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Limb Reconstruction

REVIEW

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Biomechanics of the anterior cruciate ligament and implications for surgical reconstruction

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Abstract Injury to the anterior cruciate ligament (ACL) is regarded as critical to the physiological kinematics of the femoral-tibial joint, its disruption eventually causing long-term functional impairment. Both the initial trauma and the pathologic motion pattern of the injured knee may result in primary degenerative lesions of the secondary stabilisers of the knee, each of which are associated with the early onset of osteoarthritis. Consequently, there is a wide consensus that young and active patients may profit from reconstructing the ACL. Several factors have been identified as significantly influencing the biomechanical characteristics and the functional outcome of an ACL reconstructed knee joint. These factors are: (1) individual choice of autologous graft material using either patellar tendon-bone grafts or quadrupled hamstring tendon grafts, (2) anatomical bone tunnel placement within the footprints of the native ACL, (3) adequate substitute tension after cyclic graft preconditioning, and (4) graft fixation close to the joint line using biodegradable graft fixation materials that provide an initial fixation strength exceeding those loads commonly expected during rehabilitation. Under observance of these factors, the literature encourages mid- to long-term clinical and functional outcomes after ACL reconstruction.

Key words Anterior cruciate ligament • ACL reconstruction • Biomechanics • Graft fixation • Graft tension

Introduction

The anterior cruciate ligament (ACL) is one of the most frequently injured structures of the knee joint [1]. Because of its key function as the primary restraint against anterior tibial translation, ACL disruption inevitably causes alterations in knee kinematics which are most likely to result in secondary degenerative changes and long-term functional impairment [2, 3]. As the ACL fails to heal in a manner that would restore normal knee kinematics, reconstructive techniques have been emphasised for patients who desire restoration of knee function and stability as well as return to high-level physical performance [4]. Although current concepts in knee ligament repair are reported to be clinically successful in most trials, ACL reconstruction has failed from a biomechanical point of view to both fully restore normal knee kinematics and to anatomically mimic the native ACL. Therefore, it may be postulated that surgi-

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cal ACL reconstruction would not adequately prevent secondary lesions or early degenerative changes of the injured knee joint. The purpose of this paper is to systematically review the basic research on ACL anatomy and biomechanics and to discuss its implications on current concepts in surgical ACL repair.

Kinematics of the knee

Description of motion about the knee implies sagittal plane motion and rotational components of the femorotibial joint. It is best described by the "instantaneous centre of motion" theory, which suggests that motion occurs about any instant point that acts as the centre of rotation and therefore does not move [5]. Mapping of successive instant centres throughout the range of motion, however, does not generate a single motion axis. It rather allows determination of an "instant centre pathway", which is shaped semicircularly and is located close to the joint surface in flexion and distant to the joint line in extension.

Relative surface motion between the tibia and the femur during flexion and extension occurring about the centre pathway appears to be that of gliding and rolling, the ratio of which is determined by the geometry of the articulating joint surfaces [6]. It changes from flexion to extension, with rolling of the femorotibial joint predominating near extension and gliding predominating as the knee is flexed [7]. The ratio of gliding and rolling, however, differs between the medial and the lateral condyle. While the medial femoral condyle sagittally is composed of two functional facets, the radius of which decreases from anterior to posterior, the lateral femoral condyle usually is composed of a single circular facet. This morphological principle implicates that rolling of the femur relative to the tibia would rather occur within the lateral compartment, whereas gliding would predominantly occur within the medial femorotibial compartment [8–10].

The combined movement of the medial and the lateral compartment finally equates to external rotation of the tibia with extension and internal tibial rotation with flexion of the knee joint. The axis of external and internal rotation, which is primarily determined by the geometry of the articulating surfaces, generally passes through the medial intercondylar tubercle of the tibial plateau [11]. Due to the incongruency in the radius of the medial and lateral condyle, the total range of tibial rotation depends upon the degree of knee flexion and sharply decreases near extension. Although surface geometry has been identified as a major guide of femorotibial joint motion, the complex kinematics of the knee under physiological loading *in vivo* have yet to be completely investigated. It appears to be obvious that the constraints provided by the femorotibial joint surface are insufficient for functional knee stability, though a combination of resisting structures such as the menisci, the joint capsule, the muscles, and the intra- and extraarticular ligaments account for both functional range of motion and joint stability under loading conditions.

