the tension of the ACL is greater in the extended position than at 30° of flexion (Amis & Dawkins, 1991; Blankevoort *et al.*, 1991; Markolf *et al.*, 2008a; O'Connor & Zavatsky, 1993). The recruitment patterns of various ligaments (and their respective bundles) were illustrated as functions of flexion by Blankevoort *et al.* (1991); with no external loading, the posterolateral bundle of the ACL exhibited near maximum strain with the knee in full extension.

Knowing that the distance between ACL insertions increases with internal rotation, we can deduce that the tension under an additional internal torque would only increase. This hypothesis is supported by Markolf *et al.* (2009) who showed that the *in situ* ACL force increased in both full extension and 30° of flexion with the addition of a 5 Nm load (with the force magnitude greater in the extended than in the flexed position).

The collateral ligament forces that would have permitted a prediction of the relative contribution of the various ligaments under these loading conditions were not presented in the study by Markolf *et al.* (2009). However, Blankevoort *et al.* (1991) assumed the recruitment of the ACL to be less than that of the collateral ligaments in full extension and to decrease even further throughout the first 30° of flexion. This assumption was based on the relative length changes of the cruciate and collateral ligaments through 90° of flexion. While the length (and so the tension) of the ACL decreased, the overall lengths of the collateral ligaments generally decreased to a lesser extent between 0° and 30° of flexion. Even with the addition of a 3 Nm internal torque at 20° of flexion, the overall recruitment of the ACL was assumed to be minimal when compared to the MCL (Blankevoort *et al.*, 1991). (No data for recruitment patterns were presented at 0° and 30° of flexion under torsional loading conditions.)

Amis & Dawkins (1991) also demonstrated that, despite an increase in length of the ACL under 1 Nm of applied torque when compared to the no-load condition, its length under a fixed torsional load still decreased (i.e. the ligament tension decreased) between 0° and 30° of flexion. It is therefore reasonable to assume that the overall contribution of the ACL to rotational restraint is less at 30° of flexion than in the fully extended position.

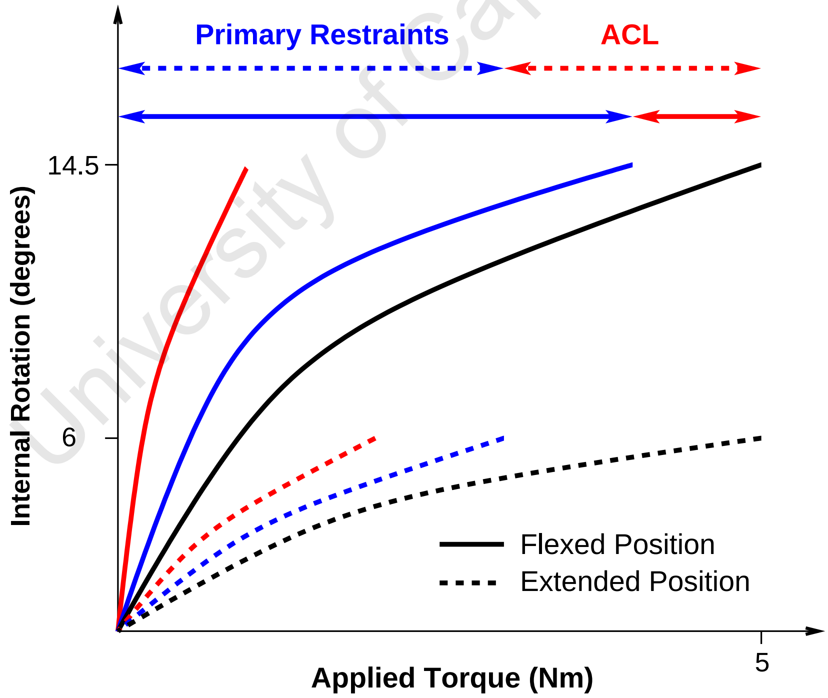


Figure 5.4: Contribution of the primary restraints and ACL to internal rotational restraint in the extended and flexed positions. (Black lines indicate the combined contribution of the primary restraints with the intact ACL.)

This concept is illustrated in Figure 5.4 which shows the proposed recruitment of joint structures based on a typical ligament load-deformation curve during physiological loading conditions (Musahl *et al.*, 2007). In the extended position under internal torsional loading, the primary restraints make the major contribution to the overall restraint required to resist the applied torque; although, the ACL also provides substantial constraint. In the flexed position, all ligaments tend to relax, increasing the slopes of the torque-rotation curves; however, the ACL slackens to a greater extent than the primary restraints (Blankevoort *et al.*, 1991). Its contribution to the overall joint restraint is, therefore smaller at 30° of flexion than in the extended position. Consequently, the rotation resulting from an applied torque in the ACL-deficient knee at 30° of flexion is the same as that of the contralateral knee, while in the fully extended position, there is an increase in rotation in the injured knee (Figure 5.5 and Figure 5.3).

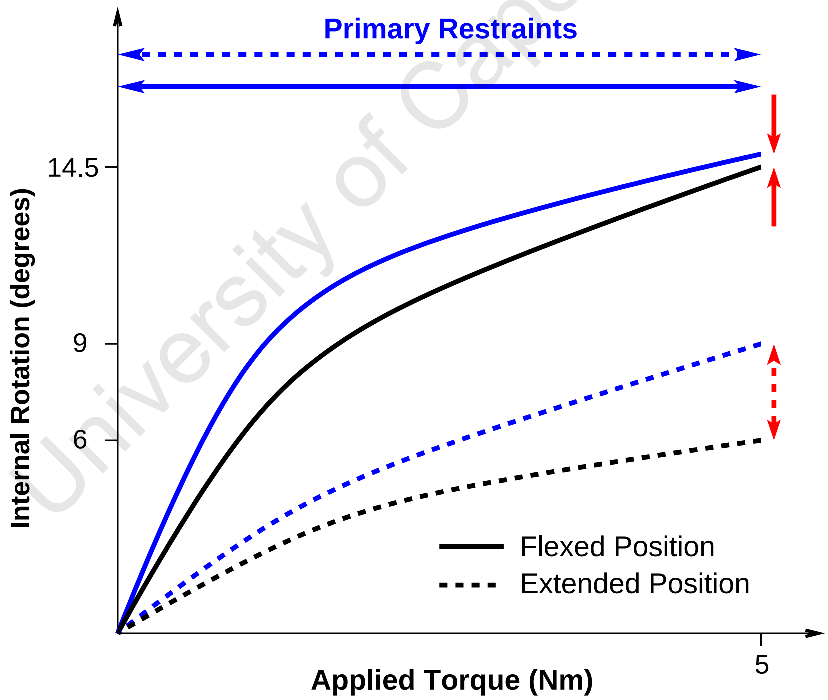


Figure 5.5: In the ACL-deficient knee, the primary restraints restrict the entire load. The change in internal rotation from the ACL-intact to ACL-deficient knee is indicated by the red arrows. (Black lines indicate the combined contribution of the primary restraints with the intact ACL.)

The specific surgical technique, i.e. single or double-bundle, affected rotational restraint in only one loading condition: internal torsion at 30° of flexion. The DB technique limited rotation with respect to the injured knee; however, since the injured knee laxity was actually closer to that of the contralateral knee, this may be considered an overconstraint of rotation. Extending the previous theory to the results obtained in the flexed position, the excessive restraint provided by the DB graft induces less internal rotation with the same applied torque (Figure 5.6).

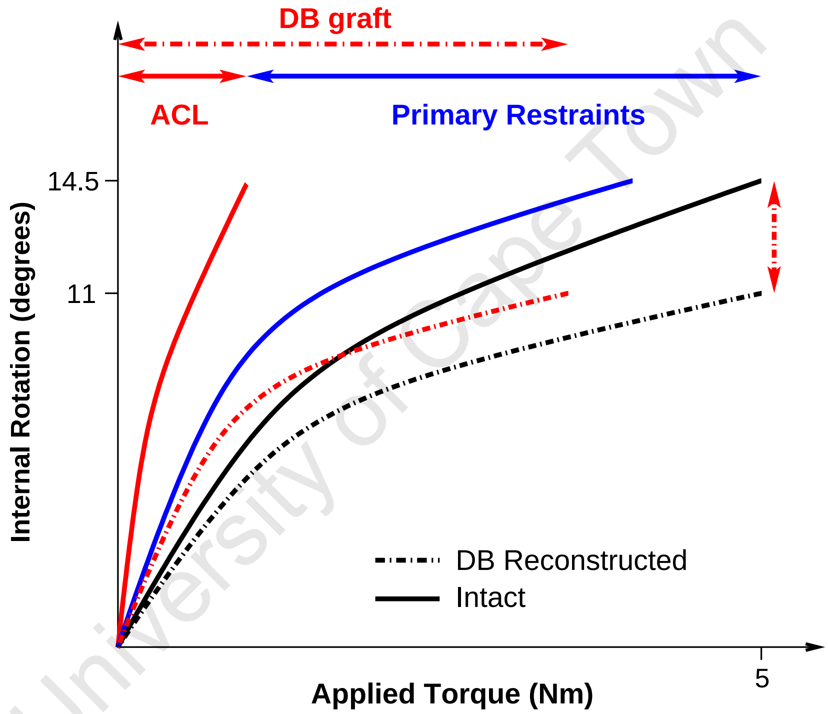


Figure 5.6: In the DB-reconstructed knee, the DB graft restricts more load than the native ACL. The change in internal rotation from the ACL-intact to DB-reconstructed knee is indicated. (Black lines indicate the combined contribution of the primary restraints with the intact or reconstructed ACL.)

This finding was substantiated by several cadaveric studies that examined both rotational laxity and graft tension under torsional loading (Markolf *et al.*, 2008b, 2009; Steckel *et al.*, 2007a). Steckel *et al.* (2007a) found that the DB technique overcorrected internal-external rotation with respect to the intact knee at 15°, 60°, 75°, and 90° of flexion, while rotation following SB reconstruction was not significantly different from the intact knee at all angles except 60° of flexion (where it also overcorrected knee laxity). In a study that compared rotation and graft tension under 5 Nm of internal torque in the ACL intact, sectioned, SB, and DB-reconstructed knee (using four different graft-tensioning protocols), Markolf *et al.* (2009) found that two of the DB techniques overconstrained rotation at higher flexion angles, specifically 50° to 120° of flexion while no differences were found with the SB technique. The resultant force in the posterolateral graft was markedly higher than both the SB graft and the intact ACL, although, similar to our results, no significant differences in rotation were shown between surgical techniques in full extension. In another study conducted by the same group, significant decreases in rotation resulted with the clinical pivot shift test in three of the four DB techniques when compared to the intact knee, while the SB technique restored rotational stability to normal (Markolf *et al.*, 2008b).

Whereas the DB technique reduced internal tibial rotation in the flexed position to a greater extent than the SB technique when comparing pre- and post-operative laxity, no significant differences in knee rotation were found between the two techniques (Figure 5.2). Results in the literature comparing the two techniques vary: while some studies have demonstrated differences in joint laxity between the SB and DB reconstruction (Adachi *et al.*, 2004; Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Lopomo *et al.*, 2009; Markolf *et al.*, 2009; Petersen *et al.*, 2006; Seon *et al.*, 2007; Steckel *et al.*, 2007a; Yagi *et al.*, 2002, 2007; Yamamoto *et al.*, 2004; Yasuda *et al.*, 2006; Zantop *et al.*, 2006), others have shown similar results with both techniques (Ferretti *et al.*, 2008; Steckel *et al.*, 2007a; Streich *et al.*, 2008). Of those that have shown an improvement in rotational restraint with respect to the uninjured knee using the DB technique (rather than an overcorrection), the majority have done so under anterior or pivot shift (i.e. combined valgus and torsional) loading (Adachi *et al.*, 2004; Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Lopomo *et al.*, 2009; Petersen *et al.*, 2006; Steckel *et al.*, 2007a; Yagi *et al.*, 2002, 2007; Yasuda *et al.*, 2006; Zantop *et al.*, 2006).

It is not possible to directly compare our results with these studies since the loading conditions differ from our study in which laxity was examined under an isolated tibial torque. It has been well-established that the ACL is the primary restraint to anterior loading; therefore, an anterior force of 134 N (typically used in the aforementioned studies) would recruit the ACL to a greater extent than torsional loading conditions in which the primary restraints (such as the collateral ligaments or menisci) shield the ACL. Similarly in the pivot shift test, the addition of a valgus moment to a torsional load has been shown to significantly increase ACL forces and strain (Kanamori *et al.*, 2000; Shin *et al.*, 2005), thereby likely recruiting the ACL to a greater extent than with a simple isolated torque.

By increasing the tension of the ACL its contribution to the torsional restraint of the joint will theoretically also increase according to equation 4.2, since the torque provided by the other structures would either decrease to constrain the same overall load or the total restraint provided by the joint structures would increase, thereby decreasing the resulting rotation. This is demonstrated by the following equations (the first of which was derived from equation 4.2):

$$\Sigma T_{restraint} = (F_{ACL} \times r_{ACL}) + \Sigma (F_{Si} \times r_{Si}) \tag{5.1}$$

where T is the restraining torque, F is the force of a specific joint structure, r is the distance between the point of application of the force and the torque axis, and the subscripts refer to the ACL or specific joint structure Si contributing to rotational restraint. As the force of the ACL increases, the forces of the other contributing structures decrease to provide the equivalent overall torque as long as the radius of rotation remains the same. (If the location of the axis of rotation changes, this distance could change.)

This theory is supported by the computational study conducted by Suggs *et al.* (2003) in which an increase in simulated graft tension decreased rotation resulting from an anterior load. With the ACL providing a greater contribution to overall joint laxity in the anterior and pivot shift loading conditions, it is reasonable that differences between SB and DB grafts could be detected.

Despite the possibility of greater overall contribution of the ACL to rotational restraint, two *in vivo* studies demonstrated no differences between graft types with anterior or pivot shift loading (Ferretti *et al.*, 2008; Streich *et al.*, 2008). Ferretti *et al.* (2008) measured anterior and rotational laxity at 30° of flexion under isolated maximum anterior and torsional loading intra-operatively in 20 patients who received either the SB or the DB reconstruction. By applying a subjective measure of manual maximum force, it is possible that the variation in applied load may have resulted in high standard deviations in measured rotation making it difficult to compare the surgical techniques. Since our protocol normalized load according to subject mass, objective comparability across subjects was established and it is more likely that standard deviations of our rotation results were reflective of actual individual subject variation.

Similar to Ferretti *et al.* (2008) in a prospective randomised control trial, Streich *et al.* (2008) found no significant differences in anterior or pivot shift laxity between SB and DB groups. They attributed this inconsistency with the literature to the subjective (and, possibly inaccurate) assessment of the pivot shift and to the placement of the femoral tunnel which permitted a more horizontal orientation of the SB graft.

In fact, several studies have shown that changing tunnel placement will affect rotational laxity in both SB and DB techniques; in general, a more anatomical tunnel placement which allows grafts to attain a more horizontal rather than vertical orientation in the joint, has been shown to improve rotational constraint (Musahl *et al.*, 2005; Scopp *et al.*, 2004; Yamamoto *et al.*, 2004; Zantop *et al.*, 2008). In our study, the tunnels for the SB graft were positioned midway between the native AM and PL bundle insertions, in other words in the SB anatomical position, which may have resulted in similar rotational laxity to the DB technique. Both postoperative graft quality (defined by its thickness and apparent tension) and tensioning during initial fixation of the graft have also been shown to affect joint laxity (Kondo & Yasuda, 2007; Suggs *et al.*, 2003). With similar tensioning protocols used for both surgical techniques and a relatively brief follow-up period of only 5 months in which minimal graft relaxation may have occurred, it is foreseeable that both of our patient groups would have similar graft tension at follow-up, resulting in similar joint laxity.

Our results do not support findings from some studies in which rotational laxity examined at 30° of flexion under isolated torsional loads in cadavers and intra-operatively showed significant differences in rotation between the ACL-reconstructed with the intact or deficient knee. These studies all applied higher loads (6.5 Nm, 10 Nm and maximum manual force) than those used in our *in vivo* study (Kanamori *et al.*, 2000; Martelli *et al.*, 2007; Monaco *et al.*, 2007; Scopp *et al.*, 2004). Furthermore, our study normalized the applied torque to individual subject mass, ensuring that the correlation between the amount of torque and measured rotation would not affect the inter-subject comparison (Almquist *et al.*, 2002).

With a smaller magnitude of only 5 Nm of internal torque, Markolf *et al.* (2008a) showed that cutting the posterolateral bundle did not affect knee rotation. (The effect of cutting both bundles was not examined in that study.) Diermann *et al.* (2008) also found no significant differences of internal rotation in ACL-deficient, intact, and reconstructed knee when a combined 10 Nm valgus and 4 Nm rotational load was applied. Conceivably, in these studies as with ours, in which the applied load was comparatively smaller, the primary restraints were able to control rotation and the torsional load did not stress the intact or reconstructed ACL enough to warrant its contribution to overall joint restraint. In addition, in one of the intraoperative studies that displayed differences between injured and reconstructed knees, 16 of the 30 subjects presented with associated injuries to collateral ligaments (Martelli *et al.*, 2007). In these patients, the capacity of the primary restraints may have been exceeded, with the ACL consequently providing secondary support under high torque conditions, thereby demonstrating rotational differences in ACL-deficient, reconstructed, and intact knees.

To ensure that the uneven distribution of male and female patients between groups (with seven of the eight females allocated an SB reconstruction) did not affect the laxity outcome, further statistical analyses were performed. Specifically, significant difference in rotation between gender subgroups was assessed using an independent samples t-test in each of the four loading conditions (extended knee with external torque, extended knee with internal torque, flexed knee with external torque, and flexed knee with internal torque). The patients were furthermore divided into the following categories: all patients contralateral knees, SB patients injured knees pre-operatively, and SB patients injured knees post-operatively. There were no significant differences in rotational laxity between genders in any subject category or in any loading condition examined. (All p-values were greater than 0.158.) Therefore, the imbalance in gender distribution did not account for differences in laxity.

Another possible limitation of our study was that the post-operative rehabilitation regime was not regulated. Although, it has been shown that rehabilitation has an effect on clinical outcome, several studies have shown that there has been no significant effect on knee laxity, specifically (Beynnon *et al.*, 2005; Grant *et al.*, 2005; Shelbourne & Davis, 1999). Furthermore, due to the brief follow-up time following surgery, differences in rehabilitation protocol would likely not have had a great effect on measured knee rotation.

This limited mean follow-up period of only 5 months may alternatively be viewed as a limitation of the study in that certain structures of the ACL-injured joint require two years to fully recover (Risberg *et al.*, 2004). All patients, however, suffered isolated ACL injury; without concomitant damage to surrounding structures, recovery time should be reduced (Anderson *et al.*, 1992). Moreover, all patients were able to walk pain-free at follow-up; general observation by the primary investigator (AH) found that the variability in patients' perceived comfort during the pre-operative testing session was greater than post-operatively. (No knee scores or other functional tests were available to confirm this observation.) An additional study with a longer follow-up period in which the same rehabilitation protocol is followed by all subjects would be beneficial to confirm these findings following complete ACL recovery.

Important clinical implications for surgeons performing anterior cruciate ligament reconstruction are identified by the results of this study. In determining whether the DB technique would benefit or, in fact, hinder a particular patient, the extent of the injury should be considered. In this study, for isolated ACL rupture with negligible damage to the surrounding soft tissues, a DB reconstruction had no advantage over the SB technique and may even have overconstrained the knee in some cases. However, if the primary restraints to rotation are debilitated and cannot be reconstructed surgically, a DB technique may provide the additional restraint that could prevent or minimize further injury.

This study provides insight into differences in surgical technique under a specific loading condition. However, comparisons of SB and DB reconstructions under different loading conditions must also be considered before determining the best treatment for a particular patient. Further research is required to evaluate these techniques in functional weight-bearing tasks, at higher flexion angles, and in patients with concomitant injury. For this reason, we have conducted another investigation into the outcome of single and double-bundle reconstruction under physiological loading conditions in Chapter 6.

Minimal difference in outcome of SB and DB reconstruction was demonstrated under torsional loading conditions in a group of patients with isolated rupture of the ACL. Since subjects had negligible concomitant injury to structures that have been shown to provide rotational restraint such as the collateral ligaments and menisci, sufficient constraint was likely provided by these structures. The overall evidence presented by this study suggests that the intact ACL does not restrict external rotation, but provides internal rotational restraint when knee conditions generate greater tension and substantial recruitment of the ACL. The rotational laxity that results from isolated ACL injury is restored by both SB and DB surgical techniques.

Chapter 6

Kinematics of the ACL-deficient and reconstructed knee during dynamic activities

6.1 Introduction

The anterior cruciate ligament is the most commonly injured ligament of the knee (Widuchowski *et al.*, 2007). Persistent laxity of the joint following reconstruction of the anterior cruciate ligament (ACL) is thought to lead to osteoarthritis (Chaudhari *et al.*, 2008; DeFrate *et al.*, 2006; Georgoulis *et al.*, 2003); therefore, research has recently concentrated on improving reconstruction techniques to restrict laxity, primarily in the transverse plane (Stergiou *et al.*, 2007). The double-bundle (DB) surgical technique, which reconstructs both anteromedial and posterolateral bundles of the ACL, has been shown to limit rotational laxity to a greater extent than the single-bundle (SB) technique (Colombet *et al.*, 2007; Järvelä, 2007; Kondo *et al.*, 2008; Yagi *et al.*, 2007).

However, little data is available evaluating transverse plane restraint following SB versus DB reconstruction techniques during physiological loading conditions *in vivo*. Jordan *et al.* (2007) demonstrated that the behaviour of the anteromedial and posterolateral bundles of the ACL throughout the range of flexion during weightbearing was inconsistent with observations made in cadaveric studies. They found that both bundles were longest near extension and decreased in length with increasing flexion, while previous non-weightbearing investigations had shown a reciprocal functioning of the length of the two bundles (Amis & Dawkins, 1991).

Conflicting results with respect to the position of the knee joint centre of rotation in the transverse plane were also illustrated by Koo & Andriacchi (2008) when comparing walking with non-ambulatory activities. Whereas passive flexion-extension and squatting activities exhibited a centre of rotation on the medial side (Dennis *et al.*, 2005; Hill *et al.*, 2000; Iwaki *et al.*, 2000), the average centre of rotation was found to be on the lateral side in all subjects during normal walking (Koo & Andriacchi, 2008).

ACL deficiency has been shown to alter three-dimensional (3D) knee kinematics during gait (Andriacchi & Dyrby, 2005; Georgoulis *et al.*, 2003; Ristanis *et al.*, 2003; Tashman *et al.*, 2004; Zhang *et al.*, 2003). Traditional methods of ACL reconstruction using the single-bundle approach are not able to restore normal kinematics (Brandsson *et al.*, 2002; Georgoulis *et al.*, 2007; Ristanis *et al.*, 2005). The objective of this study, therefore, was to confirm whether the findings of our passive loading study (Chapter 5) applied under physiological loading conditions. A randomised control trial would determine whether a double-bundle ACL reconstruction is better able to restore 3D knee kinematics with respect to those of the healthy knee than the single-bundle surgical technique during dynamic, weightbearing activities.

