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Reconstruction of the anterior cruciate ligament with anatomic double-bundle technique

A randomized controlled clinical study comparing ACL reconstructive techniques and an experimental in vitro study

PhD thesis

Marie Bagger Bohn

Health
University of Aarhus
2017

"When we started out current ACL reconstruction succeeded in stabilizing the knee, but they neither fully restored normal knee kinematics nor reproduced normal ligament function" (Sakane et al. 1997) [1].

"There is a considerable subset of patients with knee instability, especially rotational stability, and athletes not able to return to their pre-injury level of sports activity, which might profit from a different surgical approach" (Steckel et al. 2007) [2].

"Anatomic ACL reconstruction has changed the paradigm of traditional ACL surgery"
(Freddie Fu 2015) [3].

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List of papers

This thesis is based on the following papers:

Paper A

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"Mechanical stability of the femoral fixation for single- and double-bundle ACL reconstruction in an in vitro experimental model,"
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Paper B

Bohn MB, Petersen AK, Nielsen DB, Sørensen H, Lind M.
"Three-dimensional kinematic and kinetic analysis of knee rotational stability in ACL-deficient patients during walking, running and pivoting,"
J Exp Orthop. 2016;3(1):27.

Paper C

Bohn MB, Sørensen H, Petersen MK, Søballe K, Lind M.
"Rotational laxity after anatomical ACL reconstruction measured by 3-D motion analysis: a prospective randomized clinical trial comparing anatomic and non-anatomic ACL reconstruction techniques,"
Knee Surg Sports Traumatol Arthrosc. 2015;23(12):3473-81.

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English summary

Anterior cruciate ligament (ACL) rupture is one of the most frequent sports-related injuries in orthopedic surgery. Young and physically active people are especially prone to sustain an ACL injury, and most injuries are sustained during contact or pivoting sports. An ACL rupture leads to complaints of knee instability and is often treated with surgical reconstruction in order to regain knee biomechanics. Even with surgical reconstruction, long-term clinical problems such as meniscal damage and osteoarthritis (OA) development are often observed in these young patients.

In the past decade, an anatomic approach to ACL reconstruction has gained acceptance. Several factors have motivated this change, including disappointing results from traditional techniques, a rediscovery (or revision) of ACL anatomy, research on ACL biomechanics and new surgical techniques. The purpose of anatomic reconstruction techniques is to restore knee biomechanics by placing the ACL graft(s) in native insertions. Anatomic reconstructions can be performed as single-bundle (SB) or double-bundle (DB) reconstructions, and these techniques emphasize restoration of the rotational stability of the knee. Rotational stability of the knee has been brought into focus recent decades, as several studies have shown that excessive tibial rotation persists after reconstruction with traditional techniques. Additionally, excessive laxity has been proposed as an initiating factor of OA.

The overall aim of this thesis was to assess rotational stability in patients with ACL injuries before ACL reconstruction and 1 year after ACL reconstruction using 3-D motion analysis. We hypothesized that anatomic ACL reconstruction would result in better rotational stability than non-anatomic SB ACL reconstruction. This was investigated in a randomized controlled single-blinded study. Furthermore, the biomechanical properties of two different fixation methods (endobutton [EB] and bioabsorbable interference screw [BIS]) used for femoral fixation in both SB and DB anatomic reconstructions were compared in an experimental study.

In the experimental study (Paper A), both SB (using 6 mm or 9 mm grafts) and DB (2 x 6 mm grafts) reconstructions were performed for both fixation methods using porcine material. First, a cyclic test was performed and then a load-to-failure test was conducted. Elongation and load-to-failure scores were recorded, stiffness of the constructs determined and mode of failure observed. Overall, the study showed that

EB fixation was stronger than interference screw fixation. Furthermore, SB 9 mm reconstructions were found to have equal biomechanical properties as DB 2 x 6 mm ACL reconstructions, regardless of the fixation method used. Our study also demonstrated that SB 6 mm constructs were 40% weaker than SB 9 mm constructs. These findings indicate a potential higher risk of graft failure in double-bundle reconstructions where only one graft strand is loaded during range of motion.

In the clinical studies, 3-D motion analysis was used to obtain an objective measure of knee rotation. Hence, knee rotation (in degrees) and the corresponding knee joint moment (Nm) were measured during walking, running and pivoting in ACL-injured patients (pre- and one-year postoperative). These parameters were used to calculate knee rotational stiffness, which is defined as change in knee joint moment divided by change in knee rotation. Thus, in this thesis, rotational stability of the knee is expressed as rotational stiffness.

In Study 2.1 (Paper B), we set out to determine the rotational stiffness in both knees of ACL-injured patients (i.e., preoperative measures) and in a group of healthy controls during walking, running and pivoting. To our knowledge, rotational stiffness during natural movements has not been reported before. Overall, no significant difference in rotational stiffness was found between groups during walking, running or pivoting. Additionally, tibial rotation was seen to increase in all groups as the tasks got more strenuous. During pivoting, however, a subgroup of ACL-deficient patients had a lower tibial rotation of their ACL-deficient knee compared to their healthy knee. These findings indicate the use of different gait strategies to stabilize the ACL-deficient knee within our patient group, which implies that rotation measured in degrees (laxity) cannot stand alone when reporting knee stability.

In the Randomized Controlled Trial (RCT) (Study 2.2; Paper C), three different ACL reconstruction techniques were compared: transtibial SB, anatomic SB, and anatomic DB. The outcome measures were: clinical tests, patient-reported outcome measures and 3-D motion analysis. From preoperative status to one-year follow-up, an improvement in clinical tests and patient-reported outcome measures were seen in all three groups. However, at one-year follow-up, the included outcome measures did not show any difference between the three ACL reconstruction groups. The size of our groups might, however, be too small, as our sample-size calculation was based on measures of rotation from former studies which differed from our actual findings.

Danish summary

Forreste korsbåndsskader rammer primært yngre mennesker mellem 15 og 30 år. Skaden opstår typisk ved sports udøvelse og fører til et ustabilt knæ. Oftest behandles forreste korsbåndsskader med en operativ rekonstruktion. Dette foregår ved en kikkert-operation (artroskopi), hvor det overrevne korsbånd erstattes med en sene-graft, der fastgøres i lårbensknoglen (femur) og skinnebensknoglen (tibia). Traditionelt er rekonstruktionen blevet udført med den såkaldte 'transtibiale' teknik. Opgørelser af denne operations teknik har dog vist skuffende resultater på lang sigt, så som brusk og menisk skader.

Det seneste årti har nye rekonstruktions teknikker således vundet indpas. En udvikling båret af blandt andet de ovennævnte skuffende resultater af den traditionelle operationsteknik, en 'genopdagelse' af forreste korsbånds (ACL) anatomi, forskning i ACL's biomekanik samt nye operations teknikker. Der tales ligefrem om et paradigme skift. De nye rekonstruktions teknikker kaldes 'anatomiske' og kan udføres både som en single-bundle eller en double-bundle rekonstruktion. Teknikken tilstræber at placere graften (eller grafterne) i ACL's oprindelige udspring (footprints) på tibia og femur. Målet er at genskabe knæets oprindelige biomekanik og herved forebygge brusk og menisk skader. Skader der kan have vidtrækkende konsekvenser for denne unge og aktive patient gruppe.

Da denne ph.d. blev beskrevet i 2007, stod ACL kirurgien således midt i en opbrudsfase, hvor den traditionelle 'transtibiale' teknik blev udfordret af de nye anatomiske teknikker. Fokus var især på at genskabe knæets rotations stabilitet og herved forbygge menisk skader og slid på brusken. Vi satte os derfor for at undersøge, om knæ rekonstrueret med anatomisk single- og double-bundle teknik blev mere rotations stabile end knæ rekonstrueret med den traditionelle transtibiale teknik. Dette blev undersøgt i et klinisk randomiseret studie. Yderligere satte vi os for at undersøge de biomekaniske egenskaberne af to forskellige metoder til fastgørelse af sene-grafter i femur (Endobutton [EB] og Interferens skrue [IF]) i et eksperimentelt studie.

Det eksperimentelle studie (study 1; Paper A) blev udført med knogler og sener fra grise. Både single-bundle (graft diameter 6 og 9 mm) samt double-bundle (2x6 mm)

rekonstruktioner blev testet. Først blev rekonstruktionerne testet cyklisk og derefter 'to failure', alt imens elongation, stiffness og load to failure blev registreret. Studiet viste, at EB var den stærkeste måde at forankre grafter i femur på. Ligeledes fandt vi, for begge fiksations metoder, at elongation, stiffness og load to failure ikke var signifikant forskellig mellem double-bundle (DB) og single-bundle (SB) rekonstruktioner (9 mm graft). Yderligere viste studiet, at en enkelt 6 mm graft var op til 40 % svagere end en 9 mm graft i en SB rekonstruktion. Dette fund der kan give anledning til bekymring, da double-bundle rekonstruktioner ofte anvender grafter med en diameter mindre end 6 mm. Belastes en enkelt graft således under knæets bevægeudslag (ROM) kan der således være en risiko for en fornyet overrivning af denne del af korsbåndet.

I de kliniske studier anvendte vi ganganalyse, også kaldet tredimensionel (3-D) bevægeanalyse, til at opnå et objektivt mål for knæets rotation. Både rotationen (målt i grader) og tilhørende led moment (Nm) blev målt præ- og post-operativt under gang, løb og trappenedgang efterfulgt af en pivoterende bevægelse. Knæets rotations stivhed blev udregnet fra disse parametre og er defineret som ændring i led moment divideret med ændringen i rotation. Rotations stivheden anvendes således i denne ph.d. afhandling som et mål for knæets stabilitet og er ikke rapporteret hos ACL patienter under frie dynamiske bevægelser før.

Studie 2.1 (Paper B) rapporterer den præ-operative rotations stivhed af begge knæ ved patienter med en ACL skade, dvs. både det ACL skadede og det intakte knæ, samt rotations stivheden ved en rask kontrol gruppe. Ved ingen af øvelserne fandt studiet nogen forskel i rotation eller rotations stivhed mellem det ACL skadede og intakte knæ. Den tibiale rotation blev dog mere udtalt i både de ACL skadede, ACL intakte og kontrol knæ som øvelserne blev mere krævende. Ved den pivoterende øvelse fandt vi, at cirka en tredjedel af patienterne roterede mindre på det ACL skadede knæ sammenlignet med deres intakte knæ. Dette tyder på, at ACL skadede patienter bruger forskellige strategier til at stabilisere deres knæ og herved kompensere for det skadede ACL. Rotation målt i grader kan derfor ikke stå alene, når stabiliteten af ACL skadede knæ rapporteres.

I det randomiserede studie (studie 2.2; Paper C) blev tre forskellige ACL rekonstruktions teknikker sammenlignet (transtibial SB, anatomisk SB og anatomisk DB). Effektparametre var: kliniske undersøgelser, patient rapporterede spørgeskemaer, styrke og hop tests, samt 3-D bevægeanalyse. Studiet påviste en forbedring i kliniske tests og spørgeskemaer fra præ-operativ status til 1 års follow-up for alle tre grupper. Ved 1 års follow-up viste de anvendte undersøgelses metoder ikke nogen forskel mellem de tre operations teknikker. Størrelsen på vores grupper kan dog meget vel tænkes at være for lav, idet de målte værdier ved vores 3-D bevægeanalyse afveg fra de målinger (fra tidligere studier) der indgik i vores styrke beregning.

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Acronyms

ACL	Anterior cruciate ligament
ACLD	ACL deficient
ACLI	ACL intact
ACLR	ACL reconstructed
ADL	Activity of daily living
AM	Anteromedial
AMP	Anteromedial portal
AP	Antero-posterior
ATT	Anterior tibial translation
BIS	Bioabsorbable Interference Screw
BMD	Bone mineral density
BPTB	Bone-patellar tendon-bone
CI	Confidence interval
DB	Double-bundle
DXA	Dual X-ray Absorptiometry
EB	Endobutton
EMG	Electromygraphy
HT graft	Hamstring graft
IF	Interference screw

IKDC	The International Knee Documentation Committee
KOOS	Knee injury and Osteoarthritis Outcome Score
OA	Osteoarthritis
PL	Posterolateral
PROM	Patient-reported outcome measures
PT graft	Patellar tendon graft
Qol	Quality of life
SB	Single-bundle
STG	Soft tissue graft (Semitendinosus and gracilis)
RCT	Randomized clinical trial
RSA	Radiostereometric Analysis
TT	Transtibial
3-D	Three-dimensional

1. Introduction

Surgical reconstruction of the anterior cruciate ligament (ACL) has evolved significantly in the last 20-30 years. It was only at the beginning of this century that an anatomic approach to ACL reconstruction emerged and began to gain acceptance [3]. This approach is based on new research on ACL anatomy but also on the development of new reconstructive techniques. Furthermore, disappointing outcomes from previous reconstructive techniques and the development of osteoarthritis (OA) in a young and active patient group supported the development of new surgical initiatives [4, 5].

When planning this thesis in 2007, many different ACL reconstruction techniques were reported in the literature. In particular, several new double-bundle (DB) techniques were emerging based on the newly rediscovered two-bundle anatomy of the ACL. Additionally, a new surgical technique for placement of femoral tunnels was introduced. Critics, however, claimed that the DB technique would be double trouble, because the surgical procedure is more complex and the theoretical advantage of the technique had not been proven. Overall, it was hoped that a more anatomic approach would better restore knee biomechanics and result in improved rotational stability of the knee. This was highly interesting, as it had been proposed that the excessive tibial rotation seen after reconstruction with traditional techniques could cause OA [6].

In 2007, the gold standard at our institution was the traditional transtibial technique and the anatomic approach was fairly new. Interestingly, early biomechanical in vitro results of the anatomic reconstruction techniques showed promising results in terms of improved rotatory laxity [7, 8]. Data on in vivo dynamic kinematics of anatomic reconstruction techniques were, however, sparse [9]. On this background, we set out to use advanced motion capture to assess and compare dynamic rotational knee laxity in knees reconstructed with anatomic techniques and in those reconstructed with traditional technique in a randomized controlled study (RCT). Three-dimensional motion analysis was used to obtain an objective measure of pre- and postoperative rotational laxity. Furthermore, an experimental in vitro study was conducted in order to test and compare the biomechanical properties of two different femoral fixation devices used in single- and double-bundle ACL reconstructions.

2. Aims and hypotheses

Study 1

The aim of the first study was to test and compare femoral fixation principles used for ACL reconstruction in an in vitro experimental study. Specifically, we set out to compare the biomechanical properties of different graft diameters and femoral fixation principles used in single-bundle (SB) and double-bundle (DB) ACL reconstructions (Paper A).

We hypothesized that SB 6 mm graft constructs had inferior biomechanical properties compared to SB 9 mm graft constructs or DB 2 × 6 mm graft constructs. Furthermore, we hypothesized that interference screw fixation would demonstrate less elongation and higher stiffness than EB fixation.

Study 2

The overall aim of the second study was to compare the rotational knee stability, expressed as rotational stiffness, between anatomic (SB and DB) and traditional (non-anatomic) SB ACL reconstruction techniques 1 year after ACL reconstruction using 3-D motion analysis.

Study 2.1:

In this study, we aimed to quantify and compare the functional in vivo knee rotational stability between ACL-deficient (ACLD) knees and intact knees during walking, running and 90° pivoting. We hypothesized larger rotation, lower rotational moments and, therefore, lower rotational stiffness in the ACLD knees compared to the contralateral uninjured knee and a healthy control group (Paper B).

Study 2.2:

The aim of this study was to compare knee rotational stability in ACL-reconstructed individuals (ACLR) after anatomic DB ACL reconstruction, anatomic SB ACL reconstruction and traditional SB ACL reconstruction in a prospective RCT. We hypothesized that anatomic ACL reconstructions would result in better rotational stability than traditional SB ACL reconstruction (Paper C).

3. Background and definitions

3.1 Anterior cruciate ligament (ACL)

3.1.1 Anatomy

In recent decades, the anatomy of the ACL has been subjected to a critical review which has brought the two-bundle anatomy of the ACL into focus. Hence, functionally and from a gross appearance, the ACL has been divided into two major fiber bundles; the anteromedial (AM) and the posterolateral (PL) bundle, named according to the orientation of their tibial insertions (Fig. 1) [10]. Namely, the native insertions of these bundles have been studied as well as their function.

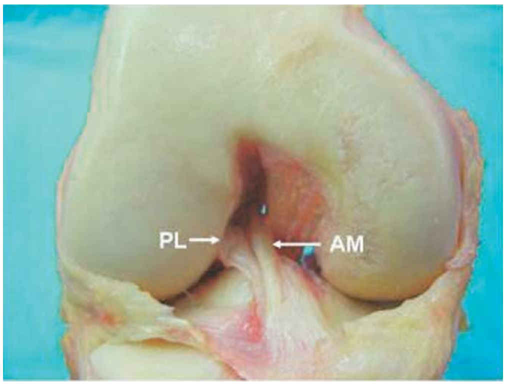

Figure 1: Anterior view of the right knee joint. The patellar tendon and the surrounding soft tissue has been removed to inspect the ACL. Note the two distinct bundles, the AM and PL bundles. Illustration from Petersen and Zantop[10].

Femoral insertion

The native femoral insertion originates from deep within the intercondylar notch. More specific, the proximal fibers of the ACL are attached to the posterior part of the medial surface of the lateral femoral condyle (Fig. 2). Anatomical studies have shown the bony femoral insertion to be in the shape of a crescent. The most anterior border of the insertion is marked by a bony ridge named the lateral intercondylar ridge (or resident’s ridge) (Fig. 2). This ridge is described as a distinctive change in slope of the femoral roof that occurs just anterior to the femoral attachment of the ACL [11-13].

Posteriorly and inferiorly, the ACL fibers extend to the border of the articular surface of the lateral femoral condyle. Several authors have shown that the size of the femoral insertion site varies greatly [12-16].

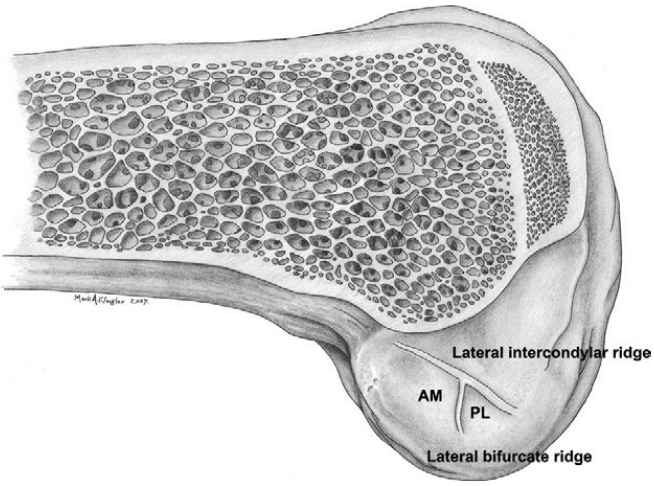

Figure 2: The lateral wall of the intercondylar notch. When the axis of the femur is parallel to the floor, the lateral bifurcate ridge runs anteroposterior, dividing the posterolateral (PL) and anteromedial (AM) femoral attachments, whereas the lateral intercondylar ridge (resident's ridge) runs proximodistal along the entire anterior cruciate ligament attachment [12].

In the extended knee, the AM bundle is described as located proximal and anterior in the insertion, whereas the PL bundle is located in the distal and posterior aspect of the insertion. The location of the AM and PL bundles in the femoral footprint is shown in Figure 2 (flexed knee). In the anterior part of the insertion, the bundles are separated by a bony ridge, the lateral bifurcate ridge (Fig. 2) [12]. Additionally, a change of slope in the femoral attachment topography has been shown by Ferretti et al. This study describes how the attachment of the PL bundle has no changes in its plane, whereas the AM bundle has a concave shape with a significant change of slope (Fig. 3) [12]. These findings suggest that the two bundles of the ACL have two distinct anatomic femoral attachments. According to Siebold et al., the distribution of the AM and PL bundles in the overall femoral ACL insertion is 52% and 48%, respectively [16]. Interestingly, this study also showed that the average femoral insertion areas of the ACL and of the AM and PL bundles were significantly larger in men compared with woman and in left knees compared to right knees [16].

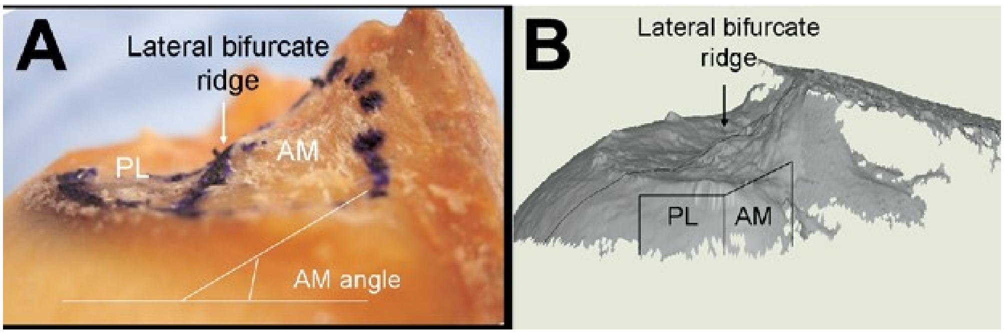

Figure 3: Posterior view of the lateral femoral condyle in a right knee depicting a change of slope of the anterior cruciate ligament femoral attachment. (A) Gross observation in a knee specimen. *White lines* show the angle formed between the posterolateral and anteromedial femoral attachments (average of 27.6° ± 8.8°). Note the lateral bifurcate ridge. (B) Three-dimensional laser picture demonstrating the topography of the femoral anterior cruciate ligament attachment. Note the change of slope forming the anteromedial angle [12].

Tibial insertion

At the tibial side, the ACL inserts into a depression or fovea between the tibial condyles, named the area intercondylaris anterior (AIA). Several anatomical studies have described the tibial insertion of the ACL without reaching a total agreement on its shape, extension and bony borders[10, 13, 17-20]. Hence, the tibial ACL footprint has been described as oval or triangular (Fig. 4) [10, 20] and, accordingly, the size of the insertion shows large variation too [10, 13, 15, 19, 20].

Descriptions of the placement of the two bundles in the tibial footprint also vary slightly among authors [10, 17, 19, 20]. Figure 4 illustrate both tibial insertion shape and placement of the two bundles according to Petersen et al. [10]. The relationship between the tibial area of the AM and PL bundles has, likewise, been reported by Siebold et al. who found it to be 56% to 44%, respectively, in both genders [20]. Comparable to the femoral insertion, the average male tibial ACL insertion area was found to be significantly larger compared to female knees [20].

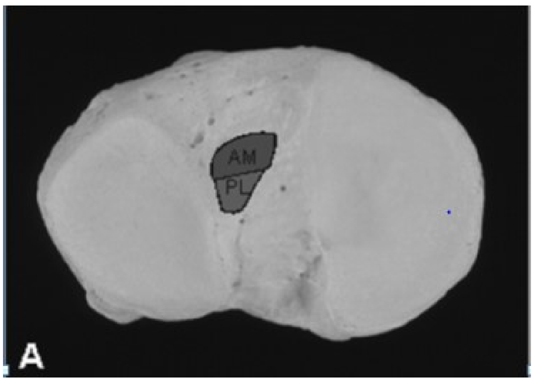

Figure 4: The tibial insertion of the ACL with regard to its two bundles [10].

General anatomical behavior of the ACL

The general anatomical behavior of the ACL is a result of its bony attachments. In full extension, the ACL is flat. As the knee flexes, the femoral origin of the ACL moves posteriorly and inferiorly, decreasing the anatomic angle made by the ACL to the tibial plateau [1]. During flexion of the knee, the ligament seems to turn itself in a lateral spiral. This external rotation is approximately 90º as the fibers approach the tibial surface [10, 18].

3.1.2 Function of the ACL

Overall, the ACL acts as the primary restraint to anterior tibial translation and as a secondary restraint to internal rotation of the weightbearing and non-weightbearing knee [10, 21]. However, it has long been understood that the ACL does not function as a simple band of fibers with constant tension as the knee moves. Historically, it was reported in 1920 by Hey Groves that the ACL was made tense by extension of the knee. In 1941, Brantigan and Voshell refined this description by differentiating between the actions of the anterior and posterior fibers of the ACL. This description was confirmed by Girgis in 1975, who described the bony attachments of the ACL and how a reciprocal tightening and slackening of the anterior and posterior fiber groups was seen during flexion and extension of the knee [22]. However, Odensten and Gilquist opposed this claim in 1985, as their histologic examination of the ACL found no evidence to separate the ligament into 2 bundles [23]. Yet another approach was taken by Amis and Dawkins, as well as other authors, who divided the ACL into three bundles namely an AM, intermediate, and PL bundle [22]. Despite these controversies, and the fact that a two-bundle description might be an oversimplification, the two-bundle description has been widely accepted as a basis for understanding the function of the ACL [10].

Function of the AM and PL bundle of the ACL

Function of the two bundles of the ACL has been shown in several biomechanical in vitro studies. In early laboratory studies, the response of the two bundles to an anterior draw was examined. In 1991, Amis and Dawkins showed the contribution of the PL bundle in resisting anterior draw in extension, and how the AM bundle became dominant at 90° of flexion [22]. Further, Sakane and Fu measured the in situ force in the ACL and its two bundles in response to an applied anterior tibial load (Fig. 5) [1]. This study showed that the in situ force in the posterolateral bundle was highest at full extension and decreased with increasing flexion. Additionally, the in situ force in the anteromedial bundle was lower than the posterolateral bundle at full extension, remained relatively constant during flexion and reached a maximum at 60° of flexion. These results imply that each bundle plays a separate, but equally important, role in the complex function of the ACL.

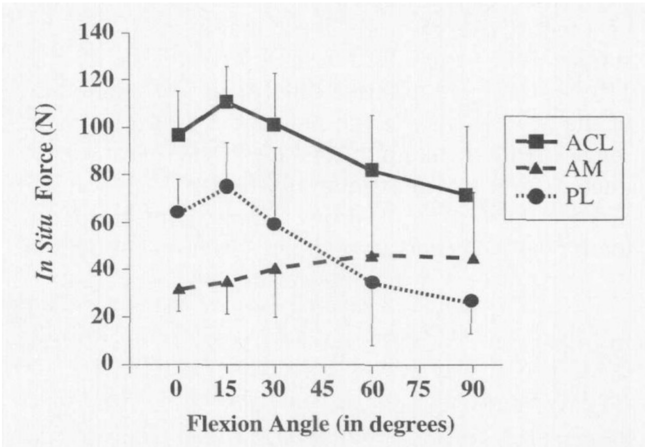

Figure 5: Magnitude of the mean in situ force in the intact ACL, anteromedial bundle (AM), and posterolateral bundle (PL) under 110 N of applied anterior tibial load [1].

Further laboratory studies added the examination of the rotatory response of the two bundles [7, 24-26]. In some studies a biomechanical model to simulate the pivot shift (anterior translation under a combined rotatory load of valgus and internal rotation) was used to measure rotational stability. Gabriel et al. used this model and found that: (1) the in situ force of the posterolateral bundle was higher at 15° and lower at 30° of flexion and (2) the in situ force in the anteromedial bundle was similar at 15° and 30° of knee flexion, under a combined rotatory load (Fig. 6) [24].