Ligamentous stability of the knee

As the knee derives its stability from ligament structures rather than from its osteochondral surfaces, disruption of any of the supporting ligamentous structures will likely alter the overall motion characteristics of the knee. Moreover, damage to the central ligamentous column (i.e., the anterior and posterior cruciate ligaments), is found to physiologically stress other joint structures due to the pathologic shift in the sagittal and longitudinal axis of rotation [7]. The ligamentous structures of the knee provide stability as they compensate for tensile stresses acting in line with the axis of collagen fibres. As motion of the knee is mechanically complex, with several simultaneously changing axes of rotation, forces are not restricted by one ligament acting alone. Rather, they are restricted by a combination of ligament fibre bundles that are being recruited depending on both the flexion angle of the knee and the loading condition.

The amount that a specific structure contributes to the absorption of deforming forces has been described by the concept of primary and secondary stabilisers of the knee joint [12]. The cruciate ligaments, by guiding the motion of the femoral and tibial joint surfaces past one another, have been clearly identified as the primary stabilisers of anteroposterior translation when the knee is flexed. Studies have shown that the ACL provided more than 80% of anterior restraining force from 30° to 90° of knee flexion, while other ligamentous structures such as the medial joint capsule, the iliotibial tract, and the medial and lateral collateral ligaments provided no relevant secondary restraint to this motion [12]. Markolf et al. were able to demonstrate that anterior tibial translation was greatest between 20° and 45° of knee flexion, while beyond 90° of flexion both the superficial and the deep medial collateral ligaments secondarily contributed to anteroposterior stability [13].

Secondly, the ACL forces the tibia to internally rotate during anterior tibial translation, indicating that the ACL primarily restrains internal rotational moments during anteroposterior translation [14]. After sectioning the ACL, a significant increase of internal tibial rotation has been reported while subsequent sectioning of the collateral ligaments produced no progressive increase in internal tibial rotation near extension [15, 16]. With the knee flexed, anterolateral and posteromedial capsular structures are recruited during internal rotation as the ACL slackens and the posterior cruciate ligament tightens.

Functional anatomy of the ACL

The ACL is a ligamentous structure composed of dense connective tissue containing parallel rows of fibroblasts and type I collagen, which has been shown to be the predominant structural component [17]. It originates from the posterior medial aspect of the lateral femoral condyle and inserts to the anterior and lateral aspect of the medial tibial spine [18]. The area of origin and insertion of the ACL is reported to average 113 and 136 mm^2, respectively. The cross-sectional area at midsubstance varies between 36 and 44 mm^2, while the length of the anterior and posterior aspect of the ligament is reported to vary between 22 and 41 mm [19–22].

The ACL does not function as a simple tube of fibres with a constant tension, but rather consists of fibre groups that are subjected to episodes of lengthening and slackening throughout the range of motion (Fig. 1) [23]. This has advocated the functional subdivision of the ACL into an anteromedial and a posterolateral bundle, although histological investigations suggested a subdivision of fibre bundles to be rather arbitrary [24, 25]. According to functional observations, the fibres of the anteromedial bundle originate most anteriorly on the femoral side and insert anteriorly and medially at the tibial attachment. The fibres of the posterolateral bundle run from the posterior part of the femoral attachment to the posterior and lateral aspect of the tibial ACL footprint (Fig. 2). With the knee in full extension, the fibres of the smaller anteromedial and the larger posterolateral bundle run parallel, while during knee flexion the anteromedial fibres tighten and twist around the slackened posterolateral fibres, leaving the anteromedial fibres as the primary restraint against anterior tibial displacement at 90° of knee flexion. With both internal and external rotation, the ACL tightens so that it may operate as a major restraint against rotational moments acting about the knee joint [7].

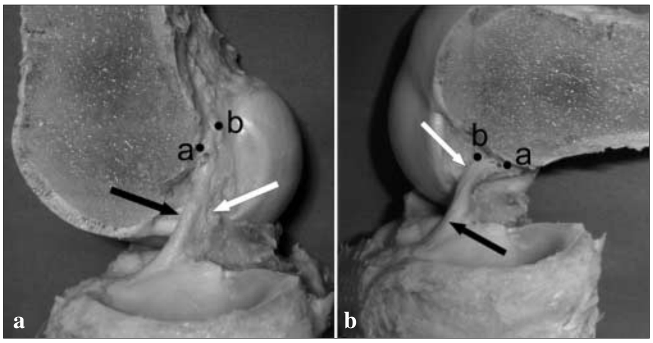

Fig. 1 Fibre arrangement of the anteromedial (*a, black arrow*) and the posterolateral (*b, white arrow*) during extension (**a**) and flexion (**b**) of the knee