6.2 Methods

6.2.1 Participants and interventions

Thirty-three subjects from either the patient or healthy control groups of the passive knee laxity studies described in Chapters 4 and 5 agreed to participate in this additional study to test dynamic knee laxity. The twenty-two patients were randomly allocated either a single or double-bundle surgical reconstruction (with 11 subjects in each group); 11 age- and gender-matched Control subjects were also selected to continue in the additional trial.

Testing was completed at the Sports Science Institute of South Africa in Cape Town between April 2007 and July 2008; testing of patients was conducted prior to and following ACL reconstruction, while Control subjects were tested once only. Eligibility criteria, surgical procedures, randomisation and blinding are described in detail in Chapter 5 in sections 5.2.1, 5.2.1.1, 5.2.1.2, and 5.2.3 respectively.

6.2.2 Data collection protocol

Subjects' gait during low- and high-demand activities was recorded at 250 Hz in six degrees of freedom using an eight-camera motion analysis system (Vicon Motion Systems, Oxford, UK). Anthropometric data were recorded and fifteen retro-reflective markers were secured to anatomic landmarks based on the modified Helen Hayes marker set (Vaughan *et al.*, 1999). A minimum of five trials were collected for each of the following activities:

Walk – A standard walking trial (low-demand activity) was used as baseline data with which to compare results to other studies. Subjects were instructed to walk along a 10 m walkway at their self-selected pace.

Ninety-degree cut – A cutting activity was designed to actuate a 90° change in direction to simulate a typical game situation in which an offensive player tries to get open from a defender. Three cones were used to mark the start, cut point, and end locations of the cut in order to guide the subject (Figure 6.1). Approximately 3 m of space was available on either side of the cut point cone for approach and termination, which limited the speed and intensity at which this activity could be performed. A demonstration of the activity was performed; however, subjects were allowed to execute the task in the manner most natural to them (i.e. they were not required to follow a specific step-sequence). The cutting activity was repeated on both sides, with subjects instructed to cut first to their right for an acceptable number of trials and then to their left to ensure that both injured and contralateral limbs would be on the inside and outside of the body during the activity.

Jump – Subjects were asked to perform a maximum distance two-foot jump from which a full recovery could be made. A practice trial in which subjects jumped as far as they felt comfortable was used to mark take-off and landing positions so that the same distance would be covered for each recorded trial. If a subject was not able to return to an upright position without losing their balance, the landing marker was brought closer to the take-off position in increments of 5 cm until a two-foot jump and full recovery landing could be achieved. For the patient group, this distance was recorded so that the same distance could be used in the follow-up test session.

6.2.3 Data analysis

The first phase of data processing was completed using Vicon's Workstation software (Oxford Metrics, England). Angles were defined according to the Joint Coordinate System (Grood & Suntay, 1983). The optimised lower-limb gait analysis (OLGA) method as described by Charlton *et al.* (2004) was used to improve the quality of the kinematic output; specifically, this method which uses a Kalman filter, has been shown to reduce variability across trials by minimising artefact due to soft tissue movement and kinematic cross-talk (Charlton *et al.*, 2004; DeGroote *et al.*, 2008). Anthropometric measurements, together with the marker positions recorded during the static trial were used to improve calculations of bone lengths. In combination with the walking trial used for the dynamic calibration, a better estimate of joint centres and segment orientations could be determined, thereby improving the reliability of the joint angle output (Charlton *et al.*, 2004; Roren, 2005). The default settings in Vicon were used for all activities.

Up to five trials from each activity were selected for further processing based on minimum kinematic fit residuals calculated in OLGA. The following gait cycle events for a minimum of three good trials were marked manually for export with the kinematic data: left and right foot strike over 1.5 strides (i.e. 3 steps = 4 foot strikes) during the walking and cutting trials, as well as heel off, toe off, and foot strike for the jump task. Foot strike was defined as the frame at which the first of either the heel or toe marker reached a local minimum during the stance or landing phase.

The cutting activity was split into two components. The first component included the initial step leading up to the change of direction and the step following the 90° cut (e.g. for a right-cut, this included the right, left, and right heel strikes). The second component included the step following the 90° cut and the step concluding the change in direction (e.g. for a right cut this included the left, right, and left step sequence). In other words, the step immediately following the initiation of the change in direction was included in both the first and last components of the activity. This was done to ensure the complete change of direction was captured in the gait cycles; while some subjects were able to finish the rotation in the second step of the three-step analysis, others performed a more rounded cut in which two steps were required to complete the 90° change of direction.

Each component of the cutting task and one stride of the walk activity was normalized to a 100% cycle. The trial time for the jump activity was calculated from first heel off to initial foot contact on landing times; in order to include the recovery period on landing, the whole jump cycle was defined as 65% longer than the trial time from initial heel off. The airborne (i.e. swing) phase of the jump, defined as final toe off to first contact landing, was also analysed as a separate 100% cycle.

The three-dimensional kinematic data and marked gait cycle events were exported from Workstation into Matlab where gait cycle normalisation was accomplished. Individual flexion-extension, add-abduction, and internal-external rotation curves were plotted for each trial, from which outliers were excluded and the remaining trials were used to calculate subject mean curves. Maximum, minimum, and range of rotation data were determined for rotations in each of the three anatomical planes from each subject mean curve for further statistical analysis. Range midpoint, defined as the mean of the maximum and minimum values, was also analysed. Activity mean curves were generated from individual subject mean curves.

Since the cutting task requires asymmetric behaviour of left and right legs, the activity was further divided to examine inside or outside (based on the turning direction) limbs. The inside limbs (i.e. right leg for the right-cut and left leg for the left-cut trials) were subsequently combined, with corresponding groupings carried out for the outside limbs. The activities were consequently categorised and described as follows:

Walk – left heel strike to heel strike,

Cut123Inside – inside limb during initial three foot strike events (i.e. limb that is in stance, then swing),

Cut123Outside – outside limb during initial three foot strike events (i.e. limb that is in swing, then stance),

Cut234Inside – inside limb during final three foot strike events,

Cut234Outside – outside limb during final three foot strike events,

JumpFull – the complete jump task from heel off to standing,

JumpSwing – the airborne/swing phase of the jump task.

6.2.4 Objectives and outcome

The primary objective of this study was to determine differences in transverse plane knee rotation from pre- to post-operative testing sessions in patients who had undergone either single or double-bundle ACL reconstruction during weight-bearing activities. The hypothesis with respect to the primary outcome was that both surgical procedures would reduce the overall range of rotation; however, the DB reconstruction would reduce it to a greater extent than the SB surgical technique. The secondary objectives were to compare the mean control and contralateral knee data with those of the injured knees pre- and post-operatively.

6.2.5 Statistical analysis

Interactions between the single and double-bundle groups from the pre- to post-operative test sessions were determined using a linear mixed model for repeated measures in SPSS 15.0 (SPSS Inc). Post-hoc analysis was conducted using a two-tailed paired samples t-test (or Wilcoxon signed-rank test if data were not normally distributed). The linear mixed model was also used to compare secondary outcomes including contralateral and injured knees of the patients in their subgroups (SB and DB tested pre- and post-operatively) and differences between

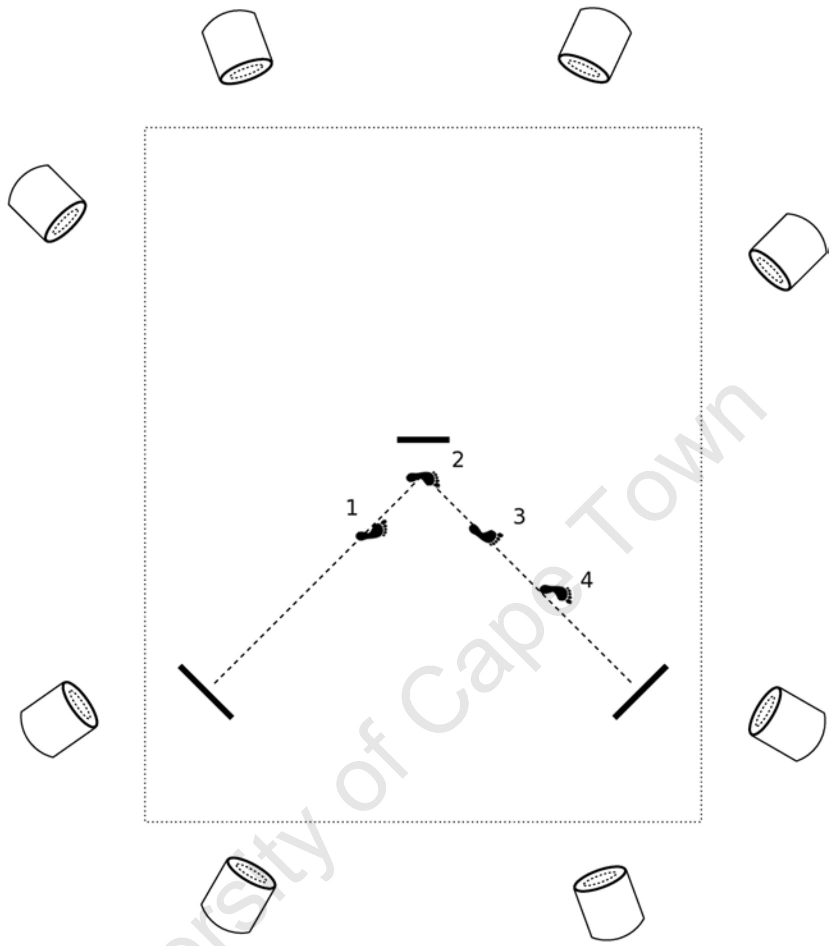


Figure 6.1: Gait lab setup showing typical foot strike positions for the Cut-Right activity and the numbering system used to define the first and second parts of the activity, Cut123 and Cut234, respectively. (Figure is not to scale.)

healthy left and right knees in the Control group. Control left-right averaged and ACL-deficient (ACLD) knee data were compared using independent samples t-tests (or Wilcoxon sum-rank test for non-normally distributed data). P-values less than or equal to 0.05 were considered statistically significant.

6.3 Results

The baseline data presented in Table 6.1 show that all four female patients were randomly allocated a single-bundle reconstruction. The SB group was also slightly older, and mean height and mass were slightly less than the DB group. Furthermore, although the differences were not statistically significant, there was a tendency towards a lower cutting activity cadence post-operatively in the SB group, while the mean cadence in the DB group remained approximately equal between testing sessions. The cadence of the ACLD patients was also significantly higher than that of the Control group for the Cut-Right activity; the difference between Controls and ACLD patients for the Cut-Left activity approached significance.

Table 6.1: Baseline demographic and clinical subject data (mean ± SD) for control and patient groups. ACL all includes subjects from both single-bundle (SB) and double-bundle (DB) groups.

Variable	**Session**	**Control**	**ACL all**	**SB**	**DB**
Sex (F:M)		3:8	4:18	4:7	0:11
Age (yrs)		29.5 ± 5.4	29.0 ± 5.7	32.1 ± 4.9	25.9 ± 4.9
Height (cm)		176.3 ± 9.3	174.3 ± 8.0	170.5 ± 7.5	178.0 ± 6.8
Mass (kg)		73.0 ± 12.0	81.8 ± 13.9	79.1 ± 14.4	84.5 ± 13.5
Time Injury-Pre (mos)		n/a	6.9 ± 10.4	6.7 ± 13.8	7.0 ± 6.0
Time Surg-Post (mos)		n/a	4.6 ± 1.6	3.6 ± 0.7	5.8 ± 1.6
Walk cadence	Pre	114.3 ± 7.9	110.0 ± 7.0	110.4 ± 7.0	109.7 ± 7.4
(steps/min)	Post	n/a	112.8 ± 6.4	114.4 ± 7.4	110.4 ± 3.8
Cut-Right cadence	Pre	149.1 ± 21.9	169.2 ± 28.1	166.9 ± 32.8	171.1 ± 25.1
(steps/min)	Post	n/a	162.9 ± 23.9	156.0 ± 11.2	171.6 ± 32.8
Cut-Left cadence	Pre	150.9 ± 25.4	169.9 ± 27.3	168.6 ± 32.8	170.6 ± 23.4
(steps/min)	Post	n/a	161.8 ± 21.3	156.5 ± 14.7	168.5 ± 27.1

6.3.1 Protocol deviations

Of the 22 patients in this randomised control trial, three allocated to the double-bundle reconstruction group were lost to follow-up (Figure 6.2). Two subjects in the single-bundle group were designated outliers for at least one of the activities. This definition was based on maximum and minimum flexion and rotation angles:

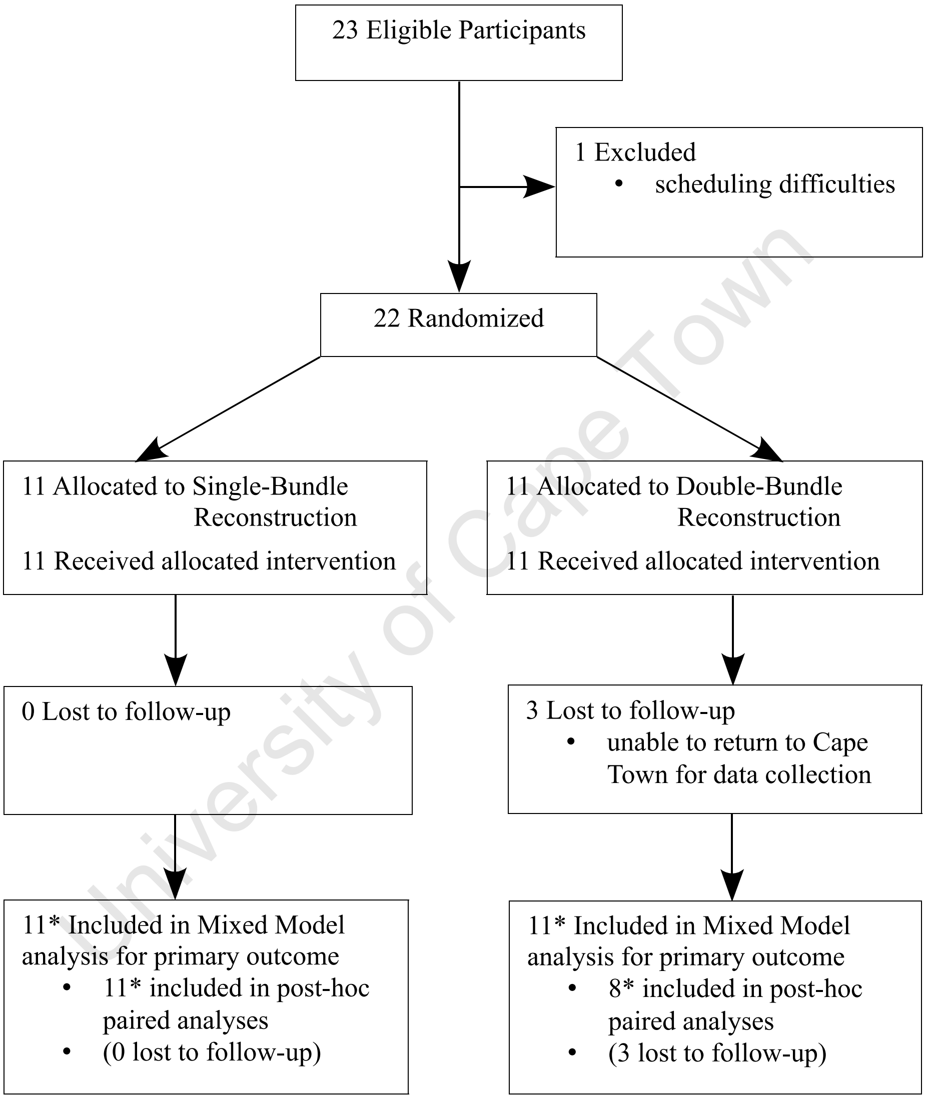


Figure 6.2: Flow diagram of participants through each stage of the randomised control trial for the primary outcome comparing single and double-bundle ACL reconstruction. * Some data were excluded from analyses if subjects were classified as outliers for a particular activity. (See explanation in section 6.3.1.)

if, for a particular activity, both flexion and rotation angles were greater than two standard deviations away from the mean of the overall group including all subjects, the data collection notes were examined for an explanation. The two patients for whom this was the case both experienced discomfort during weight-bearing activities due to their knee injuries. Their data was subsequently omitted from the group analyses for those particular activities in the pre-operative session.

6.3.2 Adverse events

Adverse events were described in detail in section 5.3.2. The two subjects who experienced problems following surgery (one requiring an additional surgery and one with swelling and discomfort) were both able to participate in the follow-up session of the gait study without any problems or physical limitations. Both of these subjects were allocated a SB reconstruction.

6.3.3 Gait activity kinematics in three planes

Maximum knee flexion angles for both Control and ACLD subjects performing the high-demand activities (cutting and jumping) exceeded maximum angles during walking (Table 6.2, Figures 6.3 to 6.5). In general, the inside knee demonstrated greater flexion than the outside knee while performing the 90° cut, with Control subjects consistently demonstrating greater maximum flexion than ACLD subjects for this and the JumpFull activities.

Similar to flexion, maximum adduction angle tended to increase for the more demanding activities; abduction angles (i.e. minimum negative adduction) varied to a lesser extent across activities for the ACLD knees than for the healthy control knees and compared to the maximum adduction angles.

Greater maximum and smaller minimum flexion and adduction angles were achieved throughout the full jump activity (JumpFull) when compared to just the airborne swing phase (JumpSW); however, rotation angles were only different for the maximum values. The minimum internal rotation angles, equivalent to maximum external rotation, were actually slightly less during the swing phase of the activity due to a more consistent alignment of the subjects' individual curves owing to more distinct activity start and end points.

No significant differences were demonstrated in maximum or minimum transverse plane rotation between the ACLD and healthy control knees, although ACL-deficient knees tended to have lower maximum and minimum internal rotation values (Table 6.2).

Since the second part of the cutting activity (Cut234) generally showed similar trends to the first part (Cut123), these data are presented in Appendix G. The following discussion will focus on the first part of the activity.

6.3.4 Outcomes: Transverse plane rotation

No significant interaction was found in range of rotation when comparing SB and DB injured knees during the pre- and post-operative test sessions. Significant interactions *were*, however, found in all activities except walking when analysing the rotation midpoint defined as the mean of the maximum and minimum rotations (Table 6.3). The rotation midpoint indicated the shift in the range of rotation (or rotational alignment) between test sessions shown in Figures 6.7 and 6.13. During both cut and jump activities, the transverse plane rotational alignment of the double-bundle knees was significantly closer to that of the healthy control group than that of the single-bundle reconstructions.

ACLD knees tended to have a more external shift in the range of rotation than did the Control subjects' knees during cutting; this shift, however, was not statistically significant. No significant differences were found in rotation midpoint between left and right knees of the Control subjects in any activities.

No differences in the shift of range of rotation of the patients' contralateral and injured knees were found from the pre- to post-operative testing session (Figures 6.14 and 6.15, Appendix G), i.e. the contralateral knees followed the same pattern as the injured knees from one test session to the next.

6.4 Discussion

This study measured 3D knee kinematics during dynamic weightbearing activities in healthy subjects and in ACL-injured patients prior to and following surgical reconstruction. With patients randomly allocated either a single or double-bundle

Table 6.2: Maximum and minimum joint angles (degrees) in three planes for all activities for the Control and ACL-deficient (pre-operative) knees. (Mean ± SD)

	Control		ACL-deficient	
Activity	**max**	**min**	**max**	**min**
Flexion				
Walk	62.7 ± 4.0	5.3 ± 4.1	59.9 ± 6.5	5.7 ± 7.6
Cut123Inside	101.2 ± 10.6	27.4 ± 7.2	87.7 ± 9.1	22.9 ± 8.8
Cut123Outside	81.8 ± 8.8	18.3 ± 6.7	72.2 ± 15.1	17.2 ± 7.5
Cut234Inside	101.7 ± 10.8	18.9 ± 5.9	87.9 ± 9.2	19.0 ± 6.9
Cut234Outside	81.5 ± 7.5	14.4 ± 5.0	77.7 ± 8.1	14.4 ± 8.0
JumpFull	82.6 ± 14.7	9.8 ± 6.5	72.8 ± 11.0	10.4 ± 7.8
JumpSW	49.9 ± 5.1	13.0 ± 7.5	50.4 ± 11.6	14.4 ± 6.5
Adduction				
Walk	14.4 ± 7.2	-2.1 ± 4.8	16.1 ± 9.2	-1.6 ± 9.2
Cut123Inside	19.3 ± 9.1	-1.8 ± 8.6	24.2 ± 12.6	0.5 ± 8.5
Cut123Outside	15.2 ± 8.9	-5.1 ± 7.9	19.6 ± 8.9	-1.1 ± 8.3
Cut234Inside	19.7 ± 9.2	-3.2 ± 7.7	24.2 ± 12.7	-0.8 ± 7.7
Cut234Outside	17.7 ± 7.8	-8.2 ± 10.0	20.4 ± 7.9	-1.2 ± 7.6
JumpFull	13.6 ± 6.8	-6.8 ± 9.7	16.6 ± 9.3	-3.8 ± 9.6
JumpSW	9.8 ± 5.6	-3.0 ± 5.4	14.0 ± 8.9	0.3 ± 6.7
Internal Rotation				
Walk	2.6 ± 7.9	-18.9 ± 7.9	3.2 ± 8.6	-18.9 ± 10.2
Cut123Inside	11.8 ± 8.1	-14.8 ± 7.6	7.4 ± 6.8	-18.9 ± 8.4
Cut123Outside	11.3 ± 5.5	-15.3 ± 6.4	8.9 ± 8.5	-16.8 ± 8.5
Cut234Inside	12.6 ± 8.3	-18.4 ± 6.4	7.5 ± 7.6	-19.8 ± 7.6
Cut234Outside	13.2 ± 6.3	-18.8 ± 8.0	10.7 ± 8.4	-19.4 ± 9.6
JumpFull	12.3 ± 6.2	-14.4 ± 6.3	10.7 ± 8.2	-13.4 ± 7.0
JumpSW	6.8 ± 7.1	-15.5 ± 6.3	4.2 ± 7.3	-14.3 ± 7.4

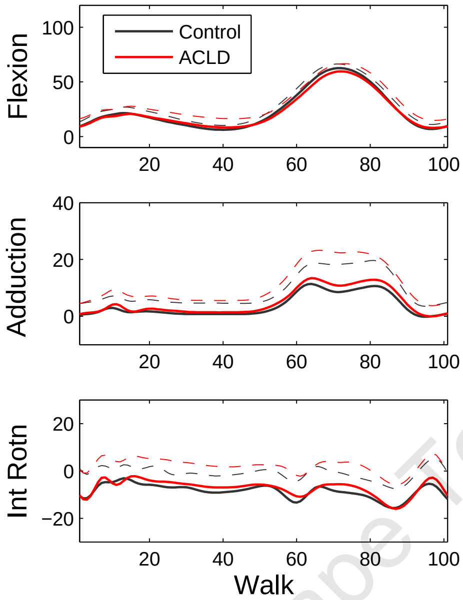


Figure 6.3: Walk three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

reconstruction, an objective comparison of these two treatment methods could be made under physiological loading conditions.