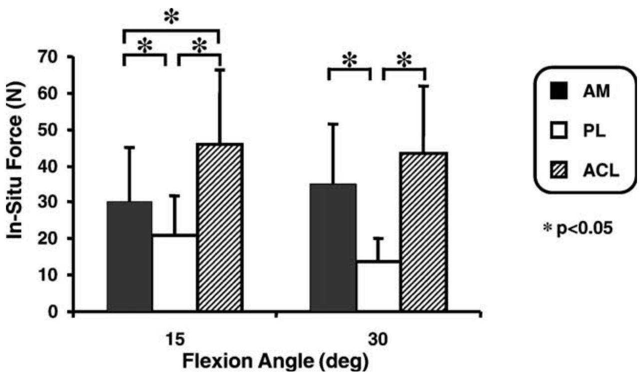

Figure 6: In situ force in the intact ACL and its AM and PL bundles in response to combined rotatory load (10 Nm valgus and 5 Nm internal tibial torque) [24].

Another approach was taken by Zantop et al. [26], who focused on knee kinematics in AM-deficient and PL-deficient knees. They found that resectioning of the PL bundle increased anterior tibial translation (ATT) significantly in response to a combined rotatory load at 0° and 30° of knee flexion when compared to the intact knee, whereas no significant increase in ATT was seen after transection of the AM bundle.

Thus, both studies demonstrated the essential role of the PL bundle as a constraint to rotatory loads in joint positions close to extension [24, 26].

In conclusion, these results suggest that a reciprocal relationship exists in the length and tension patterns between the two bundles of the ACL and that none of the bundles are isometric throughout flexion and extension [27]. Furthermore, these biomechanical data demonstrate that both the AM bundle and the PL bundle play important roles in stabilizing the knee joint. The PL bundle as an important stabilizer against rotatory and anterior loads especially when the knee was near extension. Taking our anatomical knowledge of the femoral and tibial insertions of the two bundles into account, this biomechanical behavior makes sense [19]. Thus, as the knee is flexed, the femoral attachment of the ACL moves to a more horizontal orientation, causing the AM bundle to tighten and the PL bundle to loosen up [1, 10, 22]. Furthermore, anatomical studies have shown the insertion points of the AM bundle to be close to the central axis of the human knee, a position which makes it difficult for the AM bundle to restrain rotatory loads [26].

3.2 ACL reconstruction

3.2.1 Historical view

Surgical treatment of ACL injuries date back to 1895 when the first suturing of a torn ACL was described using an open technique [28]. Over time, surgical techniques for ACL reconstruction progressed from open techniques to arthroscopic surgeries. As a curiosity, it should be mentioned that a recently discovered paper describes arthroscopic activity before World War I. Thus, already in 1912, at the 41st Congress of the German Society of Surgeons in Berlin, a Danish surgeon from Aarhus named Severin Nordentoft presented his self-built “trocart-endoscope” (Fig. 7). In addition to suprapubic cystoscopy and laparoscopy, he advised the use of such an endoscopic device in the knee joint, especially for early detection of meniscal lesions. Dr. Nordentoft baptized the procedure “arthroscopy” and the primacy of arthroscopy should be attributed to this Danish surgeon [29].

Figure 7: Severin Nordentoft, 1866-1922, “the first arthroscopist,” St Joseph’s Hospital, Aarhus, Denmark [29].

However, it took almost another seventy years before Dandy reported the first arthroscopic ACL reconstruction in 1981 [21, 30]. Subsequently, ACL surgeons rapidly adopted this new minimally invasive arthroscopic technique and the number of primary ACL reconstructions increased [3]. At first, arthroscopic ACL reconstruction was performed using a two-incision technique, in which the femoral tunnel was drilled from the outside in. Over time, a one-incision technique - the transtibial technique (TT) - was adopted, where the femoral bone tunnel was drilled from the inside out through a tibial tunnel [31]. These early arthroscopic ACL reconstructions were single-bundle techniques.

As an alternative to the traditional TT approach, O'Donnell described a modified technique for femoral tunnel placement in 1995. O'Donnell drilled the femoral tunnel through an accessory anteromedial portal (AMP) in order to avoid the constraints imposed by working through the tibial tunnel [32]. This study was followed by Bottoni in 1998, who also inserted the femoral guide through the AMP, aiming at a more anatomic femoral tunnel placement [33]. This new technique made it possible to place the femoral tunnels more inferior on the femoral condyle (in the flexed knee).

The first arthroscopic method for double-bundle ACL reconstruction was described in 1994 by Rosenberg and the double-bundle procedure was since popularized in Japan by Yasuda et al. and Muneta et al. in the late 1990s [3, 28, 34-36]. Their work and efforts allowed others to take a more critical look at ACL anatomy and subsequent ACL reconstruction, as stated in a recent paper by Freddie Fu [3].

3.2.2 ACL reconstruction in the beginning of the 21st century

Overall, major advancements were made with the introduction of arthroscopic ACL surgery. In addition, the advent of transtibial drilling of femoral tunnels simplified the procedure, enabling further reduction in surgical time and trauma by means of a single-incision approach [37]. The advancements were, however, also associated with new problems, including loss of knee range of motion [38], impingement of the ACL graft [39] and an increasing number of ACL revisions. Hence, follow-up studies (on the traditional transtibial single-bundle ACL reconstruction) showed less than optimal results in up to 25% of the patients [2, 27, 40-42]. Furthermore, development of osteoarthritis (OA) was reported in patients with ACL reconstructions [5, 7, 43].

Simultaneously, cadaver studies were performed to compare different surgical techniques for ACL reconstruction. Firstly, the inferiority of the traditional transtibial single-bundle reconstruction (TT) to restore the in situ force of the native ACL was shown in several studies [7, 44]. Additionally, it was realized that traditional ACL reconstruction techniques had been focusing on recreating the AM bundle only, while the PL bundle not had been addressed [27]. At the same time, several different double-bundle (DB) reconstructions were studied in vitro [7, 45]. These early DB reconstructions were promising in terms of biomechanical outcomes and marked the first steps toward a more anatomic approach. However, the promising in vitro biomechanical results of early double-bundle (DB) techniques were not reflected in improved clinical outcomes, when compared to the traditional single-bundle technique [46-48]. A careful review of the literature reveals that these early double-bundle techniques were rather a reconstruction of the AM bundle with two bundles [49]. Thus, in the beginning, the so-called ‘anatomic’ approach to ACL reconstruction focused more on the two-bundle anatomy of the ACL than on placing the tunnels in their native insertions [48].

The clock-face reference

As the anatomic approach was introduced, femoral tunnel placement was most often described as an imaginary face of a clock placed in the notch (Fig. 8). With the TT technique, most femoral tunnels were placed at 11.00 o’clock (right)/ 01.00 o’clock (left). Placement in a lower position (10 o’clock/2 o’clock) was reported too [41, 46, 50-54]. Moreover, an experimental study by Loh et al. showed that a low-position (10 o’clock/2 o’clock) of the femoral tunnel could better resist the rotational load of the knee than a high position of the femoral tunnel (11 o’clock/1 o’clock) [8].

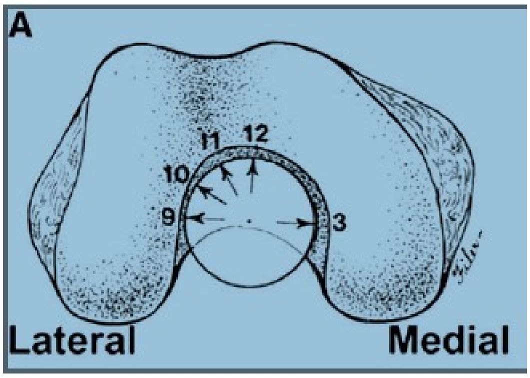

Figure 8: Clock markings around the posterior outlet of the femoral intercondylar notch, with the 9–3 o’clock axis parallel to the epicondylar axis [8].

Interestingly, an anatomical study from 2007 showed that, in the frontal plane, the AM bundle origin was in the 10:30 clock position while the PL bundle origin was situated in the 9:30 clock position [10]. Furthermore, the study showed that the bundles were more horizontally aligned on the femoral side when the knee was flexed to 90°, with the AM bundle insertion site deeper than the PL bundle insertion site (Fig. 9) [10]. These findings reinforced the concerns regarding non-anatomic femoral tunnel placement mentioned above. Additionally, incorrect tunnel placement was described as the most common cause of clinical failure [33, 55-57].

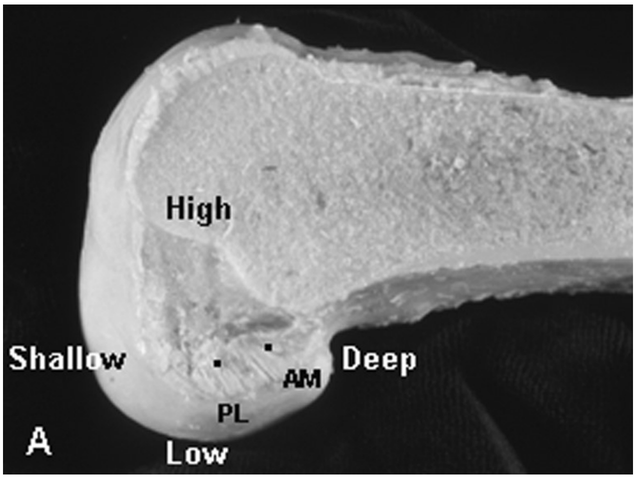

Figure 9: The lateral femoral condyle with the knee in 90° of flexion. The AM bundle is located high (anatomic description, anterior) and deep in the intercondylar fossa, whereas the PL bundle is located more low (anatomic description, posterior) and shallow [10].

Anatomic ACL reconstruction

As mentioned above, a new surgical technique to place the femoral tunnel(s) emerged in the late 1990s and several technical notes were published [33, 55, 57-59]. The technique was named the anteromedial portal (AMP) technique, as the femoral tunnel was drilled through the anteromedial arthroscopic working portal with the knee in hyper-flexion (120°) (Fig. 10). The AMP technique had several advantages. First of all, it offered the possibility of a more anatomical placement of femoral tunnel(s) due to a better view of the medial wall of the lateral femoral condyle and because the femoral and tibial tunnels were placed independently of each other. Another advantage was that tunnel placement was independent of graft type, fixation devices or tunnel guides. Finally, the AMP technique allowed easy preservation of any remaining ACL fibers (augmentation) [58].

In the following years, the AMP technique was adopted by many surgeons and further researched [60, 61]. Several different DB techniques using the AMP technique were reported and a tendency toward improved clinical outcomes was found [62-64]. In addition, the introduction of the AMP technique gave rise to a new modified technique for single-bundle (SB) ACL reconstruction. However, several different placements of the femoral tunnel were reported in these early anatomic SB reconstructions. Jarvela et al., for example, placed their femoral tunnel in the footprint of the AM bundle [62] while Harner et al. advocated that the femoral tunnel should be placed between the anatomic AM and PL tunnel positions, thus representing a SB anatomic compromise [55]. Interestingly, Chhabra et al. studied the SB AMP technique and found a significantly lower femoral tunnel expansion for the AMP technique when compared to the traditional TT SB technique [65]. Otherwise, clinical outcome data on this new SB technique was sparse.

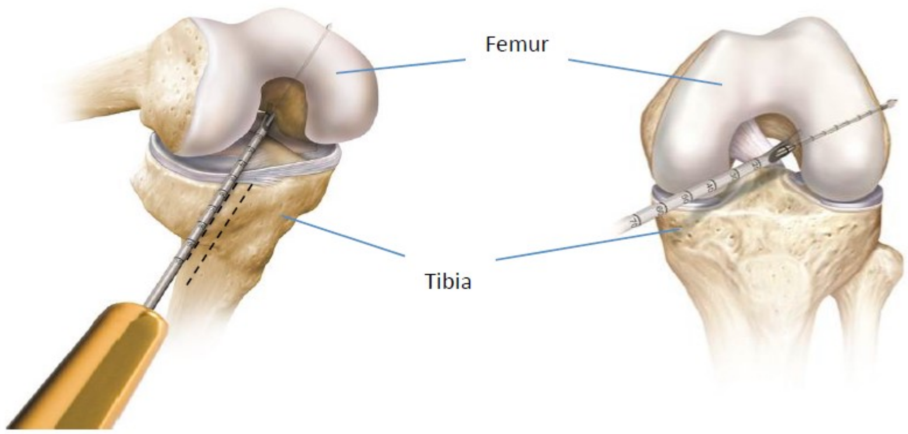

Figure 10: Left: Transtibial drilling of the femoral tunnel. Right: Drilling of the femoral tunnel through the anteromedial working portal (AMP technique) (with kind permission of Arthrex).

In 2007, current tendencies in DB ACL reconstruction were presented in a survey by Zantop et al. [42]. The survey was conducted among 22 experienced surgeons from Europe (50%), Asia (45%) and North America (5%). The study showed that most of the surgeons preferred a technique that used two femoral and two tibial tunnels for the reconstruction of the AM and PL bundles. The TT technique was preferred for AM tunnel placement by most of the surgeons while all surgeons starting with the femoral PL bundle used the AMP technique.

As shown in this survey from 2007, the AMP technique had gained interest. The introduction of the AMP technique did, however, mark a movement toward a more anatomical placement of tunnels in ACL reconstructions. As stated by Schindler: "the beginning of the twenty-first century saw a movement away from the concept of isometry toward ACL reconstruction focusing on more physiological and anatomical principles" [28]. Thus, new anatomic techniques emerged that focused on the importance of restoring any injured anatomic structure to its normal functional position and tension. In particular, Freddie Fu and his group from Pittsburgh led this "anatomic evolution of ACL reconstruction" [27, 28, 66]. This group introduced the anatomic double-bundle concept, which aimed to replicate the native ACL anatomy more closely (i.e., in its dimensions, insertions and fiber arrangement) [28, 67]. Clinical outcome data on this new anatomic technique were limited at the time and there was a call for prospective randomized clinical trials comparing SB to DB techniques [2, 49, 68], studies which should include the evaluation of rotational stability of the knee joint as an outcome measure [27].

3.2.3 Fixation methods in ACL reconstructions using soft tissue grafts

The 2007 survey by Zantop et al. showed that most surgeons used hamstring grafts for ACL reconstruction and that several different fixation methods were used on the tibial and femoral side at the time [42]. On the femoral side specifically, 80% of the surgeons preferred a suspensory button technique while the remaining 20% used an interference screw (IF screw) [42].

All though popular, various disadvantages were shown to be associated with the femoral suspensory fixation. Hence, graft tunnel motion and a bungee cord effect, which was further associated with tunnel widening, was reported [69]. It was feared that these complications could potentially lead to delayed biological incorporation and secondary rotational and anterior instability. Therefore, the IF screw was preferred by surgeons who advocated that this aperture fixation might overcome the aforementioned biomechanical disadvantages associated with suspensory fixation, as the graft was fixed closer to the articular surface and, hence, more anatomic [69-72]. Interestingly, a simultaneous clinical study did not show any difference in clinical outcomes between the two fixation methods after two years and, furthermore, significant tunnel enlargement was seen in both groups [73].

At the time, several studies had compared the biomechanical properties of different fixation devices in experimental setups [74-78]. However, data on suspensory fixation was sparse [76]. Furthermore, the biomechanical studies published at the time were all tested in a single-bundle reconstruction setup. Thus, biomechanical studies on the four-tunnel double-bundle technique were needed to address the issues of a smaller graft diameter, two tunnels versus one and fracture risk of the lateral femoral condyle [2].

3.3 Outcome measures

3.3.1 Subjective and objective outcome measures

Multiple outcome measures are used in the assessment of the ACL-injured patient [79]. Several of these outcome measures are used both pre- and postoperatively. Hence, they are used to diagnose an ACL injury and, furthermore, to evaluate the ACL-reconstructed knee. Some of these outcome measures are applicable in the clinic while others require a more specialized setting.

In the clinic, several objective knee stability tests are performed in order to assess both the anterior-posterior (AP) translation (sagittal stability) and the rotational stability of the knee. AP translation can be assessed either manually (Lachman test) or instrumentally (Rollimeter, KT1000 or KT2000) while the pivot shift test is used to assess the rotation and dynamic laxity associated with ACL insufficiency [80]. These measures are, however, highly dependent on the clinician doing the observations as well as the ability of the patients to relax their muscles during testing [81-83].

Furthermore, several subjective outcome measures are used in the ACL literature. These subjective measures contain both self-administrated standardized questionnaires (often referred to as PROMs) and clinical performance-based tests. PROMs measure the patient's own opinion on how the knee joint affects daily life and sports activities and are especially effective in comparing the results of an intervention from the patient's perspective. Interestingly, subjective variables of symptoms and function have been shown to display a robust association with patient satisfaction [84], and in recent years, patient satisfaction has become an outcome measure with great clinical and economic implications [84].

Typically, there are two types of PROMs: general health or disease-specific [85]. The general health measure evaluates a range of parameters, both mental and physical, and the most commonly used in orthopedics is the SF-36 [86]. The disease-specific measure, on the other hand, reveals patients' perceptions of a specific disease or condition and is useful for measuring clinically important changes in response to treatment [87]. Several disease- or condition-specific PROMs are used in the ACL literature. Examples are the KOOS score, IKDC subjective knee evaluation form, Lysholm, Tegner, Marx activity scale, WOMAC, Cincinnati knee rating scale and the ACL quality of life [79, 85, 88, 89]. Often, the use of both a general and a disease-specific instrument is recommended [85]. Unfortunately, a substantial variability in outcome reporting patterns are seen in the literature, which creates challenges in interpreting results [79].

Another category of subjective outcome measures are the clinical performance-based tests, which include functional hop testing and quadriceps strength. Several hop tests are described in the literature. The single-leg hop tests are, however, commonly used in functional assessment after ACL injury and have shown to be a valid and reliable functional performance measurement [90, 91]. Further hop tests are the triple hop for distance, 6 m timed hop and crossover hops for distance. Quadriceps strength is usually assessed with an isokinetic dynamometer or in a leg extensor power rig [90]. Performance-based tests capture different aspects of function than the self-reported assessments do and have gained increased interest in recent years [90, 92]. Interestingly, the functional status of the knee at the time of surgery, as assessed by clinical performance-based tests, has been shown to affect the final outcome. Thus, it has been advocated that functional tests should be taken into account in the decision for ACL reconstruction [90, 92].

3.3.2 Three-dimensional motion analysis

Gait analysis, or 3-D motion analysis, is the systematic study of human motion. Typically, 3-D motion analysis is carried out in a motion analysis laboratory, where dynamic biomechanical data (kinematic, kinetic and/or electromyography) are collected and processed in order to analyze a subject's ability to walk, run or even perform more complex tasks. Normally, a natural symmetry is seen between the left and right sides during a normal gait, while an asymmetrical pattern very often exists in a pathological gait [93]. Thus, asymmetrical gait patterns and other movement-related problems in people with injuries can be identified and assessed using 3-D motion analysis, which offers an objective measure of joint laxity and the forces acting upon the moving body.

The typical motion analysis laboratory consists of coupled infra-red cameras positioned around a walkway, where one or two force-plates are embedded. Reflective markers are placed on the patient's skin aligned with bony landmarks to define anatomic planes and joint centers. The markers can be placed separately or mounted on plates (clusters). As the patient performs tasks on the walkway, the cameras track the motion of the reflective markers while the force plates measure ground reaction forces beneath one or two feet [94]. Inputs from the cameras and force plates are sent to and processed by a computer.

Fundamental in motion analysis is the definition of the relationship between the markers placed on the skin surface and the underlying bony geometry. Hence, the operator is able to establish a "technical coordinate system" associated with the externally placed markers and an "anatomical coordinate system" associated with the underlying bony structure for each body segment under examination (Fig. 11) [95]. Thus, an embedded or body-fixed coordinate system may be determined for any body segment (assumed to be rigid) that has at least three markers attached to it.

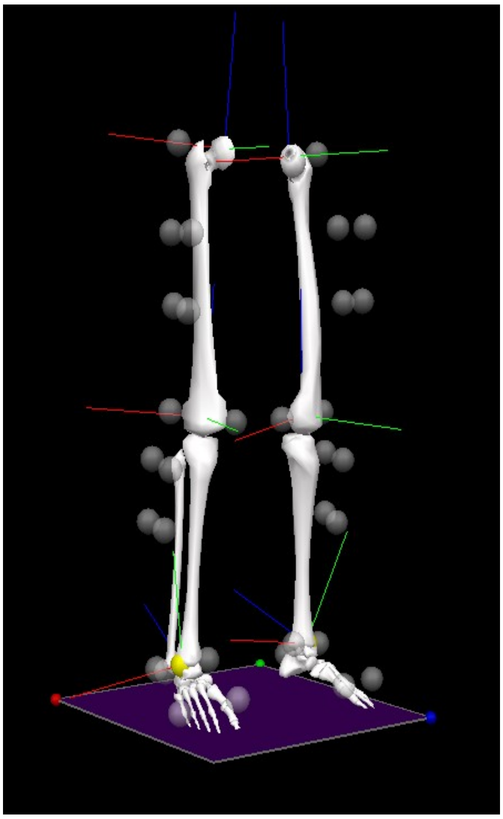

Figure 11: this picture shows the computed lower extremities and reflective markers (with kind permission of Dennis B Nielsen).

Techniques

Three-dimensional motion analysis involves measurements of several parameters such as kinematics and kinetics. In short, kinematic is the study of movement and measures the dynamic range of motion of a joint (or segment). Kinematics requires the recording of time and distance data, joint angles and accelerations over time (temporal-spatial data). Kinetics describes the forces acting on a moving body while Electromyography (EMG) identifies the timing and relative intensity of muscle activation [94, 96].

The combination of kinematic data, anthropometric data and data from the force plate enables the mechanical analysis of the gait to be performed. Inverse dynamics are used to calculate the force at each joint (joint moment) and power [96]. In practice, inverse dynamics is used to compute the internal moments and forces from measurements of the motion of limbs and external forces, such as ground reaction forces, under a special set of assumptions.

3.3.3 Rotational stability of the knee measured by 3-D motion analysis

Planning this study, several authors had used 3-D-motion analysis to assess knee biomechanics in both ACL-deficient (ACLD) and ACL-reconstructed (ACLR) patients during various low- and high-demand activities [6, 97-107]. Interestingly, several studies from a Greek research group had been focusing on the rotational laxity of the ACL-injured knee [97-100]. These studies reported that tibial rotation was not restored after ACL reconstruction with current techniques. Furthermore, the studies presented a novel pivoting task and reported the angular displacement (in degrees) while doing so. This task included stair descending and pivoting on the landing leg, and aimed to force the tibia into a maximum internal rotation. This high-demand task was included in our study in order to assess rotational laxity and stability.

4. Study design, material and methods

In this thesis, an experimental study (Study 1) and a clinical study (Study 2) have been performed. This section summarizes the general study design, the material used and the most important methods applied. A detailed description of the studies is provided in the original manuscripts.

4.1 Study 1 – Experimental study

4.1.1 Study design and material

The experimental in vitro study of this thesis was performed at the Orthopedic Research Laboratory, Aarhus University Hospital, using porcine legs from five-month-old pigs. The animal material was purchased from a local slaughter house and dissected at the laboratory. Hence, porcine flexor tendons and femora were used for ACL reconstructions in order to test the biomechanical properties of different femoral fixation principles.

In this study, two different femoral fixation techniques were tested: (1) a cortical fixation (EB CL 20 mm from Smith & Nephew®) and (2) an aperture fixation (BIS [IF], length 25 mm from Inion Hexalon™) (Fig. 12). Both fixation principles were tested while using different graft diameters and reconstruction techniques: (1) SB ACL reconstruction using a graft diameter of 6 mm (SB 6-mm), (2) SB ACL reconstruction using a graft diameter of 9 mm (SB 9-mm) and (3) DB ACL reconstruction using two grafts with a graft diameter of 6 mm (DB 2×6-mm). Ten specimens were tested for every diameter and technique (60 setups in total).

Figure 12: The figure illustrates the two different fixation methods used; Two BISs in 9 and 6 mm diameters, length 25 mm (Inion Hexalon™) and, at the right, an EB (Smith & Nephew® CL 20 mm)[108].

4.1.2 Testing procedure

Tensile testing of the femur/graft complex was performed in an 858 Mini Bionix material testing machine and the construct was mounted as shown in Figure 13. First the complex was cyclically preconditioned. Then a cyclic test was performed (1000 cycles between 50-250N at 1 Hz) and, finally, a load-to-failure test was performed. During the cyclic test, elongation of the graft construct was recorded. Finally, the graft/femur complex was tested to failure and the ultimate failure load, elongation and mode of failure were documented. The stiffness of the complex was defined as the slope of the linear region of the first and most steep part of the load-displacement curve during the failure test.

Figure 13: Mounting of the experimental ACL reconstruction. The porcine femur was embedded in a steel cylinder with bone cement, which was secured in a custom device (with six degrees of freedom), while the tendon was fixed in a custom cryoclamp. Both devices were fastened to the material testing machine [108].

4.1.3 Statistics

An analysis of variance test was used to evaluate the overall differences between the different study groups for elongation after 1000 cycles, stiffness and maximum load. The Bonferroni correction was used for pairwise multiple comparisons. Significance was set at 5% ($P < 0.05$).

4.2 Study 2 – Clinical study

4.2.1 Patients and ethics

Patient recruitment for the clinical study was carried out at the Division of Sports Trauma, Department of Orthopedics, Aarhus University Hospital. Patients scheduled for ACL reconstruction were assessed for eligibility.

The eligibility criteria were: age 18–50 years, MRI-verified ACL injury with symptoms of instability, no previous knee ligament surgery, no concomitant knee ligament injuries, and an uninjured contralateral knee. Exclusion criteria were: cartilage injuries of International Cartilage Research society grade 3 or 4 and/or meniscus injury requiring resection of more than 50% of the meniscus.

In the study period, 60 patients were identified as potential participants and completed the preoperative tests. Of these 60 patients, a total of 45 patients met all the inclusion criteria and were randomized at surgery. However, there was reason to believe that one of these patients had a bilateral ACL injury at the time of inclusion which became apparent during the postoperative period. Therefore, the cohort investigated in study 2.1 comprises 44 subjects.

Follow-up was performed at approximately one year post-operation (13 months on average). At this time, 36 patients (80%) of the 45 included were available. Details of patient flow are shown in the CONSORT flowchart in Figure 14.

The control group consisted of 16 age- and sex-matched healthy subjects with no history of lower limb pathology or trauma. Most of these subjects were students at the Section of Sports Science, Aarhus University, except one who was a former international high jumper.

The study was approved by the ethical committee of Region Midtjylland (M-AÅ-20060198).