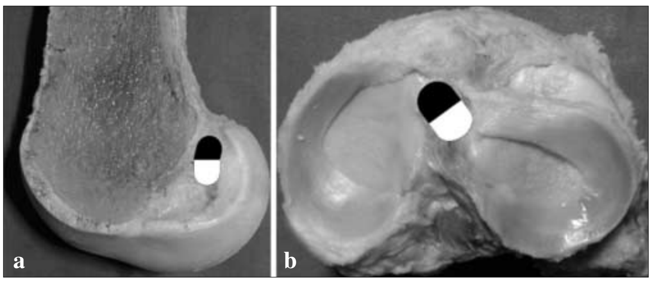

Fig. 2 Anatomical footprint of the ACL. **a** Femoral origin of the anteromedial (*black*) and posterolateral (*white*) bundle. **b** Tibial insertion of the anteromedial (*black*) and posterolateral (*white*) bundle

Biomechanics of the ACL

Like any other ligamentous structure, the biomechanical properties of the ACL are determined by the geometry of the ligament as well as the tensile characteristics of both ligament midsubstance and the ligament-to-bone insertion site. Basically, they can be characterised by the relationship between ligament length and ligament tension, which can be determined when simultaneously measuring load and the corresponding elongation during experimental uniaxial tensile testing.

The resultant load-elongation curve can be divided into four distinct regions according to the structural properties of the ACL. A first nonlinear region, the so-called 'toe region', is described as collagen fibres, which are arranged in varying degrees of crimp, easily extend under low axial forces [26–28]. The toe region is followed by a quasilinear region where collagen fibres reversibly deform. The slope of the linear region allows for reproducible determination of ligament stiffness (measured in Newtons per millimetre) and corresponds to the loads acting on the ACL during daily activities. In the intact knee, both the toe region and the linear region of the ACL loading curve will allow the tibia to translate anteriorly for 3–5 mm during knee motion as well as during an anterior drawer manoeuvre. With additional loading, the slope of the load-elongation curve decreases (yield-point) as plastic deformation of the collagen fibres occurs. Finally, the curve reaches the ultimate load, which is described as failure of the bone-ligament-bone complex. It may be derived from the load-elongation curve, that applying high loads to a ligament will increase the stiffness and may therefore sufficiently restrict excessive joint motion when high external loads are applied. Even more accurately, the biomechanical properties of a ligament are represented by the relation of stress and strain, where stress is defined as deformation per unit length (%) and where strain is defined as load per unit cross-sectional area (N/mm^2) [26].

When a constant load is applied to a ligament, the increase in ligament length is called 'creep', whereas the decrease in load with the ligament constantly elongated is called 'relaxation'. *In vivo*, cyclic loading of the ACL will cause gradual creep and relaxation, which results in increased knee laxity after physical activity. However, it will return to the original stiffness after a period of rest. The parameters derived during experimental ligament loading are considered to be essential to understanding ligament function and evaluating the appropriateness of different graft materials and fixation techniques commonly used in ACL reconstruction [26–28]. In addition, visual or apperative control of the mode of failure during tensile testing allows identification of either graft slippage or deterioration of graft material under mono- or polycylic loading conditions, thus indicating what amount of graft tension loss should be expected during the postoperative rehabilitation period.

Estimations of ACL forces during activities of daily living calculated by Morrison revealed that ACL loads of 169 N may be expected during normal level walking, while descending stairs generated 445 N of *in situ* force due to the activation of the knee extensor apparatus [29–31]. In contrast, ascending stairs as well as ascending or descending a ramp generated *in situ* forces below 100 N.

While estimation of *in vivo* ACL forces during normal activity have been subject to computational analyses, the biomechanical properties of the native ACL have extensively been analysed in *ex vivo* studies. Measurements using a universal force-moment sensor revealed that the *in situ* force of the intact ACL was largest at 15° of knee flexion and continuously decreased until 90° of knee flexion [23]. Focusing on the ACL force and ligament deformation during anterior tibial translation, Sakane et al. demonstrated that near full extension, the *in situ* force of the anteromedial bundle significantly differed from that of the posterolateral bundle with 110 N of anterior tibial load applied [23]. While there was no significant difference in the *in situ* force of the anteromedial bundle throughout the range of knee motion, the *in situ* force of the posterolateral bundle was significantly lower at 90° of knee flexion when compared to full extension, similar to what was measured for the entire ACL. This suggests that the role of the posterolateral bundle in response to anterior tibial loads near extension may be of more importance than was previously thought (Fig. 3).