6.4.1 Three-dimensional knee kinematics during low and high demand activities

The walking activity showed comparable knee kinematics between the Control subjects and previously published data (Kadaba *et al.*, 1990), thereby verifying the methods used for this study and providing an acceptable baseline data set with which to compare the results of the other activities and patient outcomes.

No previously published data could be found in which identical methods of cutting or jumping activities were performed with which to compare the kinematic data; however, reasonable comparisons could be made with studies investigating similar tasks. McLean *et al.* (1999) and Sigward & Powers (2006) investigated kinematics during the stance phase of side-step cutting in healthy

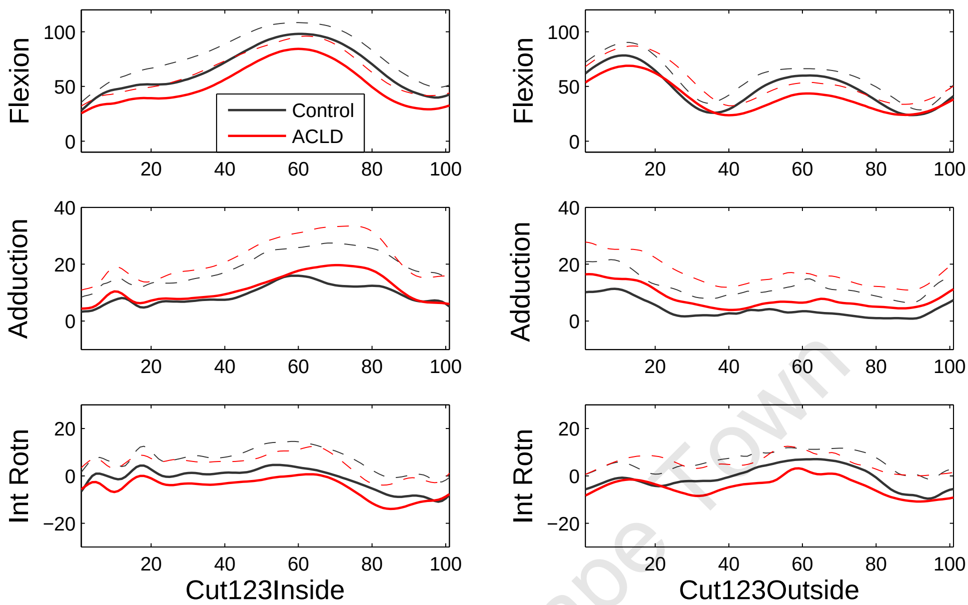


Figure 6.4: Cut123Inside and Cut123Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

subjects. Despite a less pronounced cut angle (45° rather than 90° used in this study), maximum flexion angles reported during the stance phases were slightly lower (approximately 46°) and greater (approximately 55°), respectively (McLean *et al.*, 1999; Sigward & Powers, 2006) than our results of just over 50° during the Cut123Inside stance phase (Figure 6.4).

In the frontal and transverse planes, results from the Control subjects in this study more closely matched those of Sigward & Powers (2006) and Nagano *et al.* (2009) in which subjects demonstrated between 0° and 10° of adduction, as well as initial external rotation followed by approximately 6° of internal rotation. The results of McLean *et al.* (1999) on the other hand, displayed joint angle curves in the abduction and external rotation domains (rather than primarily adduction and internal rotation); these reciprocal findings may be attributed to a simple shift in the curves as a result of differences in the anatomical landmarks chosen and segment coordinate system definitions. A change in the orientation of the flexion-extension axis can alter the calculated ab-adduction and internal-external

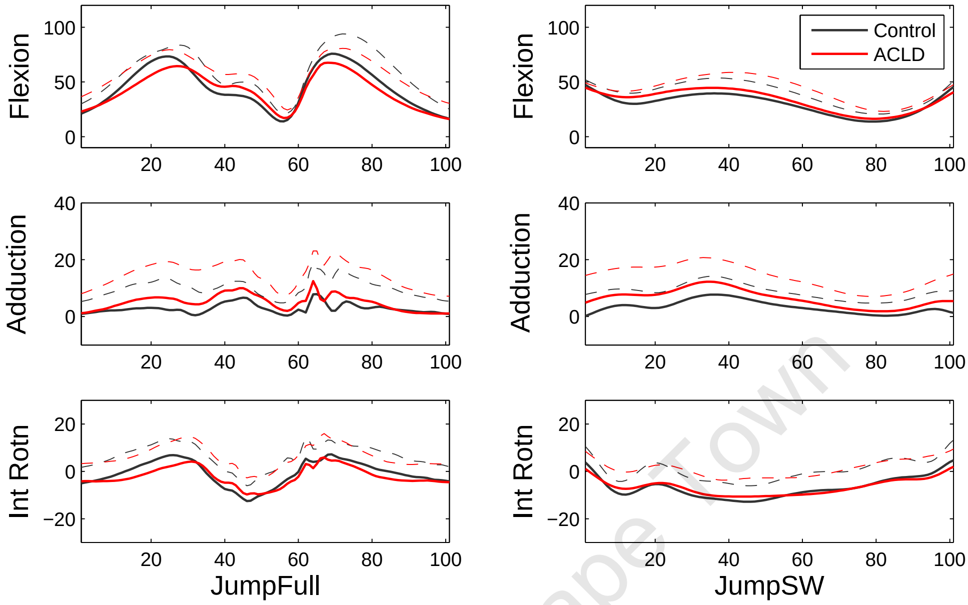


Figure 6.5: JumpFull and JumpSW three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

rotation angles by as much as 15° (Kadaba *et al.*, 1990; Piazza & Cavanagh, 2000). In both this study and that of Sigward & Powers (2006), the Vicon model was used as a basis from which to calculate joint angles.

Range of internal-external rotation over the 100% Cut123Inside cycle was greater than those presented by previous studies for similar activities (McLean *et al.*, 1999; Nagano *et al.*, 2009; Sigward & Powers, 2006); however, the greatest change between maximum and minimum values occurred between 55% and 100% of the cycle (i.e. the swing phase), which was not analysed in these other studies. As changes in kinematics with ACL injury have been observed during the swing phase of gait (Andriacchi & Dyrby, 2005; Georgoulis *et al.*, 2003), we considered it important to include this data in the analysis.

The jump activity cycle demonstrated an increase and then decrease in knee flexion prior to toe-off (discerned by comparing the JumpFull and JumpSW curves). Maximum extension was reached just before landing at which point the knee flexed to absorb the impact from the landing. Adduction and rotation

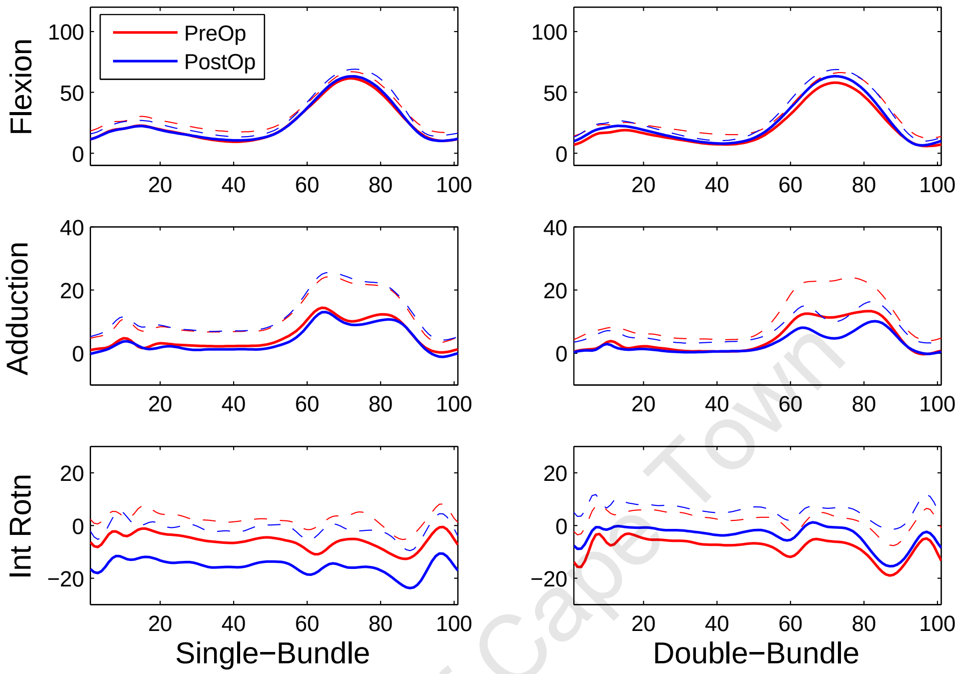


Figure 6.6: Walk three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

curves displayed relatively high frequency changes upon landing, which can most likely be attributed to the reverberations of the wand markers during this high impact stage of landing. Maximum adduction and internal rotation values must, therefore, be interpreted with caution.

The general pattern of the internal-external rotation curve during the take-off and landing (i.e. stance) phases of the JumpFull activity conformed to those of similar activities in other studies, including two-foot vertical jump landing (Nagano *et al.*, 2009) and squatting (Hemmerich *et al.*, 2006; Yamaguchi *et al.*, 2009). The tibia rotated internally with respect to the femur with increasing knee flexion, demonstrating coupled screw-home motion (Benoit *et al.*, 2007; Hill *et al.*, 2000). External rotation accompanied knee extension during the swing phase of this activity (Figure 6.5).

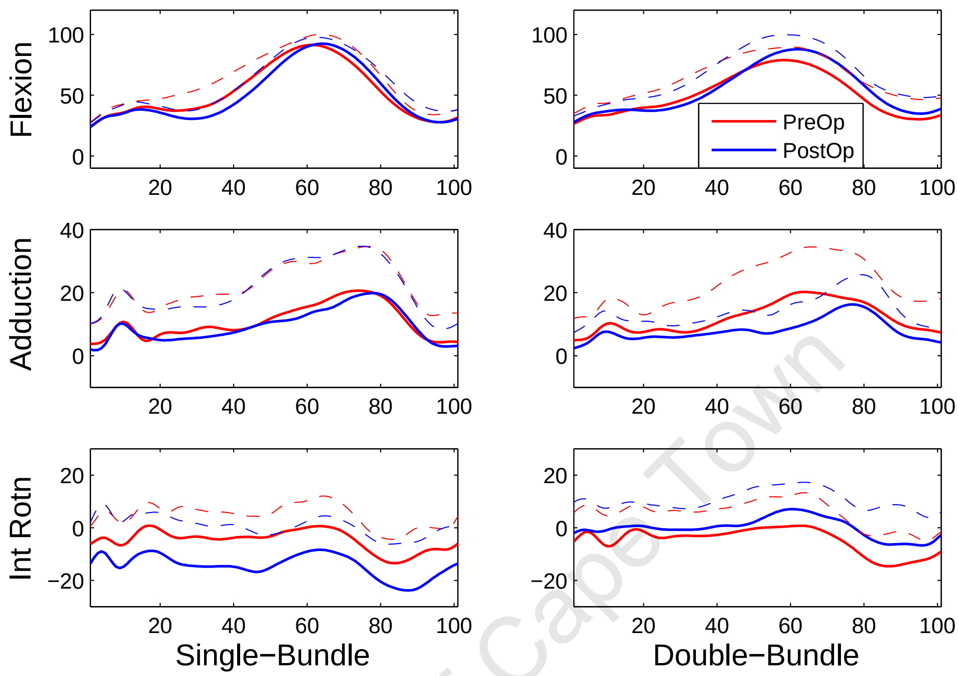


Figure 6.7: Cut123Inside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

6.4.2 Differences in kinematics in healthy Control and ACL-ruptured knees

Joint kinematic curves in all three planes demonstrated similar patterns between the injured and healthy knees. However, the slower cadence of the self-selected walking pace and a decrease in maximum flexion angles during all three activities indicated a possible protection mechanism adopted by the ACLD subjects (Table 6.2). Similar reductions in flexion were observed during the stance phase of walking gait in ACLD subjects when compared with healthy control subjects by several other investigators (Chmielewski *et al.*, 2005; Georgoulis *et al.*, 2003; Rudolph *et al.*, 2001). By increasing knee stiffness and thereby limiting degrees of freedom, a subject with deficient afferent feedback due to injury is able to stabilise the joint (Chmielewski *et al.*, 2005; Georgoulis *et al.*, 2003).

This decrease in flexion angle was more pronounced during cutting and jump-

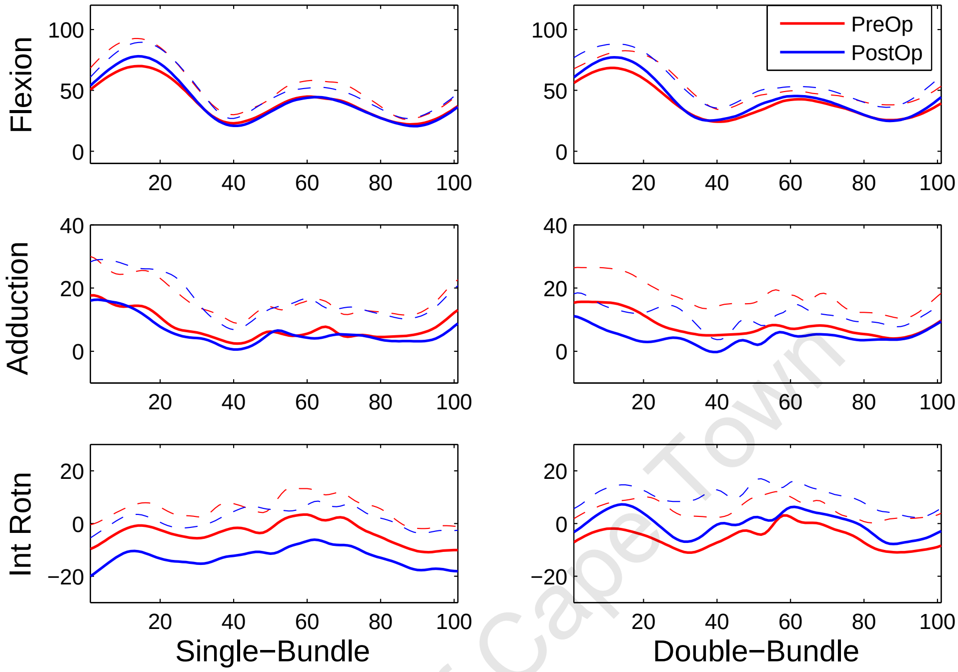


Figure 6.8: Cut123Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

ing than walking with a difference of up to 13.8° between mean maximum flexion between the two groups during the Cut234Inside activity (Table 6.2). The reduced flexion may not be attributed to a more cautious technique by which this activity was performed by the patient group as evidenced by a slight increase in cadence when compared to the the Controls. This shows that, whereas the Control group adopted a relaxed, moderate intensity level, the patient group made greater effort to perform this activity at the maximum intensity level at which they felt comfortable in their injured state.

No significant differences in range of internal-external rotation were found between the ACLD and the healthy Control knees. The ACLD subjects, however, demonstrated a general tendency toward greater adduction and less internal rotation than the Controls during the high demand activities (Table 6.2). Again, this supports the theory that the patients adopted a knee protection strategy.

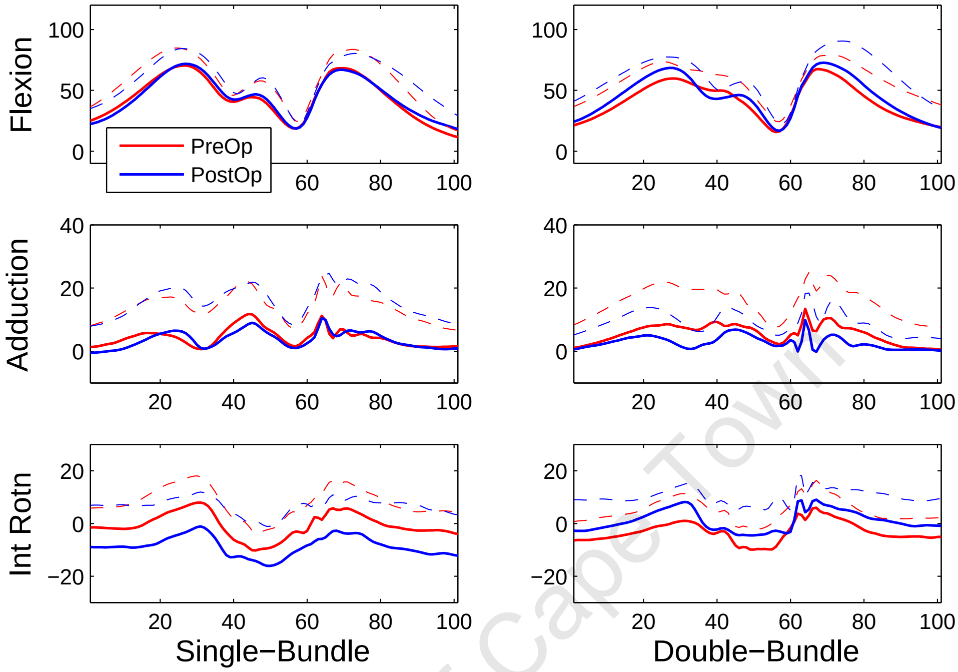


Figure 6.9: JumpFull three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

Passive abduction (valgus moment) combined with internal rotation manifests the pivot shift phenomenon in the ACLD knee and correlates to a patient's sense of instability during gait (Amis *et al.*, 2005; Yamaguchi *et al.*, 2009). In order to prevent a possible 'giving way' occurrence, someone with an injured ACL may actively adduct and externally rotate the tibia through muscle activation. Since the ACL is a secondary restraint to internal rotation, co-contraction of the medial hamstrings would not only stiffen the joint, but would rotate the tibia externally with respect to the femur, reducing the risk of damage to other structures in the absence of restraint of the ACL.

Similar shifts in rotation and adduction have been observed in ACL-deficient subjects during walking and running (Tashman *et al.*, 2004; Zhang *et al.*, 2003). When compared with the normal knees in the healthy control group (Zhang *et al.*, 2003) or contralateral uninjured knees (Tashman *et al.*, 2004), the ALCD knees

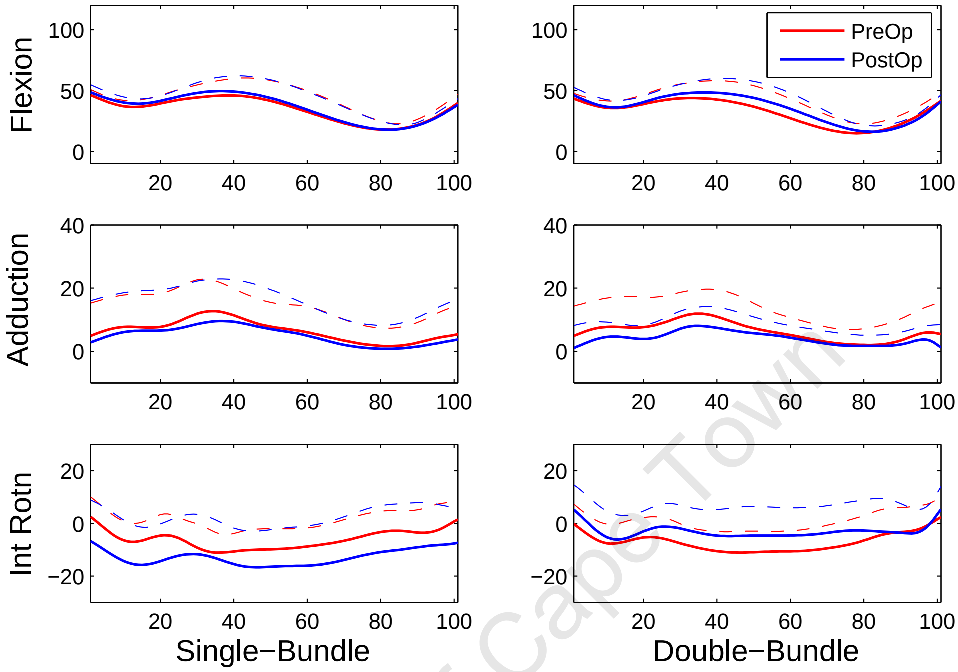


Figure 6.10: JumpSW three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

showed a similar ranges and patterns of rotation, but with a roughly 4° increase in external rotation and 3° increase in adduction.