4.2.2 Study design

Study 2.1
Cross sectional study

Study 2.2
Prospective randomized controlled trial

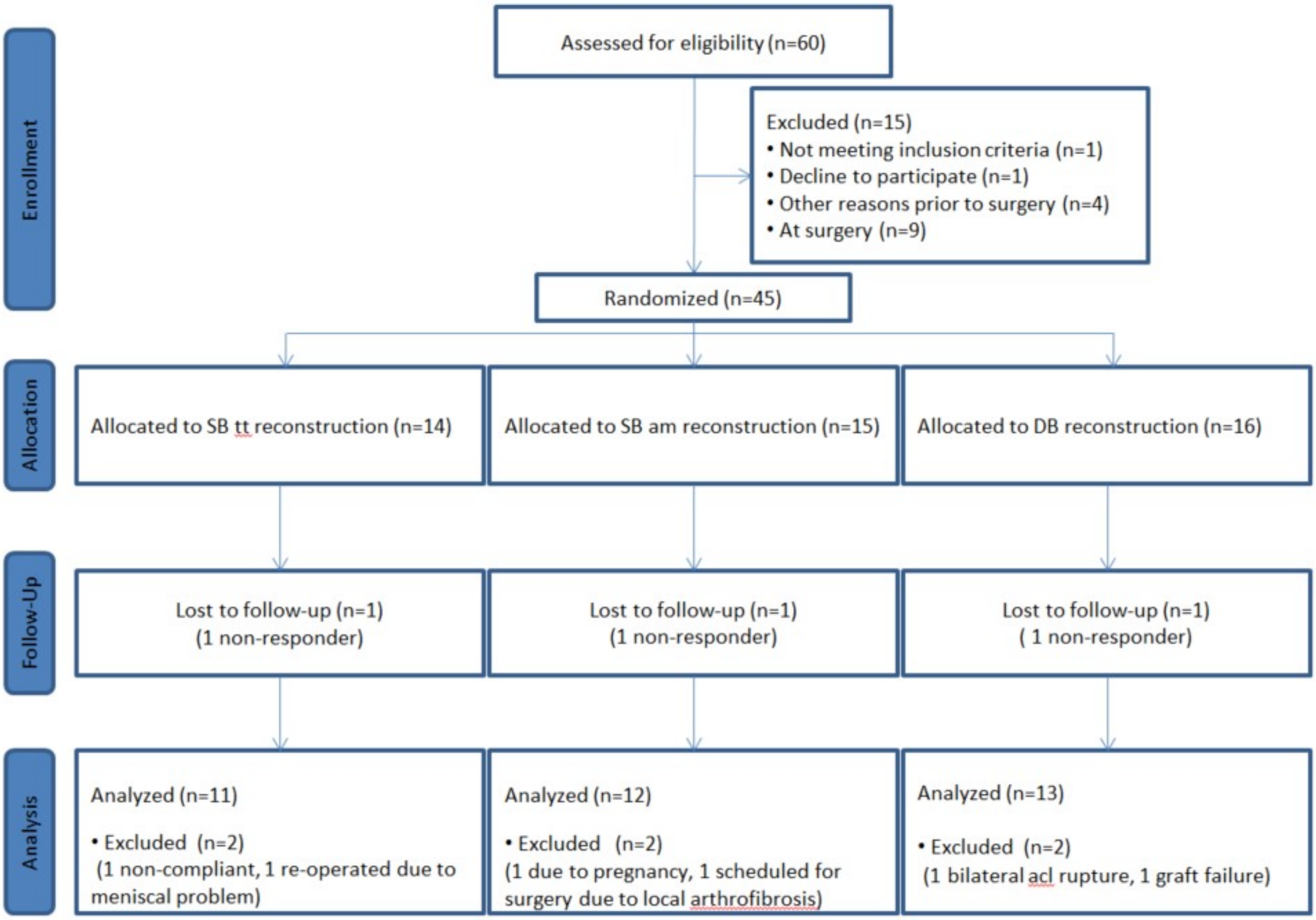

Figure 14: CONSORT flowchart used in study 2.2 (Paper C)[109] .

4.2.3 Randomization and reconstructions

Randomization was done at surgery. A diagnostic arthroscopy was performed initially and, if no exclusion lesions were present, the patients were randomized with the closed envelope method to one of three specific ACL reconstruction techniques:

1) Traditional single-bundle transtibial technique (SB-TT)

2) Anatomic single-bundle anteromedial technique (SB-AM)

3) Four-tunnel anatomic double-bundle technique (DB)

In the anatomic reconstructions the femoral tunnels were drilled using the AMP technique aimed at the native footprint. More specifically, placement of the femoral tunnel in the SB-AM was aimed at the middle of the ACL footprint while the femoral tunnels in the DB technique were aimed at the native insertion of each bundle.

4.2.4 Rehabilitation

Rehabilitation was the same in the three reconstruction groups. Pre-operatively and 14 days post-operation the patients were instructed by a physiotherapist at the hospital. One month post-operation, rehabilitation was continued for patients living in Aarhus at the Rehabilitation Center at Marselisborg Hospital (Marseliborg Centret).

4.2.5 Outcome measures

At the time of inclusion and one year post-operation, each patient was examined clinically with the Lachman test and the Pivot shift test and instrumentally with a KT1000. On the basis of this clinical examination, the patients were graded as normal (Grade A), nearly normal (Grade B), abnormal (Grade C), or severely abnormal (Grade D) according to the International Knee Documentation Committee objective knee examination form 2000 (IKDC 2000) (Appendix 1).

Four disease- or condition-specific questionnaires were completed pre- and postoperatively: KOOS, IKDC, Lysholm and Tegner (Appendix 2-5). Clinical performance tests were only performed at the one-year follow-up (Table 1).

Table 1: Pre- and postoperative outcome measures.

Outcome measures	Pre-operative	Post-operative
Objective clinical tests	Pivot shift Lachman test KT-1000	Pivot shift Lachman test KT-1000
PROMs	IKDC KOOS Tegner Lysholm	IKDC KOOS Tegner Lysholm
3-D motion analysis	Walk, run and pivot	Walk, run and pivot
Functional tests		Quadriceps strength Single-leg one-hop test Single-leg triple-hop test

4.2.6 Three-dimensional motion analysis protocol

Three-dimensional motion analysis was performed preoperatively and at one-year follow-up at the biomechanical laboratory at the Section of Sports Science, Aarhus University. Motion data were captured by eight optoelectronic motion capture cameras placed around a walkway (Fig. 15). Qualisys Tracking Manager (QTM) software was used to record the motion data and, subsequently, the data were exported to and analyzed by the Visual3-D software. Ground reaction forces (GFR) were recorded using a force plate, which was embedded in the walkway (and covered by a long carpet while walking and running).

Figure 15: The biomechanical laboratory at the Section of Sports Science, Aarhus University (private photo).

Two sets of markers were used during recordings. The first model was used as a static model to define anatomic planes and joint center positions relative to the clusters. The second marker model was used to measure dynamic movements (Fig. 16).

During the entire study, placement of markers on the participants was done by the author. At the beginning of the study, the system calibration and recordings were performed by the author, aided by two graduate students from the Section of Sports Science. Later on, a biomechanical engineer working at the Section of Sports Science carried out the calibrations, recordings and data analysis.

Fig 16: Left picture shows the marker model used for the static trial. Right picture shows the pivoting maneuver and the marker model used while performing the tasks [109, 110].

The motion analysis protocol was identical for patients and control subjects and consisted of three different tasks performed in the following order: (1) level walking, (2) running/jogging, and (3) stair descend followed by a 90° pivoting maneuver. The tasks were all performed at the participant's self-selected speed. Walk and run was performed along the runway and at least five successful trials (clean force plate contact) were recorded for each side and exercise.

The third task was conducted on a staircase placed just beside the force plate in the middle of the runway (Fig. 16, right picture). The staircase was constructed according to Andriacchi et al. [111] and had no handrail. Prior to this task, the subjects were carefully instructed in how to perform the pivoting maneuver: following contact with the force plate at the bottom of the staircase, the swing leg should be moved through a 90° arc in front of the stance leg and, finally, the foot of the swing leg should contact the ground in a 90° angle relative to the stance foot. Approval of the attempt required full plantar-side force plate contact while pivoting (Fig. 16, right picture). After pivoting, the subjects walked a few steps away from the force plate in one continuous movement. At least 10 successful trials for the ACL-intact knee, followed by at least 10 successful trials for the ACLD knee, were recorded. For the control group, ten successful trials were recorded for both legs.

4.2.7 Data reduction and definitions

For each trial and subject, the maximum tibial internal rotation was determined. To avoid potential outlier bias in further analyses, the second highest value of internal rotation was used together with the corresponding rotational moment. These values were used to calculate the rotational stiffness of each knee. Stiffness in a ligament is defined as the amount of force it takes to stretch the ligament a certain length and is calculated as follows:

Rotational stiffness: Δ Moment of force (Nm)/ Δ Angle rotation (degree)

Thus, each exercise provided a rotation, a rotational moment and a rotational stiffness for each knee tested. The results reported are means of the individually calculated values.

4.2.8 Statistics

The power calculation for the RCT was performed on the primary outcome (i.e., the tibial rotation measured by 3-D motion analysis). Previous studies demonstrated the tibial rotation in ACL-reconstructed knees (non-anatomic reconstruction technique) to be in the range 23–24° while intact knee tibial rotation was in the range 16–17° [97, 98, 100, 107]. Furthermore, the standard deviation for rotational measurements during 3-D motion analysis in these studies was around 4°. We hypothesized that an anatomic ACL reconstruction could reduce tibial rotation better than a non-anatomic ACL reconstruction with a 4° rotation reduction. Given these assumptions, nine subjects were needed per group to demonstrate significant difference with a power of 0.8. We chose to include 15 subjects per group to account for dropouts due to lack of follow-up and problems with data retrieval from 3-D motion analysis.

The number of patients included in Study 2.1 (the cross sectional study) was, therefore, based on the power calculation shown above. In this study, three parameters were evaluated: maximum internal tibial rotation, rotational moment and rotational stiffness. Each parameter was compared between the three groups (ACLD, ACLI and control group) for each of the three exercises performed (walking, running and pivot). Outcomes were analyzed in a repeated measurement ANOVA (mixed model). The analysis was performed with the STATA software, version 14 (StataCorp LP, Texas, USA). The significance level was set at $P < 0.05$. Bonferroni adjustments were not conducted in this study [112].

In study 2.2, the Chi-square test was used to evaluate differences between groups in the pivot-shift and Lachman test. Pre- and postoperative objective IKDC data were evaluated using the Fisher's exact test while the change in ratings from pre- to postoperative state was evaluated using a Wilcoxon signed rank test. An ANOVA test was used to evaluate differences between groups in terms of subjective outcome scores, KT-1000 measurements, quadriceps strength and one-leg hop tests. The pre-/postoperative comparison of these parameters was performed using a paired t-test. Three-dimensional motion analysis data were analyzed with a repeated measurement ANOVA test, and the Bonferroni correction was used when a significant difference was found. The significance level was set at $P < 0.05$.

4.2.9 Papers

This clinical study gave rise to two papers (Papers B and C). Paper B deals exclusively with preoperative data while Paper C presents the data from the RCT.

5. Results

5.1 Study 1 – Experimental study

Ultimate failure load

Comparing the ultimate failure load of the cortical fixation to the aperture fixation, we found that the average ultimate failure load of the EB groups was 30% higher than the interference screw (IF) groups. This difference was statistically significant ($P < 0.001$) [108].

Table 2 shows the ultimate failure load of both fixations and the different diameters tested. Comparing graft diameters within each fixation, the ultimate failure load of the SB 6 mm reconstructions were found to be significantly less than both the SB 9 mm and DB 2 × 6 mm reconstructions for both fixation methods. No significant difference was found between the ultimate failure load of the DB reconstructions and the SB 9 mm reconstruction for both fixation methods [108].

Furthermore, comparisons of equal diameters between fixation groups were conducted (EB 6 mm vs. IF 6 mm, etc.). None of these comparisons were significantly different. The EB 9 mm graft constructs showed a tendency of being stronger than the IF 9 mm construct ($P = 0.069$) [108].

Table 2: Failure load (RAW DATA)[108]

Fixation and bone tunnel diameter	Failure load (Mean ± SD (N))	95% CI (Range;N)
Endobutton 6 mm	568 (±169)	447-689
Endobutton 9 mm	969 (±177)*	843-1096
Endobutton 2×6 mm	1071 (± 244)++	896-1246
IF screw 6 mm	432 (±97)	362-502
IF screw 9 mm	708 (±230)#	544-872
IF screw 2×6 mm	806 (±167)§	686-926

Significant differences (EB, endobutton; IF, interference):
* EB 9 mm vs. EB 6 mm ($P < 0.001$),
++ EB 2 × 6 mm vs EB 6 mm ($P < 0.001$),
\# IF 9 mm vs IF 6 mm (P = 0.004),
§ IF 2 × 6 mm vs IF 6 mm ($P < 0.001$).

Elongation

Table 3 shows the elongation of the grafts after the cyclic test and load to failure test. Overall, no significant difference was found between the two fixation methods [108].

Within the IF group, the elongation after 1000 cycles of the DB reconstruction was shown to be significantly less than that of the SB 6 mm reconstruction. No significant difference was seen between the IF 9 mm and the IF 2 × 6 mm (P = 0.15). Within the EB group, no significant difference was seen between graft diameters (P = 0.78) [108].

Comparing equal diameters, a significant difference was seen between DB reconstructions (P = 0.05) [108].

Table 3: Elongation (RAW DATA)[108]

Fixation and bone tunnel diameter	Displacement after 1000 cycles (mm)	95% CI (Range;mm)	Displacement at failure (mm)	95% CI (Range;mm)
Endobutton 6 mm	3.5 (0.8)	2.9-4.0	11.3 (2.5)	9.7-12.9
Endobutton 9 mm	3.4 (0.8)	2.9-4.0	14.4 (1.8)	13.2-15.6
Endobutton 2×6 mm	3.2 (0.7)	2.7-3.7	16.3 (6.9)	12.0-20.6
IF screw 6 mm	4.1 (2.1)*	2.6-5.6	9.9 (1.7)	8.8-11.0
IF screw 9 mm	3.3 (1.5)	2.2-4.4	11.4 (2.5)	9.8-13.0
IF screw 2×6 mm	2.1 (0.6)§	1.7-2.5	10.6 (1.5)	9.7-11.5

Table 3 shows elongation/displacement after 1000 cycles; mean ± SD and 95% CI for each fixation and diameter. Displacement at failure includes displacement after preconditioning + displacement after a 1000 cycles + displacement after the load to failure test. Displacement at failure; mean ± SD and 95% CI for each fixation and diameter.
Significant differences:
*IF 6 mm vs IF 2 × 6 mm (P = 0.002).
§ IF 2 × 6 vs EB 2 × 6 mm (P = 0.05).

Stiffness

In general, stiffness is defined as the amount of force it takes to displace a given material a given length. In this study, stiffness is a measure of how much force (N) it takes to displace the graft/femur complex one mm and is defined as the slope of the linear region of the first and most steep part of the load-displacement curve in the failure test.

Overall, the stiffness in the IF group was shown to be 22% higher than the EB group (P < 0.001) [108].

Within the IF group, no significant differences were seen, although the IF 6 mm reconstructions showed a tendency to be less stiff than the IF 9 mm and IF 2 × 6 mm reconstructions (P = 0.112 and P = 0.126, respectively). Within the EB group, no significant differences were seen (Table 4) [108].

Comparing equal diameters, a significant difference was found between the SB 9 mm reconstructions (P = 0.009) [108].

Table 4: Stiffness (RAW DATA)[108]

Fixation and bone tunnel diameter	Stiffness (Mean ± SD (N/mm))	95% CI (Range; N/mm)
Endobutton 6 mm	241.7 (24.9)	226.3 - 257.1
Endobutton 9 mm	253.4 (64.1)*	213.6 - 293.2
Endobutton 2×6 mm	285.1 (51.2)	253.3 - 316.9
IF screw 6 mm	274.3 (65.5)	233.7 - 314.9
IF screw 9 mm	353.6 (77.8)	305.4 - 401.8
IF screw 2×6 mm	351.5 (70.9)	307.6 - 395.4

*EB 9 mm vs. IF 9 mm (P = 0.009)

Mode of failure

Graft slippage along the screw was seen in all IF mountings, while the cortical fixation most frequently was pulled though the cortex as the failure mode. Combined failure modes were seen in both groups (Table 5) [108].

Table 5: Failure mode[108]

Fixation and bone-tunnel diameter	Slippage past the screw	Graft failure	EB through cortex	Condyle fracture	Graft slippage cryoclamp	Bone torn out of cement
EB 6 mm		4	6			
EB 9 mm			10			
EB 2×6 mm		2	5	1	2	2
IF 6 mm	10					
IF 9 mm	10					
IF 2×6 mm	10			2		

EB: endobutton, IF: interference screw.

5.2 Study 2 – Clinical study

5.2.1 Study 2.1

Baseline findings

In total, 44 patients with a unilateral ACL lesion and 16 healthy subjects took part in this cross-sectional study. As shown in Table 6, no significant difference was seen in demographics between the patients and controls[110].

Table 6: Demographics of the ACLD patients and the control group [110]

	ACLD	Control	P-value
Sample size	44	16	
Sex (F/M)	18/26	6/10	0.12
Age (y)	25.7 ± 6.1	25.6 ± 3.6	0.98
Height (cm)	177.4 ± 9.0	178.6 ± 8.5	0.63
Weight (kg)	76.5 ± 14.1	73.7 ± 7.5	0.34
BMI (kg/m^2)	24.2 ± 3.2	23 ± 1.5	0.06

F: female. M: male. Age (years), height, weight and body mass index (BMI) are shown as mean ± SD.

Clinical evaluation

Clinical evaluation of all patients was done preoperatively. Objective IKDC rating showed that none of the patients were graded A (normal), 43% were graded B (nearly normal), 39% were graded C (abnormal) and 18% were graded D (severely abnormal) [110].

Knee laxity measured by KT-1000 showed an average difference between ACL-deficient and ACL-intact knees of 3.5 mm ± 2.2 mm [110]. Results from the pivot shift test were distributed as follows: 26% equal, 26% glide, 45% clunk and 3% gross. The Lachman test showed: 29% normal, 32% nearly normal, 21% abnormal and 18% severely abnormal.

Patient-reported outcome measures

Four questionnaires were completed by the patients and the average scores were 61 ± 11 for the IKDC; 73 ± 13 for the KOOS4, 72 ± 13 for the Lysholm and 3.8 ± 1.4 for the Tegner. The average score for the knee healthy controls were 97 ± 4 for the IKDC, 97.5 ± 2.8 for the KOOS4, 95 ± 9 for the Lysholm and 7.5 ± 1.9 for the Tegner [110].

Three-dimensional motion analysis

No significant difference was seen in tibial rotation between the ACL-deficient and ACL-intact knees during any of the three tasks performed (Table 7). During walking, the tibial rotation of the control group was significant higher than both the ACL-deficient and the ACL-intact knees while the control group displayed a significantly smaller tibial internal rotation during running compared to both the ACL-deficient and the ACL-intact knees. Pivoting did not show any difference in tibial internal rotation between the ACL-deficient, ACL-intact and control group knees [110].

Rotational moments were not significantly different between the ACL-deficient, ACL-intact and control group knees during walking and pivoting. During running, the rotational moment of the ACL intact knees were significantly higher than both the ACL-deficient knees and the control group (Table 7) [110].

Rotational stiffness was not significantly different during walking and running. A tendency toward a higher rotational stiffness in the ACL-deficient knees compared to the control group knees was seen during walking (P = 0.098). During running the ACL-intact knees showed a tendency towards a higher rotational stiffness compared to the control group (P = 0.062). No significant difference was seen in rotational stiffness between the ACL-deficient, ACL-intact and control group knees while pivoting (Table 7) [110].

Table 7: Kinematic and kinetic data from the 3-D motion analysis[110]

		ACLD Mean (95% CI)	**ACLI** Mean (95% CI)	**Control** Mean (95% CI)
WALKING	**Rotation**	9.2[1] (7.8-10.7)	8.6[2] (7.1-10.1)	14.2 (11.9-16.5)
	Moment	0.038 (0.02-0.56)	0.028 (0.01-0.044)	0.04 (0.02-0.059)
	Stiffness	0.0067 (0.002-0.01)	0.006 (0.002-0.01)	0.0027 (0.001-0.004)
RUNNING	**Rotation**	18.9[3] (15.3-22.5)	20.5[4] (16.9-24.1)	13.9 (12.1-15.6)
	Moment	0.14[5] (0.09-0.19)	0.25[6] (0.17-0.32)	0.1 (0.07-0.13)
	Stiffness	0.008 (0.003-0.013)	0.016 (0.007-0.024)	0.0075 (0.005-0.0096)
PIVOTING	**Rotation**	28.2 (23.9-32.4)	27.9 (25.2-30.6)	29.7 (28.4-31.1)
	Moment Median (range)	0.164 (0.06-0.33)	0.168 (0.11-0.44)	0.164 (0.05-0.43)
	Stiffness Median (range)	0.0061 (0.002-0.015)	0.0062 (0.004-0.016)	0.0056 (0.002-0.012)

ACLD: ACL-deficient knee. ACLI: contralateral ACL-intact knee. Control: healthy knee control group. Rotation: tibial internal rotation, expressed in degrees (deg). Moments: net knee joint external moments, expressed as Nm/kg. Stiffness: rotational stiffness, expressed as (Nm/kg)/deg. Means and 95% confidence intervals (CI) are reported. Pivoting/rotational moments and pivoting/stiffness were log transformed during statistical analyses; therefore, median and range are reported for these parameters.
([1]ACLD vs. Control p < 0.001, [2] ACLI vs. Control p < 0.001, [3]ACLD vs. Control p = 0.014, [4]ACLI vs. Control p = 0.001, [5]ACLD vs. ACLI p= 0.015, [6] ACLI vs. Control p < 0.001).

5.2.2 Study 2.2

Baseline findings

In total, 45 patients with a unilateral ACL lesion and 16 healthy subjects took part in this RCT. No significant differences were seen in demographics between the control and randomization groups (Table 8) [109].

Table 8: Demographics of the randomized groups and the control group [109].

	DB	SB-AM	SB-TT	Control group	*P*-value
Sample size	16	15	14	16	
Sex (F/M)	5/11	7/8	6/8	6/10	ns
Age (y)	26.5 ± 6.4	24.3 ± 4.9	27.5 ± 7.2	25.6 ± 3.6	ns
Height (cm)	179 ± 8	174 ± 8	179 ± 9	178 ± 8	ns
Weight (kg)	78.7 ± 13.7	75.7 ± 15.1	74.9 ± 14.4	73.7 ± 7.5	ns
BMI (kg/m^2)	24.7 ± 3.3	24.6 ± 3.1	23.1 ± 3.1	23 ± 1.5	ns
Injury-surgery (months)	6 (2-26)	6 (3-16)	12 (4-42)		ns

F: female. M: male. Age (years), height, weight and body mass index (BMI) are shown as mean ± SD. Time from injury-surgery is shown as mean and (range).

Clinical knee laxity

No significant difference in objective IKDC was found between the three reconstruction groups preoperatively (p = 0.08) or at follow-up (p = 0.78). A significant difference was seen in objective IKDC grading from preoperative to follow-up (p < 0.001) (Table 9).

No significant differences were seen in clinical knee laxity between reconstruction groups, as evaluated by the KT-1000, the Lachman test or the pivot-shift test at follow-up (Table 9). Comparisons of preoperative and follow-up data for the pivot-shift and Lachman test revealed significant improvements for all three reconstruction groups (Table 9). Furthermore, KT-1000 measurements showed a significant difference in the DB group when comparing preoperative and follow-up status [109].

Patient reported outcome measures

Four different PROMs were completed preoperatively and at follow-up (subjective IKDC, KOOS4, Tegner and Lysholm scores). Significant improvements were reported for all reconstruction groups in all of these PROMS at follow-up (P < 0.01). No significant differences were seen between reconstruction groups at follow-up in either of these PROMs (Table 9) [109].

Quadriceps strength and hop test

No significant difference was seen between the reconstruction groups in terms of postoperative quadriceps strength (P = 0.12), single hop-test (P = 0.84) or triple hop-test (P = 0.2) (Table 9) [109].

Table 9: Preoperative and follow-up results of objective IKDC, clinical findings and subjective outcome scores [109].

	Preoperative DB	SB-AM	SB-TT	Follow-up DB	SB-AM	SB-TT
IKDC A Normal (%)	0	0	0	38	25	27 (*)
IKDC B Nearly normal (%)	33	73	21	54	75	73
IKDC C Abnormal (%)	47	20	50	8	0	0
IKDC D Severely abnormal (%)	20	7	29	0	0	0
KT-1000 (max) mm	3.7 ± 1.9	3.9 ± 2.6	2.8 ± 2.1	1.6 ± 2.1*	2.3 ± 1.9	2.0 ± 1.7
Pivot-shift test Normal (%)	13	38	29	92*	75*	82*
Lachman test Normal (%)	20	47	21	85*	75*	73*
Subjective IKDC	63 ± 11	58 ± 13	62 ± 11	76 ± 11*	71 ± 15*	76 ± 13*
KOOS4	64 ± 14	57 ± 13	64 ± 12	78 ± 13*	73 ± 18*	73 ± 13*
Tegner score	3.6 ± 1.0	3.7 ± 2.1	3.9 ± 0.9	5.5 ± 1.4*	5.6 ± 1.2*	5.5 ± 1.0*
Lysholm score	73 ± 15	70 ± 9	73 ± 14	87 ± 14*	81 ± 14*	86 ± 12*
Quadriceps strength (% of normal leg)				103 (15)	88 (18)	94 (16)
Single hop (% of normal leg)				91 (13)	95 (17)	93 (14)
Triple hop (% of normal leg)				92 (6)	91 (7)	97 (10)

(*) The objective IKDC grading improved significantly from preoperative to follow-up (P < 0.0001).
* Significant differences from preoperative to follow-up state (P < 0.05).

NB! The headings SB-AM and SB-TT have been switched in Paper C (Table 2)!

Three-dimensional motion analysis

No significant difference was found between the three reconstruction groups regarding tibial rotation or stiffness for walking, running and pivoting (Table 10). Furthermore, no significant differences were seen among the three different reconstruction groups and the two control groups during any of the three tasks [109].

Table 10: Three-dimensional motion analysis data at follow-up [109].

		SB-TT	SB-AM	DB	Intact knee	Control group
Walking	**Rotation**	13.7 (6.1)	12.7 (4.9)	16.4 (4.6)	15.7 (4.9)	13.6 (6.2)
	Stiffness	0.0035 (0.0049)	0.0043 (0.0034)	0.0033 (0.0029)	0.0042 (0.0053)	0.0029 (0.0052)
Running	**Rotation**	20.5 (7.8)	15.7 (3.2)	20.4 (3.6)	17.3 (5.6)	18.9 (5.3)
	Stiffness	0.0057 (0.0062)	0.0062 (0.0055)	0.0059 (0.0032)	0.0057 (0.0050)	0.008 (0.0069)
Pivot	**Rotation**	30.4 (5.4)	31.7 (7.7)	31.1 (5.5)	31.4 (6.2)	27.9 (5.3)
	Stiffness	0.0051 (0.0014)	0.0038 (0.0019)	0.0042 (0.0015)	0.0042 (0.0016)	0.0071 (0.0035)

This table presents data for the three reconstruction groups, intact knees of the patients and the control group. Tibial rotation is the maximal internal tibial rotation during walk, run or stair descend/pivot expressed in degrees. Stiffness is defined as moment of force/rotation and is presented with the unit (Nm/kg)/deg.