Performing tensile testing of the bone-ligament-bone complex, Woo et al. reported the ultimate failure load of the native femur-ACL-tibia complex in younger cadaveric specimens to average 2160±157 N, while mean ACL stiffness was 242±28 N/mm [32, 33]. They were also able to demonstrate that both the ultimate failure load and the linear stiffness significantly decreased with age and with the axis of loading. Ultimate failure loads in older specimens being loaded in an anterior drawer mechanism averaged 496±85 N with a mean stiffness of 124±16 N/mm. Given this data, Noyes et al. concluded that the initial fixation

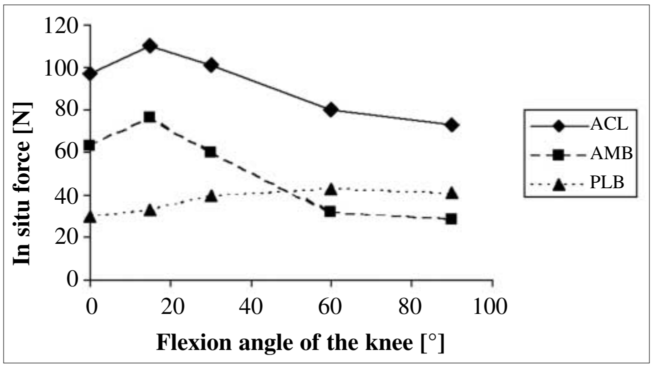

Fig. 3 *In situ* force of the intact ACL, the anteromedial bundle (*AMB*) and the posterolateral bundle (*PLB*) under 110 N of anterior tibial load. Adapted from [23]

strength of an ACL graft required for sufficient knee stability during daily activities should exceed 450 N, although earlier studies performed by Shelbourne and Gray reported excellent clinical results using graft fixation techniques with a significantly lower initial failure strength of only 248 N [34–36].

The ACL deficient knee

ACL injuries commonly occur during sport activities with sudden stresses applied to the knee joint while the tibia is in contact with the ground. Typically, the ACL is torn in a non-contact deceleration situation that produces valgus and internal rotational moments on the knee joint that begins to flex from near extension. This usually occurs in high-risk pivoting sports when the athlete lands on the leg and starts rotating to the opposite direction [37].

Active quadriceps pull is considered to play an important role in the pathomechanism of ACL injury. Reflective eccentric quadriceps contraction accompanied by apparent weakness of the hamstring muscles allows the extensor mechanism to strain the ACL during anterior tibial translation. This mechanism can be observed during jump stop landings. Less frequently, direct contact injuries occur as a result of extensive valgus stress or hyperextension of the knee joint.

Isolated disruption of the ACL is a rather rare condition, while complete or incomplete ruptures accompanied by traumatic lesions to the medial or lateral meniscus as well as to the medial collateral ligament are reported in 80% of all cases [3, 37–39]. Those concomitant lesions are reported to significantly influence the long-term functional outcome. Additionally, in both isolated and combined ACL injuries, minor or major bruising of chondral and subchondral structures may be present. Histologic analyses of bruised bone performed by Johnson et al. and others demonstrated areas of necrotic osteocytes and chondrocytes, indicating severe damage to the local osteochondral tissue homeostasis after ligament injuries of the knee [38–43].

Disruption of the ACL inevitably results in alterations in knee kinematics as a transfer of loads can be effective only if the joint is mechanically stable. ACL insufficiency causes deterioration of the physiologic roll-glide mechanism of the femorotibial joint and results in an increased anterior tibial translation as well as an increased internal tibial rotation. In the advent of muscular fatigue or deficient neuromuscular control, the patient will experience this combined anterior and rotatory instability as a subluxation of the femorotibial joint. According to the concept of primary and secondary restraints, failure of a primary restraint will cause recruitment of secondary structures in order to resist external forces and to stabilise joint motion. The increase in loads applied to secondary structures may render them more susceptible to degeneration or secondary failure [44–46]. Studies investigating the long-term functional outcome after conservative treatment of ACL ruptures, though not characterised by well designed prospective cohort studies, reported on the prevalence of radiographic osteoarthritis in 60%–90% of subjects 10–15 years after injury [2, 3, 44, 47].