Opposite trends were observed by Georgoulis *et al.* (2003) and Andriacchi & Dyrby (2005); their ACLD subjects demonstrated less external rotation with differences reaching significance during the swing phase of gait. Interestingly, Andriacchi & Dyrby (2005) observed a simultaneous decrease in anterior translation just prior to heel strike and found that the shift in rotation was correlated with the magnitude of the flexion moment during the initial stance period. In fact, our data also indicates a minimal shift towards internal rotation (although not statistically significant) of the ACLD group during the walking activity (Figure 6.3) with a greater difference occurring at the peak just before heel strike. It is possible that these subjects had developed a different compensation strategy with priority placed on minimizing anterior displacement of the tibia by increasing the

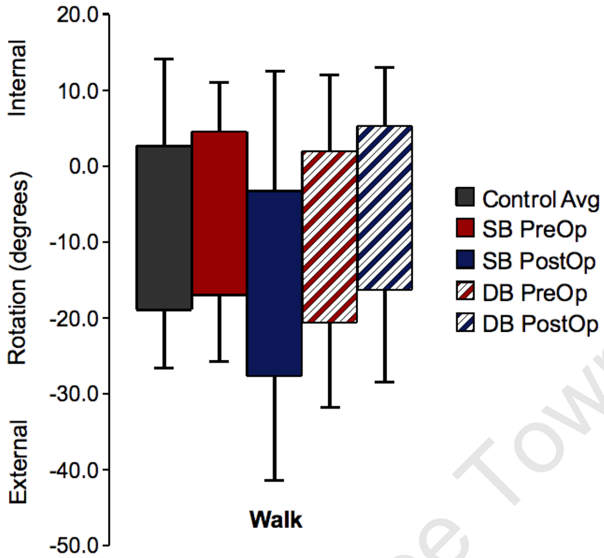


Figure 6.11: Walk rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

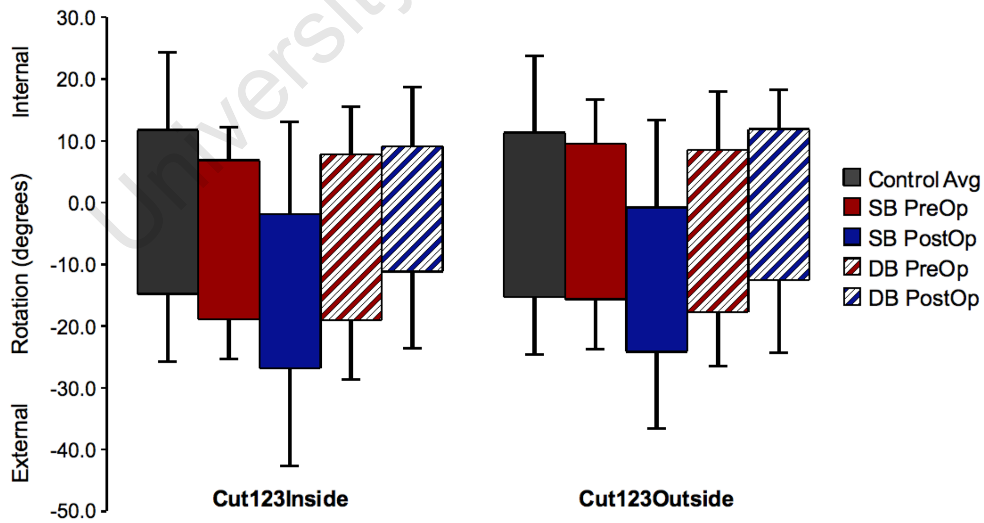


Figure 6.12: Cut123Inside and Cut123Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

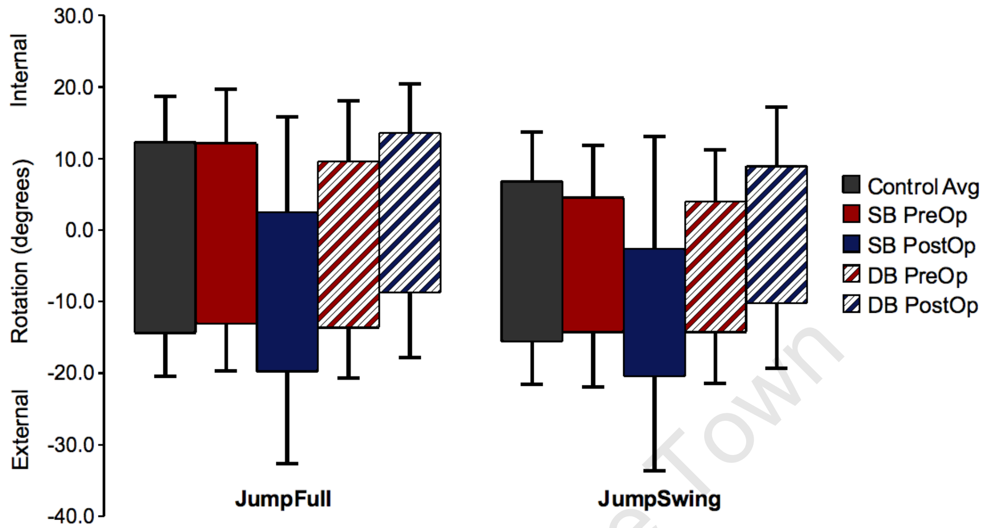


Figure 6.13: JumpFull and JumpSW rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

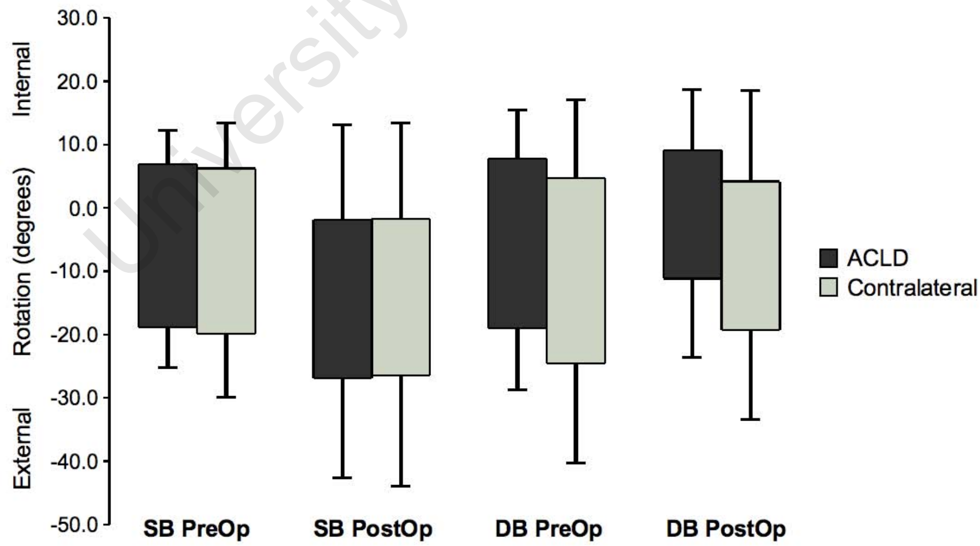


Figure 6.14: Cut123Inside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

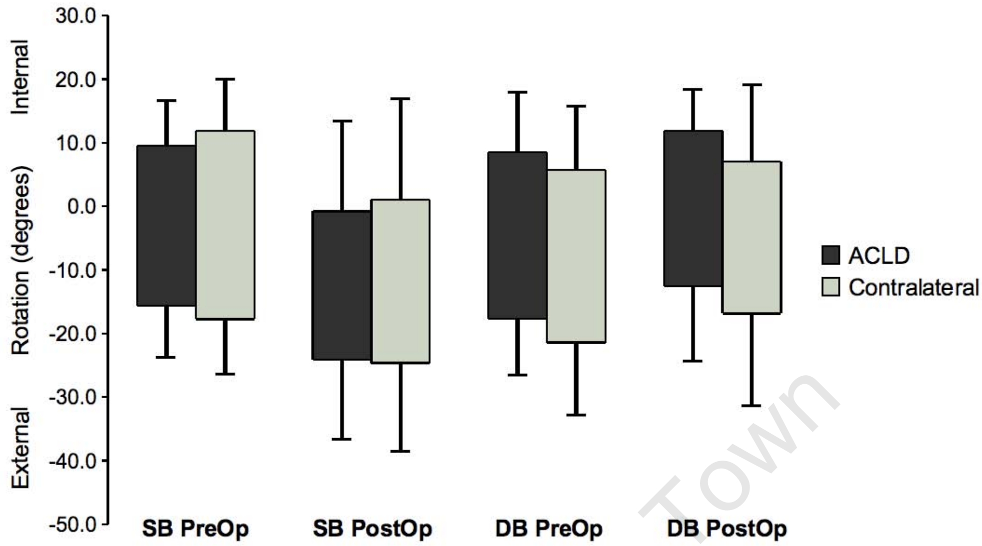


Figure 6.15: Cut123Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

flexion moment during the less demanding activity of walking.

It has been shown that co-contraction of the hamstrings is more effective at reducing the strain on the ACL between 15° and 60° of flexion and reducing internal rotation of the tibia at flexion angles greater than or equal to 30° (Li *et al.*, 1999). Since maximum flexion angles during the weightbearing phase of walking are smaller than during cutting or jumping, a co-contraction strategy would have less effect in reducing the strain on the ACL during this activity. With conflicting reports in the literature, however, further investigation is required to draw accurate conclusions regarding causes for differences in transverse plane rotation between ACLD and healthy knees during walking and high-demand activities.

Table 6.3: Means and standard deviations (in degrees) of rotation midpoint for all subject groups during each dynamic activity. P-values for surgery by test-time interaction are listed for the injured knee groups. (Statistically significant interactions are highlighted.)

Activity	**Test**	**ControlAVG**		**SB Contralat**		**DB Contralat**		**SB Injured**		**DB Injured**		**Srg x Time**
	Time	mean	SD	mean	SD	mean	SD	mean	SD	mean	SD	**p-value**
Walk	PreOp	-8.2	7.8	-7.9	6.7	-12.0	12.3	-6.2	7.2	-9.3	10.4	0.070
	PostOp			-16.9	13.7	-11.0	14.4	-15.5	14.7	-5.5	9.9	
Cut123	PreOp	-1.5	6.9	-6.8	8.6	-9.9	13.6	-6.0	5.7	-5.6	7.1	**0.041**
Inside	PostOp			-14.1	16.1	-7.5	14.2	-14.4	15.4	-1.0	10.0	
Cut123	PreOp	-2.0	5.7	-2.9	7.1	-7.8	10.7	-3.1	7.3	-4.6	8.7	**0.019**
Outside	PostOp			-11.8	14.6	-4.9	13.3	-12.5	13.2	-0.3	8.6	
Cut234	PreOp	-2.9	6.0	-7.5	8.4	-10.3	13.5	-5.5	5.8	-6.8	7.2	**0.023**
Inside	PostOp			-14.9	16.6	-8.3	13.5	-15.1	14.4	-1.9	8.8	
Cut234	PreOp	-2.8	6.6	-5.0	5.8	-8.5	11.0	-4.0	8.0	-4.7	8.8	0.082
Outside	PostOp			-13.0	14.5	-6.4	13.1	-13.4	15.4	-2.1	10.4	
JumpFull	PreOp	-1.1	5.9	-1.3	8.5	-4.8	11.5	-0.4	6.9	-2.0	7.1	**0.018**
	PostOp			-6.8	14.5	-2.0	12.6	-8.6	13.2	2.4	7.9	
JumpSW	PreOp	-4.4	6.2	-3.5	7.1	-7.7	12.0	-4.9	6.9	-5.1	7.2	**0.033**
	PostOp			-10.1	14.3	-4.2	12.9	-11.5	14.6	-0.6	8.7	

6.4.3 Differences in rotational laxity in single versus double-bundle reconstructed knees

Although maximum and minimum rotation values varied between ACLD and Control group knees and between the SB and DB injured knees from pre- to post-operative testing sessions, the overall range of rotation remained relatively constant between groups for all activities; similar results were found by Zhang *et al.* (2003) and Tashman *et al.* (2004) when investigating ACL-deficient knee kinematics. Because maximum and minimum values shifted in the same direction when comparing groups or test sessions (Figures 6.11 to 6.13), the rotation range midpoint gave a better indication of differences concerning transverse plane rotations between groups. The rotational midpoint gives an indication of the rotational alignment of the tibia with respect to the femur throughout the activity. By using the mean of maximum external and internal rotations, the centre of the envelope of rotation for each activity was established. The clinical importance of this quantity is associated with the mechanics of the joint; in this case the potential for joint degeneration caused by increased normal or shear forces in areas where they normally do not occur.

The greater maximum flexion angles and higher cadences of the 90° cut for both the patient and Control groups are clear indications that this activity was more demanding than walking; it was, therefore, not surprising that the trends observed following reconstruction with regard to internal-external rotation midpoint during walking (Figure 6.11) became statistically significant during cutting (Figure 6.12).

Similarity in range of rotation between the SB and DB groups is evidence that the the different surgical reconstruction techniques provided little difference in knee transverse plane laxity; however, the shift in range midpoint in opposing directions from pre- to post-operative testing sessions when comparing groups is a clear indicator that there is a difference in outcome between surgical techniques. The fact that the contralateral (uninjured) knees demonstrated the same shift of range midpoint between test sessions as their injured counterpart (Figure 6.14) suggests that the contrast between SB and DB outcomes may be attributed to more than simply differences in the mechanics of the surgical procedures. Berchuck *et al.* (1990) observed comparable homogeneity between injured and contralateral knees in their group of unilateral ACL-deficient patients when measuring knee flexion angles and moments. They attributed the similarities in injured and uninjured knees to their reprogramming theory, in which the locomotor process adapts to the deficient ACL by avoiding excessive motion of the tibia in order to protect the joint. The fact that the shift in range of rotation occurred over the entire cycle rather than during a specific phase (e.g. stance or swing) supports the concept that this is a neuromuscular adaptation due to poor stability rather than an instantaneous response that would likely show a variation in rotation at the point in the stride cycle following displacement of the tibia (Berchuck *et al.*, 1990).

What is unusual about the interaction between SB and DB groups with respect to the shift in rotation midpoint is that post-operatively, the SB group demonstrated an even greater disparity from the Control group than pre-operatively. In other words, the injured knees tended to exhibit slightly more external rotation than the healthy knees. The DB reconstructed knees' range of rotation then shifted internally with respect to the pre-operative state back to that of the normal knees, while the SB reconstructed knees demonstrated a *further* increase in external rotation range shift of approximately 10° for the cutting activity (Figure 6.12). This response was more pronounced for the high-demand activities than for walking.

Both of these observations indicate the involvement of the muscles around the knee used to stabilise the joint during dynamic gait, specifically, those involved in the ACL surgery. Several studies, including this one, have shown an increase in knee stiffness with ACL injury, described as an 'immature stabilization strategy' as observed by a decrease in maximum knee flexion angle and an increase in co-contraction of the muscles around the joint (Chmielewski *et al.*, 2005; Rudolph *et al.*, 2000, 2001). Co-contraction has been shown to unload the anterior cruciate ligament at higher flexion angles, thereby protecting the injured knee (Li *et al.*, 1999; O'Connor, 1993). Since the semitendinosus and gracilis tendons are in part responsible for internal rotation of the tibia, weakness of these muscles following harvesting of the tendons for the ACL graft may have resulted in an insufficiency in their ability to oppose the external torque produced by the biceps femoris during co-contraction (Viola *et al.*, 2000).

The return to normal of the DB range of rotation midpoint implies that this group did not require the same co-contraction strategy for stabilisation used by the SB group. A tendency to return to normal magnitudes of flexion and adduction in the DB group, while these rotations in the sagittal and frontal planes remained similar to pre-opeative values in the SB group (Figures 6.6 to 6.10), further supports this hypothesis.

Additionally, it is possible that subjects experiencing instability due to injury implemented a reprogramming strategy similar to that suggested by Berchuck *et al.* (1990). Since the ACL is secondarily responsible for restraint of internal torsional loads at the knee (Amis *et al.*, 2005; Blankevoort & Huiskes, 1996), a protection mechanism providing greater overall external rotation of the tibia may have been used to reduce the possibility of further injury that could occur with extreme internal rotation resulting from a ruptured or inadequate ACL ligament or graft. The significant shift towards external rotation observed in the SB group following ACL reconstruction may be a combined effect of the locomotor system adapting to an unstable joint and the inability of the compromised medial hamstring muscles to oppose the external torque produced by the biceps femoris during co-contraction.

6.4.4 Study limitations

One limitation of this study was that all (4 out of 22) female patients were randomly selected to receive the single-bundle reconstruction. It has been found that females suffer a higher incidence of ACL ruptures when compared to their male counterparts; however, knee kinematics were not found to be significantly different between men and women during side-step cutting tasks (McLean *et al.*, 1999; Sigward & Powers, 2006).

Nonetheless, to ensure gender did not influence our findings, additional analyses were carried out to compare test session by rotation range midpoint interaction between the female and male subjects for each activity. Female subjects displayed the same movement toward greater external rotation following ACL reconstruction as male subjects and no statistical differences were found between gender subgroups. (All p-values were greater than 0.31.) We therefore, concluded that this did not influence our overall results.

The mean follow-up period for the DB group was approximately two months longer than the SB group, which may have permitted further healing of the graft in these subjects and differences in the outcome. However, two months is a negligible duration when compared with the overall time of several years required for recovery from this injury with some patients *never* returning to a pre-injury sense of joint stability (Risberg *et al.*, 2004).

Furthermore, if the findings may, in part, be explained by muscle co-contraction and weakness at the donor site, one must take into account the time required for healing of the harvested tendons. In an intra-operative investigation, Ferretti *et al.* (2002) found distinct differences in the collagen fibre bundles of the regenerated semitendinosus tendon at 6 versus 24 months post-reconstruction, indicating that two years or longer are required for complete regeneration. Viola *et al.* (2000) moreover observed significant reductions in internal tibial rotation strength over four years post-operatively. The two-month difference in the follow-up testing period between groups is, therefore, considered clinically negligible. In order to verify this deduction and to additionally determine long-term differences in outcome between these two surgical techniques, further follow-up testing should be conducted at least two to five years post-operatively.

Soft tissue artefact is a common concern when interpreting results in gait analysis. In a recent study in which kinematics from skin markers were compared with those calculated from bone-pin markers, Benoit *et al.* (2006) illustrated up to 4.4° and 13.1° differences for walking and cutting activities, respectively. In this study, an optimization algorithm (OLGA) was used to minimize the effects of soft tissue motion especially in the determination of transverse plane rotation (Charlton *et al.*, 2004; DeGroote *et al.*, 2008; Roren, 2005). Most importantly, the same methods were implemented for all subjects (those receiving single and double-bundle reconstructions) at both pre- and post-operative testing sessions. The statistically significant interaction established between intervention groups and sessions, therefore, could not have been a result of soft tissue artefact.

6.4.5 Conclusions

In this study, the outcome of single and double-bundle reconstruction of an isolated ACL rupture was investigated during dynamic activities. The 3D knee kinematics measured during walking, cutting, and jumping compared well with those of similar activities reported in the literature; high-demand activities revealed greater differences between the healthy Control and ACLD knees pre-operatively, as well as the SB and DB groups post-operatively. Although no significant differences were found between Control and ACLD transverse plane rotations, findings in all three planes suggested the use of a protection mechanism by the injured patient group. The observation that contralateral knee kinematics followed the same directional shift in rotation following ACL reconstruction as the injured knees supported the theory by Berchuck *et al.* (1990) that the locomotor process is reprogrammed to adapt to the deficient joint.

The hypothesis that range of rotation would be affected by ACL reconstruction was not confirmed by this study. However, the interaction of the rotation range midpoint between the SB and DB groups from pre- to post-operative testing sessions indicated persistent laxity in the SB group, while the DB three-dimensional kinematics returned closer to those of the healthy control subjects.

The possibility that a co-contraction stabilization strategy was used with subsequent graft harvest site weakness contributing to changes in transverse plane kinematics may be an important consideration for post-operative rehabilitation. The external rotation shift demonstrated by the SB group following reconstruction (and to a smaller extent by all ACLD subjects) could have significant implications for long-term joint degeneration as different structures within the joint will experience compressive or shear forces that previously were unloaded; concomitant unloading of other tissues will occur in other areas of the joint.

Additional long-term follow-up studies are required to determine the effects on joint laxity, donor site morbidity, and joint degeneration following single and double-bundle ACL reconstruction for a more comprehensive comparison of these surgical techniques; however, these preliminary *in vivo* results under physiological loading conditions indicate improved joint constraint following double-bundle reconstruction.

Chapter 7

Conclusions and recommendations

7.1 Rotational laxity outcome: Does the double-bundle ACL reconstruction provide better restraint than the single-bundle technique?

In this thesis I have addressed the question of whether the double-bundle (DB) reconstruction of a deficient anterior cruciate ligament (ACL) is more advantageous in providing rotational restraint at the knee than single-bundle (SB) surgery. The term '*rotational* laxity' used in the literature is somewhat ambiguous. It has been interpreted as the subjective degree of instability as assessed by a clinician, resulting from a load that incorporates a torsional element, such as the pivot shift test. The instability measured by this test is not confined to transverse plane rotation, however. Rotational laxity has also been assessed in terms of tibiofemoral internal-external rotation. While the kinematics ensuing from daily physiological activities *may* be comprised of a certain degree of axial rotation, the loading conditions may not necessarily incorporate a torsional component at all.

In evaluating the outcome of the two surgical techniques, our measure of laxity focussed on the transverse plane rotation occurring at the tibiofemoral joint; the applied loading conditions varied, however, in the two studies involving ACL-deficient patients. In the first study (Chapter 5), we wished to compare the proficiencies of the single and double-bundle reconstruction against the ACL-deficient and healthy knee to restrain a known isolated torsional load. Previously existing methods of assessing rotational laxity at the knee were judged to be inadequate to accurately and objectively measure knee kinematics under these specific loading conditions. An innovative device was therefore designed to apply a precise static torque about the long axis of the tibia, incorporating a 'relaxation' period to ensure minimal depreciation of applied load once the foot position was fixed. With the custom-built apparatus keeping the knee position stationary under load, the knee was imaged using a magnetic resonance imaging (MRI) scanner, thereby avoiding soft tissue artefact and providing reliable kinematic data (Chapter 3). Knowing that the complex structure of the joint may yield motion in more than one degree-of-freedom, the image analysis methodology was furthermore developed to measure rotations and translations in all three anatomical planes that could result from the applied torque.

The second study that assessed rotational knee laxity in ACL patients (Chapter 6), did so under dynamic weightbearing conditions. Although the specific loads applied at the knee via the ground reaction forces at the foot were not known or precisely controlled as with the first investigation, this study provided a means by which to compare the two surgical techniques under realistic loading conditions. The effects of not just the passive restraints of the knee were taken into account, but also the influence of compressive loads and muscle activation on the measured outcome.

Given the difference in loading conditions between the two clinical studies, it is not surprising that the findings were comparatively at odds: under passive torsional loading at 30° of flexion, the patients who had been allocated the double-bundle reconstruction demonstrated a trend toward less internal rotation than the contralateral uninjured knee, while the rotation measured in the single-bundle group was closer to normal. Alternately under dynamic conditions, while the range of rotation did not differ between the two groups, the transverse plane rotational alignment of the double-bundle reconstructed knees was significantly closer to that of the healthy control group than the transverse-plane position of the single-bundle reconstructions during the cut and jump activities.

In addition to the difference in loading conditions, the other important distinction between the two studies was the flexion angles at which rotation was being measured: under passive loading, tibiofemoral rotation was measured at 0° and 30° of flexion, while under dynamic loading during the high-demand activities flexion angles ranged from 10° to over 100°.

The results from these two studies are not necessarily contradictory and illustrate the importance of the different methods of assessing *in vivo* rotational laxity. To reconcile the outcomes, we must examine how the variations in study conditions could affect the results.

The intact ACL was better able to control the applied torsional load in only the extended position under passive isolated torque; at 30° of flexion there was no difference in either internal or external rotational laxity between the ACL-deficient and the contralateral knees under isolated passive loading conditions. The effect of surgical technique was only observed in the flexed position, however, where the DB reconstruction restricted rotation to a greater extent than both the SB reconstruction and the intact knee. Therefore, the addition of the second graft bundle was able to restrain (or perhaps overconstrain) a torsional load more than both the single-bundle graft and native ACL at a higher angle of flexion.

On the other hand, the divergent rotation shift displayed by the dynamic kinematics in the two patient groups following reconstruction was reasonably consistent over the entire activity cycle and was not restricted to a specific flexion angle. The shift in the knees of the SB group away from the normal rotational alignment following reconstruction, which suggested inferior kinematic constraint when compared to the pre-operative state, was evidence that the graft donor site – compromised with surgery – was involved in providing joint restraint. This, therefore, gave an indication (albeit speculative) that an active stabilisation strategy involving the hamstrings was used by the patients in the SB, but not the DB group under dynamic loading conditions over the range of flexion. A greater sense of security in the DB patients would make any additional co-contraction of the hamstrings unnecessary, resulting in the observed rotational shift towards the normal control group. Further investigation via electromyography could objectively assess this hypothesis.

Since the passive loading investigation demonstrated that there was no difference in rotational laxity with ACL rupture compared with the intact knee at the greater flexion angle, the sense of stability perceived by the DB patients under physiological loading conditions may be attributed to the capability of the double-bundle graft to restrain those forces and moments accompanying torsional loading during dynamic tasks, such as compressive axial force, varus-valgus moments, and anterior-posterior forces. This theory is again supported by the findings of the passive loading study at full extension in which the ACL and other rotational restraints are in greater tension (Amis & Dawkins, 1991; Blankevoort *et al.*, 1991). In this extended position, the ruptured ACL *did* cause an increase in laxity and the reconstruction improved transverse plane restraint. Similarly, with additional loads at the knee that would strain the supporting structures including the ACL during dynamic tasks, the effect of an inferior graft would become apparent.

Axial compression in the absence of any additional external forces at the knee joint has been shown to cause both anterior tibial translation and transverse plane rotation (Liu-Barba *et al.*, 2007; Meyer & Haut, 2008). The augmented ACL strain and further increase in anterior and rotational laxity that has been observed in the ACL-deficient knee under joint compression (Fleming *et al.*, 2001; Liu-Barba *et al.*, 2007; Meyer & Haut, 2008) is consequently logical.