The difference in maximum tibial internal rotation during the pivoting task from before to after surgery, expressed as a mean of the differences for every subject (calculated as Diff = ACLD - ACLR), revealed no significant difference between the three reconstruction groups (P = 0.35). The mean differences were 0.8° in the DB group, -2.8° in the SB-AM group, and 2.9° in the SB-TT group. Furthermore, the mean difference in tibial internal rotation between the ACLI knees (Diff_ACLI = pre-ACLI – post-ACLI) was not significantly different among the reconstruction groups (P = 0.82) [109].

6. Discussion

6.1 Methodological considerations and limitations

Study 1

In Study 1, porcine bones and tendons from five-month-old animals were used in an experimental setup. In Denmark, pigs are slaughtered when they weigh 95 kilograms, which corresponds to an age of approximately five months. The pigs are not full grown at this age. In contrast, Nagarkatti et al. tested 24-month-old pigs and found that they had a bone mineral density (BMD) (measured by Dual X-ray Absorptiometry = DXA) comparable to young humans [113]. Consequently, one could speculate that the BMD of the porcine bones used in our study (which was measured by peripheral quantitative computed tomography = pQCT) was lower than the specimens used by Nagarkatti. Comparison of BMD measurements from a pQCT scanner and a DXA scanner is, however, not directly comparable. To our knowledge, BMD means for specific age groups are only available for DXA scans. Interestingly, Nagarkatti et al. showed that BMD is related to fixation strength. Thus, the ultimate failure loads measured in our study might be underestimated due to the potentially lower BMD in the five-month-old pigs.

The loading protocol used in this study represents early rehabilitation loads placed on the knee in the initial postoperative period, and finally, an event causing failure of the ligament [114]. The same loading protocol is used in several other studies [74, 114]. The graft is, however, only pulled in one direction in this specific setup. It would be interesting to test the mountings when subjected to rotational forces as well. Thus, the results should be interpreted with care, as they do not entirely reflect real-life biomechanical conditions.

Study 2

Study design and population

In this study, 60 patients were assessed for eligibility and 54 of these patients were tested in the motion analysis laboratory. Nine patients were excluded at surgery, mostly due to suturing or damage to the meniscus (resection of more than 50%). The remaining 45 individuals were randomized. Therefore, the cohort in Study 2.1 was selected by the inclusion criteria, which ensured that the included patients did not have extensive damage to other stabilizing structures in the knee.

Patient inclusion was carried out at the Clinic of Sports Traumatology, Aarhus University Hospital while 3-D motion analysis was carried out in the motion analysis laboratory at the Section of Sports Science, Aarhus University. These institutions are situated at two different addresses. Selection bias might have occurred as the patients had to accept the motion analysis and transportation to the motion analysis laboratory. However, as the patient group was young and mobile, transportation to the motion analysis laboratory was not a significant issue for this group. Four patients were, however, excluded as it was not possible to schedule the motion analysis before surgery.

The ACL reconstructions for this study were performed by six different surgeons. Four of these were senior surgeons while two were staff specialists. Conducting our study, the AMP technique was quite new at our institution and the DB technique was not offered as a standard operation. Hence, a learning curve could be expected on these two methods, which possibly could affect the outcome in a negative way.

In this study, the patients and the physiotherapist who conducted the follow-up examinations were blinded to the intervention. Blinding is a critical methodological feature of RCTs. Although randomization minimizes selection bias and confounding, its use does not prevent subsequent differential co-interventions or biased assessment of outcomes [115]. At follow-up, the clinical and performance-based tests were conducted by the blinded physiotherapist in cooperation with a senior surgeon. Only when all tests were completed, were the patients (and the physiotherapist) told which ACL reconstruction they had received.

The number of patients lost to follow-up was equal in each of the three groups. The reasons for discontinuation were, however, different between the groups.

Power calculation

The power calculation for the RCT in this PhD thesis was based on former studies from a Greek research group, who had measured tibial rotation in ACL-reconstructed subjects by use of 3-D motion analysis [97, 98, 100, 107]. However, our results at follow-up differed considerably from the means and SD used for the power calculation. Hence, our measurements of tibial rotations and standard deviations were higher for all three ACL reconstruction methods and intact knees than those reported by the Greeks. Thus, it is possible that our sample size is too small.

The SD retrieved from our data could reflect a higher biological variation among our study population. Furthermore, we defined the neutral position of the knee (0° knee rotation) as the knee angle five frames before foot contact. In this position, the knee is close to full extension and no external moments affect the limb. This definition of the neutral position is different from other studies, where the neutral position is recorded during a static trial, and this could possibly affect the range of angular displacement.

Patient-reported outcome measures

In a recent survey, questionnaires of patient-reported outcome measures (PROMs) have been identified as an important tool for measuring outcomes after ACL reconstruction [116]. However, a wide range of PROMs are used in ACL research, which makes it difficult to compare results across trials. As none of these PROMs are ACL-specific, we chose to include several PROMs in our study in order to cover most possible aspects of the disabilities associated with ACL injury. Thus, anatomic-specific (IKDC, KOOS, Lysholm) as well as a sport-specific (Tegner Activity Scale) PROMs were included. Preoperatively, the PROMs were collected and/or completed at the time of the preoperative motion analysis and the response rates of the four PROMs ranged from 96–100%. At follow-up, the PROMS were completed in the clinic (digitally and on paper) with a response rate of 81–97% (81% completed the KOOS while 97% completed the IKDC, Lysholm and Tegner scores).

In our study, functional performance tests were performed at follow-up only. It would have been interesting to have tested the patients preoperatively also, as knee function at the time of surgery is an important indicator of the expected outcome after ACL reconstruction [90, 92].

Three-dimensional motion analysis

Three-dimensional motion analysis has the advantage of being able to measure dynamic movements in all three planes. However, a known drawback of 3-D motion analysis is the use of skin markers, which are displaced by the elastic human skin relative to the underlying anatomic structure during movements. To reduce the influence of skin motion on kinematic data, clusters of markers mounted on light-weight, thin plates were used in our study [94, 117]. Cluster markers are especially useful for measuring optimized rotational measurements in 3-D motion analysis.

In Study 2, the placement of skin markers and the supervision of all of the patients were performed by the author. Recordings were performed by four different persons, which might have affected the quality of the recordings. Hence, the number of recordings usable for analysis increased postoperatively, with only one lab worker performing all of the recordings.

The assessment of the rotational laxity and the rotational stiffness of the knee using 3-D motion analysis were emphasized in our clinical study. Although the reliability of transverse plane measures using 3-D motion analysis has been questioned [118], Tranberg et al. showed that, during simultaneous knee motion measurement using an optical tracking system and dynamic radiostereometric analysis (RSA), the internal/external rotations were fairly similar up to 25° of flexion [119]. RSA is considered the 'gold standard' in the investigations of joint motions due to its high accuracy, high resolution and detailed documentation. Furthermore, RSA is not influenced by soft tissue artifacts [120]. Additionally, Webster et al. showed that the within-day and between-day measures of tibial rotation were repeatable during a pivoting task similar to ours. This study also found that rotational excursion was more repeatable than peak rotation [118].

In this study, all participants conducted the tasks at self-selected speeds. Walking and running (and stair descending) speeds do, however, affect joint moments. Hence, with a given joint stiffness, provided by the joint structures, an increased moment will cause an increased rotation. By expressing stability as stiffness, i.e., normalizing the magnitude of the moment causing the rotation, comparisons among different moments (caused by different movement speeds) becomes possible.

Finally, the rehabilitation compliance of the patients was not documented. Thus, the amount of training performed potentially differed a lot among the included patients, which could have affected their stabilizing strategies during movements and hence their rotational stiffness. Additionally, it would have been interesting to record EMG activity during all three tasks to describe possible gait adaptions.

6.2 Results and findings

In 2017, the treatment and assessment of ACL injuries remain a matter of debate and the object of ongoing evolution and research. Historically, ACL reconstruction has developed extensively from open surgery to minimally invasive arthroscopic techniques [28], from non-anatomic to anatomic techniques [3] and even toward individualized ACL reconstructions [121]. According to Freddie Fu, there has been a paradigm shift in ACL reconstruction [3]. Thus, current trends in ACL reconstruction have changed since this thesis was described [122].

Fixation of soft tissue grafts in ACL reconstruction

The biomechanical properties of two fixation methods used for femoral soft tissue graft fixation were investigated in Study 1. This study showed that the ultimate failure load of the EB was significantly higher than that of the interference screw, that elongation after 1000 cycles did not differ significantly between fixations and that the stiffness of the interference screw fixations were significantly higher than that of the EB. In addition, Study 1 showed that failure loads of 6-mm graft constructs were up to 40% less than those of 9-mm and 2x6-mm graft constructs regardless of fixation. Furthermore, we found that the failure load, elongation and stiffness did not differ significantly between 9-mm graft and 2x6-mm reconstructions for both fixations. Thus, our hypothesis was partly confirmed.

In the literature, the biomechanical properties of the BIS and EB used as femoral fixation have been investigated by several other authors in experimental settings [74, 76, 123, 124]. Similar to our findings in Study 1, these studies showed a higher ultimate failure load of the EB fixation compared to that of the BIS [76, 123]. Furthermore, Lehmann et al. compared single-bundle (SB) to double-bundle (DB) ACL reconstructions and reported significantly better outcomes (higher ultimate failure loads and stiffness, and lower elongation) in DB reconstructions with a cortical button fixation compared to SB reconstructions using an 8-mm graft construct [123]. The same trend was seen in our study for both fixation methods. Additionally, a trend toward the poorer structural properties of the graft constructs of smaller diameters (BIS and EB) was seen in our study, which raises concern about the survival of grafts of smaller diameters for DB reconstructions. The failure loads of the 6-mm-graft-construct tested in our study did, however, exceed rehabilitation loads placed on the entire knee for both BIS and EB [114]. Thus, Trump et al. estimated the applied loads on the knee of a 70-kg individual during walking, jogging and stair descent to be 150 N, 250 N and 340 N, respectively [114].

It is well recognized that graft fixation is the weakest link in the early postoperative period after ACL reconstruction [125]. The overall goal of the graft fixation technique is to ensure a stable and strong graft fixation so that effective graft-to-bone healing occurs. Meanwhile, the fixation should be able to withstand strains posed on it during the postoperative rehabilitation [126]. Rehabilitation after ACL reconstruction has become more “aggressive” in the past decade and therefore potentially challenges the healing process of the graft in the bone tunnel. Thus, additional aspects of biomechanical properties are important in the evaluation of fixation methods, such as the ingrowth of grafts, clinical outcomes and the effects of reconstruction techniques.

Recent research has addressed the issue of graft-to-bone healing with an emphasis on soft tissue grafts fixed with BIS or EB in a large animal model [126]. Interestingly, histologic assessment from this study showed a significantly better graft incorporation with four-zone direct healing to bone for the grafts using suspensory fixation compared with grafts using interference screw fixation (12 weeks postoperative) [126]. The findings argue against the bungee or windshield-wiper effect of suspensory fixation.

Furthermore, the clinical results of the interference screw and the extra-tunnel fixation of soft tissue grafts have been reported in several studies [127-129]. Overall these studies do not report any difference between fixation methods in terms of instrumented knee laxity, subjective IKDC scores or objective IKDC examination ratings. Interestingly, Lubowitz et al. reported that 100% of the patients reconstructed with a suspensory fixation were graded normal after 24 months, whereas only 86% were graded normal in the interference screw fixation group [128]. Although this difference was not statistically significant, it might be clinically important. Furthermore, Colvin et al. reported a trend toward fewer surgical failures with the use of an interference screw for the femoral fixation of hamstring autografts, whereas Ilahi et al. reported no difference in the graft failure rate between the fixation methods [127, 129].

Finally, one has to keep in mind that surgical reconstructive techniques have changed over the years, especially the placement of the femoral tunnel. This is important, as fixation strength has been shown to increase with increasing divergence between the tension angle and the femoral tunnel [130]. In our study, a pull/loading angle of 60° was chosen. This setup reduces the force imposed on the intra-articular graft as the corner of the tunnel creates a frictional effect. Theoretically, both fixation techniques used in Study 1 would benefit from this effect. One could speculate if this is also the case in an anatomic reconstruction technique in vivo, where the femoral tunnel(s) are placed more inferiorly on the lateral femoral condyle. Additionally, pre-tensioning of the graft before tibial fixation has become good practice. Pre-tensioning allows the graft to reach its final elongation before fixation, and this could potentially augment the stiffness of the ACL reconstruction [113]. In theory, femoral fixation with both BIS and EB would benefit from these initiatives.

In summary, advantages and disadvantages exist for both femoral fixation methods compared in Study 1. The suspensory extra-cortical fixation technique has, however, gained popularity in recent years [122]. It is a fixation technique that has the advantage of being easy to perform and furthermore can be performed by a single surgeon during ACL reconstruction. Additionally, recent research has rejected some of the concerns attached to the suspensory fixation as outlined above.

Rotational stability of the knee

Dynamic biomechanical data on knee rotation are most often reported as laxity, i.e., in degrees [97, 99, 131]. In Study 2 (Paper B & Paper C), we used the parameter of "rotational stiffness," which is calculated as: Δ rotational moment (N/m)/ Δ angle rotation (degrees). Rotational stiffness combines kinematic and kinetic measures and expresses the ability of both passive and active structures of the knee to resist rotation. To our knowledge, this measure has not been used before in biomechanical in vivo studies using 3-D motion analysis. Rotational stiffness in a static setup is, however, reported by Louie et al., who showed an increase in rotational stiffness of more than 400% though the activation of the muscles around the knee [132].

Our main findings in Study 2.1 were that the rotational stiffness did not differ significantly between the ACL-deficient knees, the ACL-intact knees and the control group during walking, running or pivoting. Furthermore, the tibial internal rotation did not increase in the ACL-deficient knees compared to the ACL-intact knees during any of the three tasks. Rotational moments did only differ significantly between knees during running, where the ACL-intact knees displayed a higher rotational moment than did both the ACL-deficient knee group and the control group. Thus, our hypothesis was not confirmed in this study.

In Study 2.2, the most important findings were that no difference in tibial internal rotation or rotational stiffness was found between the non-anatomic and the two anatomic ACL reconstruction techniques during walking, running or pivoting at one-year follow-up. Furthermore, the tibial internal rotation did not differ from the preoperative state to one-year postoperative for any of the three reconstruction techniques compared. In addition, clinical tests, subjective outcome measures and objective IKDC ratings did not show any difference at follow-up between the non-anatomic and the two anatomic ACL reconstructions compared. Thus, our findings did not confirm our hypothesis in this study, either. A significant difference from pre- to postoperative state was seen in the Lachman test, the Pivot shift test, the objective IKDC rating and the four PROMs, which all improved significantly. KT1000 measurements improved significantly only in the DB group from the preoperative state to follow-up.

In terms of the rotational laxity measured by 3-D motion analysis, several studies have compared SB to DB reconstructions[131, 133-137]. Similar to our findings, no difference in rotational laxity between SB and DB reconstruction are reported in these studies. However, these studies did not analyze the same biomechanical tasks as in our study. Interestingly, Hantes et al. reported DB ACL-reconstructed knees to demonstrate significantly better control of tibial rotation when fatigued [134]. Moreover, Wang et al. reported the biomechanical impact of TT and AMP femoral tunnel drilling in SB ACL reconstructions during walking [138]. This study reported a better normalization of AP translation and tibial rotation during walking for knees reconstructed with the AMP technique. We could not replicate the findings in our study. Finally, Tsarouhas et al. reported moments of force during rotation in healthy controls, ACL-deficient knees and SB and DB reconstructed knees[135, 136] and found the applied rotational moment on the affected side to be constantly lower than that on the unaffected side in all groups, which differs from our results.

Study 2 also showed a tendency toward a decrease in rotational stiffness from the preoperative to postoperative status for all three tasks (walking, running and pivoting) and in both legs of the patients. This is partly explained by a tendency toward higher tibial internal rotation postoperatively for all tasks and both legs. Additionally, lower rotational moments were measured during running and pivoting at follow-up. These findings were quite surprising. A possible biomechanical explanation could be that, due to the surgical reconstruction, the patients feel more confident about the reconstructed knee and therefore do not activate the stabilizing muscles around the knee to the same extent. As a consequence, a higher tibial rotation occurs compared to the preoperative state. According to this theory, the patients reconstructed with the TT technique have the most unstable knees during pivoting, as they develop the highest rotational moments (and therefore the highest stiffness) to prevent symptoms of instability. Interestingly, Hofbauer et al. also reported kinematic changes in both knees of ACL-reconstructed subjects similar to ours [139]. Hofbauer states that "these kinematic adaptions could have important implications for postoperative care, including evaluating optimal timing of return to sports and the development of bilateral neuromuscular rehabilitation programs that may improve patient outcome and reduce re-injuries in both short and long term" [139].

Interestingly, a subgroup of ACL-deficient patients were identified in Study 2 (Paper B) based on 3-D motion analysis measurements. We found that approximately one-third of our patients rotated less with the ACLD knee compared to the ACLI knee during pivoting, and furthermore, they used a smaller rotational moment to do so. It could be speculated that these patients were so-called copers, with copers being defined as ACL-deficient persons with no symptoms of knee instability even with sports involving cutting and pivoting. Copers can, however, be quite difficult to differentiate from patients at risk of continued instability symptoms (non-copers), who would benefit from ACL reconstruction [140]. Interestingly, Rudolph et al. states that, if copers are included in the mix of all subjects, the genuine differences in movements would be obscured [101]. Hence, in our RCT, it was assumed that all patients were non-copers, as they presented with symptoms of instability. It is possible that the aforementioned subgroup used different muscle activation patterns in their ACLD and ACLI legs, indicating some kind of adaptive or stabilizing strategy, which produced an overall equal stiffness of the knees. This is, however, clearly speculative, as we did not measure muscle activation patterns.

Typically, muscle activation patterns are measured by Electromyography (EMG). EMG measurements were not obtained in Study 2, but such data could have added important information on adaptive gait strategies. Functional adaptions to ACL injury are often observed as alterations in the patterns of extensor and flexor muscle activations (timing and magnitude) at the knee during ambulation [141]. Thus, various and complex gait strategies related to the ACLD are described in the literature, based on 3-D motion analysis measurements during walking [101-103, 142]. An example is the "quadriceps avoidance gait," as described by Georgoulis, which is characterized by an important reduction or absence of the external knee flexor joint moment during the mid-stance phase of the gait cycle. Theoretically, this reduces the anterior force applied to the tibia by the eccentric contraction of the quadriceps (net extensor force) at low knee flexion angles that could lead to the anterior displacement of the tibia in relation to the femur [103]. However, more recent studies have proposed that this reduction in the external flexor moment could instead be explained by increased hamstring activity. These findings demonstrate different adaptive strategies in the ACL population and indicate that adaptations to the patterns of muscle firing could compensate for the loss of the ACL [102, 142, 143].

Yet another gait strategy was introduced by Fuentes et al., who reported a significantly reduced internal rotation knee joint moment and larger knee flexion angles during the terminal stance phase of the gait cycle in the ACLD group compared to the control group [142]. According to Fuentes et al., these biomechanical compensations allowed the patients to avoid a condition mimicking the first part of the lateral pivot-shift maneuver, and they were named the "pivot-shift avoidance gait." Fuentes et al. concluded that the ACLD subjects possibly adopted the "pivot-shift avoidance gait" to prevent anterolateral rotational knee instability.

Additionally, Rudolph et al. studied ACLD subjects (copers and non-copers) using 3-D motion analysis and EMG [101]. This study showed that non-copers achieved peak hamstring activity later in the weight acceptance phase and used a strategy involving more generalized co-contraction. Furthermore, this study showed that both copers and non-copers had high levels of quadriceps femoris muscle activity and concluded that the reduced knee moment in the involved limbs of the non-copers did not represent "quadriceps avoidance" but rather represented a strategy of general co-contraction with a greater relative contribution from the hamstring muscles. Interestingly, a recent study by Zabala et al. has shown that knee mechanics change over time in ACL-deficient patients with no sign of osteoarthritis, and it concludes that the time since injury is an important factor [144]. These findings might explain the different gait adaptions mentioned above.

Anatomic ACL reconstruction

In recent years, the placement of femoral tunnels for ACL reconstruction has been intensively debated and researched. Interestingly, better rotational stability of the knee has been reported when the femoral tunnel is placed with the AMP technique compared to the TT technique, in both cadaveric and clinical studies [145-147]. In the review by Chen et al., the anatomic SB technique (using the AMP technique) was found to be superior in terms of surgeon-recorded stability (Lachman test, pivot-shift test and objective IKDC) [147]. These findings differ from our results in Study 2 (Paper C). However, a recent national cohort registry-based study reports an increased risk of revision in SB reconstructions performed with the AMP technique compared with the TT technique [148]. This study by Rahr-Wagner et al. concludes that their findings might be explained by technical failures resulting from the introduction of the new and more complex AM portal technique. The data of Study 2.2 are, in fact, included in this study. Updated register-based studies of the anatomic single-bundle technique have not been published since.

Interestingly, the aforementioned review from Chen et al. does not report any difference in PROM (Lysholm score), which is similar to our results. This mismatch between PROMs and the clinical results are seen in several other studies, too [140, 146], and calls for a gold-standard definition of a successful outcome after ACL injury and reconstruction. According to Lynch et al., who conducted a survey to define a successful outcome one and two years after ACL injury or reconstruction, consensus was reached on six measures of knee function. These measures were: the absence of knee joint effusion, the absence of knee joint giving way, symmetrical quadriceps and hamstring muscle strength, patient satisfaction measured by PROMs and return to sports [116]. Study 2 (Paper C) reports on four of these six measures. Surprisingly, laxity measures, functional testing and measures of osteoarthritis did not achieve consensus in the survey [116].

In the past decade, several clinical studies have compared double-bundle to single-bundle ACL reconstructions [146, 149-151]. Roughly two tendencies are seen among these studies. Some studies find the double-bundle technique to be superior [146, 150, 152-156], whereas others report few potential benefits of DB reconstruction in terms of the laxity measured by clinical tests or subjective patient-reported outcome measures [121, 149, 151, 157-159]. Our results from Study 2 (Paper C) are in accordance with the latter part, which do not find any significant differences in knee stability between SB and DB reconstructions. We did, however, find a tendency of the DB reconstructions to display a better outcome than both SB techniques in several of the parameters measured.

As mentioned earlier, anatomic ACL reconstruction techniques have been studied extensively during the past decade [3, 35]. However, it is crucial to keep in mind that many different techniques of anatomic ACL reconstructions have been presented over the years, which makes it difficult to compare the literature on this subject[159]. As a consequence, a survey-based “anatomic ACL reconstruction scoring system” was published by Van Eck et al. in 2013 [160]. This scoring system is designed to grade ACL reconstruction procedures for individual patients and furthermore for the comparative evaluation of the descriptions of surgical methods in published studies on anatomic single- and double-bundle ACL reconstruction [3]. The surgical technique used in the RCT included in this thesis was conducted before the creation of this “anatomic ACL scoring system.” Thus, the anatomic technique used in Study 2 (Paper C) does not fulfill the entire “anatomic checklist” [160].

In summary, the superiority of one anatomic reconstruction method over another has not yet been established in clinical studies. Additionally, recent experimental studies comparing anatomic SB and DB techniques reveal no difference in AP and rotational stability [161, 162]. Furthermore, the presence of OA five years after reconstructive surgery has been shown not to differ between anatomic SB and DB in a recent study [159]. In fact, this study reports a significant increase in OA within the DB group at five-year follow-up [159]. Randomized clinical trials performed according to the aforementioned anatomical scoring system and with longer follow-ups are, however, still lacking.

7. Conclusion

Study 1

In conclusion, the EB fixation of tendon grafts in ACL reconstruction was stronger that interference screw fixation. Furthermore, single-bundle 9-mm reconstructions were found to have equal biomechanical properties as double-bundle 2 x 6-mm ACL reconstructions regardless of the fixation method used. These findings indicate that double-bundle reconstruction can be used as an alternative to single-bundle reconstruction based on biomechanical properties. Furthermore, our study demonstrated that a single 6-mm double-stranded graft was 40% weaker than a single 9-mm four-stranded graft. These findings indicate a potential higher risk of graft failure in double-bundle reconstructions where only one graft strand is loaded during range of motion.

Study 2.1

In this cross-sectional study, dynamic 3-D motion analysis during walking, running and pivoting did not display a uniform picture of the ACL-deficient knee being more lax and less stiff than the ACL intact knees (contra-lateral intact knee and a knee healthy control group).

Study 2.2

This randomized controlled study did not reveal any significant differences in tibial internal rotation and stiffness measured by 3-D motion analysis, clinical laxity tests, patient-reported outcome measures or functional performance tests between anatomic single-bundle, anatomic double-bundle and non-anatomic single-bundle ACL reconstruction techniques at one year follow-up. Patient-reported outcome measures, objective IKDC grading, the Lachman test and the pivot-shift test improved significantly from the pre- to postoperative state in all three reconstruction groups.

These results should, however, be interpreted with caution because the study population might be too small (the assumptions related to angular displacement and standard deviations used in the sample size calculation differed from the final measurements).

8. Perspectives and future research

The findings of the present PhD thesis indicate that the stiffness or elasticity of the knee is modulated by other structures around the knee, which are partly able to compensate for the loss of the ACL's stability. In addition, Study 2.1 implied that different gait strategies were used among the ACL patients, and it was speculated that our cohort consisted of both copers and non-copers. Furthermore, Study 2.2 showed a tendency of lower stiffness and a higher rotation in both legs of the ACL-reconstructed patients at one-year follow-up compared to preoperative measures. At the same time, subjective outcome measures and clinical tests improved significantly from the preoperative period to the one-year postoperative period. Depending on the interpretation of the objective 3-D motion analysis findings, one could speculate about whether the patients had gained rotational stability after ACL reconstruction and if anatomic techniques are of clinical importance. Further thoughts are if knee stability can be improved in other ways than with ACL reconstruction.

Recent research has brought the progressive rehabilitation of ACL injuries into focus [92,140]. Frobell et al. states that it appears that a proportion of the ACL-deficient patients might be able to cope with their ACL deficiency if they receive adequate progressive rehabilitation [140]. Moreover, Grindem et al. showed a superior patient-reported outcome (KOOS) at two-year follow-up after ACL reconstruction in patients who performed progressive pre- and postoperative rehabilitation compared to patients receiving usual care[92]. Although these studies report their findings only in patient-reported outcome (KOOS score), their findings are quite interesting and underline the importance of progressive rehabilitation in the treatment of ACL injuries. In addition, rehabilitation allows the patient to influence his or her own recovery, which is especially important and motivating for athletes and active patients. Attendance to preoperative rehabilitation might also indicate the level of motivation in each patient.

On the basis of our findings, new questions for future research were raised:
1) Would progressive rehabilitation improve objective 3-D motion analysis measures before and after ACL reconstruction, and would it be possible to reveal a difference in knee laxity and stiffness between reconstructive techniques?
2) Would the use of RSA in a motion analysis setup (cameras and force plate) offer an even greater accuracy of joint range of motion than 3-D motion analysis[120]?

3) Would EMG measurements reveal any difference in muscle activation patterns among ACL patients?
4) What is the impact of progressive rehabilitation and/or anatomic ACL reconstruction on long-term OA development?
5) What impact do fixation methods have on the clinical outcome after ACL reconstruction?