The magnitude of rotation that resulted from isolated compression was less than 10°, however, even at the point of catastrophic failure in the cadaver specimens (Liu-Barba *et al.*, 2007; Meyer & Haut, 2008). This degree of transverse plane rotation, still within normal limits of knee motion, would not cause damage to the ACL or other structures without coupled motion in another degree of freedom. Mean anterior translation at failure under compressive loading was found to be 27 ± 15 mm (Meyer & Haut, 2008) and likely provided a greater contribution to ACL strain than did rotation under axial loading condtions. This anterior tibial displacement is ascribed to the anterior component of the compressive force resulting from the posterior slope of the tibial plateau (Blankevoort & Huiskes, 1996; Liu-Barba *et al.*, 2007; Meyer & Haut, 2008).

With concomitant loads at the knee such as internal or external torques, however, joint compression has actually resulted in decreased rotational laxity when compared with isolated torsional loading conditions; the geometry of the tibiofemoral contact surfaces were thought to provide this additional restraint to the joint (Blankevoort & Huiskes, 1996; Wang & Walker, 1974). Activation of the muscles around the joint provide further stability, not only through the additional stiffness supplied by the muscle fibers, but also by taking advantage of the frictional and normal force contribution of the joint contact surfaces with joint compression (Wang & Walker, 1974). Li *et al.* (1999) demonstrated this experimentally in 10 knee specimens; a decrease in ACL force with co-contraction of the quadriceps and hamstrings was observed when compared with the isolated quadriceps force during a simulated isometric extension of the knee.

During dynamic activities such as the cutting task performed by our subjects, substantial joint moments occur in all three anatomical planes of motion (Sigward & Powers, 2006) and AP forces have been demonstrated during even low-demand activities such as walking (Andriacchi & Dyrby, 2005). While the combined forces at the knee would have undoubtedly added stress to the intact ACL or reconstructed graft in the absence of axial compressive loading, under weightbearing conditions with muscle co-contraction it would not be unreasonable for the measured range of rotation to be equal to or less than the range under isolated loading conditions.

These distinctions have been described as the 'operating point' and 'limits of passive stability' (Tashman *et al.*, 2004); in our subject groups the range of passive transverse plane rotation at 30° of flexion was approximately 25° (Figures 4.2, 5.2, and 5.3), while the range of dynamic rotation over the cutting activity cycle was generally between 20° and 30° with the maximum and minimum rotation peaks typically occurring at or above 30° of flexion (Table 6.2 and Figures 6.4, 6.7, 6.8, 6.12, and 6.14). Given the consistency in range of rotation across subject groups and test times during the dynamic activities, the perceived instability in the SB as compared to the DB group was conceivably due to motion other than axial rotation. The observed external rotation shift in the SB group was therefore not due to the inefficacy of this surgical procedure to restrain rotation, but rather an indirect consequence of joint laxity caused by ACL graft deficiency and the co-contraction stabilisation strategy employed to control it.

The regularity of the operating range of rotation under dynamic loading is highlighted again when comparing it with the asymmetrical limits of passive laxity in left and right knees of the healthy control group under isolated internal and external torque (Chapter 4). With the knee in flexion, a substantial degree of 'secondary' rotation - easily up to or beyond 10° in each direction – occurs with relatively small (1 to 2 Nm) initial torque values (Musahl *et al.*, 2007; Wang & Walker, 1974). Rotation in excess of this initial laxity requires disproportionately more torque due to the non-linear stiffness of the ligaments (Woo *et al.*, 2006). While the difference in left and right passive rotation was statistically significant at 3° and 5°, this quantity may not be clinically relevant under passive loading conditions.

The neutral resting position of the subject in which the high resolution MRI scan was performed with the knee in full extension was *not* assumed to be at 0° of rotation in our study. Instead, the degree of neutral position rotation was subtracted from the torqued measure of rotation to calculate the net rotation under load. At full extension, the degree of secondary laxity is probably less than 10° in each direction, however, it is possible that the 5° left-right difference in total range of rotation could nonetheless be attributed to an imbalance in the neutral knee position, which is not clinically relevant within this range of secondary laxity.

The divergence in the rotation shift between the SB and DB groups during the post-operative testing session of the cutting activity was approximately 14° (Figure 6.12), well beyond the asymmetry observed in the healthy control subjects under passive loading. Combined with the relatively consistent range of rotation under dynamic loading conditions, this rotation shift becomes even more clinically prominent.

The hypothesis that the difference in joint restraint following SB or DB reconstruction is in their capacities to constrain combined loads, rather than simply rotational laxity, is also supported by the studies that have evaluated the two procedures using the pivot shift test. The evidence in the literature review was ambiguous when all studies with conclusions reporting measures of 'rotational' laxity were assessed. However, when segregating only those studies that used a pivot shift to evaluate joint laxity, the evidence was clearly in favour of the double-bundle procedure with Järvelä (2007); Kondo *et al.* (2008); Muneta *et al.* (2007); Siebold *et al.* (2008); Yagi *et al.* (2007) finding better outcome in the DB group, while only Streich *et al.* (2008) found no significant difference between the two techniques and Markolf *et al.* (2008b) demonstrated the potential for overcorrection of joint laxity with the DB reconstruction.

Due to the difficulty in quantitatively evaluating the pivot shift test, no *in vivo* study has quantitatively compared SB and DB reconstructions by the associated degree of rotation. It has moreover been demonstrated that the combined effects of valgus and internal rotational moments on ACL strain are greater than either load in isolation (Shin *et al.*, 2005). Therefore, the conclusion that the DB reconstructive technique provides superior 'rotational' stability due to better pivot shift outcome is inaccurate.

The illustration by Pearle *et al.* (2008) of the AM and PL bundle obliquity throughout the range of flexion (Figure 7.1) may shed some light on both the reason for the apparent improved constraint of the DB technique under combined loading situations, as well as the outcome of the isolated torsional load investigation. The bundles are parallel throughout the first 30° of flexion (Jordan *et al.*, 2007; Pearle *et al.*, 2008). Their contributions to the restraint of isolated torsional loading would, therefore, theoretically be comparable and may wholly account for the equal reduction of internal rotation in both the SB and DB reconstructions in the extended knee position (Chapter 5).

With greater knee flexion, the angles of the two bundles were found to change non-uniformly in the three anatomical planes (Jordan *et al.*, 2007; Pearle *et al.*, 2008). The more oblique orientation of the PL bundle in the transverse plane is used to explain its enhanced ability to restrain rotation (Blankevoort & Huiskes, 1996; Pearle *et al.*, 2008). However, the difference in AM-PL angle obliquity in the sagittal plane is more substantial than in the axial plane (Figure 7.1) and would theoretically favour an AM bundle graft reconstruction – rather than the PL bundle – in rotation restraint depending on the location of the axis of rotation. The more vertical orientation of the PL bundle in the coronal plane would furthermore better resist joint distraction or varus-valgus rotation.

The overconstraint of rotation that was observed in the DB group when a passive internal torque was applied to the knee in the flexed position would have

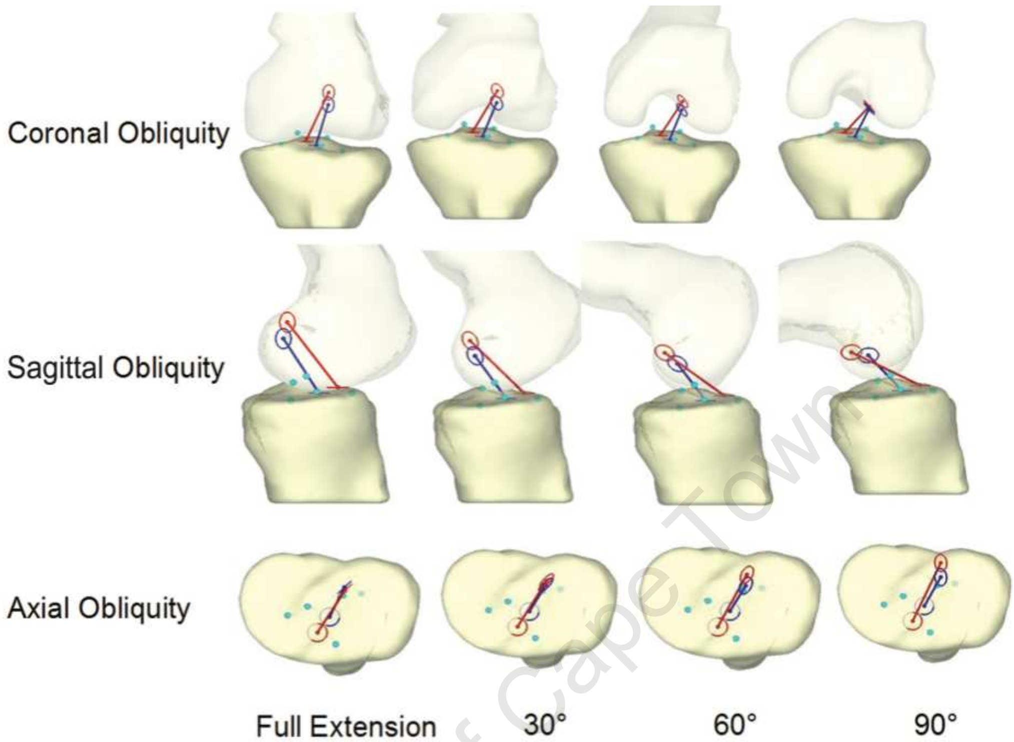


Figure 7.1: Anteromedial and posterolateral bundle obliquity in the three anatomical planes at four angles of knee flexion (Pearle *et al.*, 2008).

occurred at the point at which the AM-PL bundle orientation changed from parallel to oblique with respect to one another. Although the PL bundle loosens to a greater degree than the AM bundle throughout the first 30° of flexion, internal rotation of the tibia counteracts this slackening to a certain extent (Amis & Dawkins, 1991). The additional tension that would be provided by the PL bundle during internal torsional loading would actually have been in a more vertical direction when viewed in the sagittal or coronal planes. In other words, the addition of the PL bundle would essentially pull the tibia and femur closer together than the SB reconstruction could. The distal-proximal direction of force may then have permitted the congruency of the joint contact surfaces to further contribute to joint constraint at the higher angles of flexion experienced throughout the dynamic activities.

If the proposed theory that the improved overall joint constraint is a result of supplemental PL bundle tension normal to the tibial surface is in fact correct, it is still uncertain whether the tension in each graft bundle is similar to that of the native ACL bundles or moreover, whether the graft strain is at a healthy level. It is believed that the contribution of the native PL bundle to joint constraint is greatest at low flexion angles because this bundle slackens to a greater extent than the AM bundle with knee flexion (Markolf *et al.*, 2009; Yagi *et al.*, 2002). If reconstructed grafts are tensioned differently from their native counterparts, the effects on joint stability may appear to be similar under certain loading conditions, but actually have undesirable consequences. It is possible, for example, that excessive tensioning of the *anteromedial*, rather than the additional posterolateral bundle, may account for the observed trend towards overconstraint in the DB group under passive internal torque in the flexed position (Chapter 5), while the lack of vertically directed (distal-proximal) tension in the SB group could simultaneously account for the laxity under dynamic loading conditions (Chapter 6).

Most studies to date have measured ligament strain as an indicator of tension *in vivo* without knowing the initial recruitment length or actual ligament tension. Although equivalent force was employed in both SB and DB surgical techniques to pull the grafts taut, the direct effect of the order of bundle fixation and angles at which the tension was applied on overall graft tension is unknown. Furthermore, after several months of healing and physiological strain, tension in the individual bundles may have changed. In the absence of the ability to measure the force of each native and reconstructed graft bundle directly, it can only be theorized that extreme tensioning of one or both bundles during the DB technique brought about the reduced rotation relative to the contralateral knee under isolated torque in the flexed knee position.

7.2 Research implications and recommendations for the future

The findings presented in this thesis specifically concern the capacity of two techniques of ACL reconstruction to restore *rotational* restraint to the joint *in vivo*. The broader questions are *'Does the double-bundle technique restrict joint laxity* *to a greater extent than the single-bundle technique? If so, how?'* The results from our dynamic study suggest that the DB technique does, in fact, constrain the joint more adequately than the SB reconstruction, allowing patients to apply normal knee kinematics during physiological weightbearing tasks. However, the outcome from the passive torsional loading study indicates that the superior restraint provided by the DB technique is not primarily in opposition to axial rotation, but to another direction of rotation or translation. We have therefore begun to answer the question of *'how?'* by finding that it is *not* as simple as stating that the DB reconstruction provides superior rotational restraint.

Fortunately, we have the means to go back to the drawing board to find the answer to the question of how the DB technique may improve patient outcome after ACL reconstruction. The MRI-compatible loading device has been designed to permit relatively simple modifications in order to apply static loading about or along another axis of rotation/translation. The methodology by which six degree-of-freedom motion can be analysed will facilitate investigations into motion in any anatomical plane under varying loading conditions.

The three-dimensional analysis under torsional loading in our healthy subject group has already provided valuable information with respect to coupled joint motion (Chapter 4); for example, the increase in flexion accompanying external rotation demonstrated a contrasting paired movement under torsional loading from the typical screw-home motion observed with flexion-extension. Therefore, there are either independent or additional structures that control knee kinematics under these different loading conditions, or those structures that contribute to motion restraint behave differently with changes in loading direction. Once these differences are correctly interpreted, the changes in three-dimensional joint kinematics due to knee pathology such as ACL rupture may be more effectively measured. Ultimately, a better understanding of the six degree-of-freedom joint motion will allow surgeons to better predict and assess modifications made to existing surgical reconstructive techniques.

Not only is the measurement of joint motion becoming more accurate with the ability to track tibiofemoral movement in all three anatomical planes, but the number of journal articles describing aspects of surgical techniques, such as the precise three-dimensional location and orientation of the graft bundles, are also continuing to grow (Agneskirchner *et al.*, 2004; Doi *et al.*, 2009; Luites *et al.*, 2007; Zantop *et al.*, 2007). The need for revision surgery has primarily been attributed to incorrect tibial or femoral bone tunnel placement (Crawford *et al.*, 2007; Harner & Poehling, 2004); enhanced accounts of proper surgical technique may, therefore, already improve the outcome of ACL reconstruction without resorting to the latest unsubstantiated surgical trend. For less experienced surgeons, it is probably more valuable to accurately perform a simpler procedure rather than to attempt a more complex technique that may eventually require revision if the initial surgery is poorly executed.

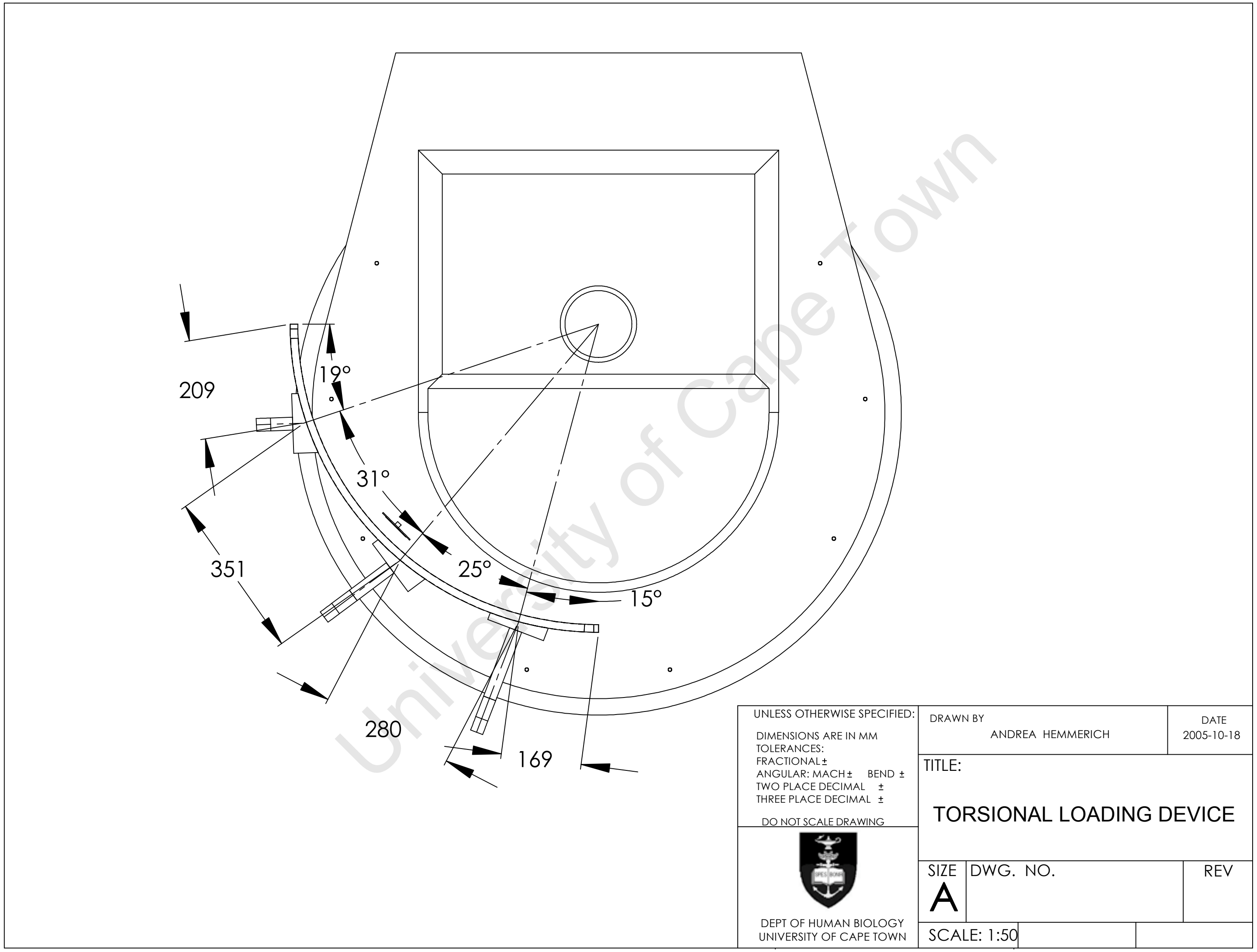
While our dynamic activity randomised control trial did demonstrate kinematics closer to normal in the DB group, the findings examined together with those of our passive torsional loading investigation indicate that the improved constraint is not simply about the tibial axis of rotation. Additionally, the disproportionate reduction in internal rotation in the DB group in the flexed, isolated torque condition may indicate an overconstraint due to excessive graft tensioning. Further research is required into the exact mechanism by which the DB technique may provide superior constraint, as well as the long-term effects this will have at the joint and on the reconstructed graft. Studies using inverse dynamics and electromyography to further investigate the contributions of the muscles on knee function will also add to our understanding of the benefits and detriments that could arise after either type of surgical reconstruction.

It is presumed that kinematics resembling the healthy knee joint, such as those measured in the DB group under dynamic loading conditions, will precipitate fewer long-term complications. However, it has been shown that joint degeneration following ACL reconstruction can be more extensive than in the ACL-deficient knee (Fu *et al.*, 2000; Kessler *et al.*, 2008). It is accordingly possible that a similar counterintuitive finding may be demonstrated with SB and DB reconstructions. Long-term follow-up studies comparing not just the SB and DB techniques, but also the effects of varying tunnel placement, graft tension, concomitant soft tissue injury, and any other factors that influence joint constraint should be undertaken to determine whether a specific technique is associated with a lower incidence of osteoarthritis or disabling joint disease.

This thesis presents new insights regarding the biomechanics of two techniques of ACL reconstruction using either a single or double-bundle graft. The different outcomes of the passive and dynamic loading studies demonstrate that conclusions about the ability of these reconstruction techniques to reduce rotational knee laxity cannot be based on one test alone. While the patients who received the double-bundle reconstruction showed improved performance compared with those who were allocated the single-bundle surgery, further investigation is required to determine whether the evidence of overconstraint of rotation may be detrimental to joint function in the future.

Appendix A

Technical drawings used for manufacturing main components of MRI-compatible torsional loading device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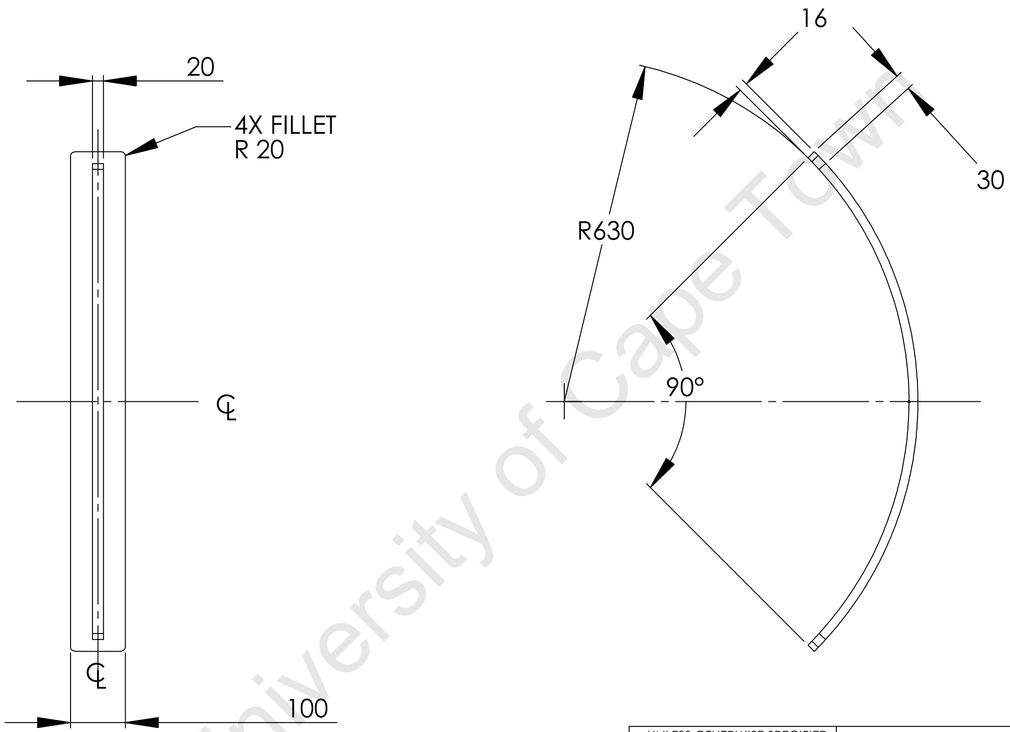

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DEPT OF HUMAN BIOLOGY
UNIVERSITY OF CAPE TOWN

DRAWN BY ANDREA HEMMERICH

DATE 2005-09-07

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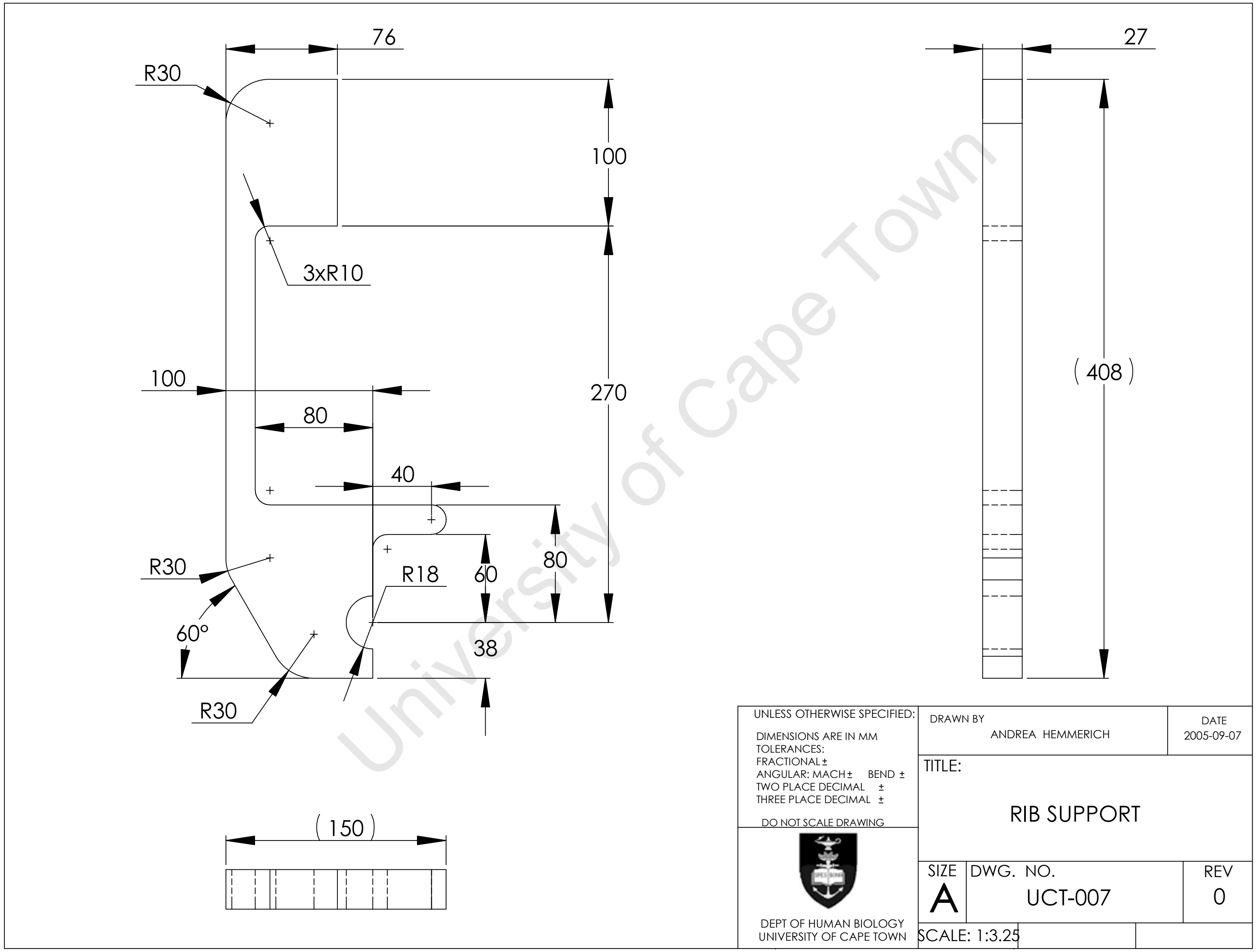
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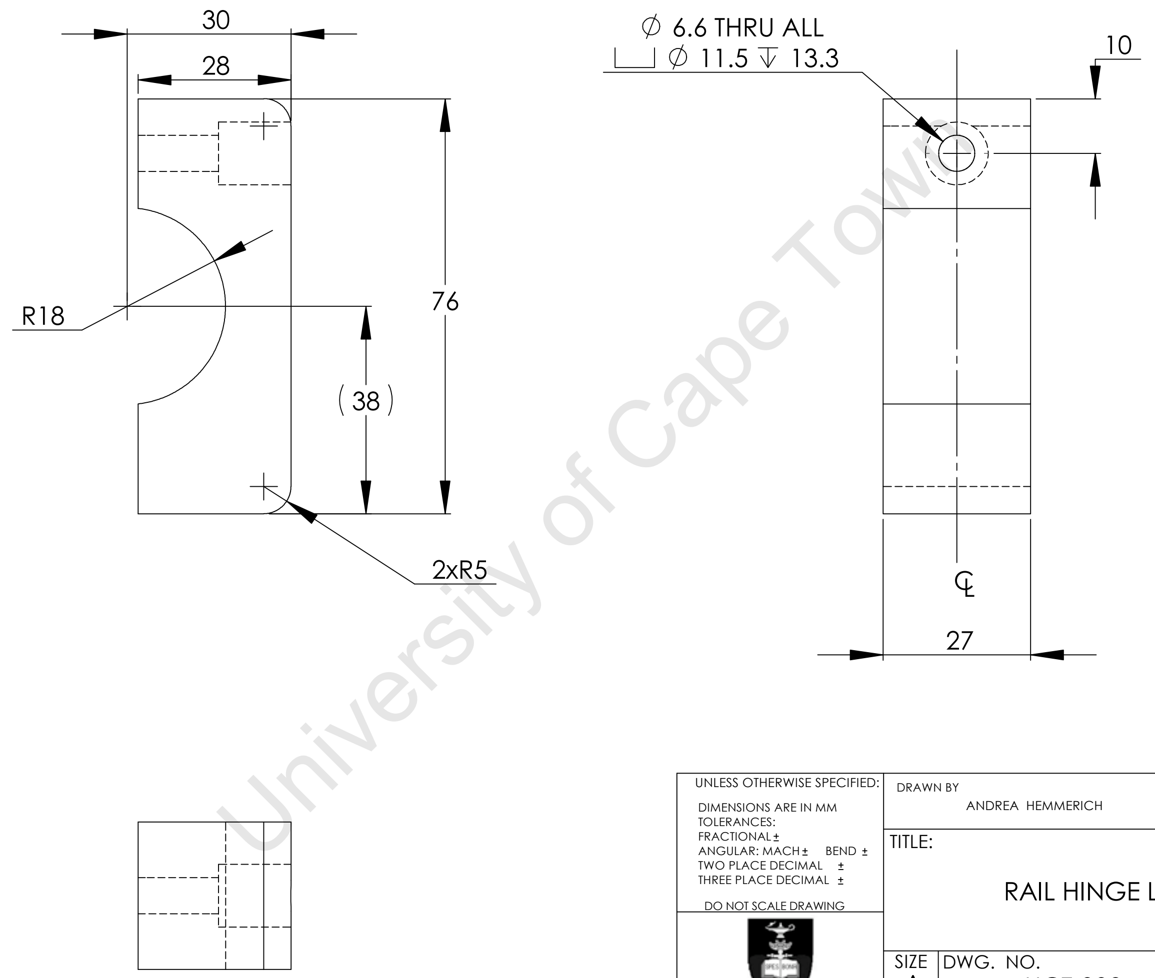

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FRACTIONAL ±
ANGULAR: MACH ± BEND ±
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DO NOT SCALE DRAWING

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DRAWN BY ANDREA HEMMERICH

DATE 2005-09-07

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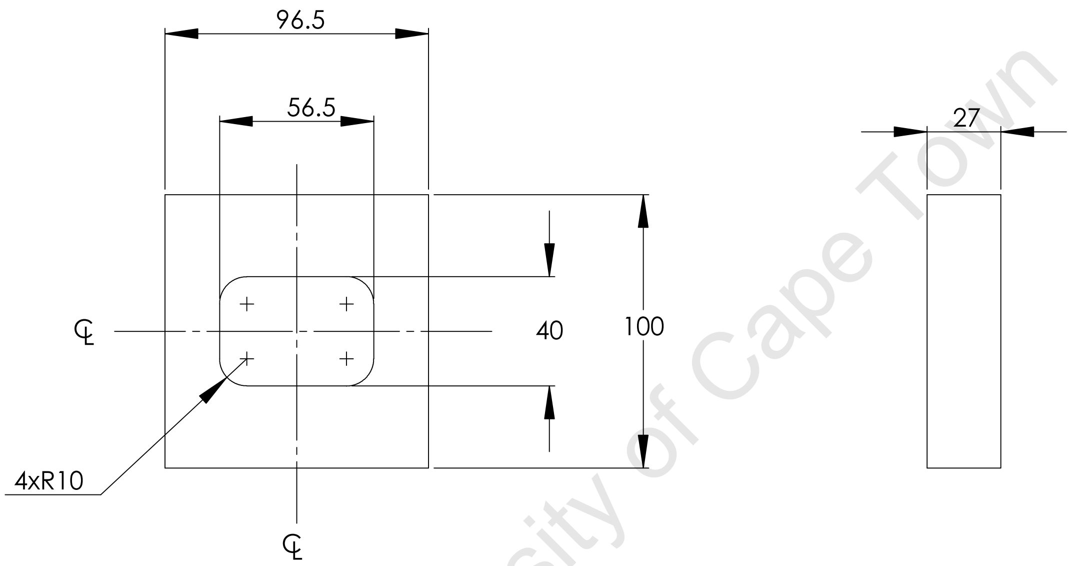

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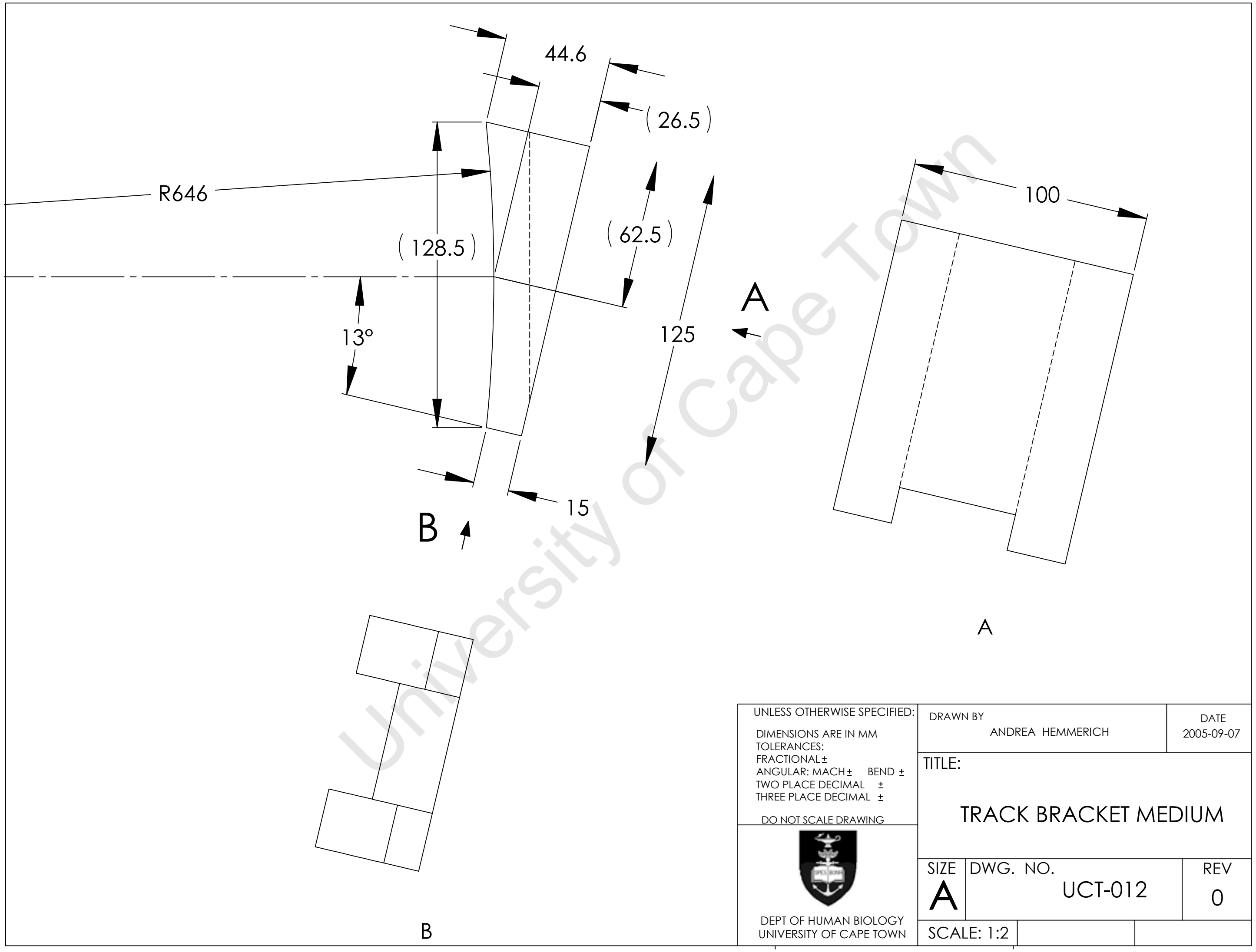
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ANDREA HEMMERICH
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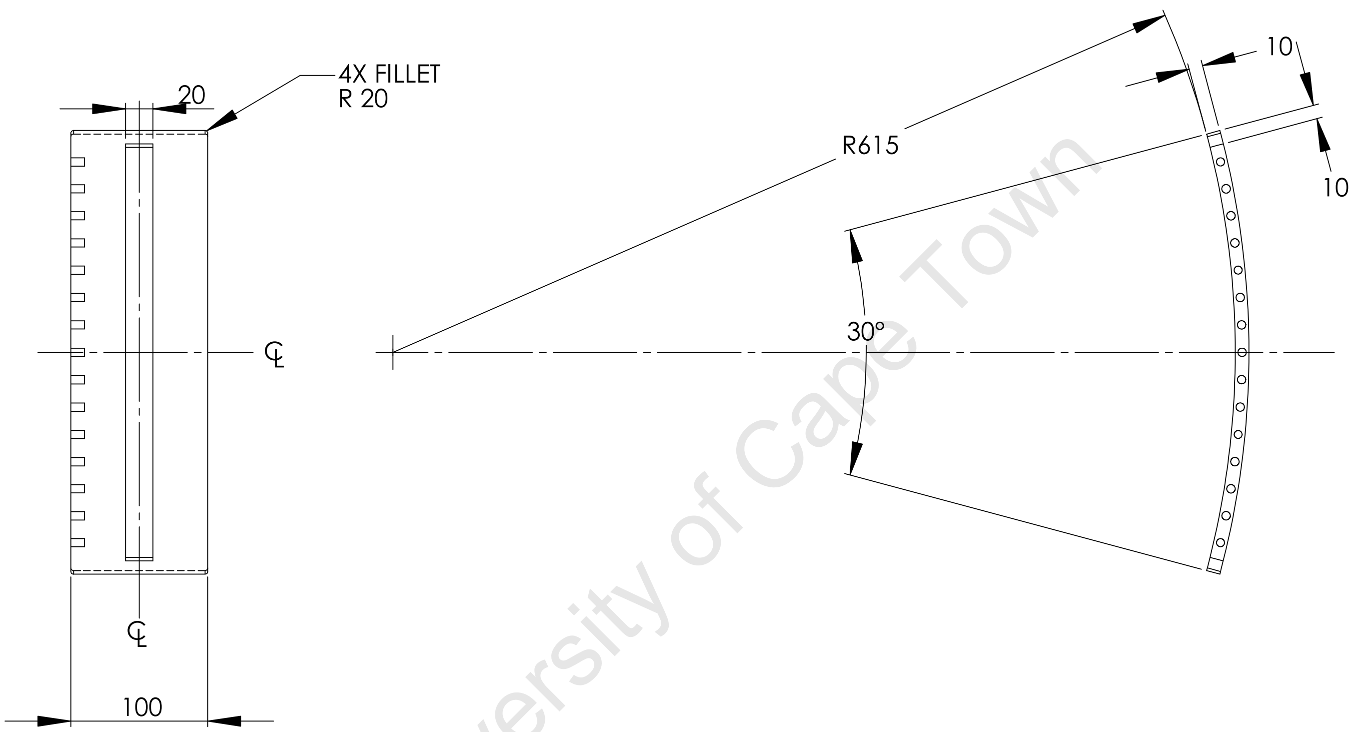

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DRAWN BY ANDREA HEMMERICH

DATE 2005-10-18

TITLE:

AB-ADDUCTION TRACK

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A	UCT-016	0

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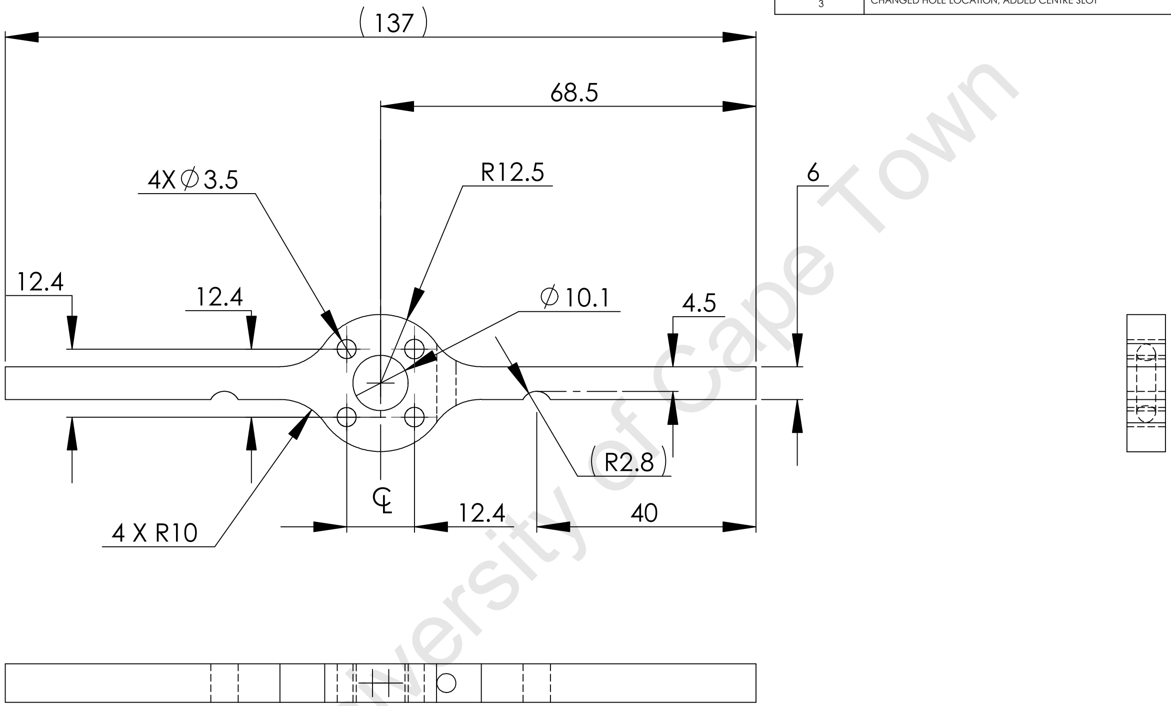
2	DECREASED NOTCH DEPTH; INCREASED PART THICKNESS TO 6 MM; INCREASE FILLET RADIUS	12-04-2006
3	CHANGED HOLE LOCATION; ADDED CENTRE SLOT	06-06-2007