Finally, recent research on ACL anatomy questions the double-bundle anatomy concept [163, 164]. These studies describe the ACL as a ribbon-like structure without clear separation between AM and PL bundles and places the functional femoral ACL insertion along the intercondylar ridge, i.e. higher in the notch than previous anatomic technique. In theory, this might explain why the TT SB technique actually performs surprisingly well in clinical studies.

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10. Appendices

List of Papers

Paper A

Bohn MB, Vestergaard R, Dalstra M, Jakobsen BW, Søballe K, Lind M.
"Mechanical stability of the femoral fixation for single- and double-bundle ACL reconstruction in an in vitro experimental model,"
Scand J Med Sci Sports. 2013;23(3):263-70 [108].

Paper B

Bohn MB, Petersen AK, Nielsen DB, Sørensen H, Lind M.
"Three-dimensional kinematic and kinetic analysis of knee rotational stability in ACL-deficient patients during walking, running and pivoting,"
J Exp Orthop. 2016;3(1):27 [110].

Paper C

Bohn MB, Sørensen H, Petersen MK, Søballe K, Lind M.
"Rotational laxity after anatomical ACL reconstruction measured by 3-D motion analysis: a prospective randomized clinical trial comparing anatomic and non-anatomic ACL reconstruction techniques,"
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Mechanical stability of the femoral fixation for single- and double-bundle ACL reconstruction in an in vitro experimental model

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Anterior cruciate ligament ACL reconstruction using the double-bundle (DB) technique is gaining popularity. A possible weak link in the DB technique could be that two tendon grafts of smaller diameters are used. The purpose of this study was to test different femoral fixation methods and graft diameters representing single-bundle (SB) and DB ACL reconstructions and compare their biomechanical properties. We hypothesized that SB 6-mm graft constructs had inferior biomechanical properties than SB 9-mm grafts or DB 2 × 6-mm grafts. Furthermore, we hypothesized that interference (IF) screw fixation would demonstrate less elongation and a higher stiffness than Endobutton (Smith & Nephew®, Inc., Andover, Massachusetts, USA) fixation (EBF). We performed an *in vitro* study using porcine knees and extensor tendons. The mechanical test consisted of a cyclic test followed by a load-to-failure test. We found that 6-mm graft constructs had an ultimate failure load that was up to 40% less than both the 9-mm and 2 × 6-mm graft constructs, despite the fixation method (*P*-values ≥ 0.004). Comparing fixation methods, EBF was superior to IF concerning maximum load to failure ($P < 0.001$); IF resulted in a higher stiffness of the femur/graft complex than the EBF ($P < 0.001$) but no significant difference in elongation between fixation methods. Since the two graft strands are subjected to different loads in different knee flexion angles, the reduced strength of the individual graft strands in DB ACL reconstruction could be a concern.

Recent research on the anterior cruciate ligament (ACL) anatomy and biomechanics have lead to changes in ACL reconstruction principles. During the last two decades, the ACL has been reconstructed using a single-bundle (SB) technique. It has been known for at least 100 years that the ACL can be divided into two functional bundles, i.e. the anteromedial bundle (AM) and the posterolateral bundle (PL) and a more anatomic reconstruction approach using the double-bundle (DB) reconstruction is gaining popularity. Biomechanical studies have suggested that an ACL reconstructed knee can obtain more normal biomechanical properties if two separate graft bundles are used for the reconstruction, especially by improving rotational stability (Sakane et al., 1997; Chhabra et al., 2006; Siebold et al., 2008). The DB technique has been extensively studied in the recent years, with several studies showing that separate restorations of the AM and PL functional bundles of the ACL result in more normal knee kinematics when compared to SB reconstruction both *in vitro* and *in vivo* (Yagi et al., 2002; Jarvela et al., 2008; Tsai et al., 2010; Aglietti et al., 2010).

Biomechanical studies of the ACL have shown that the tension of the two bundles varies during the arch of motion of the knee. While the AM bundle is relatively isometric with an almost constant tension that increases at maximal knee flexion, the PL bundle is tight in extension and at full flexion, with reduced tension in the mid-part of the arc of motion. The PL bundle also accepts rotational loads and for such loading modality only parts of the ACL fibers participate in accepting the load (Sakane et al., 1997; Zantop et al., 2007). This knowledge is essential for the study of possible failure mechanisms of the ACL and raises concern that one bundle might fail in knee flexion angles where it is subjected to the majority of the knee load.

An anatomic DB ACL reconstruction is performed using two tibial and two femoral tunnels. Knowledge of the biomechanical properties of tendon grafts and different fixation techniques in smaller diameter/drill holes is therefore of great importance when comparing SB to DB ACL reconstruction. Typically, the hamstring tendons are used as graft material and one double-stranded hamstring graft is placed in each tunnel. The diameters of the double-stranded hamstring grafts vary. Most commonly, the two-strand folded semitendinosus tendon with a diameter range from 5 to 7 mm is used for the AM bundle. The two-strand folded gracilis tendon used for the PL bundle have diameters ranging between 4 and 6 mm.

The soft tissue grafts used in anatomic ACL reconstructions are often secured with a cortical fixation on the femoral side using cortical button implants (Zantop et al., 2007). But also an aperture fixation with interference (IF) screws for the fixation in both femur and tibia has been presented (Brucker et al., 2006; Jarvela et al., 2008). A recent study has focused on the influence of different fixation techniques and the bony bridge between the two femoral tunnels used in anatomic DB reconstruction. Results from this study indicate that cortical fixation using a button technique results in higher ultimate failure load compared to an aperture fixation using an IF screw. Additionally, the study showed that the ultimate failure load in DB reconstruction was significantly higher than was the ultimate failure load in SB reconstruction (Lehmann et al., 2009).

The aim of this study was to compare biomechanical properties of different graft diameters and femoral fixation principles used for SB and DB ACL reconstruction. Different graft diameters representing SB and DB reconstructions were used. We hypothesized that SB 6-mm graft constructs had inferior biomechanical properties than SB 9-mm grafts or DB 2 × 6-mm grafts. Furthermore, we hypothesized that IF screw fixation would demonstrate less elongation and a higher stiffness than Endobutton fixation (EBF).

Materials and methods

Materials

Porcine knees from animals with a mean age of 5 months were used for this study. The material was obtained from a local slaughter house. Muscles and soft tissue were removed leaving the femur intact before its proximal part of the femur was removed using an oscillating saw. The bones were then scanned in a peripheral quantitative computed tomography (pQCT) scanner to determine bone mineral density (BMD) variation in the used bone specimens. The porcine femora were subsequently frozen at −20°C. Superficial porcine extensor tendons were harvested from the porcine hind leg to be used as grafts during the experiments. All tendons were kept moist in gaze soaked in saline irrigation before they were frozen at −20°C.

Study groups

ACL reconstructions were performed using two different fixation techniques in porcine femora: (1) Endobutton (EB) CL 20 mm from Smith & Nephew®, and (2) IF screw, length 25 mm from Inion Hexalon™, Inion Oy, Tampere, Finland (Fig. 1). Each technique was performed for diameters 6, 9, and 2 × 6 mm (DB).Ten specimens were tested for every diameter and technique, giving a total of 60 setups.

Mounting

Before testing, the bone and graft material was thawed for 16 h at room temperature. Bones and tendons were kept moist during the mounting by wrapping them in gauze soaked in saline irrigation. All tests were performed at room temperature.

First, the grafts were prepared. All tendons were sutured in both ends with Ethibond, EXCEL 2, ETHICON, Johnson & Johnson AB, Birkeroed, Denmark using whip stitch sutures. For the 6-mm graft, only one tendon was used, whereas two tendons were used for the 9-mm graft. The tendons were folded over an Ethibond suture and their diameter was measured. All tendons used for screw fixation were folded over an Ethibond suture and whip stitch sutures over a length of 30 mm was used to connect the strands to each other at the site of the loop.

The femur was fixed with bone cement in a steel cylinder with a roughened inner surface. Once the bone cement had hardened, the ACL reconstruction was performed. All bony tunnels were drilled from the notch area of the femoral condyle aiming toward the outer cortex of the femur securing a tunnel depth of 35–40 mm. When conducting the SB technique and IF fixation, a K-wire was drilled into the condyle and through the outer cortex, secondly, the K-wire was overdrilled with a cannulated 6- or 9-mm reamer leaving a tunnel depth of 25–30 mm. Performing the DB technique, two K-wires were drilled almost parallel into the condyle,

Fig. 1. The figure illustrates the two different fixation methods used; Two interference screws in diameter 9 and 6 mm, length 25 mm (Inion Hexalon™) and on the left an Endobutton (Smith&Nephew® CL 20 mm).

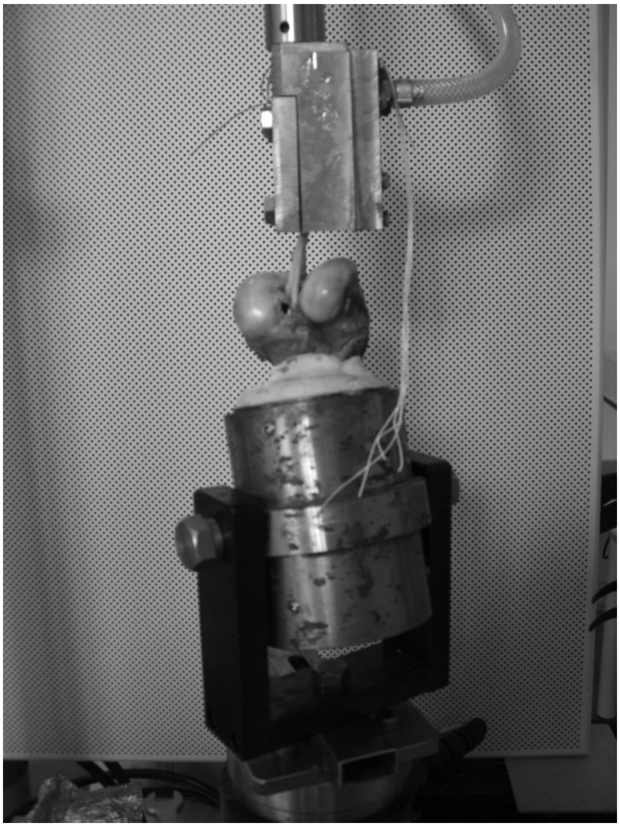

Fig. 2. The figure illustrates the bone cement fixed femur with the tendon grafts (ACL) friction locked in the cryofixation clamp. This setup was tested in an 858 Mini Bionix material testing machine (MTS).

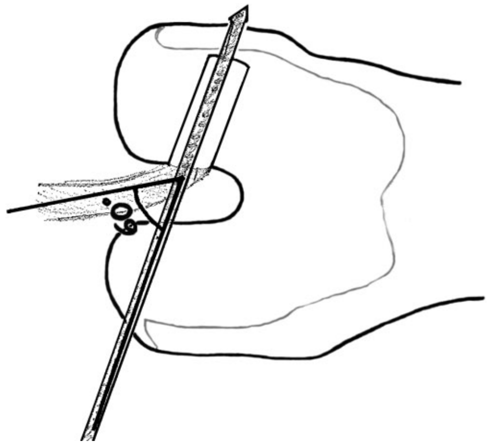

Fig. 3. The figure illustrates the angle between the bone tunnel and the tendon graft which is 60°.

overdrilled with a cannulated 6 or 9 mm reamer, leaving a bony bridge of 2–3 mm between tunnels, using free hand technique. When an EB was used for the reconstruction a 4.5 mm cannulated Endobutton drill was used to overdrill the K-wire and penetrate the outer cortex of the femur. Subsequently, the K-wire was overdrilled with a cannulated 6- or 9-mm reamer and care was taken that a 10-mm bony bridge was preserved between the tunnel end and the cortex. Endobutton CL 20 mm was used for all the tests.

The steel cylinder containing the femur/graft complex was placed into a custom-made fixation device with six degrees of freedom. The tendon graft was friction locked in a custom-made cryofixation clamp (Riemersa & Schamhardt, 1982) and care was taken to measure the exact length of the free graft length (between the bone and the clamp). Free graft length was 30 mm in each mounting (both fixation methods) in order to compare the stiffness of the grafts. This setup was tested in an 858 Mini Bionix (MTS systems Corp., Minneapolis, USA) material testing machine. All loads were applied in 60° direction to the bone channel to imitate an anatomical loading scenario and to standardize positioning (Figs 2 and 3). The loading angle in DB reconstructions was measured from one of the tunnels, and as the tunnels were approximately parallel, the two grafts were almost equally loaded. It is shown in the study made by Sakane et al. that the two bundles of the ACL are equally loaded at a knee flexion angle at 50–60° (Sakane et al., 1997). In the DB setups, the grafts were fixed conjoined in the cryofixation clamp.

Tensile testing

A preload of 5 N was first applied to the specimen. The grafts were cyclically preconditioned between 10–50 N at a rate of 1 Hz and then subjected to a cyclic loading protocol of 1000 cycles between 50–250 N at 1 Hz. The elongation of the graft construct was recorded during the cyclic test. Finally, the graft/femur complex was tested to failure (Weimann et al., 2006; Lehmann et al., 2009; Trump et al., 2011). Stiffness of the graft/femur complex and ultimate failure load was documented. In this study, stiffness is defined as the slope of the linear region of the first and most steep part of the load-displacement curve during the failure test.

Statistics

One way analysis of variance was used to compare data between the IF fixation group and the EB fixation group. The analysis was performed on log-transformed data for elongation after a 1000 cycles (in each trial the elongation was calculated as displacement after 1000 cycles – displacement after preconditioning), stiffness and maximum load. The Bonferroni correction was used for pairwise multiple comparisons. Significance was set at 5% ($P < 0.05$).

Results

Bone mineral density in test specimen

Bone mineral density measurement by pQCT scanning demonstrated that femoral condyle trabecular bone density average were in the range of 294–315 mg/ccm and cortical bone density average ranged 823–841 mg/ccm in the six groups. No significant differences were seen between BMD of femora used in the different test groups.

Ultimate failure load (Table 1)

The average ultimate failure load of the EB groups was 30% higher than the IF groups ($P < 0.001$). Comparing different diameters within each fixation group, we found that the ultimate failure load of the SB 6-mm reconstructions were significant less than both the SB 9-mm reconstructions (IF 6 mm vs IF 9 mm; $P = 0.004$, EB 6 mm vs EB 9 mm; $P < 0.001$) and the DB reconstructions (IF 6 mm vs IF 2×6 mm; $P > 0.001$, EB 6 mm vs EB 2×6 mm; $P > 0.001$). No significant difference was found between the ultimate failure load of the DB reconstruction and the ultimate failure load of the SB 9 mm reconstruction for both fixation methods. Comparing equal diameters between fixation groups (IF 6 mm vs EB 6 mm, etc.), no significant difference was found, although the comparison between the 9-mm graft constructs showed a tendency ($P = 0.069$).

Elongation (Table 2)

Overall, no significant difference was found in elongation between the two fixation methods.

Within the IF group, a significant difference was seen between the IF 6 mm and IF DB; hence, the average elongation after a 1000 cycles of the DB reconstruction was significantly less than the SB 6-mm reconstruction ($P = 0.002$). Comparing the IF 9 mm to the IF DB reconstruction, no significant difference was seen ($P = 0.15$). Between graft diameters in the EB group, no significant difference was seen ($P = 0.78$). Comparing equal diameters, a significant difference was seen in the DB reconstructions ($P = 0.05$).

The parameter "displacement at failure" includes the displacement after preconditioning + displacement after 1000 cycles + displacement after the load-to-failure test. Displacement at failure reveals a "bedding-in" elongation, which is highest for EB.

Stiffness (Table 3)

The average stiffness of the femur-graft complex when an IF fixation was used for the reconstruction was 22% higher than when an EB fixation was used ($P < 0.001$). Comparing the different graft diameters within the EB groups, no significant differences were found. Within the IF group, a tendency was seen toward the SB 6-mm reconstruction being less stiff than the 9-mm and 2 × 6-mm reconstructions ($P = 0.112$ and $P = 0.126$, respectively). Comparing equal diameters between fixation groups, a significant difference was found between the SB 9-mm reconstructions ($P = 0.009$).

Table 1. Failure load

Fixation and bone tunnel diameter	Failure load (N)	95% CI (Range; N)
Endobutton 6 mm	568 (±169)	447–689
Endobutton 9 mm	969 (±177)*	843–1096
Endobutton 2 × 6 mm	1071 (±244)†	896–1246
IF screw 6 mm	432 (±97)	362–502
IF screw 9 mm	708 (±230)‡	544–872
IF screw 2 × 6 mm	806 (±167)§	686–926

Table 1 shows failure load; mean ± SD and 95% CI for each fixation and diameter. RAW DATA.
Significant differences (EB, Endobutton; IF, interference):
*EB-9 mm vs EB-6 mm ($P < 0.001$),
†EB-2 × 6 mm vs EB-6 mm ($P < 0.001$),
‡IF-9 mm vs IF-6 mm ($P = 0.004$),
§IF-2 × 6 mm vs IF-6 mm ($P < 0.001$).
No significant differences when equal diameters compared.
CI, confidence interval.

Table 3. Stiffness

Fixation and bone tunnel diameter	Stiffness (N/mm)	95% CI (range; N/mm)
Endobutton 6 mm	241.7 (24.9)	226.3–257.1
Endobutton 9 mm	253.4 (64.1)*	213.6–293.2
Endobutton 2 × 6 mm	285.1 (51.2)	253.3–316.9
IF screw 6 mm	274.3 (65.5)	233.7–314.9
IF screw 9 mm	353.6 (77.8)	305.4–401.8
IF screw 2 × 6 mm	351.5 (70.9)	307.6–395.4

Table 3 shows stiffness of the femur-graft construct; mean ± SD and 95% CI for each fixation and diameter. RAW DATA.
*EB-9 mm vs IF-9 mm ($P = 0.009$).
EB, Endobutton; IF, interference; CI, confidence interval.

Table 2. Elongation

Fixation and bone tunnel diameter	Displacement after 1000 cycles (mm)	95% CI (range; mm)	Displacement at failure (mm)	95% CI (range; mm)
Endobutton 6 mm	3.5 (0.8)	2.9–4.0	11.3 (2.5)	9.7–12.9
Endobutton 9 mm	3.4 (0.8)	2.9–4.0	14.4 (1.8)	13.2–15.6
Endobutton 2 × 6 mm	3.2 (0.7)	2.7–3.7	16.3 (6.9)	12.0–20.6
IF screw 6 mm	4.1 (2.1)*	2.6–5.6	9.9 (1.7)	8.8–11.0
IF screw 9 mm	3.3 (1.5)	2.2–4.4	11.4 (2.5)	9.8–13.0
IF screw 2 × 6 mm	2.1 (0.6)†	1.7–2.5	10.6 (1.5)	9.7–11.5

Table 2 shows elongation/displacement after 1000 cycles; mean ± SD and 95% CI for each fixation and diameter. Displacement at failure includes displacement after preconditioning + displacement after a 1000 cycles + displacement after the load-to-failure test. Displacement at failure; mean ± SD and 95 % CI for each fixation and diameter. RAW DATA.
Significant differences:
*IF-6 mm vs IF-2 × 6 mm ($P = 0.002$),
†IF-2 × 6 mm vs EB-2 × 6 mm ($P = 0.05$).
EB, Endobutton; IF, interference; CI, confidence interval.

Table 4. Failure mode

Failure mode	Fixation and tunnel diameter					
	Endobutton			Interference screw		
	6 mm	9 mm	2 × 6 mm	6 mm	9 mm	2 × 6 mm
Slippage past the screw				10	10	8
Slippage past the screw and condyle fracture						2
Condyle fracture			1			
Graft failure (GF)	5		2			
Endobutton (EB) through cortex	3	10	1			
GF and EB through cortex	2		1			
GF and EB through cortex and bone torn out of cement			1			
Bone torn out of cement			2			
Graft slippage cryoclamp and EB through cortex			2			

Table 4 shows the different and various failure modes of each femur-graft construct after the failure test. Ten specimens were tested for each fixation and diameter. Graft failure = shearing of the tendon fibers at the polyester loop of the Endobutton.

Mode of failure (Table 4)

Using IF fixation, the mode of failure was graft slippage along the screw in all cases. In two of the DB reconstructions using IF fixation, an additional fracture of the condyle was seen.

The Endobutton being pulled through the cortex was the most frequent mode of failure in the EB groups (SB 9-mm and DB). In the Endobutton 6-mm reconstruction group, the main failure mechanism was graft tissue rupture at the polyester loop.

In the EB-DB reconstruction group, various and combined failure modes were seen; the button being pulled through cortex, tendon extension at the loop, fracture of the condyle, bone torn out of the cement, and graft failure at the clamp (Table 4). We realized that these mountings could accept very high loads and that failure at the clamp or cement appeared before the buttons were pulled through the cortex. This indicates that the strength of the EB-DB technique in five cases was higher than the measured values.

Discussion

Two different femoral fixation methods used for ACL reconstruction were tested for different graft diameters, representing graft-loading situations in SB and DB ACL reconstructions in the initial post-operative period. First, a cyclic test was performed, which simulates early rehabilitation activities, and finally, a load-to-failure test was performed, which mimics an event causing failure of the ligament. Comparing the fixation methods tested, we found that cortical fixation using an Endobutton with a closed loop accepted significant higher ultimate loads than did an aperture fixation using an IF screw. We also demonstrated that the cortical fixation method resulted in a significant lower stiffness of the femur-graft complex than aperture fixation. No significant difference in elongation was seen between fixation methods. Another important finding was that both the DB 2 × 6-mm reconstruction and the SB 9-mm reconstruction accepted a significant higher ultimate load than the SB 6-mm reconstruction for both fixation methods tested. Hence, the 6-mm graft construct was found to accept 40% lower ultimate loads than both the 9-mm and 2 × 6-mm graft constructs irrespective of fixation method used.

As shown by Sakane et al., tension forces shift between the bundles of the ACL during the range of motion of the knee. In full extension, the PL bundle takes the majority of the load and between 30 and 100 degrees of flexion, the AM bundle is mainly loaded (Sakane et al., 1997). On this basis, concern rises for failure of one bundle in knee flexion angles where only one bundle takes the majority of the load after an anatomic DB reconstruction. A study by Kaz et al. presents three cases of revision surgery after DB anterior cruciate ligament reconstruction (Kaz et al., 2007). In each case presented, the patient had a new traumatic event playing sports. In two cases, the AM was completely torn and the PL was stretched and nonfunctional. In the third case, the rupture pattern presented an intact and functional PL and a midsubstance tear of the AM. The failure mechanism in these cases could be caused by overloading an insufficiently sized AM bundle reconstruction. The study also indicates that the AM is mostly at risk after DB reconstruction and it support our concern about loading pattern of the two bundles and graft failure.

It is well known, that the graft-fixation site of soft tissue grafts is the weakest point in ACL reconstruction during the initial post-operative period. Physiologic loads placed on the ACL under dynamic *in vivo* conditions are difficult to measure (Beynnon et al., 1992; Fleming et al., 1993; Heijne et al., 2004). A biomechanical *in vitro* study on human specimens by Noyes et al. has estimated the force on the graft fixation during normal activities to be approximately 450 N (Noyes et al., 1984).

By the use of 3D motion analysis and a force plate, the net moments of force acting on the entire knee can be measured. A recent study by Trump et al. estimates the loads applied on the knee of a 70-kg individual during walking, jogging, and stair to be 150 N, 250 N, and 340 N, respectively (Trump et al., 2011). Pivoting places a load of approximately 0.3x body weight on the knee (not yet published data from the authors PHD). In this study, we demonstrated pullout strengths of the grafts ranging from 446–806 N (means, IF) and 568–1071 N (means, EB) for the different diameters; hence, these pullout strength is much higher than rehabilitation loads placed on the entire knee.

However, this *in vitro* study does not entirely mimic real life and the way ACL most often is torn where compressive and rotational forces acts on the knee. This study reflects absolute values for the different fixation and diameters in this specific *in vitro* setup where the grafts have been pulled in one direction. In that context, these results should be interpreted with reservations.

The results from the present study are consistent with the results of a recent biomechanical study by Lehmann et al. (2009). This study tested the impact of different widths of the bony bridge between the two femoral tunnels used in DB reconstruction. The study demonstrated that DB cortical fixation compared to DB aperture fixation resulted in (1) a higher ultimate load; (2) more elongation; and (3) less stiffness. Additionally, the study showed that DB reconstructions using a cortical fixation had a significant higher ultimate load, a significant smaller elongation, and a significant higher stiffness than SB 8-mm reconstruction using a cortical fixation. We compared a 9-mm graft construct to a DB reconstruction for both an aperture and a cortical fixation technique. In our study, the DB reconstruction demonstrated the same or better structural properties than the 9-mm SB reconstruction for both fixation techniques but the differences were not significant.

The literature concerning the optimal fixation method for soft tissue grafts is still controversial. DB ACL reconstructions are most commonly performed using a cortical fixation technique at the femur for soft tissue grafts (Zantop et al., 2007). Other authors, however, prefer an aperture fixation with an IF screw for the fixation of the AM and PL bundle in DB ACL reconstructions (Brucker et al., 2006; Jarvela et al., 2008). These authors advocate that an aperture fixation with an IF screw near the joint line is more anatomic, although this technique compresses the graft toward the borders of the tunnel and hereby reduces the footprint of the ACL. However, another concern, when using cortical fixation, is graft motion within the bone tunnel. This phenomenon could potentially lead to delayed graft incorporation and subsequent tunnel enlargement (Brucker et al., 2006; Jarvela et al., 2008). If, however, the graft fixed with a cortical method gets a good in-growth in the bone tunnel, the biomechanical properties would, theoretically, change and probably lead to a higher stiffness over time. This study only imitates the initial post-operative situation before any healing/in-growth between the graft and bone tunnel.

When choosing a graft/fixation combination, it is important to utilize both the knowledge of biomechanical properties of graft fixation and the knowledge of soft tissue graft-to-bone healing and graft remodeling. A hybrid fixation technique using a cortical button and an IF screw might be a biomechanical advantageous solution as described in a study by Weimann et al. (2006). This study demonstrated that if an IF screw fixation was secured by an additional cortical fixation, structural properties of the graft/implant/bone complex were superior to both aperture and cortical fixation, even when an undersized screw was used. To date, no data concerning hybrid fixation in DB reconstruction is available.

With the increasing number of DB ACL reconstructions being performed, specific DB failure mechanism leading to revision cases is expected. Our results suggest that the reduced diameter of the grafts is a potential weak link in DB reconstruction with specific failure mechanism that can be related to insufficient graft tissue material in relation to the implant used. Endobutton fixation of a 6-mm graft resulted in some cases (Table 4) of shearing of graft fibers at the polyester loop of the implant and was seen in both SB and DB setups. Shearing of graft fibers at the polyester loop was not seen to the same extend in the SB 9-mm setup where the failure mechanism were related to bone tissue failure such as the button being pulled through cortical bone. Failure mechanism when using IF screws was grafts slippage between the screw and bone tunnel surface. Condyle fractures were seen in few DB setups (EBF and IFF).