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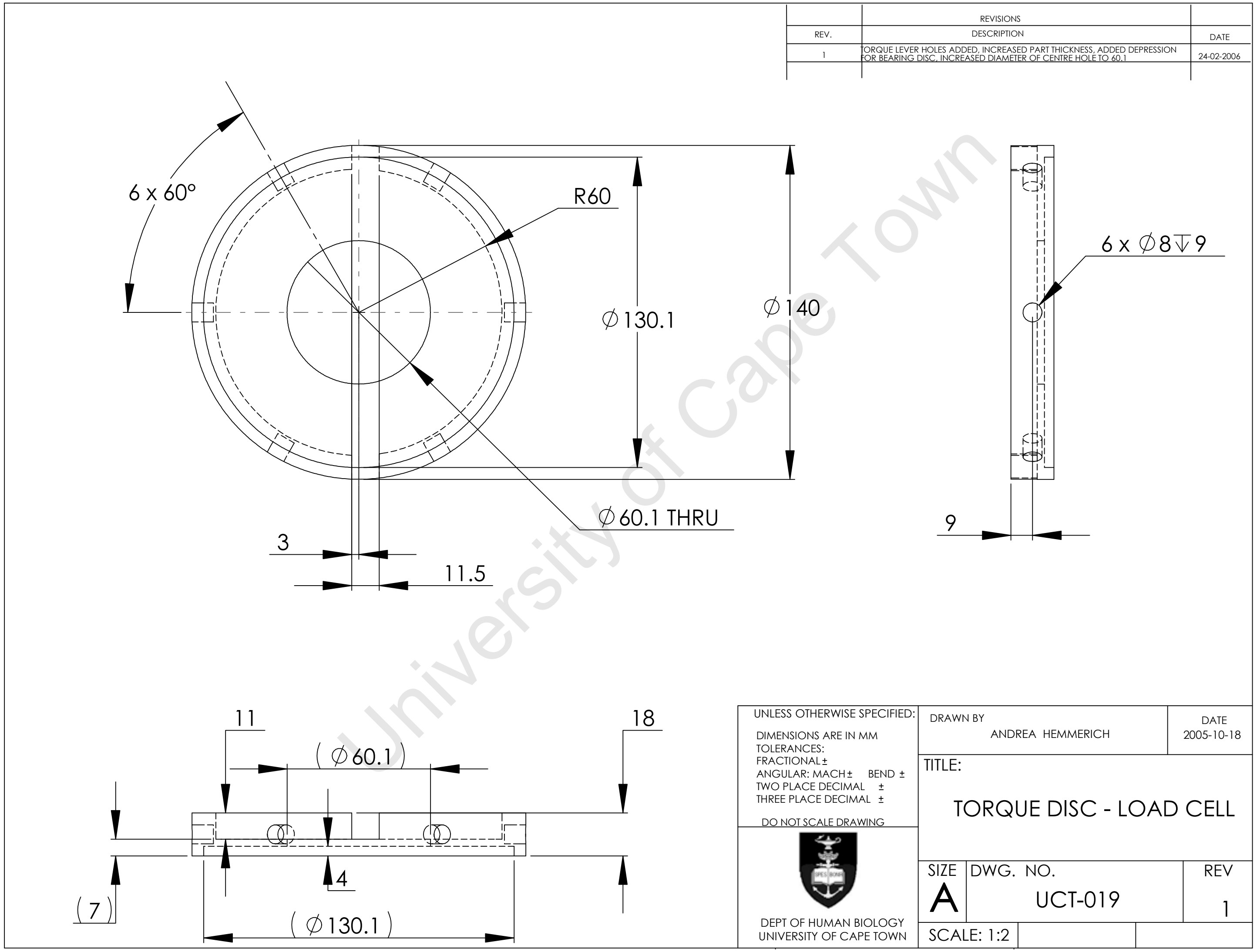
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DESCRIPTION
DATE
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24-02-2006
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DEPT OF HUMAN BIOLOGY
UNIVERSITY OF CAPE TOWN
DRAWN BY
ANDREA HEMMERICH
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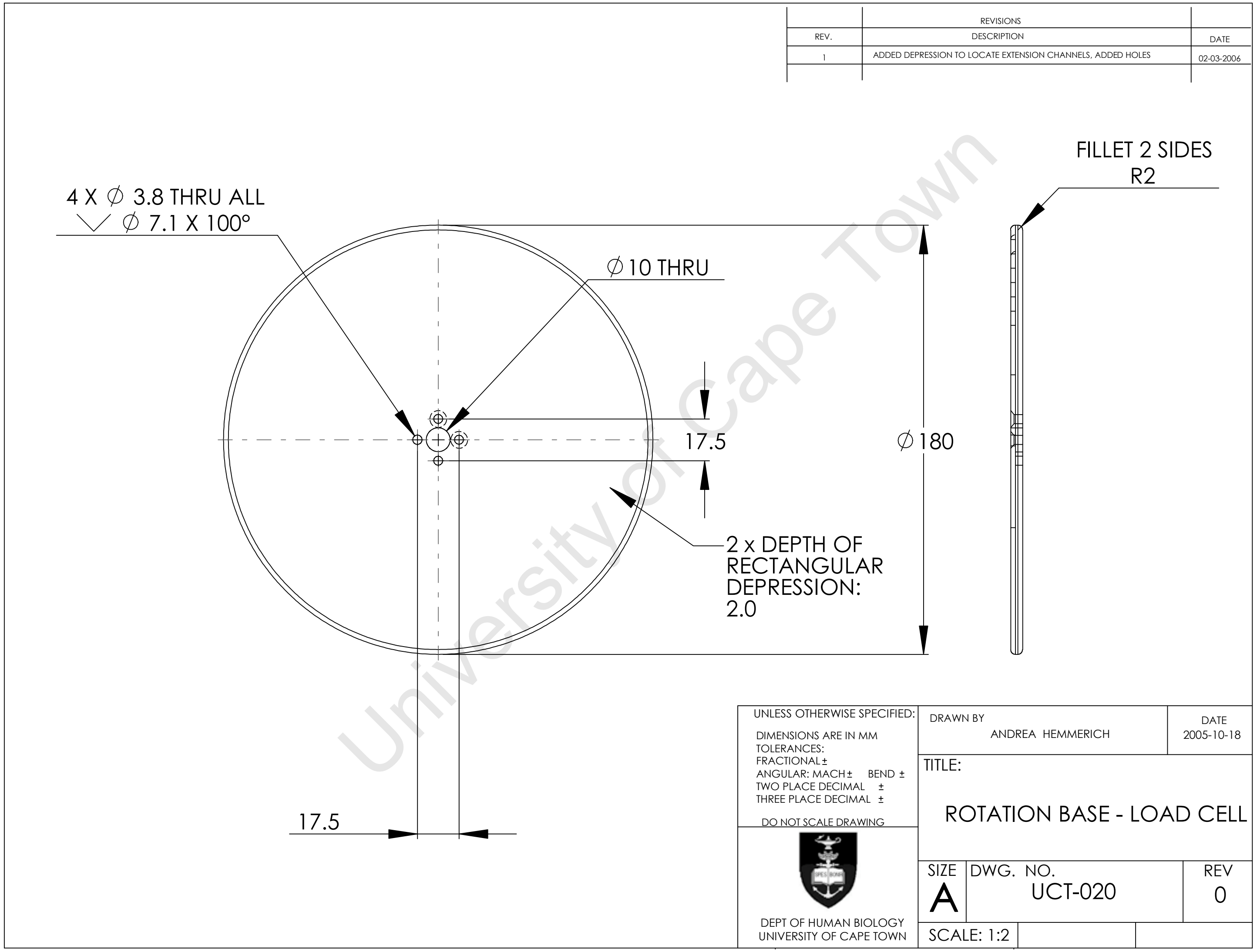
REVISIONS
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ANDREA HEMMERICH
DATE
2005-10-18
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ROTATION BASE - LOAD CELL
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UNIVERSITY OF CAPE TOWN
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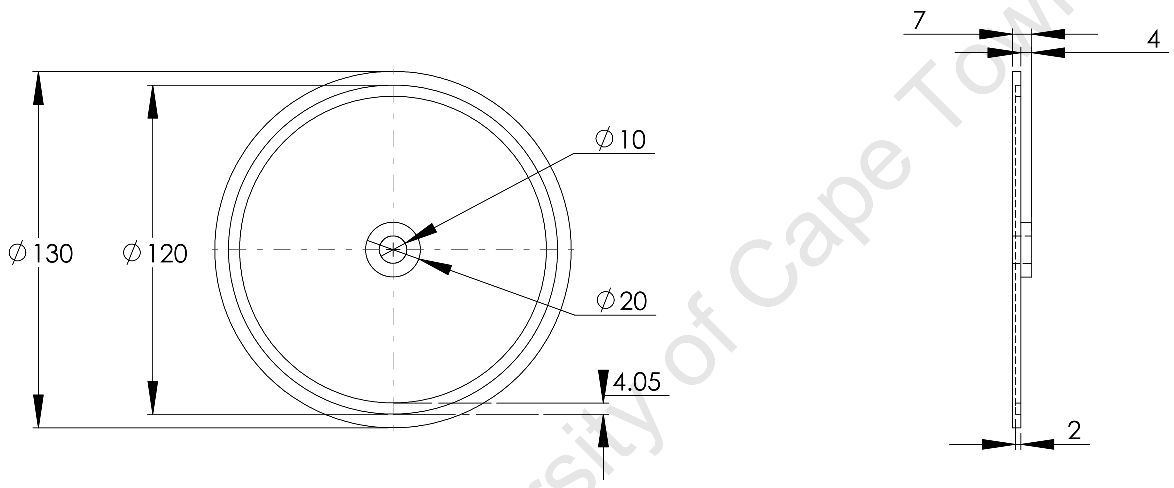

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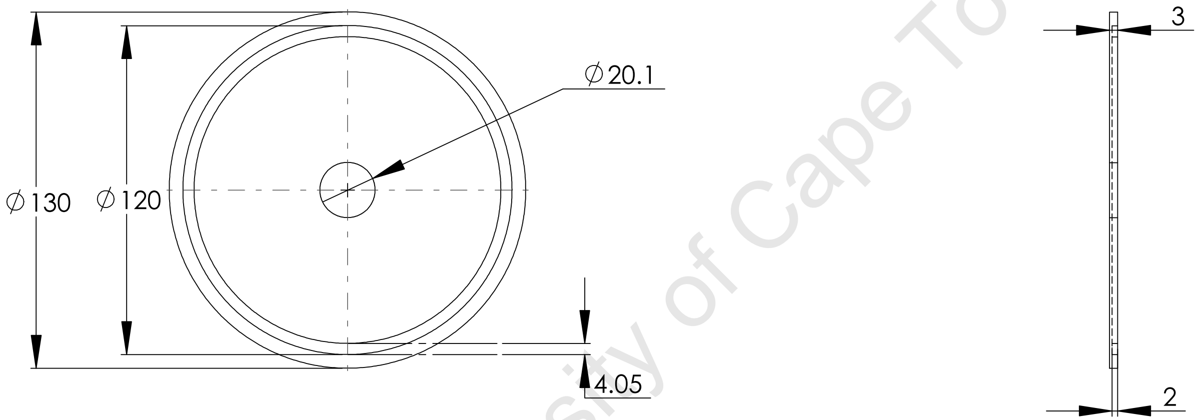

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ANDREA HEMMERICH

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2006-02-24

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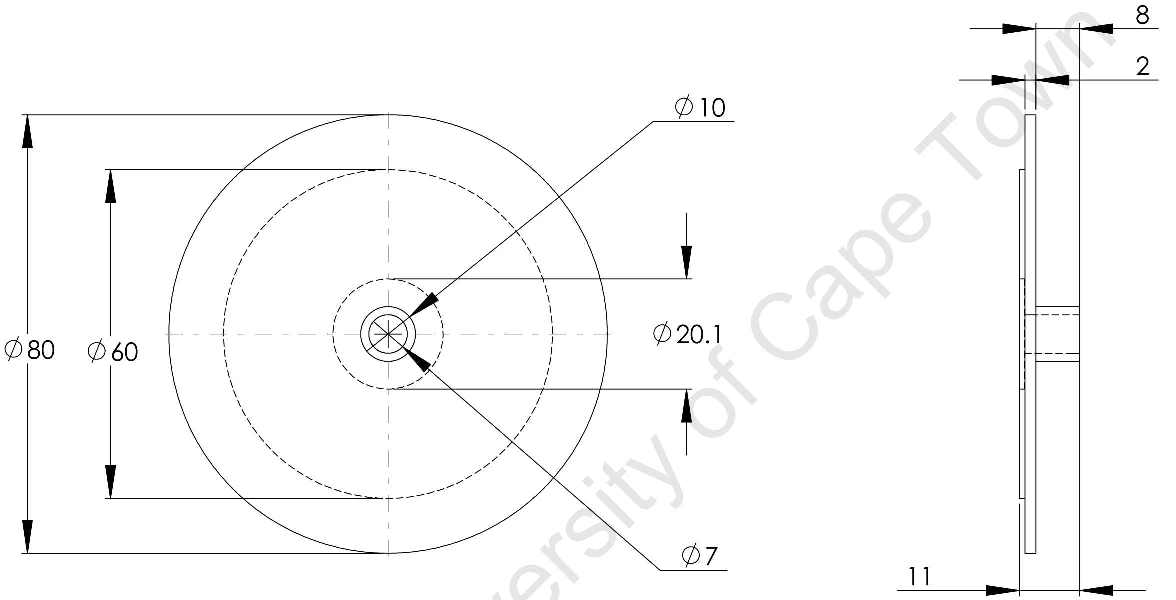

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DATE 2006-02-24

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COMPRESSION PLATE

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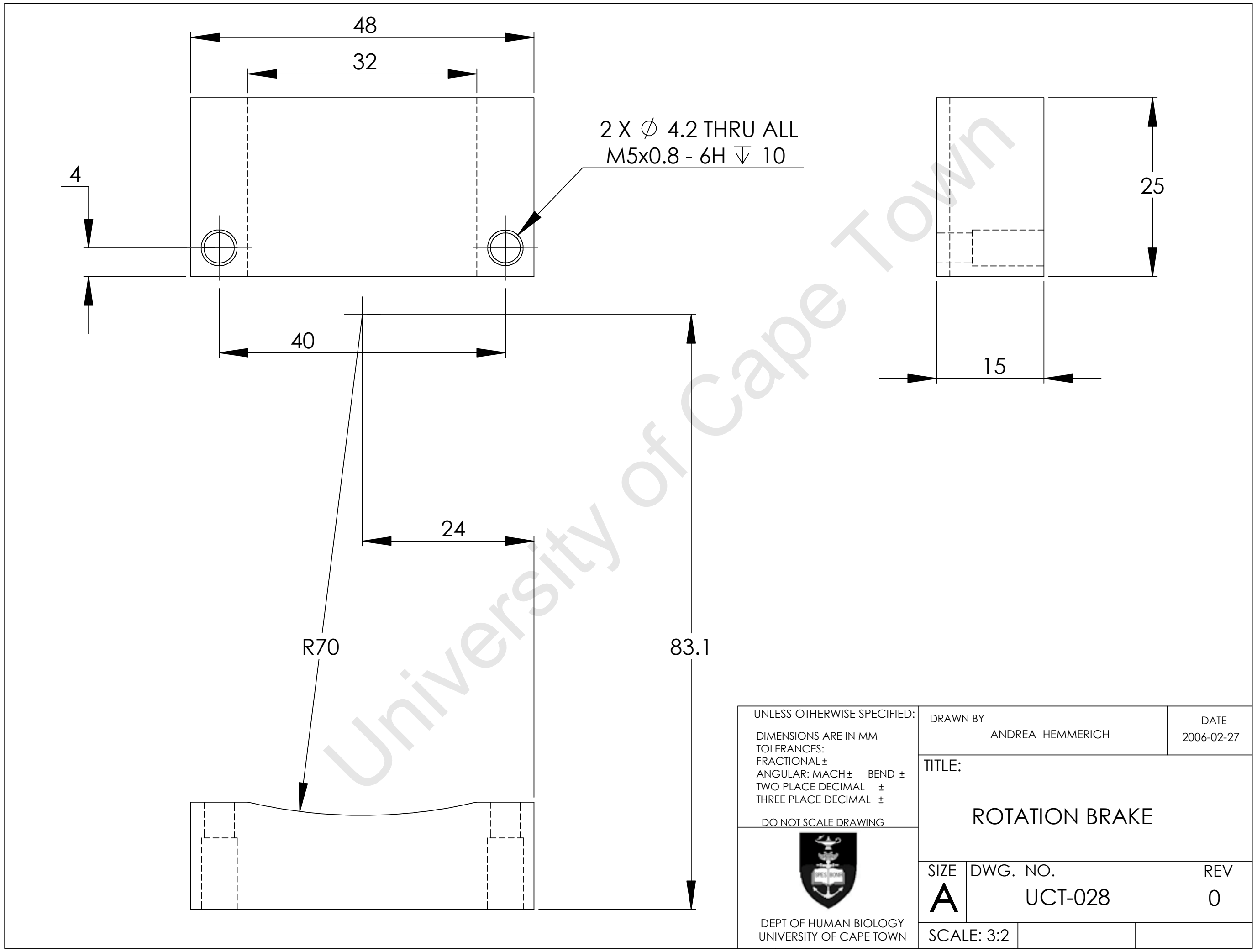
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UNIVERSITY OF CAPE TOWN
DRAWN BY
ANDREA HEMMERICH
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Appendix B

Details of the strain gauge circuit used to measure torque

A strain gauge bridge circuit, as shown in Figure B.1, was used to measure the torque applied to the load cell. With rotation of the torque disc, a normal force was exerted at the end of the cantilevered load cell, thereby applying tension to the strain gauge and changing its resistance. The response to the mechanical strain was indicated by the voltmeter. A calibration procedure in which fixed weights applied known loads to the strain gauge, allowed the resulting voltage to be linearly correlated with the torque in LabVIEW^{TM}.

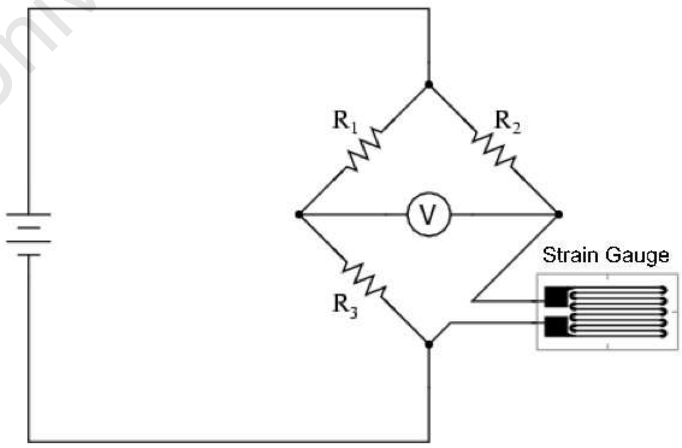

Figure B.1: A quarter-bridge strain gauge circuit was used to measure torque applied to the load cell.

While a half-bridge circuit (using two strain gauges and two resistors) is more typically used in this type of application to compensate for temperature change or other sources of resistance-induced error, a quarter-bridge circuit was considered sufficient for the purposes of this study since the LabVIEW^{TM} software had the capability of 'zeroing' the measured torque before the load was applied. Since the voltage-torque relationship was linear, the initial torque reading that was subtracted by the software did not affect the final measurement.

Appendix C

Torque relaxation measured at fixed rotation angle for one subject

Stress relaxation is used to describe the behaviour of viscoelastic materials such as the ligaments of the knee when subjected to a constant load over time (Woo *et al.*, 2006). In our study, a torque applied manually to the distal end of the shank was kept constant by locking the rotated position of the foot once the specified torque was reached. Figure C.1 shows the change of measured torque as the load was applied and then adjusted to meet the desired magnitude of 4.25 Nm calculated from equation 3.1 for a 60 kg subject. The specified torque was attained at approximately 100 seconds, at which point the rotation brake was applied; a spike in the signal at approximately 120 seconds was likely due to a muscle spasm and the subject adjusting to the applied load. The torsional load measured by the strain gauge was then recorded for an additional 7 minutes before the brake at the boot was released.

The regression curve fit to the data over the 7-minute period of constant loading resembles a typical stress-relaxation curve measured for joint ligaments (Woo *et al.*, 2006). In our experimental setup, the subject's thigh was free to move; consequently, the tibia could rotate with respect to the femur and the femur could move with respect to the pelvis. The observed relaxation, therefore, may reasonably be attributed to the soft tissues at both the knee and hip joints.

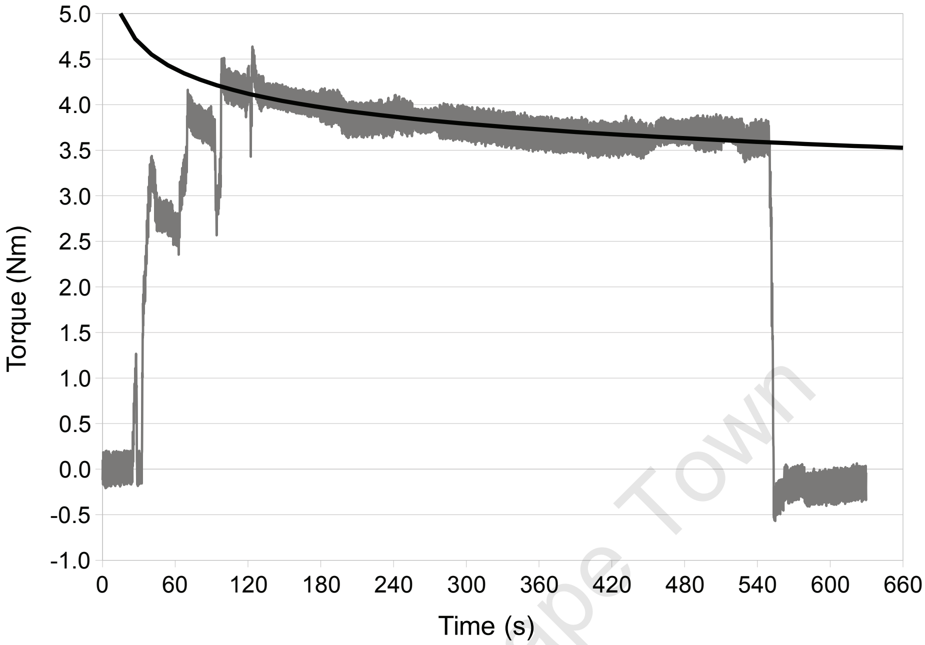

Figure C.1: Torque measured with manual adjustment of initial load and following fixation of rotation brake (shown in grey). Black regression curve displays general trend of time-torque curve, believed to be primarily a result of stress relaxation of the soft tissues and ligaments at the knee and hip joints.

The greatest decrease in measured torque occurred within the first 120 seconds of constant loading. Waiting approximately two minutes after applying the torque not only permitted relaxation of the soft tissues around the joints, but also allowed the subject to become accustomed to the load, thereby minimising the possibility of image artefact resulting from muscle activation and movement. Once sufficient stabilization of the measured torque value was achieved, the correct torque could be reapplied without significant reduction in load because the soft tissues had been 'pre-strained.' For this reason, the two-minute 'relaxation' period was incorporated into the protocol before MR imaging was performed.

Appendix D

Ethics approval letter and subject informed consent form

UNIVERSITY OF CAPE TOWN

Health Sciences Faculty
Research Ethics Committee
Room E53-24 Groote Schuur Hospital Old Main Building
Observatory 7925
Telephone [021] 406 6338 • **Facsimile** [021] 406 6411
e-mail: preaward@curie.uct.ac.za

15 November 2005

REC REF: 392/2005

Prof CL Vaughan
Human Biology

Dear Prof Vaughan

PROJECT TITLE: THE EFFECTS OF THE INTACT DEFICIENT, AND RECONSTRUCTED ANTERIOR CRUCIATE LIGAMENT ON ROTATIONAL STABILITY OF THE KNEE JOINT

Thank you for submitting your study to the Research Ethics Committee for review.

It is a pleasure to inform you that the Ethics Committee has **formally approved** the above-mentioned study on the 11 November 2005.

Please quote the REC. REF in all your correspondence.

Yours sincerely

PROFESSOR T. ZABOW
CHAIRPERSON, HSF HUMAN ETHICS

lemjedi

Informed Consent

I, ________________________________, agree voluntarily to participate in the research project titled "The Effects of the Intact, Deficient, and Reconstructed Anterior Cruciate Ligament on Rotational Stability of the Knee Joint" conducted at the Department of Human Biology of the University of Cape Town. The data for this project will be collected at the Sports Science Institute of South Africa (Cape Town).

The following procedures and concepts have been explained to me in full:

I. Measurement of Knee Instability using Magnetic Resonance Imaging (MRI) Compatible Device

1. The lower limb will be placed in a neutral position (full extension with no rotation). A high resolution knee scan lasting just over ten minutes will be taken in this position.
2. The lower limb will be positioned passively at a specific knee flexion angle within the magnetic resonance imaging (MRI) coil. A set torque (rotational stress) will be applied to the foot with the aid of a specially designed stress-testing device. The torque will be applied gradually by the investigator and can be stopped if any discomfort should occur.
3. The lower limb will be held in place for a period of approximately three minutes by the stress-testing device while an MRI scan is performed.
4. This procedure will be repeated for 4 rotated positions for each limb (i.e. 8 MRI scans in total).

II. Gait Analysis

5. Several retro-reflective markers will be placed on the thigh and shank of each leg.
6. You will walk with your customary gait along a 10 metre walkway.
7. During walking, your gait will be captured by a six-camera motion analysis system and ground reaction force plate.

III. ACL Reconstructive Surgery

There are two commonly performed surgical procedures to repair the ruptured anterior cruciate ligament (ACL), usually referred to as the "single-bundle" and "double-bundle" techniques. At present, no clear evidence exists to suggest that one technique has better

Knee Torque Study – Informed Consent page 1 of 3

results than the other; orthopaedic surgeons worldwide tend to use the procedure with which they are most comfortable and have the most experience. Dr. van der Merwe has over 10 years of experience performing both types of surgery on his patients. For the current study you will be randomly selected to have either one of the surgical techniques performed by Dr. van der Merwe. If he feels that there is a clear indication one way or the other that you should have a specific surgical procedure performed in order to best treat your injury, he will do so and you will no longer be a part of the study.

Procedures I and II will be performed twice over the course of the study: once prior to surgery and once about three months after the surgery (following sufficient time for rehabilitation).

Benefits/Risks

Risks associated with this study do not exceed those associated with normal clinical assessment by the patients' orthopaedic surgeon, Dr. Willem van der Merwe, or that of normal walking.

The information obtained in this study may or may not be of direct use to you. However, it will provide important information concerning the motion of the knee joint and may be of direct importance in the future for the improvement of anterior cruciate ligament reconstruction techniques. You would benefit directly from this information if you required an ACL reconstruction.

Voluntary Participation

Participation in this study is voluntary. You may withdraw at any time without prejudice to you in any way. You may also decline to participate without any negative repercussions.

Confidentiality

All information obtained during this study is confidential. The information obtained will only be available to the investigators involved in the study. The identity of subjects will not be disclosed in any published findings of the study.

Copy

You will be given a copy of this signed Consent Form for your own information.

Contact People

If you have any questions or concerns about the study, you may contact the following people.

Principal Investigator:
Prof. Kit Vaughan, Faculty of Health Sciences, University of Cape Town 021 406 6238

Co-investigator:
Andrea Hemmerich, Faculty of Health Sciences, University of Cape Town 021 406 6549

Co-investigator:
Dr. Willem van der Merwe, Sports Science Orthopaedics Clinic 021 686 1196

Chair, Ethics Committee:
Dr. Lesley Henley, Faculty of Health Sciences, University of Cape Town 021 658 5304

I have read the preceding form and understand the testing procedures outlined therein. I understand any accompanying risks and discomforts. Knowing these risks and discomforts and having had the opportunity to pose questions answered to my satisfaction, I hereby consent to participate in this study. I understand that I may withdraw from this study at any time without further question. I have been informed that the individual data derived from my participation in these protocols will remain confidential.

Signature of Subject:_________________________ **Date:__________**

Signature of Investigator:______________________ **Date:__________**

Knee Torque Study – Informed Consent page 3 of 3

Appendix E

Subject level and passive knee rotation data for 15 healthy Control subjects

Table E.1: Subject level data and range of rotation in extended and flexed knee positions for 15 Control subjects.

Subj	Sex	Age	Height	Mass	Torque	Range of Rotation			
						Knee Extended		Knee Flexed 30°	
		(yrs)	(cm)	(kg)	(Nm)	Left	Right	Left	Right
1	M	31.8	174	73	4.9	17.1	25.2	23.1	21.0
2	M	29.3	182	75	5.0	20.0	18.8	18.0	20.0
3	F	29.7	157	54	4.0	19.3	20.2	25.4	24.0
4	M	23.8	175	62	4.4	11.7	9.0	36.5	33.1
5	M	25.9	171	75	5.0	7.5	8.5	19.9	12.5
6	F	37.6	170	62	4.4	10.0	9.0	21.8	23.3
7	M	38.0	170	79	5.2	22.4	17.4	24.4	25.7
8	F	29.8	166	64	4.5	21.1	21.9	29.4	30.3
9	F	31.5	158	50	3.8	16.3	12.5	27.4	29.2
10	M	34.0	183	90	5.8	16.5	9.2	27.4	22.0
11	M	25.8	181	76	5.1	19.5	14.3	27.5	24.5
12	M	22.2	181	85	5.5	4.7	7.6	21.8	20.3
13	M	43.0	176	70	4.8	19.6	17.5	28.5	25.8
14	M	24.1	183	80	5.3	15.8	12.8	35.0	22.9
15	M	28.1	190	80	5.3	10.3	10.3	18.2	16.8
mean		30.3	174.5	71.7	4.8	15.8	14.6	26.2	23.9
SD		5.9	9.3	11.3	0.6	5.4	5.6	5.5	5.2

Appendix F

Primary and secondary outcome results for randomised control trial under passive torsional loading

Table F.1: Means, standard deviations, and p-values for measured rotation (in degrees) of subject groups in the primary outcome analyses for the passive rotational laxity study.

						p-value	
Loading Condition	**Analysis**	**Subject Group**	N	mean	SD	Mixed Model	Paired t-test
Extended Ext Torque	Test Session	PreOp	27	9.5	3.5	0.712	
		PostOp	28	9.5	3.8		
	Interaction	SB PreOp	14	9.4	2.9	0.930	
	Test Session	SB PostOp	16	9.4	3.3		
	by	DB PreOp	13	9.6	4.2		
	Surg Technique	DB PostOp	12	9.6	4.6		
Extended Int Torque	Test Session	PreOp	28	-8.8	3.5	**0.028**	
		PostOp	29	-6.9	3.1		
	Interaction	SB PreOp	14	-8.9	4.0	0.924	
	Test Session	SB PostOp	17	-6.6	2.7		
	by	DB PreOp	14	-8.8	3.0		
	Surg Technique	DB PostOp	12	-7.4	3.6		
Flexed Ext Torque	Test Session	PreOp	27	10.0	4.8	0.535	
		PostOp	28	10.2	4.2		
	Interaction	SB PreOp	13	10.5	3.7	0.841	
	Test Session	SB PostOp	16	10.0	3.5		
	by	DB PreOp	14	9.6	5.8		
	Surg Technique	DB PostOp	12	10.4	5.2		
Flexed Int Torque	Test Session	PreOp	30	-14.0	4.3	0.032	
		PostOp	29	-12.3	6.2		
	Interaction	SB PreOp	16	-12.9	3.9	**0.012**	0.793
	Test Session	SB PostOp	17	-13.4	6.0		
	by	DB PreOp	14	-15.3	4.4		**0.002**
	Surg Technique	DB PostOp	12	-10.8	6.4		

Table F.2: Means, standard deviations, and p-values for measured rotation (in degrees) of subject groups in the secondary outcome analyses for the passive rotational laxity study.

Loading Condition	**Analysis**	**Subject Group**	N	mean	SD	**p-value** Mixed Model	**p-value** Paired t-test
Extended Ext Torque	Healthy-Uninjured	Control Average	15	9.0	2.8	0.635	n/a
		Patient Contralateral	29	9.6	3.9		
	Injured-Uninjured	Patient ACL-deficient	27	9.5	3.5	0.553	
Extended Int Torque	Healthy-Uninjured	Control Average	15	-5.9	2.9	0.551	n/a
		Patient Contralateral	28	-6.4	2.9		
	Injured-Uninjured	Patient ACL-deficient	27	-8.8	3.5	**0.054**	**0.011**
Flexed Ext Torque	Healthy-Uninjured	Control Average	15	9.4	4.3	0.756	n/a
		Patient Contralateral	27	8.9	4.5		
	Injured-Uninjured	Patient ACL-deficient	28	10.0	4.8	0.572	
Flexed Int Torque	Healthy-Uninjured	Control Average	15	-15.2	4.7	0.645	n/a
		Patient Contralateral	29	-14.4	5.3		
	Injured-Uninjured	Patient ACL-deficient	29	-14.0	4.3	0.588	

Appendix G

Additional dynamic activity data

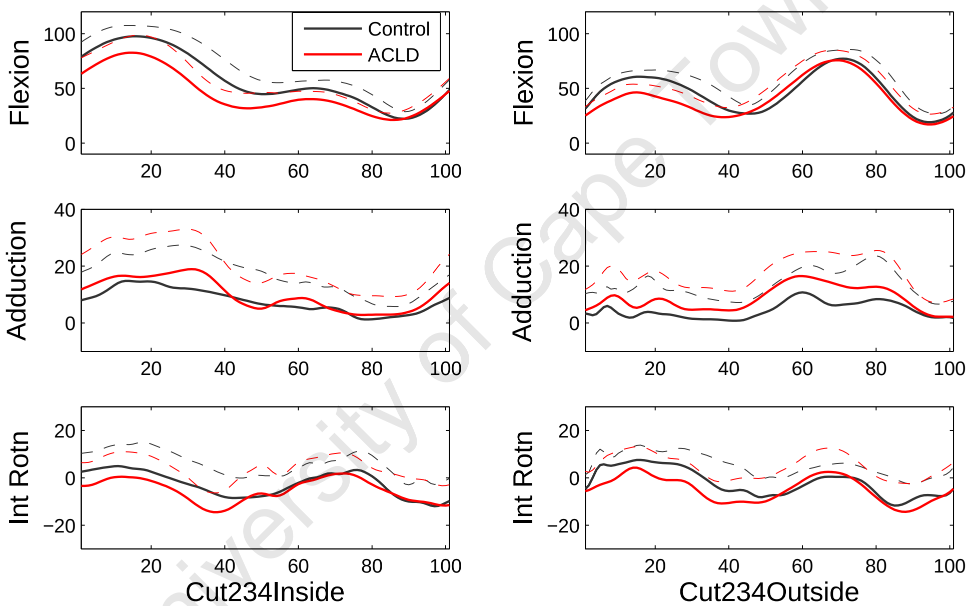

Figure G.1: Cut234Inside and Cut234Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for Control and ACL-deficient (pre-operative) knee groups.

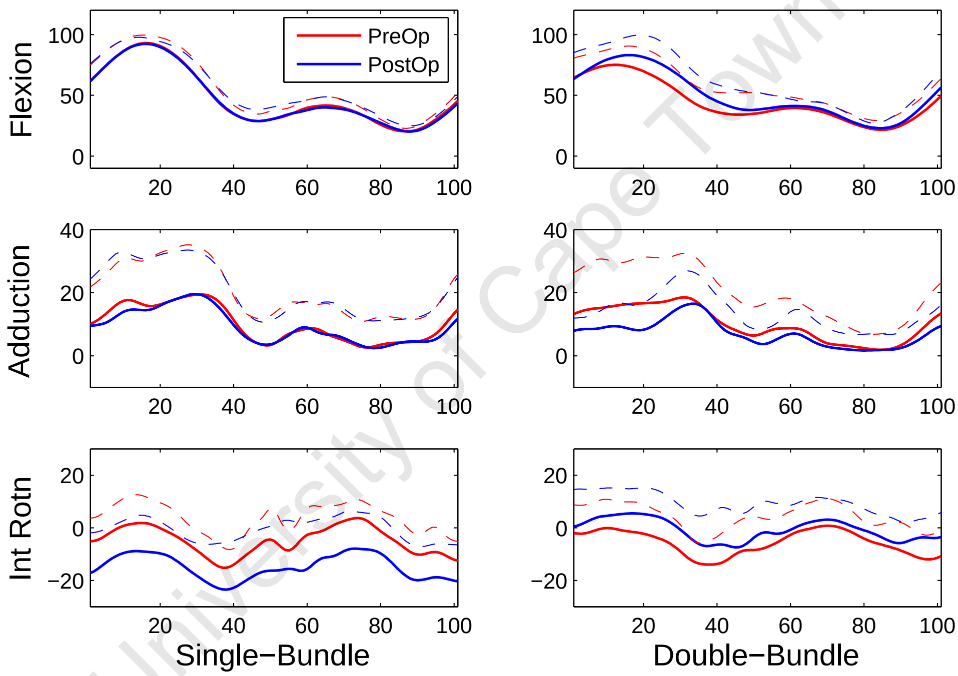

Figure G.2: Cut234Inside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

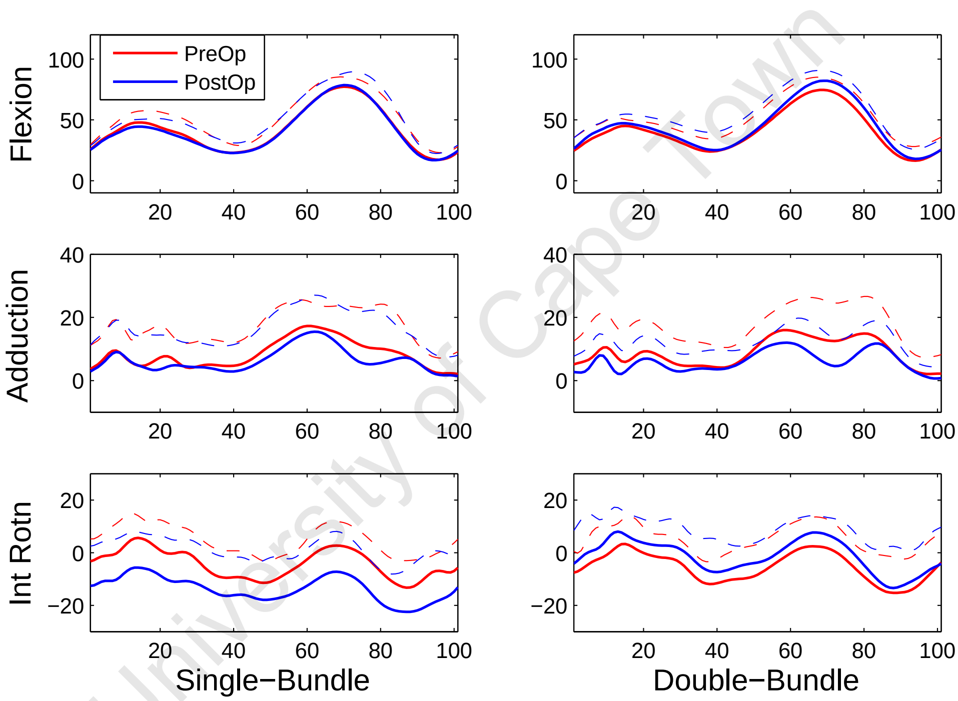

Figure G.3: Cut234Outside three-dimensional mean joint angles in degrees (solid) plus 1 standard deviation (dashed) over 100% gait cycle for SB and DB groups both pre- and post-operatively.

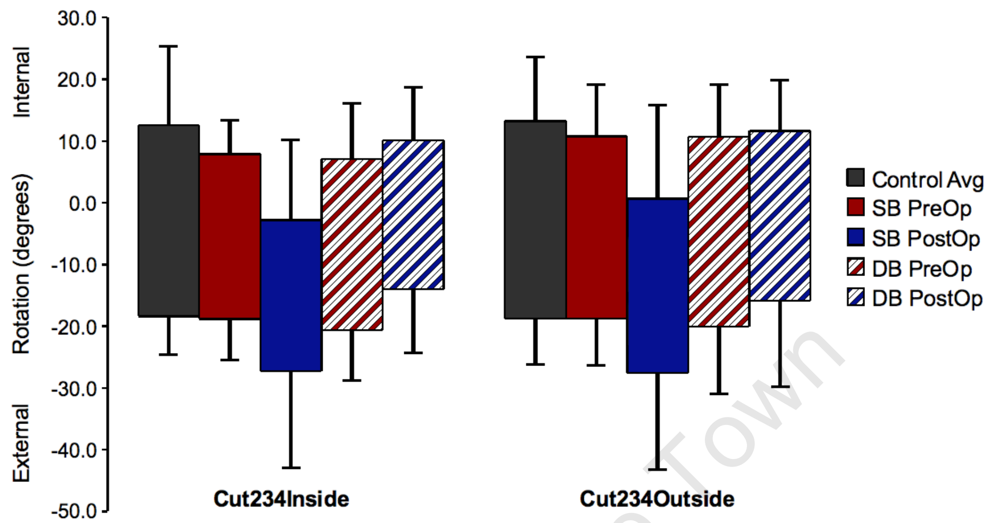

Figure G.4: Cut234Inside and Cut234Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for Control, as well as SB and DB groups both pre- and post-operatively.

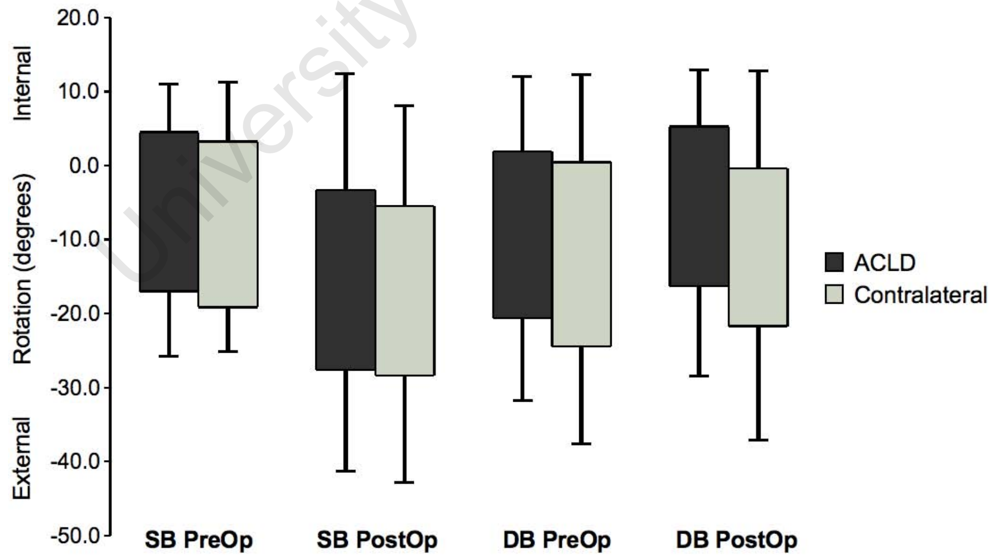

Figure G.5: Walk rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

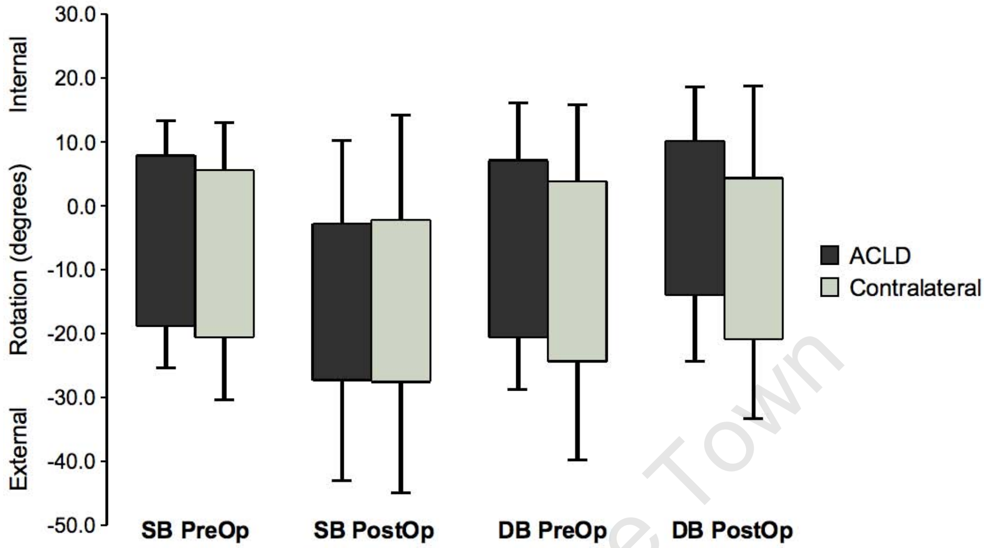

Figure G.6: Cut234Inside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

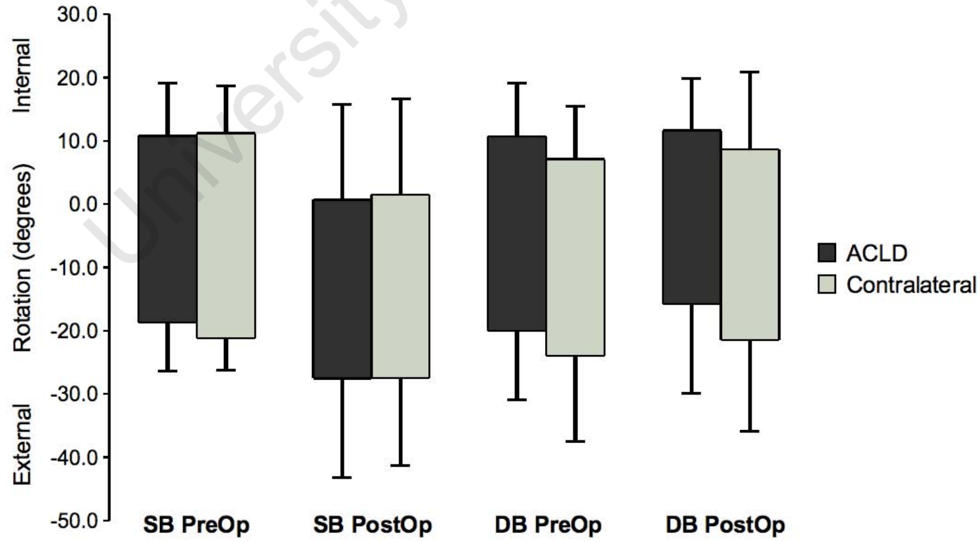

Figure G.7: Cut234Outside rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

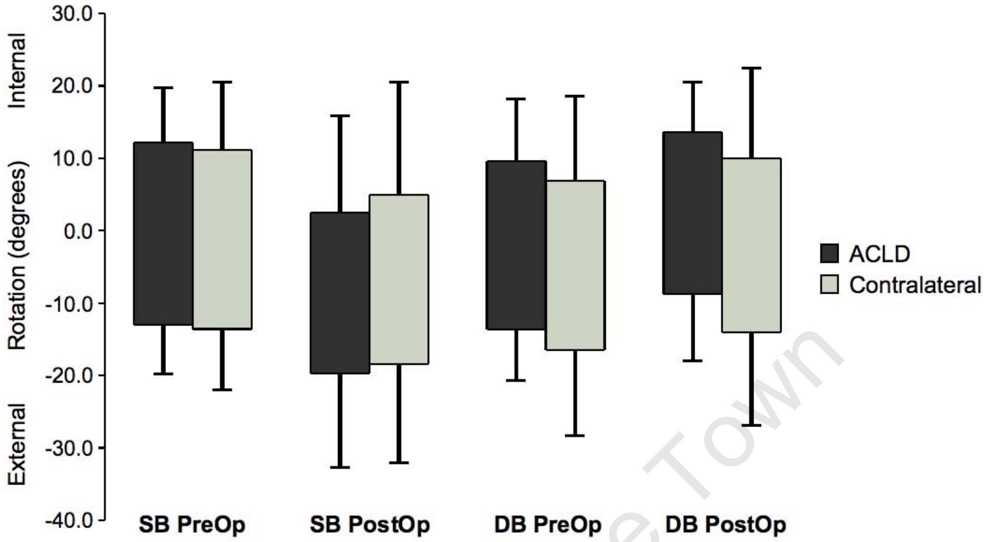

Figure G.8: JumpFull rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

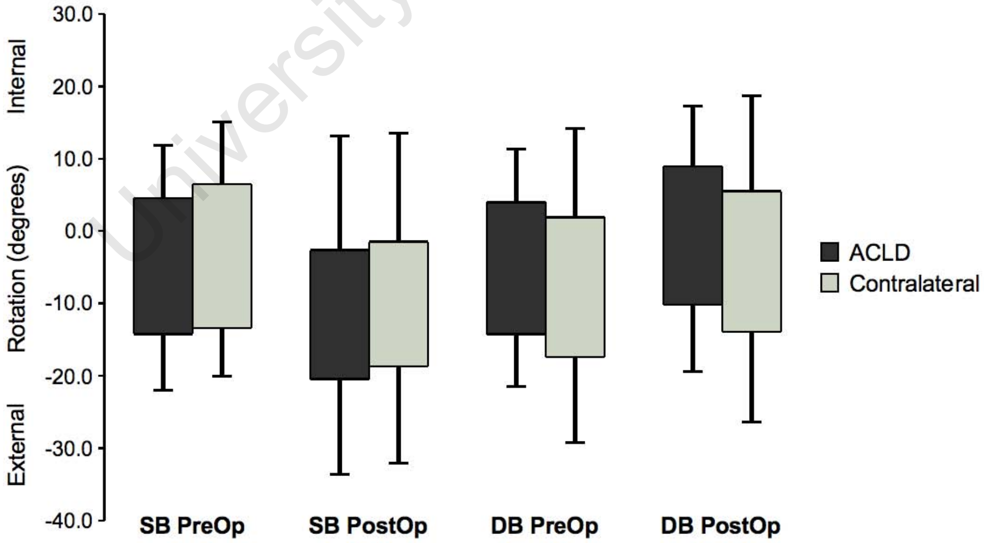

Figure G.9: JumpSW rotation ranges over gait cycle with maximum and minimum rotation standard deviations for SB injured (ACLD) and contralateral, as well as DB injured (ACLD) and contralateral groups both pre- and post-operatively.

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