The following limitation applies to this study. We tested the graft-femur complex with the pull force in 60° direction to the bony channel. This is different from other studies that most frequently use a pull force parallel to the bone channel. We chose a 60° pull direction to mimic the typical anatomic situation of the ACL graft position at the lateral notch wall on a fully extended knee. This might have an influence on ultimate load and the stiffness of the graft-femur complex, which was higher in the present study compared to recent biomechanical studies (Weimann et al., 2006; Lehmann et al., 2009). Under *in vitro* conditions, the corner of the tunnel creates a frictional effect, which reduces the force imposed on the intra-articular graft. To calculate the stiffness, only the slope of the steepest part of the linear region of the load-displacement curve from the load-to-failure test was used. In other studies, this calculation is not specified. Additionally, we used a porcine model as described by Nargakatti et al. This study shows that the bone mineral density of porcine bones is comparable to that of the human femur (Nagarkatti et al., 2001). Failure of the graft at the site of the clamp or the bone torn of the cement was the exclusion criteria. These failure mechanisms were seen five times in the DB reconstruction fixated with Endobutton at very high loads. As a result, the true ultimate failure load for this group is potentially higher than the results presented. As mentioned above, these results reflect an absolute difference in biomechanical properties for this specific experimental setup. The results should be interpreted with care when it comes to *in vivo* conditions, which is more dynamic, multidirectional and where many other structures around the knee contributes to the stability of the knee.

In conclusion, our study demonstrated that the structural properties of the DB 2 × 6-mm reconstruction were equal to the SB 9-mm reconstruction regardless of the fixation method used. Our study also showed that single 6-mm grafts are weaker than DB 2 × 6-mm reconstructions for both cortical and aperture fixation. A potential clinical implication of these results is to suggest a possible graft failure mechanism of DB ACL reconstructions when only one graft bundle accept most of the load.

Perspectives

Double-bundle ACL reconstruction, with two relatively thinner graft strands than SB ACL reconstructions, have potential higher risk of graft failure in range of motion where only one graft strand is loaded. The present study demonstrated that a single 6-mm double-stranded graft was 40% weaker than a single 9-mm four-stranded graft. In many DB reconstructions performed, a folded gracilis tendon has been used with diameters down to 4–5 mm resulting in even weaker bundle strength.

The needed graft material per bundle in DB reconstruction is unknown, but the results from the present study suggest that a diameter less than 6 mm might be insufficient. A method to increase graft diameter, and thereby enabling DB ACL reconstruction, could be to triple-fold the tendons.

Key words: anterior cruciate ligament, biomechanics, double-bundle reconstruction, *in vitro*, femoral fixation.

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Three-dimensional kinematic and kinetic analysis of knee rotational stability in ACL-deficient patients during walking, running and pivoting

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Abstract

Background: Anterior cruciate ligament (ACL) deficiency leads to altered stability of the knee. The purpose of this study was to compare the dynamic, rotational stability of the knee, expressed as rotational stiffness, between anterior cruciate ligament-deficient (ACLD) knees, their contralateral intact knees (ACLI) and a knee healthy control group during walking, running and 90° pivoting. We hypothesized a larger tibial internal rotation, a smaller knee joint external moment and a lower rotational stiffness in the ACLD group compared to the ACLI and the control group.

Methods: Kinematic and kinetic data were collected from both legs of 44 ACLD patients and 16 healthy controls during walking, running and a pivoting maneuver (descending a staircase and immediately pivoting 90° on the landing leg). Motion data were captured using 8 high-speed cameras and a force-plate. Reflective markers were attached to bony landmarks of the lower limb and rigid clusters on the shank and thigh (CASH model). Maximum internal tibial rotation and the corresponding rotational moment were identified for all tasks and groups and used to calculate rotational stiffness (= Δmoment /Δrotation) of the knee.

Results: The tibial internal rotation of the ACLD knee was not significantly different from the ACLI knee during all three tasks. During walking and running, the tibial rotation of the control group was significantly different from both legs of the ACL-injured patient. For pivoting, no difference in tibial rotation between knees of the ACLD, ACLI and the control group was found. Knee joint external moments were not significantly different between the three groups during walking and pivoting. During running, the moments of the ACLI group were significantly higher than both the knees of the ACLD and the control group. Rotational stiffness did not differ significantly between groups in any of the three tasks.

Conclusion: A high-intensity activity combining stair descent and pivoting produces similar angular rotations, knee joint external moments and rotational stiffness in ACLD knees compared to ACLI knees and the control group. During running, the ACLI knee displayed a higher external moment than the ACLD and the healthy control group. This could indicate some type of protective strategy or muscular adaptation in the ACL-injured patients.

Keywords: ACL, ACL-deficient, Motion analysis, Stiffness, Laxity, Pivoting

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Background

Anterior cruciate ligament (ACL) rupture is one of the most frequent sports-related injuries in orthopedic surgery (Fu et al., 2015). Young and physically active people are prone to sustain an ACL injury, and most injuries are sustained during contact or pivoting sports. In Scandinavia, the median age for sustaining an ACL injury is 23–27 years, and the yearly incidence is 38 per 100,000 people (Granan et al., 2009). Therefore, ACL patients are often referred to as a young patient with an old knee, as long-term clinical problems such as meniscal damage and osteoarthritis (OA) development are often observed (Frobell et al., 2010, Fu et al., 2015,Lohmander et al., 2007, Risberg et al., 2016, Stergiou et al., 2007).

It is well known that an ACL deficient (ACLD) knee can exhibit pathological laxity, which often leads to complaints of knee instability from the patient (Gabriel et al., 2004, Hasegawa et al., 2015). This condition has been proposed to contribute to the development of osteoarthritis (Stergiou et al., 2007).

Functional knee assessments pre- and post-surgery often mention laxity and stability. However, from a strictly biomechanical point of view, laxity and stability are not well defined in the clinical literature. Cross defined laxity as "… the measured amplitude of joint movement within the constraints of its ligaments", i.e. a purely kinematic measure expressed in millimeters and degrees for translational and rotational laxity, respectively (Cross 1996). Kovalski, on the other hand, defined laxity as "… the freedom of movement within a joint and is measured as joint translation at a given force load", i.e. a kinetic measure including both the magnitude of the movement (millimeters) and the force (newton) causing the movement, bringing laxity closer to the well-defined biomechanical concept compliance (Kovaleski et al., 2008). Cross further defined instability as "a complaint from the ACL injured subjects because they lose single leg stance as the joint subluxes due to the pathological laxity", i.e. a purely subjective measure (Cross 1996).

Because of the ambiguous definitions of laxity and the subjective definition of instability, we use the biomechanically well-defined concepts rotation, moment and stiffness in the present paper. Rotation, measured in degrees (deg), is the magnitude of the movement of the tibia about its longitudinal axis relative to the femur. Joint moment (of force), measured in newton-meters (Nm), is the magnitude of the turning force exerted by the ACL and other knee structures, equal in magnitude but opposite in direction to moments applied to the leg from the surroundings (typically the ground) causing the rotation. Stiffness, defined as change in joint moment divided by change in rotation, and hence measured in Nm/deg, is the knee's ability to withstand moments applied to the leg from the surroundings without rotating – the stiffer the joint, the less it rotates when exposed to a certain moment. Thus, the clinical, subjective concept stability can be precisely quantified as stiffness.

Knee joint rotation and moment, and thereby stiffness, can be assessed under both static conditions and during natural movements. Static conditions are for instance with the patient reclining on the clinician's bed or fixated in an isokinetic dynamometer (or material testing apparatus for cadaver knees), while the only restriction on assessment during natural movements is the dependency on advanced motion capture equipment, ng it takes place in a laboratory.

Static assessment has been used in several studies in both cadaver knees (Hsu et al., 2006, Kanamori et al., 2000, Yagi et al., 2002, Zantop et al., 2007) and living subjects (Louie and Mote 1987, Schmitz et al., 2008). In some of the cadaver studies, stiffness, i.e. simultaneous measurement of rotation and moment, was measured with tension in the knee joint muscles realized artificially via various pulling mechanisms, while the subjects in Louie and Mote's study could tense and relax their muscles voluntarily. However, regardless of the type of muscle tension, it always resulted in a considerable increase in stiffness; Louie and Mote, for instance, saw stiffness increase by over 400 % when their healthy subjects went from relaxing to fully activating their knee joint muscles.

Assessment during natural movements using 3D motion analysis has by definition higher external validity compared to static assessment, both due to the movement itself, and to the subject being able to activate thee knee joint muscles in a natural way. A number of studies have reported altered transverse plane kinematics in ACLD patients during various tasks, with a possible trend toward increased internal rotation of the tibia (Andriacchi and Dyrby, 2005,Gao and Zheng, 2010, Georgoulis et al., 2003, Ristanis et al., 2005, Ristanis et al., 2006, Waite et al., 2005, Zabala et al., 2015). However, most of these studies only report kinematics, either knee range of motion (ROM) or peak knee rotation. This is problematic, since knee rotation depends on both the moment of force applied by the surroundings, causing the rotation, and on the ACL's (and other knee joint structures') ability to resist the rotation by creating a counteracting knee joint moment, i.e. the stiffness of these structures. Thus, in comparative studies, rotation is a valid measure of stability only when the moment is carefully measured and reported, e.g. together as stiffness. However, moments have been reported in only a few studies (Fuentes et al., 2011, Tsarouhas et al., 2010, Tsarouhas et al., 2011), and none of these calculated the actual stiffness. Interestingly, Tsarouhas et al. found that the knee joint moment of the affected side (ACL deficient or reconstructed) was constantly lower than on the unaffected side (ACL intact knee), although a similar range of tibial rotation was seen in the affected and unaffected knees in all groups (Tsarouhas et al., 2010, Tsarouhas et al., 2011). These findings indicate some kind of protective, stiffness increasing strategy in the affected knee, and, furthermore, give rise to the question if rotation can stand alone when reporting joint stability.

To our knowledge, the rotational stiffness in ACLD knees, contralateral ACL intact knees and knees of healthy control subjects during natural movements has not yet been reported. Thus, the purpose of this study was to determine knee rotational stability expressed as rotational stiffness, in ACLD and healthy knees during simulated, natural movements. Three movement tasks were analyzed: walking, running and stair descent followed by 90° pivoting. We hypothesized larger internal rotation, lower rotational moments and therefore a lower rotational stiffness in the ACLD knees compared to the ACLD subjects' contralateral uninjured knee (ACLI) and a healthy control group (control).

Methods

Study design

This cross-sectional study was conducted between January 2009 and November 2010. The protocol was approved by the Region Midtjylland ethical committee (jr. nr. 20060198). Prior to participation, written informed consent was obtained from every subject.

Subjects

Forty-four patients (18 females and 26 males) with a unilateral ACL lesion were included from a public hospital waiting list by the first author. Inclusion criteria for entering the study were: age 18–50 years, ACL injury with symptoms of instability and an uninjured contralateral knee. Exclusion criteria were: concomitant knee ligament injuries, previous knee ligament surgery, cartilage injuries of International Cartilage Research Society (ICRS) grade 3 or 4 and meniscus injury requiring resection of more than 50 % of a meniscus. Sixty patients were assessed for eligibility. Six of these patients were excluded prior to surgery (one declined to participate, one did not meet the inclusion criteria, and 4 were excluded for other reasons) and another ten were excluded at surgery. The control group consisted of 16 age- and sex-matched healthy subjects who had no history of lower extremity pathology or trauma.

The demographic data of the ACL injured patients and the control group is presented in Table 1. No significant differences were observed between the two groups in terms of sex, age, height, weight and body mass index (BMI). The median time since injury was 11 months (range: 2–42).

Table 1 Demographics of ACL injured patients and the knee healthy control group

	Patients	Control group	*P*-value
Sample size	44	16	
Sex (Female/Male)	18/26	6/10	0.12
Age (years) (mean ± SD)	25.7 ± 6.1	26.6 ± 3.6	0.94
Height (cm) (mean ± SD)	177.5 ± 9.6	178.6 ± 8.5	0.67
Weight (kg) (mean ± SD)	76.5 ± 14.9	73.7 ± 7.5	0.35
BMI (kg/m^2) (mean ± SD)	24.1 ± 3.3	23 ± 1.5	0.08
Time since injury (months) (mean ± SD (range))	11 ± 9 (2–42)		

SD Standard Deviation, *BMI* Body mass index

There were 20 injured right knees and 24 injured left knees. The majority of the injuries were sports-related; 20 ruptures were sustained during soccer, 5 during skiing, 7 during handball (European team handball) and one each during field hockey, tennis, cheerleading, trampoline jumping and roller blading. One patient was injured as she was kicked directly on the knee by a horse while the remaining six patients had torsional traumas to their knee during activities of daily living.

Testing procedure and data analysis

Clinical evaluation

At the time of inclusion, each patient was examined clinically and graded according to the objective IKDC grading. Passive, sagittal laxity was measured using a KT-1000 arthrometer (Medmetric® Corp., San Diego, California). Furthermore, four subjective outcome questionnaires were completed; the International Knee Documentation Committee (IKDC) subjective score (Hefti et al., 1993), the Knee Osteoarthritis Outcome Score (KOOS) (Roos et al., 1998), the Tegner score and the Lysholm score (Wright, 2009).

Three-dimensional motion analysis

Protocol The protocol was identical for patients and control subjects and the patients were tested up to 3 weeks prior to ACL reconstruction. Hence, all participants performed three different tasks in the following order: 1) level walking, 2) running/jogging, and 3) stair descending followed by a 90° pivoting maneuver. Walking and running were performed at the participant's self-selected speed until at least 5 successful (clean force plate contact) trials were recorded for each side and exercise. The tasks were performed on an 8 m walkway with an embedded force plate. The entire walkway was covered with a thin carpet to conceal the position of the force plate.

Stair descending was performed on a three-step plywood staircase with no handrail, which was placed next to the force plate (Fig. 1). The staircase was constructed according to Andriacchi et al. (Andriacchi et al., 1980) (rise 21 cm, run 25 cm, width 48 cm). The participants were asked to descend the staircase at their own pace. Following contact with a force plate at the bottom of the staircase, the subjects were instructed to make a pivoting maneuver by moving the swing leg through a 90° arc across the stance leg and to contact the ground with the foot of the swing leg at a 90° angle relative to the stance foot. Full plantar side force plate contact was required. The subjects then walked a few steps away from the plate. This pivoting maneuver was designed to impart an internal rotation of the shank relative to the thigh of the stance leg (Fig. 1). Similar procedures have been described in several previous studies (Georgoulis et al., 2003, Ristanis et al., 2003, Webster et al., 2010). At least 10 successful trials for each leg of the participants were recorded. The patients always started out by pivoting on their ACLI knee followed by the ACLD knee. To ensure a constant procedure during stair descent and pivoting, each trial was carefully supervised.

Instrumentation for data collection All trials were performed with bare feet. Automatic tracking was facilitated by 16 reflective markers placed on anatomical landmarks and as clusters on rigid plates on each leg. Eight markers were fitted on bony prominences (greater trochanter, medial and lateral femoral condyle, medial and lateral malleolus, heel, and 1st and 5th metatarsal head) to define anatomical planes and joint centers, while the remaining eight markers were placed as two four-marker clusters on the shank and thigh segment, respectively (Fig. 2). All markers were placed by one investigator.

To measure the position of the reflective markers, eight optoelectronic motion capture cameras (ProReflex MCU 1000, Qualisys Medical AB, Gothenburg, Sweden) operating at 240 frames per second were used together with Qualisys Tracking Manager (QTM) software on a personal computer. The system was calibrated prior to each data collection session. To obtain a reference point for the markers, a static trial was obtained before performing the protocol with the subject in quiet standing.

The collected trajectory data were gap filled if required (gaps were rare and typically only 2–3 frames wide, in

Fig. 1 Pivoting task

Fig. 2 Reflective markers

very rare occurrences up to 10 frames) in QTM using NURBS interpolation and exported to Visual3D software (C-motion Inc., Kingston, Canada) where a Visual3D Hybrid Model for ideal rigid segments and 6-degrees-of-freedom (6DOF) was applied. The marker position data were low-pass filtered using a 2nd order Butterworth digital filter with one bidirectional pass (effectively making it a 4th order filter) and an effective cut-off frequency of 6 Hz.

Ground reaction force (GRF) was sampled simultaneously using an AMTI OR6-6 force plate (Advanced Medical Technology Inc., MA, USA) sampled at 960 Hz. The GRF data were low-pass filtered with a cut-off frequency of 30 Hz.

Data analysis From the marker positions and GRF data, the knee rotation and moment were calculated for each participant using inverse dynamics for idealized rigid segments. All moments were normalized to body mass. Anthropometric data were calculated from individual body mass and height using Dempster's regression equations (Dempster, 1955).

Knee rotation was calculated based on the joint coordinate system definition (Grood and Suntay, 1983), which described knee rotation as occurring around the shank's longitudinal axis. The neutral position of the knee (0° knee rotation) was defined as the knee angle five frames before foot contact (defined to occur at the first frame where vertical GRF exceeded 20 N), i.e., knee close to fully extended with no external moment affecting the limb (0 Nm knee joint moment).

For each trial (patients and controls), the maximum tibial internal rotation and corresponding net external knee joint moment about the tibia's longitudinal axis was determined during the stance phase. Then, to avoid potential outliers, rotation and moment values from the trial with the second highest tibial internal rotation were used for further analyses. Furthermore, these values were used to calculate the rotational stiffness of each knee by dividing the change in rotational moment with the change in tibial internal rotation; the change was taken between the mentioned values and the values from the unloaded condition, defined as 0° rotation and 0 Nm moment.

Statistical analysis

Sample size calculation was limited by the fact that in vivo rotational stiffness has not been reported in ACL deficient subjects using 3D motion analysis. Hence, measures of tibial rotation from previous studies were used in the sample size calculation (Ristanis et al., 2005, Ristanis et al., 2006). As the cohort of ACL deficient patients were to be divided into three groups afterwards, which received three different ACL reconstructions in a randomized clinical trial (RCT) (Bohn et al., 2015), the power calculation was based on the difference in tibial rotation between ACL intact and ACL reconstructed knees. According to this power calculation, a total of nine subjects were needed in each of the three groups in the RCT. To account for dropouts and problems with data retrieval from 3D motion analysis, we chose to include approximately 15 patients per group and ended up with a total of 44 ACL deficient subjects and a control group of 16 subjects.

Statistical analysis consisted of a comparison between three groups of knees: the ACLD knee, the ACLI knee and the control group (both knees). Three parameters were evaluated: internal tibial rotation, rotational moment and rotational stiffness, respectively, during each of the three tasks performed (walking, running and pivoting). The three parameters combined with the three tasks were considered as separate end points for a total of nine parameters. For each of such parameter a repeated measurements ANOVA was used with the subject ID as the repeated factor, to take into account different subjects level. The ANOVA was one way using the leg group as the factor. The residual variation, i.e. the within subject variation, was allowed to vary by group. The model was fit as a mixed model in Stata (STATA software version 14, StataCorp LP, Texas, USA). To our knowledge, it is statistically unclear which adjustment for multiple comparisons one should apply and therefore we have chosen to report the uncorrected p-values (Perneger, 1998). Significance level was set at $P < 0.05$.

Model validation was performed separately for each end point by inspecting (standardized) residuals, fitted values and random effect estimates (BLUP's). Specifically, we checked distributional assumptions and for signs of heteroscedasticity. When evaluating the model we paid attention to the fact that the model was to be used for estimating mean differences and not for e.g. obtaining predictions. When initial inspections along with the model validation gave impression of a screwed distribution, the log transform was applied to the outcome and the model validation repeated. In the case of the pivot measurements, this caused us to prefer the analysis of log transformed observations.

Results

Knee stability and patient-related outcome scores

At the time of inclusion, none of the ACL deficient patients were graded IKDC A (normal), whereas 43 % were IKDC B (nearly normal), 39 % were IKDC C (abnormal) and 18 % were graded IKDC D (severely abnormal). The average passive, sagittal laxity measured by KT-1000 was 3.5 ± 2.2 mm.

For the ACL deficient patients, the four different patient-related outcome scores were as follows (mean ± SD): IKDC

61 ± 11, KOOS4 73 ± 13, Lysholm 72 ± 13 and Tegner 3.8 ± 1.4. The average scores for the healthy knee controls were IKDC 97 ± 4, KOOS4 97.5 ± 2.8, Lysholm 95 ± 9 and Tegner 7.5 ± 1.9.

Three-dimensional motion analysis

During walk, both the ACLI and the ACLD knee rotated less than the knees of the control group and these differences were significant ($p < 0.001$) (Table 2). No significant difference was seen in moment or stiffness during walk between the three groups of knees, although a tendency was seen between the stiffness of the ACLD knee and the control group ($p = 0.098$).

Tibial rotation during running showed a significant difference between groups. Hence, the control group displayed a significantly lower tibial rotation than both the ACLD knee ($p = 0.014$) and the ACLI knee ($p = 0.001$). Moments during running were significantly higher in the ACLI knee compared to both the ACLD knee ($p = 0.015$) and the control group ($p < 0.001$). Stiffness was not significantly different between groups, although a tendency was seen between the ACLI knee and the control group ($p = 0.062$).

Pivoting displayed no significant differences in tibial rotation, moments or stiffness between the ACLD, ACLI and control knees.

Discussion

The most important findings of this cross-sectional study were that tibial internal rotation was not increased in ACLD knees compared to ACLI knees during any of the three tasks investigated. Thus, our findings did not support this part of our hypothesis. Furthermore, we found that the tibial internal rotation was significantly different between both knees of the ACL-injured group and the healthy control group during walking and running but not during pivoting. Furthermore, no difference in external knee joint moments was found between ACLD, ALCI and control knees during walking and pivoting. Interestingly, the ACLI knee displayed a significantly higher moment during running compared to both the ACLD knee and the control group, which could represent a compensatory strategy. Finally, rotational stiffness did not differ significantly between groups in any of the three tasks performed. Thus, none of our findings supported our hypothesis.

The pivoting maneuver applied in the present study has previously been used to investigate tibial rotation in ACLD knees (Ristanis et al., 2006, Tsarouhas et al., 2011). Similar to our study, these two studies compared ACLD knees to both the contralateral ACLI knee and a knee healthy control group. Thus, Ristanis et al., from whom we replicated our pivoting task, reported an increase in tibial rotation of the ACLD knee compared to both control groups (Ristanis et al., 2006). However, Tsarouhas et al. (Tsarouhas et al., 2011) did not find any difference in tibial rotation between ACLD, ACLI and control knees, which is in line with the findings of the current study. The pivoting maneuver performed in the latter study was, however, slightly different from the one

Table 2 Kinematic and kinetic data from the 3D motion analysis

		ACLD		ACLI		Control	
		Mean	(95% CI)	Mean	(95% CI)	Mean	(95% CI)
WALKING	Rotation	9.2^{1}	(7.8-10.7)	8.6^{2}	(7.1-10.1)	14.2	(11.9-16.5)
	Moment	0.038	(0.02-0.56)	0.028	(0.01-0.044)	0.04	(0.02-0.059)
	Stiffness	0.0067	(0.002-0.01)	0.006	(0.002-0.01)	0.0027	(0.001-0.004)
RUNNING	Rotation	18.9^{3}	(15.3-22.5)	20.5^{4}	(16.9-24.1)	13.9	(12.1-15.6)
	Moment	0.14^{5}	(0.09-0.19)	0.25^{6}	(0.17-0.32)	0.1	(0.07-0.13)
	Stiffness	0.008	(0.003-0.013)	0.016	(0.007-0.024)	0.0075	(0.005-0.0096)
PIVOTING	Rotation	28.2	(23.9-32.4)	27.9	(25.2-30.6)	29.7	(28.4-31.1)
	Moment Median (range)	0.164	(0.06-0.33)	0.168	(0.11-0.44)	0.164	(0.05-0.43)
	Stiffness Median (range)	0.0061	(0.002-0.015)	0.0062	(0.004-0.016)	0.0056	(0.002-0.012)

ACLD ACL deficient knee, *ACLI* contralateral ACL intact knee, Control: knee healthy control group. Rotation: tibial internal rotation, expressed in degrees (deg). Moments: net knee joint external moments, expressed as Nm/kg. Stiffness: rotational stiffness, expressed as (Nm/kg)/deg. Means and 95% Confidence intervals (CI) are reported. Pivoting/rotational moments and pivoting/stiffness were log transformed during statistical analyses; therefore, Median and range are reported for these parameters. ([1]ACLD vs. Control $p < 0.001$, [2]ACLI vs. Control $p < 0.001$, [3]ACLD vs. Control $p = 0.014$, [4]ACLI vs. Control $p = 0.001$, [5]ACLD vs. ACLI $p = 0.015$, [6]ACLI vs. Control $p < 0.001$)

conducted in this current study (stair descent and 60° pivoting (Tsarouhas) versus 90° pivoting (Ristanis)).

In general, higher absolute values of mean rotation during pivoting are measured in our study compared to other authors (28.5° (Table 2) vs. 15.3° (Tsarouhas) and 22.5° (Ristanis)) (Ristanis et al., 2006, Tsarouhas et al., 2011). These differences might be attributed to different definitions of neutral position (i.e., 0° knee rotation). Therefore, in our study, a neutral rotational position was defined as a knee angle five frames before foot contact, while others most often used a standing trial to define the neutral position (Ristanis et al., 2006, Tsarouhas et al., 2010, Tsarouhas et al., 2011, Webster et al., 2010). Thus, this neutral position was chosen to ensure that no external moment affected the limb, forcing zero rotation to correspond to zero moment, which makes sense mechanically because rotation is caused by moments. This furthermore provided us with two sets of corresponding moment-rotation values, enabling us to calculate rotational stiffness, defined as change in moment divided by change in rotation. However, absolute values are of lesser importance in studies where the outcome variable is a difference between absolute values (in our case, absolute ACLD and ACLI rotation angle values).

Existing literature on the kinematics of ACLD knees during walking and running is likewise inconsistent (Andriacchi and Dyrby, 2005, Takeda et al., 2014,Waite et al., 2005, Yim et al., 2015, Zabala et al., 2015). The discrepancies might be due to methodological differences, which make it difficult to compare gait analysis results across studies (Fuentes et al., 2011, Zabala et al., 2015). A possible trend towards an increased internal rotation of the tibia during walking was, however, described by Zabala et al. (Zabala et al., 2015). Although our results are not in line with the current trend, other authors have reported comparable kinematic results to ours while walking and running (Takeda et al., 2014, Yim et al., 2015). These studies did, however, compare only the ACLD knee to their contralateral ACLI knee. Similar to our results, no difference in tibial rotation during the stance phase of walking was found (Takeda et al., 2014, Yim et al., 2015). Additionally, Takeda et al. investigated tibial rotation during running and did not find any significant difference in rotation between knees, which is in line with our findings (Takeda et al., 2014). Waite et al. also investigated running in ACLD knees (Waite et al., 2005). This author reported the ACLD knee to be more internally rotated in the latter part of the stance phase compared to the contralateral uninjured knee, which was different from our findings (Waite et al., 2005).

Few studies have reported on the rotational moments in ACLD knees during natural movements (Fuentes et al., 2011, Tsarouhas et al., 2010, Tsarouhas et al., 2011). The moments during walking were described by Fuentes et al., who found a lower internal rotational moment in ACLD knees compared to a healthy control group (Fuentes et al., 2011). These findings are not in line with the current study. Only Tsarouhas et al. reported on rotational moments of force in ACLD knees during pivoting and, similar to the findings of this current study, no significant difference in rotational moments was found between knees (Tsarouhas et al., 2010, Tsarouhas et al., 2011). To our knowledge, rotational moments during running have not been reported in ACL-deficient subjects in the past. Surprisingly, we found that net external knee joint moments in both knees of the ACL-injured patients were higher than the control group during running. Additionally, our kinematic results during walking and running showed both knees of the ACL-injured group to be significantly different from our healthy control group. These findings indicate some type of adaptive or compensatory strategy for both knees of the ACL-injured patients and, furthermore, that the kinematics and kinetics of the contralateral limb are not necessarily unchanged or representative of healthy control knees when there is an ACL injury in the ipsilateral knee (Zabala et al., 2015).

As mentioned in the introduction section, the rotational stiffness in ACLD knees during natural movements has not been reported previously. Interestingly, the present study found an increase in mean rotational stiffness greater than 50 % between control knees and both knees of the ACL-injured patients during walking. These differences in means were not statistically significant, though. A possible explanation could be that the ACL-injured subjects, as a precaution to episodes of instability, activate their muscles more than healthy subjects. During running, the stiffness of both knees increases with the greater mechanical demand placed on the knee. Hence, running eliminates the double-support phase and reduces the effects of compensation from the contralateral limb. Surprisingly, the stiffness of the ACLI knee during running was more than 50 % higher than the stiffness of the ACLD knee and the control group in the current study, which we cannot explain. These differences in means were, however, not statistically significant either. Finally, we found that stiffness during pivoting was almost alike in all three groups (approximately 0.0056-0.0062 (Nm/kg)/deg), which was quite surprising. Thus, while rotational stiffness during internal rotation in all ACLI knees was provided by the ACL, passive structures and muscle contractions, the ACLD knees must be able to adapt their muscle activation to obtain a suitable rotation and, therefore, a rotational stiffness equal to the ACLI knees during pivoting. The latter statement is supported by Andriacchi et al., who stated that "adaptations to the patterns of muscle firing can compensate for kinematic changes associated with the loss of the ACL" (Andriacchi and Dyrby, 2005). Unfortunately, electromyography (EMG) measurements were not obtained in the present study; these measurements would have contributed important information on muscle activation patterns. Additionally, we did not differentiate between copers and non-copers in our study population, and this could potentially obscure genuine differences in movement patterns (Frobell et al., 2010, Rudolph et al., 2001).

In sum, an increase in internal tibial rotation and knee joint external moment was observed when a higher rotational demand was placed on the knees of all test subjects (progression from walking to running to stair descent/pivoting). However, ACLD knees did not demonstrate increased tibial rotation as hypothesized. The increased load during the different tasks was, however, not immediately reflected in the rotational stiffness, as pivoting displayed the same rotational stiffness as walking in the ACL-injured knees. Therefore, kinetic data are equally important, as the ACL is a passive, elastic structure, and the magnitude of rotation allowed by the ACL depends on the magnitude of the moment applied to by the surroundings to the leg about the tibia's longitudinal axis. If the moment is not carefully measured and reported (similar for the compared legs), rotation is not a valid measure of the ability of the ACL to prevent rotation of the knee, i.e., provide rotational stability.

The results of the current study should be considered in light of the study's limitations. Firstly, a cross-sectional design was used and subjects were compared at different time points since injury (Zabala et al., 2015). Because the pre-injury stiffness of both knees of the patients was not known, knee healthy subjects were selected as a control group instead. Secondly, although 3D motion analysis is widely accepted and well established for advanced functional biomechanical analysis in knee patients, numerous limitations have been described, especially the use of skin markers to predict rotational bone movements (Reinschmidt et al., 1997). To minimize this problem in the present study, marker clusters were used instead of single markers (Cappozzo et al., 1997). Cluster markers are especially useful for measuring optimized rotational measurements in 3D motion analysis. Furthermore, it has been shown that during simultaneously measured knee motion using an optical tracking system and dynamic radiostereometric analysis (RSA), internal/external rotation was fairly similar up to 25° of flexion (Tranberg et al., 2011). As shown in Fig. 1, the pivoting task in this study was performed on an almost extended leg. Thirdly, one of the pairs of corresponding moment-rotation values used for stiffness calculation was 0°, 0 Nm; we defined 0° rotation as the knee rotation angle just prior to ground contact, and assumed the corresponding knee joint moment to be 0, because the moment created by the GRF at this instance by definition was 0; however, while no GRF moment present implies that no counteracting knee joint moment is necessary, knee muscle activity might still have created a knee joint moment and affected the rotation angle. Finally, our cohort consisted of both females and males, which might increase the variability in our data, as several static studies have shown female knees to be more lax and less stiff than male knees (Hsu et al., 2006, Shultz et al., 2012).

Conclusions

In conclusion, we found that a high-intensity activity combining stair descent and pivoting produces similar tibial internal rotations, net knee joint external moments and rotational stiffness in ACL deficient knees compared to contralateral ACL intact knees and a knee healthy control group. During running, the ACL intact knees displayed a higher external moment than the ACL deficient knees and the knee healthy control group. This could indicate muscular adaption or a protective strategy in the ACL-injured patients.

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Authors' contributions

ML and MBB carried out the design of the study. MBB included all patients, conducted mounting of markers on all patients and participated in collection and analysis of data, and drafted the manuscript. MKP and HS helped out in the initial phase of data collection in the gait laboratory. DBN helped out in data collection and data analysis. HS and ML participated in analysis and interpretation of data and critical revision of the manuscript. All authors have read and approved the final manuscript.

Competing interests

The authors declare that they have no competing interests.

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Rotational laxity after anatomical ACL reconstruction measured by 3-D motion analysis: a prospective randomized clinical trial comparing anatomic and nonanatomic ACL reconstruction techniques

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Abstract

Purpose To compare the ability of three different anterior cruciate ligament (ACL) reconstruction techniques to normalize rotational knee stability 1 year after ACL reconstruction. Two of these techniques are so-called anatomic techniques.

Methods Three different ACL reconstruction techniques were tested for their ability to normalize rotational knee stability in a prospective randomized study. Forty-seven ACL-deficient (ACLD) patients were randomized to transtibial single-bundle (SB), anatomic SB, and double-bundle ACL reconstruction. Three-dimensional motion analysis was performed preoperatively and at 1-year follow-up to evaluate tibial rotation and rotational stiffness. Motion data were captured using an eight-camera motion analysis system. Tibial rotation was determined during walking, running, and a pivoting task. Other outcome parameters were KT-1000 knee laxity measurements and the subjective outcome scores KOOS and IKDC.

Results Three-dimensional motion analysis demonstrated that the tibial internal rotation and the rotational stiffness did not differ between the ACL reconstruction techniques during walking, running, and pivoting at 1-year follow-up. Objective knee stability and subjective outcome scores did not differ between the reconstruction groups.

Conclusion No significant difference in rotational stability walking, running, and pivoting was seen between anatomic and nonanatomic ACL reconstruction techniques at 1-year follow-up.

Level of evidence Therapeutic study, Level I.

Keywords ACL reconstruction · Double-bundle reconstruction · Motion analysis · Rotational stability

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Introduction

The past decade has seen a shift towards using anatomic anterior cruciate ligament (ACL) reconstruction techniques to improve knee biomechanics, especially rotational stability. Two anatomic ACL reconstruction techniques have been developed; one uses double-bundle (DB) ACL reconstruction where two graft bundles cover the native footprints of the ACL; the other uses single-bundle (SB) ACL reconstruction with graft placement at the centre of the native femoral ACL footprint with drilling through an anteromedial (AM) portal or retrograde drilling [22, 30]. Reviews of the clinical results after anatomic ACL reconstruction with the DB technique conclude that DB ACL reconstruction results in better anterior–posterior knee stability, reduced pivot-shift, and in a few studies also report better subjective outcome than with SB ACL reconstruction [1, 12, 15, 23, 24, 27, 31, 32].

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Furthermore, biomechanical in vitro studies have suggested that a DB anatomic ACL reconstruction can result in more normal rotational stability in vitro than SB reconstruction [7, 21, 29]. Clinical evaluation of rotational laxity is traditionally performed with the pivot-shift test. This test is highly subjective and observer-dependent and, furthermore, easily biased by the patient's muscle guarding [13, 14]. New quantitative methods for measuring static knee rotation with robotics or inertial sensors have recently been presented [3, 16]. Functional knee rotation can be assessed by three-dimensional (3-D) motion analysis, and several studies have analysed knee rotation during various activities and after different ACL reconstruction techniques. Traditionally, kinematic data (angular displacement) are reported, but two randomized controlled studies used 3-D motion analysis to compare SB with DB reconstruction [6, 11]. Hemmerich et al. compared anatomic SB reconstruction with anatomic DB reconstruction during a cutting manoeuvre and found no difference in overall rotation. However, they found greater external rotational shift in the SB-reconstructed group than in the DB group. These data are consistent with findings by Claes et al. who found no difference in tibial rotation between anatomic SB and DB reconstructions and therefore concluded that there is no advantage of a DB reconstruction over an anatomic SB reconstruction. Some nonrandomized studies have used 3-D motion analysis for comparison of DB reconstructions with SB reconstructions during pivoting manoeuvres (level III studies) have reported no difference in tibial rotation between the two reconstruction techniques. Interestingly, some of these studies also report kinetic data, and they found a lower moment of force on the reconstructed knee than on the contralateral, intact knee [18, 20, 25]. There is, however, a lack of randomized studies investigating the impact of different anatomic ACL reconstructions on tibial rotational biomechanics.

The aim of the present prospective randomized clinical trial was to evaluate and compare the rotational laxity and stiffness after anatomic DB ACL reconstruction, anatomic SB ACL reconstruction, and nonanatomic SB ACL reconstruction using 3-D motion analysis for evaluation of rotational biomechanics. We hypothesized that anatomic ACL reconstructions would result in better rotational stability than nonanatomic SB ACL reconstruction.

Materials and methods

Patients

In this prospective randomized clinical trial (single-blinded), 45 patients underwent ACL reconstruction. Inclusion criteria were: age 18–50 years, magnetic resonance imaging-verified ACL injury with symptoms of instability, no previous knee ligament surgery, no concomitant knee ligament injuries, and an uninjured contralateral knee. Exclusion criteria were cartilage injuries of International Cartilage Research Society grade 3 or 4, and/or meniscus injury requiring resection of more than 50 % of a meniscus. Furthermore, 16 age- and sex-matched healthy control subjects who had no history of lower limb pathology or trauma were selected as a control group, which underwent 3-D motion analysis similar to that of the ACL-reconstructed patients.

The local ethical committee approved the study, and written informed consent was obtained from every subject. The study was conducted between June 2009 and January 2012. The study was approved by Region Midtjylland ethical committee (M-AÅ-20060198).

The demographic data of the three randomized groups and the control group are presented in Table 1. There were no significant differences between the four groups in terms of sex, age, height, weight, and body mass index (BMI). The median time from injury to operation was not significantly different between the three groups, although a tendency was seen ($p = 0.053$).

Six experienced orthopaedic surgeons performed the ACL reconstructions. A diagnostic arthroscopy was performed initially during each operation to confirm the ACL lesion and to identify cartilage and meniscus lesion exclusion criteria. If no exclusion lesions were present, the patients were randomized with the closed envelope method into three different groups of ACL reconstruction: four-tunnel anatomic DB (DB group) reconstruction, anatomic

Table 1 Demographics of the randomized groups and the control group

(Mean ± SD)	DB	SB-AM	SB-TT	Control group	p value
Sample size	16	15	14	16	
Sex (F/M)	5 and 10	7 and 8	6 and 8	6 and 10	ns
Age (y)	26.5 ± 6.4	24.3 ± 4.9	27.5 ± 7.2	25.6 ± 3.6	ns
Height (cm)	179 ± 8	174 ± 8	179 ± 9	178 ± 8	ns
Weight (kg)	78.7 ± 13.7	75.7 ± 15.1	74.9 ± 14.4	73.7 ± 7.5	ns
BMI (kg/m^2)	24.7 ± 3.3	24.6 ± 3.1	23.1 ± 3.1	23 ± 1.5	ns
Injury–surgery (months)	6 (2–26)	6 (3–16)	12 (4–42)	–	ns

single-bundle anteromedial (SB-AM) group reconstruction, and single-bundle transtibial (SB-TT) group reconstruction.

Surgical technique

The semitendinosus and gracilis tendons were harvested with a tendon stripper through a horizontal tibial incision at the pes anserinus.

Four-tunnel double-bundle ACL reconstruction: The semitendinosus tendon (for the AM bundle) and the gracilis tendon (for the PL bundle) were looped over a 20-mm EndoButton CL femoral fixation implant (Smith & Nephew Endoscopy, Mansfield, MA, USA). The distal free ends of the tendons were armed with No. 2 sutures using a whipstitch technique, and the grafts were pretensioned on a suture board. The tibial and femoral ACL footprints and the intercondylar notch were cleaned from soft tissue to obtain an exact arthroscopic view of the AM and posterolateral bundle insertion sites. No notchplasty was performed. The tunnel diameter varied from 6 to 7 mm for AM bundles and from 5 to 6 mm for PL bundles depending on graft diameters. The femoral AM and PL bone tunnel were drilled using an accessory AM portal at the insertion sites of the AM and PL bundle just below the intercondylar ridge with the knee flexed to 120°. A bone bridge of at least 2 mm was preserved between the femoral tunnels in all patients. Tibial bone tunnels were positioned in the intercondylaris anterior area. The tibial AM bone tunnel was positioned in the AM aspect of the ACL footprint and the PL bone tunnel in the posteromedial aspect of the insertion area respecting the natural border of the tibial ACL footprint. The tibial and femoral bone tunnels were drilled with a conventional reamer on the tibial side and with a headed reamer on the femoral side. After positioning of the graft, the femoral EndoButton position was secured, and the grafts were tensioned by 10 cycles of knee motion. Tibial fixation was performed for each graft using a 30-mm Inion Hexalon™ interference screw (Inion Oy, Tampere, Finland). In both tibial tunnels, a screw with a diameter 1 mm larger than the tunnel diameter was used. The screws were placed in the most distal part of the bone tunnel. Both bundles were fixed in 20° of knee flexion.

Single-bundle ACL reconstructions: The tibial bone tunnel was positioned in the intercondylaris anterior area in the centre of the native tibial ACL footprint using the inner aspect of the lateral meniscus anterior insertion area as a landmark. In the anatomic SB reconstruction, the femoral bone tunnel was drilled at the centre of the femoral ACL footprint with a knee flexion angle of 120° using an accessory AM portal. For the transtibial SB reconstruction, the femoral tunnel was drilled through the tibial tunnel. A femoral drillguide with an offset of 7 mm was used for initial K-wire positioning. The drillhole was positioned at a 10 or 2 o'clock position orientating from the posterior wall of the notch. After K-wire positioning, overdrilling was first performed with a 4.5-mm drill to ensure passage of the EndoButton fixation devices and subsequently with a headed reamer with the size of the prepared graft. For both of the SB ACL reconstructions, the semitendinosus and gracilis tendons were looped over one single 20 mm EndoButton CL femoral fixation device (Smith&Nephew, Andover, MA, USA). After positioning of the graft, the femoral EndoButton position was secured, and the grafts were tensioned by 10 cycles of knee motion. Finally, tibial fixation was ensured with biodegradable interference screw (Inion Hexalon™, Inion Oy, Tampere, Finland) with a knee flexion angle of 20°. The tibial fixation screws were 30-mm-long screw; the diameter was 1 mm larger than the bone tunnel, and they were positioned in the most distal part of the tunnel.

Rehabilitation

Rehabilitation was the same in the three reconstruction groups. The subjects were allowed immediate full-weight bearing and full range of motion. Crutches were used for 2 weeks for longer walking distances. No brace was used. Closed-chain exercises were started immediately postoperatively. Cycling on a stationary bike was permitted after 2 weeks and on normal bike after 4 weeks. Running was allowed at 3 months, noncontact sports after 6 months, and contact/pivoting sports 12 months after surgery, provided that the patient had regained full functional stability. If meniscus repair was performed simultaneously with the ACL reconstruction, the range of motion was limited to 0°–90° for the first 6 weeks. Subsequently, the rehabilitation was carried out as described above. The patients were blinded to the reconstruction technique during rehabilitation.

Testing procedure and data analysis

Clinical evaluation

At the time of inclusion, each patient was examined clinically with a Lachman test, a pivot-shift test, and a passive anterior knee laxity test using a KT-1000 arthrometer (Medmetric® Corp., San Diego, California, USA). Additionally, four subjective outcome questionnaires were completed: the International Knee Documentation Committee (IKDC) subjective score, the Knee Osteoarthritis Outcome Score (KOOS), the Tegner score, and the Lysholm score. The KOOS questionnaire consists of five subscales. The KOOS4 score is the average score of the following four subscales: pain, other symptoms, function in sport and recreation, and Knee-related quality of life, which are most responsive for ACL patients. A score of 100 indicates no symptoms, and 0 indicates extreme symptoms [8]. At 1-year follow-up, one blinded physiotherapist and one senior surgeon performed a clinical examination of all the patients (Pivot, Lachman, and KT-1000), and the four subjective questionnaires were completed once again. Furthermore, a leg extensor power rig (Queens Medical Center, Nottingham, UK) was used to measure the quadriceps strength on both legs (force/speed), and single-leg one- and triple-hop tests were performed. Quadriceps strength is presented as a percentage of the strength of the nonoperated leg.

Three-dimensional motion analysis

Three-dimensional motion analysis was conducted up to 3 weeks before and 12–18 months after ACL reconstruction, while control subjects were tested only once.

The motion analysis protocol was identical for patients and control subjects. Each patient was asked to perform three different tasks in the following order: (1) level walking, (2) running/jogging, and (3) stair descend followed by a 90° pivoting manoeuvre. For the stair descend, a three-step plywood staircase with no handrail was used according to Andriacchi et al. [2] (rise 21 cm, run 25 cm, width 48 cm). The participants were asked to descend the staircase at their own pace. Following contact with a force plate at the bottom of the staircase, the subjects were instructed to make a pivoting manoeuvre by moving the swing leg through a 90° arc across the stance leg and to contact the ground with the foot of the swing leg at a 90° angle relative to the stance foot, which was required to maintain full plantar side force plate contact (Fig. 1). This pivoting manoeuvre was designed to induce an internal rotation of the shank relative to the thigh of the stance leg. The subjects then walked a few steps away from the plate. At least 10 successful trials for both the ACL-defiant and ACL-intact knee were recorded. For the control group, 10 successful trials for both legs were recorded.

Fig. 1 A study subject performing the pivoting task. The marker placement for 3-D motion analysis can be seen

Motion analysis data collection

All participants were bare-footed and fitted with 16 passive reflective markers at each limb: eight markers on bony prominences (greater trochanter, medial and lateral femoral condyle, medial and lateral malleolus, heel, and 1st and 5th metatarsal head) to define anatomic planes and joint centres. The remaining eight markers were placed as two clusters on rigid plates (one four-marker cluster on the shank and thigh segment, respectively) (Fig. 1). To measure the position of the reflective markers, eight optoelectronic motion capture cameras (ProReflex MCU 1000, Qualisys Medical AB, Gothenburg, Sweden) operating at 240 frames per second were used together with Qualisys Tracking Manager (QTM) software on a personal computer.

The collected trajectory data were gap-filled if required in QTM using nonuniform rational B-spline interpolation and exported to Visual3D software (C-motion Inc., Kingston, Canada) where a Visual 3-D Hybrid Model for ideal rigid segments and 6-degrees of freedom was applied. The marker position data were low-pass filtered using a second-order Butterworth digital filter with one bidirectional pass (effectively making it a fourth-order filter) and an effective cutoff frequency of 6 Hz.

Ground reaction force (GRF) was sampled simultaneously using an AMTI OR6-6 force plate (Advanced Medical Technology Inc., MA, USA) sampled at 960 Hz. The GRF data were low-pass filtered with a cutoff frequency of 30 Hz. The analysis period started when the vertical force exceeded 20 N and ended when it fell below 20 N.

Data analysis

For each patient, the knee angle and the moment of force were calculated from the marker positions and the GRF data using inverse dynamics for an idealized rigid segment. All moments were normalized to body mass. Anthropometric data were calculated from individual body mass and height using Dempster's regression equations. Knee rotation was calculated based on the joint coordinate system definition describing knee rotation as occurring around the longitudinal axis of the shank [10]. The neutral position of the knee (0° knee rotation) was defined as the knee angle five frames before foot contact, i.e., the knee was close to being fully extended with no external moment affecting the limb. For each trial, we determined the maximum tibial internal rotation; we used each subject's (patients and controls) second highest value (to avoid potential outlier bias) together with the corresponding rotational moment for further analyses. Furthermore, these values were used to calculate the rotational stiffness of each knee by dividing the rotational moment with the maximum tibial internal rotation.

Statistical analysis

The size of the groups was based on the power calculation described below. Previous studies have demonstrated that tibial rotation is in the range 23°–24° in ACL-reconstructed knees using nonanatomic techniques, and intact knee tibial rotation was in the range 16°–17°. The standard deviation for rotation measurements during 3-D motion analysis is around 4° [18]. We hypothesized that an anatomic ACL reconstruction could reduce tibial rotation better than a nonanatomic ACL reconstruction with a 4° rotation reduction, which is 50 % of the difference between intact and ACLD knees. With these assumptions, nine subjects per group were needed to demonstrate significant difference with a power of 0.8. We chose to include 15 subjects per group to account for dropouts due to lack of follow-up and problems with data retrieval from 3-D motion analysis.

The chi-square test was used to evaluate differences in pivot-shift test, Lachman test, and the objective IKDC between groups. An ANOVA test was used to evaluate differences between the groups in terms of subjective outcome scores, KT-1000 arthrometric measurements, quadriceps strength, and the one-leg hop tests. The pre–postoperative comparison of these parameters was performed using a paired t test. Three-dimensional motion analysis data were analysed by ANOVA test, and the Bonferroni correction was used when a significant difference was found. The significance level was set at $p = 0.05$.

Results

Of the 45 randomized and reconstructed patients, 36 (80 %) (13 in DB group, 12 in SB-AM group, 11 in SB-TT group) were available for an average of 13 months of follow-up (range 12–18 months) (Fig. 2). Of these 36 patients, 4 had been operated during the follow-up period. Two patients in the DB group had an early postoperative infection that required arthroscopic lavage and antibiotic treatment, and two patients in the SB-TT group had the tibial screw removed. These patients completed 1-year follow-up including motion analysis.

Clinical results (Table 2)

No difference in IKDC, KOOS and Tegner scores, KT-1000 knee laxity, Lachman test, pivot-shift test, and hop

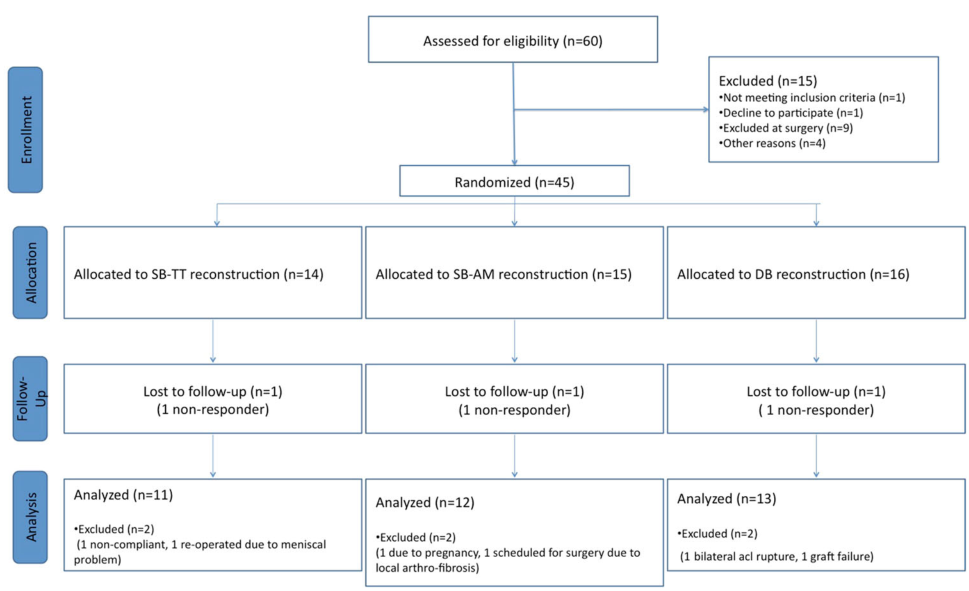

Fig. 2 Study flow and patient randomization and follow-up

Table 2 Preoperative and follow-up results of objective IKDC objective clinical findings and subjective outcome scores

	Preoperatively			Follow-up		
	Double bundle	Single-bundle transtibial	Single-bundle anteromedial	Double bundle	Single-bundle transtibial	Single-bundle anteromedial
IKDC A (normal) (%)	0	0	0	38	25	27
IKDC B (nearly normal) (%)	33	73	21	54	75	73
IKDC C (abnormal) (%)	47	20	50	8	0	0
IKDC C (severely abnormal) (%)	20	7	29	0	0	0
KT-1000 (max) mm	3.7 ± 1.9	3.9 ± 2.6	2.8 ± 2.1	1.6 ± 2.1*	2.3 ± 1.9	2.0 ± 1.7
Pivot-shift test						
Normal (%)	13	38	29	92*	75*	82*
Lachman test						
Normal (%)	20	47	21	85*	75*	73*
Subjective IKDC	63 ± 11	58 ± 13	62 ± 11	76 ± 11*	71 ± 15*	76 ± 13*
KOOS4	64 ± 14	57 ± 13	64 ± 12	78 ± 13*	73 ± 18*	73 ± 13*
Tegner score	3.6 ± 1.0	3.7 ± 2.1	3.9 ± 0.9	5.5 ± 1.4*	5.6 ± 1.2*	5.5 ± 1.0*
Lysholm score	73 ± 15	70 ± 9	73 ± 14	87 ± 14*	81 ± 14*	86 ± 12*
Quadriceps strength (% of normal leg)				103 (15)	88 (18)	94 (16)
Single hop (% of normal leg)				91 (13)	95 (17)	93 (14)
Triple hop (% of normal leg)				92 (6)	91 (7)	97 (10)

* Significant difference from preoperative to follow-up state $p < 0.05$

tests were found between the three surgical groups at follow-up. All three groups reported significantly subjective improvements in patient-evaluated functional scores at 1-year follow ($p < 0.01$).

Three-dimensional motion analysis (Table 3)

There was no difference between SB-TT, SB-AM, and DB groups regarding tibial rotation and rotational stiffness at follow-up one year after ACL reconstruction for walking, running, and pivoting. Also, no difference was found between the three different surgical groups and the two control groups for the rotational parameters.

The change in tibial internal rotation during a pivot stress test from before the operation to one year after surgery was 0.8°, −2.8°, and 2.9° for DB, SB-AM, and SB-AM groups, respectively. A reduction in tibial rotation was seen only for the SB-AM group. But the changes in rotation from before surgery to one year after surgery did not differ significantly between the three surgical groups.

Table 3 Kinematic data from pivoting task of stair descend and turning 90°

	Single-bundle transtibial	Single-bundle anteromedial	Double bundle	Intact knee	Control group	p value
Walking						
Tibial rotation (°)	13.7 (6.1)	12.7 (4.9)	16.4 (4.6)	15.7 (4.9)	13.6 (6.2)	NS
Stiffness	3.5 (4.9)	4.3 (3.4)	3.3 (2.9)	4.2 (5.3)	2.9 (5.2)	NS
Running						
Tibial rotation (°)	20.5 (7.8)	15.7 (3.2)	20.4 (3.6)	17.3 (5.6)	18.9 (5.3)	NS
Stiffness	5.7 (6.2)	6.2 (5.5)	5.9 (3.2)	5.7 (5.0)	8.0 (6.9)	NS
Pivot						
Tibial rotation (°)	30.4 (5.4)	31.7 (7.7)	31.1 (5.5)	31.4 (6.2)	27.9 (5.3)	NS
Stiffness	5.1 (1.4)	3.8 (1.9)	4.2 (1.5)	4.2 (1.6)	7.1 (3.5)	NS

Data from the three reconstruction groups and the two controls groups at 1-year follow-up are presented. Tibial rotation is maximal internal tibial rotation during the pivot task. Stiffness is defined as rotation/moment of force during pivot and is presented with the unit (Nm 10^{-3}/kg/deg). The p value indication refers to comparisons between the different surgical groups and comparisons between surgical group and the two control groups

Discussion

The main finding of this study was that there was no difference between anatomic ACL reconstruction techniques and nonanatomic technique for knee tibial rotation and rotational stiffness during walking, running, and pivoting after stair descend at 1-year follow-up. Moreover, no difference was found between the reconstruction techniques in change of internal rotation from the preoperative ACL-injured state to the ACL-reconstructed state at the 1-year follow-up.

Cross-sectional studies traditionally present kinematic data (tibial rotational excursion) and use 3-D motion analysis to measure knee stability during pivoting [4, 5, 9, 18, 19]. These studies compare the reconstructed knee to the uninjured contralateral knee at follow-up. Ristanis et al. [19], for example, found no difference in tibial rotation between normal controls and patients in patients with patella tendon graft ACL reconstruction. Only a few studies have investigated postoperative knee rotation after with different ACL reconstruction techniques using 3-D motion analysis in randomized study designs. One such study by Claes et al. compared anatomic SB and DB reconstruction. They found no difference in tibial rotational excursion between the SB- and DB-reconstructed knees at 6-month follow-up. They concluded that both anatomic SB and DB reconstruction adequately restored tibial rotational excursion and that there was no difference between the two techniques [6]. The biomechanical impact of using AM portal femoral drilling for anatomic SB ACL reconstruction has recently been investigated. In a study by Wang et al. transtibial drilling and AM drilling for SB ACL reconstruction were compared using 3-D motion analysis. They found better normalization of AP translation and tibial rotation during walking for knees with AM drilling reconstructions [28].

The present study found that the magnitude of tibial rotation was less affected by walking than more stressful manoeuvres such as running and pivoting. We found no reduction in tibial rotation as a result of ACL reconstruction for walking, running, and pivoting. We could therefore not reproduce the finding of reduced rotation during walking reported by Wang et al. The present study thus presents data that conflict with those of previous studies concerning tibial rotation after DB ACL reconstruction. Still, the data presented in the present study are valuable since it is the first randomized study testing knee biomechanics of two different types of anatomic ACL reconstruction, the DB technique and the anatomic SB technique, using AM drilling of femoral fixation holes. Moreover, it is the first randomized study comparing the results of these techniques with that of a nonanatomic transtibial ACL reconstruction technique.

The ACL is an elastic structure, and the knee's rotational stability therefore cannot be determined on the basis of the rotation angle only. The rotation (angular displacement) depends on both the moment of force causing the rotation and the ACL's (and other knee structures) ability to resist the rotation. Rotational knee stability could also be expressed as stiffness, i.e., moment divided by rotation. For this reason, we also investigated rotational moments of force and present stiffness data as well as angular rotation data. Tsarouhas et al. [25, 26] reported rotational moments of force developed during a pivoting manoeuvre. They found that DB ACL reconstruction did not reduce knee rotation moment compared with SB reconstruction and that the ACL-reconstructed knee was subjected to reduced knee rotational moments compared with the intact knee during stressful functional manoeuvres. Our kinetic findings are consistent with the finding of reduced rotational moment in ACL-reconstructed knees. Potentially neurologic mechanisms that aim to protect the reconstructed knee could result in reduced moment of force during stressful pivoting.

We found no significant differences in clinical outcome between the three different reconstruction techniques using subjective outcome instruments and objective measures for knee stability. All three groups significantly improved their subjective scores at follow-up compared with their preoperative status. Objective knee stability also improved in all three groups. Several authors have compared the clinical outcome of the anatomic DB technique to that of SB reconstruction techniques in randomized studies. Recent reviews conclude that DB ACL reconstruction results in better anterior–posterior knee stability and reduced pivot-shift, and a few studies also report better subjective outcome than after SB ACL reconstruction [1, 12, 23, 24, 32].

Since the present study was powered to investigate the use of 3-D analysis parameters, the small number of subjects is the most likely explanation for why our findings for clinical outcome parameters contradict those of other randomized studies.

Overall, we could not confirm our hypothesis that ACL reconstruction using anatomic techniques results in better normalization of functional tibial rotation than nonanatomic techniques. The clinical relevance of the present study is that functional knee rotation is only limited affected by different types of ACL reconstruction. Further studies are needed to identify reconstruction techniques that better control knee rotation in clinical situations where this is needed. This could be done by extraarticular reconstruction principles.

The present study has several strengths. First of all, it utilizes a randomized controlled design to investigate the biomechanical impact of three different ACL reconstruction techniques. All patients were recruited from a single clinic and followed the same rehabilitation protocol.

The study included two control groups: the intact nonoperated knee and knees from a healthy control group. In an ACL patient, the intact knee can be subject to compensatory forces, and the use of an independent control group for studying rehabilitation differences is therefore to be preferred. Also, 3-D motion analysis is widely accepted and well established as a gold standard for advanced functional biomechanic analysis in patients after knee surgery.

A limitation of the present study is its use of gait analysis in general and its use of skin markers to predict rotational bone movements [17]. To minimize this problem, marker clusters were used instead of single markers. Cluster markers are especially useful for measuring optimized rotational measurements in 3-D motion analysis.

Six different surgeons performed the ACL reconstructions, which could contribute to a variation in surgical technique. However, all surgeons were skilled ACL surgeons, and all details of the surgical technique used in the present study were agreed upon prior to study start.

The sample size in the study groups is small. This number was based on power calculation for the rotational parameter of the 3-D motion analysis. So, the data are underpowered for analysis of secondary outcomes such as objective knee stability and subjective knee scores. Caution is therefore advised when interpreting the result of subjective outcome parameters.

Conclusion

No significant difference in rotational stability during walking, running, and pivoting was seen between anatomic and nonanatomic ACL reconstruction techniques at 1-year follow-up.

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2000
IKDC KNEE EXAMINATION FORM

Patient Name:______________________________

Label

SEVEN GROUPS	**FOUR GRADES** A Normal	B Nearly Normal	C Abnormal	D Severely Abnormal	*Group Grade A B C D
1. Effusion	☐ None	☐ Mild	☐ Moderate	☐ Severe	☐ ☐ ☐ ☐
2. Passive Motion Deficit					
ΔLack of extension	☐ <3°	☐ 3 to 5°	☐ 6 to 10°	☐ >10°	
ΔLack of flexion	☐ 0 to 5°	☐ 6 to 15°	☐ 16 to 25°	☐ >25°	☐ ☐ ☐ ☐
3. Ligament Examination *(instrumented, KT-1000)*					
ΔLachman (25° flex) (134N)	☐ -1 to 2mm	☐ 3 to 5mm(1$^+$) ☐ <-1 to –3	☐ 6 to 10mm(2$^+$) ☐ <-3 stiff	☐ >10mm(3$^+$)	
Anterior endpoint:	☐ firm		☐ soft		
ΔTotal AP Translation (25° flex)	☐ 0 to 2mm	☐ 3 to 5mm	☐ 6 to 10mm	☐ >10mm	
ΔTotal AP Translation (70° flex)	☐ 0 to 2mm	☐ 3 to 5mm	☐ 6 to 10mm	☐ >10mm	
ΔPosterior Drawer Test (70° flex)	☐ 0 to 2mm	☐ 3 to 5mm	☐ 6 to 10mm	☐ >10mm	
ΔMed Joint Opening (20° flex/valgus rot)	☐ 0 to 2mm	☐ 3 to 5mm	☐ 6 to 10mm	☐>10mm	
ΔLat Joint Opening (20° flex/varus rot)	☐ 0 to 2mm	☐ 3 to 5mm	☐ 6 to 10mm	☐ >10mm	
ΔExternal Rotation Test (30° flex prone)	☐ <5°	☐ 6 to 10°	☐ 11 to 19°	☐ >20°	
ΔExternal Rotation Test (90° flex prone)	☐ <5°	☐ 6 to 10°	☐ 11 to 19°	☐ >20°	
ΔPivot Shift	☐ equal	☐ +glide	☐ ++(clunk)	☐ +++(gross)	
7. Functional Test					☐ ☐ ☐ ☐
One Leg Hop (% of opposite side)	☐ ≥90%	☐ 89 to 76%	☐ 75 to 50%	☐ <50%	
****Final Evaluation**					☐ ☐ ☐ ☐

* Group grade: The lowest grade within a group determines the group grade

** Final evaluation: the worst group grade determines the final evaluation for acute and subacute patients. For chronic patients compare preoperative and postoperative evaluations. In a final evaluation only the first 3 groups are evaluated but all groups must be documented. Δ Difference in involved knee compared to normal or what is assumed to be normal.

IKDC COMMITTEE AOSSM: Anderson, A., Bergfeld, J., Boland, A. Dye, S., Feagin, J., Harner, C. Mohtadi, N. Richmond, J. Shelbourne, D., Terry, G. ESSKA: Staubli, H., Hefti, F., Hoher, J., Jacob, R., Mueller, W., Neyret, P. APOSSM: Chan, K., Kurosaka,

Spørgeskema til knæpatienter

KOOS

Label

Vejledning: Dette spørgeskema indeholder spørgsmål om, hvordan du oplever dit knæ. Informationerne vil hjælpe os til at følge med i hvordan du har det og hvor godt du klarer dig i dagligdagen.
Ved hvert spørgsmål skal du sætte et kryds i det alternativ, der passer bedst på dig. Du skal kun sætte krydset ved det alternativ, der føles mest rigtigt.

Symptomer

Tænk på de knæegener du har haft i løbet af den sidste uge, når du besvarer de næste spørgsmål

S1 Har knæet været hævet?

Aldrig	sjældent	Ind i mellem	Ofte	Altid
☐	☐	☐	☐	☐

S2 Har du haft murren i knæet, hørt klik eller andre lyde fra knæet, når du bevæger det?

Aldrig	sjældent	Ind i mellem	Ofte	Altid
☐	☐	☐	☐	☐

S3 Har knæet haget sig fast eller været låst?

Aldrig	sjældent	Ind i mellem	Ofte	Altid
☐	☐	☐	☐	☐

S4 Kan du strække knæet helt?

Altid	Ofte	Ind i mellem	sjældent	Aldrig
☐	☐	☐	☐	☐

S5 Kan du bøje knæet helt?

Altid	Ofte	Ind i mellem	sjældent	Aldrig
☐	☐	☐	☐	☐

Stivhed:

Følgende spørgsmål handler om ledstivhed. Ledstivhed medfører besvær med at komme igang eller øget modstand, når du bøjer knæet. Angiv i hvor høj grad du har oplevet ledstivhed i knæet i løbet af den sidste uge:

S6 Hvor stift er dit knæ, når du lige er vågnet om morgenen?

Slet ikke	lidt	Moderat	Meget	Ekstremt
☐	☐	☐	☐	☐

S7 Hvor stift er dit knæ senere på dagen, når du har siddet eller ligget og hvilet?

Slet ikke	lidt	Moderat	Meget	Ekstremt
☐	☐	☐	☐	☐

Udfyldes af Idrætsklinikken

Præop ☐ **6mdr** ☐ **12 mdr** ☐ **36 mdr** ☐ **Kons 1 år** ☐

Smerte:

P1 Hvor ofte har du ondt i knæet?

Aldrig	Hver måned	Hver uge	Hver dag	Altid
☐	☐	☐	☐	☐

Hvor mange knæsmerter har du haft i løbet af den sidste uge, under følgende aktiviteter?

P2 Dreje/vride på belastet knæ

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P3 Strække knæet helt

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P4 Bøje knæet

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P5 Gå på jævnt underlag

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P6 Gå op eller ned ad trapper

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P7 Om natten (smerter som forstyrrer din søvn)

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P8 Siddende eller liggende

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

P9 Stående

Ingen	Lette	Moderate	Stærke	Ekstremt
☐	☐	☐	☐	☐

Funktion i hverdagen

Følgende spørgsmål omhandler dit fysiske formåen. Angiv hvilken grad af besvær du har oplevet under følgende aktiviteter løbet af den sidste uge, på grund af dine knæproblemer.

A1 Gå ned ad trapper

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A2 Gå op ad trapper

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A3 Rejse dig fra siddende

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

Angiv graden af besvær du har oplevet ved hver aktivitet i løbet af den sidste uge.

A4 Stå stille

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A5 Gå ned i knæ, fx for at samle noget op fra gulvet

	Intet	Lidt	Moderat	Stort	Ekstremt
	☐	☐	☐	☐	☐

A6 Gå på jævnt underlag

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A7 Gå ind/ud af en bil?

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A8 Tage på indkøb

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A9 Tage strømper af

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A10 Stå ud af sengen

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A11 Tage strømper på

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A12 Ligge i senge (vende dig, have knæet i samme stilling i lang tid)

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A13 Stige ind og ud af badekar/brusebad

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A14 Sidde

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A15 Sætte dig og rejse dig fra toilettet

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A16 Udføre tungt hus arbejde (vaske gulv, støvsuge, bære øl/sodavandskasser og lign.)

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

A17 Udføre let husarbejde (lave mad, tørre støv etc)

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

Funktion, sport og fritid

Følgende spørgsmål handler om din fysiske formåen. Angiv hvilken grad af bevsær du har oplevet under følgende aktiviteter i løbet af den sidste uge på grund af dine knæproblemer.

SP1 Sidde i hug

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

SP2 Løbe

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

SP3 Hoppe

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

SP4 Dreje/vride på belastet knæ

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

SP5 Ligge på knæ

Intet	Lidt	Moderat	Stort	Ekstremt
☐	☐	☐	☐	☐

Livskvalitet

Q1 Hvor ofte bliver du mindet om dit knæproblem

Aldrig	Hver måned	Hver uge	Hver dag	Altid
☐	☐	☐	☐	☐

Q2 Har du forandret din måde at leve på for at undgå at overbelaste knæet

Slet ikke	Noget	Moderat	I stor udstrækning	Totalt
☐	☐	☐	☐	☐

Q3 I hvor stor grad kan du stole på dit knæ?

Fuldt ud	I stor udstrækning	Moderat	Til en vis grad	Slet ikke
☐	☐	☐	☐	☐

Q4 Hvor store problemer har du almindeligvis med dit knæ?

Ingen	Små	Moderate	Store	Ekstreme
☐	☐	☐	☐	☐

Tak for at du har besvaret samtlige spørgsmål.

IKDC SUBJECTIV KNÆ EVALUERINGSSKEMA

SYMPTOMER*:

1. **Hvad er det højeste aktivitetsniveau du kan klare uden at få væsentlige knæsmerter**?

- ❑ 5 Meget krævende aktivitet som idræt med spring og retningsskrift som i fodbold eller basketball
- ❑ 4 Krævende aktivitet som hårdt fysisk arbejde, står på ski eller tennis.
- ❑ 3 Moderat aktivitet som let fysisk arbejde, løb eller jogging.
- ❑ 2 Let aktivitet som almindelig gang, hus eller have arbejde.
- ❑ 1 Er ikke i stand til at udføre nogle af ovenstående aktiviteter.

2. **Indenfor de seneste 4 uger, eller siden du kom til skade, hvor ofte har du smerter i dit knæ**.

	1	2	3	4	5	6	7	8	9	10	11	
konstant smerter	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	Aldrig smerter

3. **Hvis du har smerter, hvor slemme er de?**

	1	2	3	4	5	6	7	8	9	10	11	
Værst Tænkelige smerter	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	ingen smerter

4. **Indenfor de seneste 4 uger, eller siden du kom til skade, hvor stift eller hævet har dit knæ været.**

- ❑ Overhovedet ikke
- ❑ Let
- ❑ Moderate
- ❑ Meget
- ❑ Ekstremt

5. **Hvad er det højeste aktivitetsniveau du kan klare uden at få væsentlig hævelse af knæet?**

- ❑ 5 Meget krævende aktivitet som idræt med spring og retningsskrift som i fodbold eller basketball
- ❑ 4 Krævende aktivitet som hårdt fysisk arbejde, står på ski eller tennis.
- ❑ 3 Moderat aktivitet som let fysisk arbejde, løb eller jogging.
- ❑ 2 Let aktivitet som almindelig gang, hus eller have arbejde.
- ❑ 1 Er ikke i stand til at udføre nogle af ovenstående aktiviteter.

6. **Indenfor de seneste 4 uger, eller siden du kom til skade, har dit knæ låst sig fast eller har der været fornemmelse af at det var ved at låse sig fast.**

❑ 1 Ja ❑ 3 Nej

7. **Hvad er det højeste aktivitetsniveau du kan klare uden at dit knæ giver efter eller pludselig svigter under dig.**

❑ 5 Meget krævende aktivitet som idræt med spring og retningsskrift som i fodbold eller basketball
❑ 4 Krævende aktivitet som hårdt fysisk arbejde, står på ski eller tennis.
❑ 3 Moderat aktivitet som let fysisk arbejde, løb eller jogging.
❑ 2 Let aktivitet som almindelig gang, hus eller have arbejde.
❑ 1 Er ikke i stand til at udføre nogle af ovenstående aktiviteter.

SPORTSAKTIVITETER:

8. **Hvad er det højeste sports eller aktivitetsniveau du kan deltage i jævnligt**.

❑ 5 Meget krævende aktivitet som idræt med spring og retningsskrift som i fodbold eller basketball
❑ 4 Krævende aktivitet som hårdt fysisk arbejde, står på ski eller tennis.
❑ 3 Moderat aktivitet som let fysisk arbejde, løb eller jogging.
❑ 2 Let aktivitet som almindelig gang, hus eller have arbejde.
❑ 1 Er ikke i stand til at udføre nogle af ovenstående aktiviteter.

9. **Hvordan påvirker dit knæproblem din evne til at-:**

		Overhove det ikke svært 5	Let svært 4	Moderate svært 3	Ekstremt svært 2	Kan ikke gennem-føres 1
a.	Gå op ad trapper	❑	❑	❑	❑	❑
b.	Gå ned ad trapper	❑	❑	❑	❑	❑
c.	Sidde på knæ	❑	❑	❑	❑	❑
d.	Sidde på hug	❑	❑	❑	❑	❑
e.	Sidde med knæene bøjede	❑	❑	❑	❑	❑
f.	Rejse sig fra en stol	❑	❑	❑	❑	❑
g.	Løbe lige fremad	❑	❑	❑	❑	❑
h.	Hoppe og lande på det dårlige ben	❑	❑	❑	❑	❑
i.	Hurtig stop og start	❑	❑	❑	❑	❑

FUNKTION:

10. Hvordan vil du vurdere din knæfunktion på en skala fra 1 til 11 med 11 som normal, perfekt knæ funktion og 1 værende total manglende evne til at udføre dagligdagsaktiviteter inklusiv sportsudøvelse. ?

FUNKTIONS NIVEAU FØR DIN KNÆSKADE: **(tæller ikke med score)**

Kan ikke klare hverdags aktiviteter	1	2	3	4	5	6	7	8	9	10	11	Ingen begrænsning i hverdags aktiviteter
	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	

AKTUELT FUNKTIONS NIVEAU :

Kan ikke klare hverdags aktiviteter	1	2	3	4	5	6	7	8	9	10	11	Ingen begrænsning i hverdags aktiviteter
	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	❑	

Scoring Instructions for the 2000 IKDC Subjective Knee Evaluation Form

Several methods of scoring the IKDC Subjective Knee Evaluation Form were investigated. The results indicated that summing the scores for each item performed as well as more sophisticated scoring methods.

The responses to each item are scored using an ordinal method such that a score of 1 is given to responses that represent the lowest level of function or highest level of symptoms. For example, item 1, which is related to the highest level of activity without significant pain is scored by assigning a score of 1 to the response "Unable to Perform Any of the Above Activities Due to Knee" and a score of 5 to the response "Very strenuous activities like jumping or pivoting as in basketball or soccer". For item 2, which is related to the frequency of pain over the past 4 weeks, the response "Constant" is assigned a score of 1 and "Never" is assigned a score of 11.

The IKDC Subjective Knee Evaluation Form is scored by summing the scores for the individual items and then transforming the score to a scale that ranges from 0 to 100. **Note**: The response to item 10 "Function Prior to Knee Injury" is not included in the overall score. The steps to score the IKDC Subjective Knee Evaluation Form are as follows:

1. Assign a score to the individual's response for each item, such that lowest score represents the lowest level of function or highest level of symptoms.
2. Calculate the raw score by summing the responses to all items with the exception of the response to item 10 "Function Prior to Your Knee Injury"
3. Transform the raw score to a 0 to 100 scale as follows:

$$\text{IKDC Score} = \left[\frac{\text{Raw Score - Lowest Possible Score}}{\text{Range of Scores}}\right] \text{x}100$$

Where the lowest possible score is 18 and the range of possible scores is 87. Thus, if the sum of scores for the 18 items is 60, the IKDC Score would be calculated as follows:

$$\text{IKDC Score} = \left[\frac{60 - 18}{87}\right] \text{x}100$$

$$\text{IKDC Score} = 48.3$$

The transformed score is interpreted as a measure of function such that higher scores represent higher levels of function and lower levels of symptoms. A score of 100 is interpreted to mean no limitation with activities of daily living or sports activities and the absence of symptoms.

The IKDC Subjective Knee Score can still be calculated if there are missing data, as long as there are responses to at least 90% of the items (i.e. responses have been provided for at least 16 items). To calculate the raw IKDC score when there are missing data, substitute the average score of the items that have been answered for the missing item score(s). Once the raw IKDC score has been calculated, it is transformed to the IKDC Subjective Knee Score as described above.

Lysholm Score, dansk version okt. 2006

Besvares sammen med fysioterapeut/fagperson

Udfra afprøvning på knæpatienter og fysioterapeuternes erfaringer er der tilføjet enkelte kommentarer, disse skrevet med kursiv.

1. **Halten**
   - Ingen halten — 5 p
   - Periodevis eller let halten — 3 p
   - Konstant eller udtalt halten — 0 p
2. **Albuestokke / behov for støtte**
   - Ingen / Intet — 5 p
   - Stok eller albuestokke — 2 p
   - Kan ikke støtte — 0 p
3. **Aflåsninger /låsningstendens**
   - Ingen — 15 p
   - Låsningstendens men ingen aflåsninger — 10 p
   - Enkelte egentlige aflåsninger — 6 p
   - Gentagne aflåsninger — 2 p
   - Låst knæ i dag — 0 p
4. **Instabilitet** (*knæsvigt / manglende styring af knæet / knæet giver efter*)
   - Ingen knæsvigt — 25 p
   - Knæsvigt forekommer sjældent under fysisk aktivitet (*idræt / fysisk krævende arbejde*) — 20 p
   - Knæsvigt forekommer ofte under fysisk aktivitet / umuligt at deltage i skadeudløsende aktivitet — 15 p
   - Knæsvigt forekommer nu og da under almindelige aktiviteter (*ADL*) — 10 p
   - Knæsvigt forekommer ofte under almindelige aktiviteter — 5 p
   - Knæsvigt forekommer ved hvert skridt — 0 p
5. **Smerte**
   - Ingen smerte — 25 p
   - Let smerte som forekommer sjældent under kraftig fysisk aktivitet — 20 p
   - Større smerte som forekommer under kraftig fysisk aktivitet — 15 p
   - Smerte under eller efter mere end 2 kilometers gang — 10 p
   - Smerte under eller efter mindre end 2 kilometers gang — 5 p
   - Konstant smerte — 0 p
6. **Hævelse** (*både ledhævelse og/eller kapselsvulst*)
   - Ingen — 10 p
   - Ved kraftig anstrengelse — 6 p
   - Ved almindelige aktiviteter — 2 p
   - Konstant — 0 p
7. **Trappegang**
   - Intet problem — 10 p
   - Lette problemer — 6 p
   - Et skridt ad gangen (enkelt benskift) — 2 p
   - Kan ikke — 0 p
8. **Hugsiddende** (*vægtbæring på begge ben, hælene må evt. gerne løftes*)
   - Intet problem — 5 p
   - Lette problemer — 4 p
   - Kommer ikke over 90 grader — 2 p
   - Kan ikke — 0 p

Er der tvivl om besvarelsen i punkt 1, 7 og 8 kan disse funktioner afprøves.

Dansk version af den svenske Lysholm Score fra 1985 er udarbejdet af fysioterapeuterne Bente Holm og Holger Hautopp udfra oversættelse foretaget af fysioterapeuterne Nils Erik Sjöberg, Josefin Agergaard og Rina Mærsk. Information kan fås via Fysioterapien, Hvidovre Hospital, tlf 36322232, Email: Bente.Holm@hvh.regionh.dk. Manualen kan downloades via http://fysio.dk/fafo/Maleredskaber/Maleredskaber-alfabetisk/Lysholm-Score/ 07.11.2009

Tegner score

Dato___________ **Label:**

Sæt **en** cirkel omkring det tal som bedst beskriver dit aktuelle maksimale funktionsniveau indenfor idræt eller arbejde

	Idræts aktivitet	Arbejde/hverdag
10	**Konkurrence sport** Fodbold internationalt niveau	
9	**Konkurrence sport** Fodbold, lavere divisioner Ishockey Brydning Gymnastik	
8	**Konkurrence sport** Badminton Squash Atletik Alpin ski	
7	**Konkurrence sport** Tennis Løb Moto-cross, speedway Håndbold **Motionsidræt** Fodbold, ishockey Squash, atletik Orienteringsløb	
6	**Motionsidræt** Badminton, Håndbold, basketball Alpin ski Jogging > 5 gange ugentlig	
5	**Konkurrence sport** Cykling, langrend ski **Motionsidræt** Jogging på ujævnt underlag	**Arbejde** Hårdt arbejde (ex. jord og beton)
4	**Motionsidræt** Cykling Langrend ski Jogging på jævnt underlag	**Arbejde** Moderat hårdt arbejde
3	**Konkurrence eller motionsidræt** Svømning Golf	**Arbejde** Let arbejde, ex. Pleje **Gang i natur**
2		**Arbejde** Let arbejde **Gang på ujævnt terræn**
1		**Arbejde** Stillesiddende arbejde **Gang på jævnt underlag**
0		**Sygemeldt eller pension**

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3. Calcium phosphate coatings for fixation of bone implants. Evaluated mechanically and histologically by stereological methods
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