Principles of ACL reconstruction

Current research supports the concept that under observance of several key factors, arthroscopically assisted ACL reconstruction done with a biologic autograft significantly improves the stability and function of the knee in most patients. The key factors significantly influencing the functional outcome are:
- individual choice of graft material;
- anatomic bone tunnel placement;
- adequate graft (pre-)tension;
- anatomical graft fixation;
- sufficient initial graft fixation strength.

Individual choice of graft material

Currently recommended graft materials for ACL reconstruction are biologic autografts and, although rarely available in Western Europe, allografts [48]. Graft choices basically include the patellar tendon-bone graft, semitendinosus/gracilis tendon or the quadriceps tendon graft [49–56]. Although a surgeon may prefer one specific substitute for reconstruction, modern knee surgery requires individual concepts and a variability of treatment options. In their metaanalysis on the functional outcome of patellar tendon and hamstring tendon ACL reconstructions, Freedman et al. reported that patellar tendon grafts displayed significantly less failure and better knee stability but resulted in an increased rate of donor side morbidity [56]. Several other studies confirmed no significant difference in functional parameters and subjective results obtained at follow-up when comparing patellar tendon-bone and hamstring tendon grafts [57].

From a biomechanical point of view, no graft material commonly used has ultimate failure strength or stiffness comparable to the native ACL. Although ultimate failure loads of the native bone-patellar tendon-bone complex, a quadrupled hamstring tendon complex, or the quadriceps tendon-bone complex average 2977, 4140 and 2353 N, respectively, and consequently exceed the values reported for the native ACL, graft harvest and artificial graft fixation into bone significantly decreases both the ultimate strength and the linear stiffness [58–60]. Patellar tendon-bone grafts should be used for young patients and high-demand athletes who prefer early return to high-level activities, while hamstring tendons are advantageous when a large skin incision or anterior knee pain should be avoided. Quadriceps tendon grafts should primarily be used for revision surgery as they are difficult to harvest and size and location of donor-site scar are disadvantageous [61].

Anatomic bone tunnel placement

The key to proper postoperative knee function is to restore the physiologic roll-glide mechanism of the femorotibial joint, and thus avoid both increased anterior displacement and pathologic patterns of knee rotation. In order to achieve these functional demands, several studies have shown that graft positioning is one of the most important factors in ACL reconstruction [62–65] (Table 1). Investigating the anterior–posterior stability of the knee after ACL reconstruction, Rupp et al. reported that an increase in postoperative knee laxity was most likely to be due to malposition

Table 1 Malposition of the femoral tunnel and resulting functional consequences [65]

Position	Kinematic consequences
Femoral tunnel	
Anterior	Tightens in flexion/slackens in extension
Posterior	Slackens in flexion/tightens in extension
Tibial tunnel	
Anterior	Tightens in flexion/notch-impingement with extension
Posterior	Tightens in extension/impingement with posterior cruciate ligament
Medial/lateral	Impingement at ipsilateral femoral condyle

of the bone tunnels [66]. Fu et al. further reported that anterior positioning of both the femoral and the tibial tunnel was the most common technical mistake during arthroscopic surgery [61]. Consequently, avoiding nonphysiological strain patterns of a ligament graft throughout the functional range, which also avoids graft failure and any limitation in knee motion, has been emphasised for successful reconstruction [7]. A proposed isometric bone tunnel placement (i.e., the distance between the proximal and the distal insertion site remained constant throughout the entire range of motion), however, was not possible to obtain in either *in vivo* or *in vitro* observations [62, 65].

Hefzy et al. investigated the resulting changes in distance between a given point for the femoral and the tibial attachment of an ACL graft [62]. They reported that altering the location of the femoral bone tunnel had a much greater effect on the length of the substitute than did altering the tibial attachment site did. They also pointed out that the area in which tunnel placement resulted in length changes of the graft of less than 2 mm throughout the range of motion was smaller than the cross-sectional area of grafts commonly used for reconstruction of the ACL. The resulting inhomogenous tension patterns within a graft would therefore not support the concept of isometric graft placement.

Csizy and Friederich noted that with an arthroscopic view of the knee joint, the femoral tunnel is most susceptible to being placed anterior to the anatomic ACL footprint, thus resulting in excessive graft tension during knee flexion and correlating well with poor functional outcomes [65, 67]. Varying the femoral tunnel position between the anatomic ACL footprint and the most isometric position, Musahl et al. demonstrated that neither tunnel position fully restored the physiologic kinematics of the femorotibial joint [63]. However, they concluded that anatomic graft placement resulted in kinematics closer to the normal knee joint than did a tunnel placement for best isometry.

Several methods for measuring femoral graft position in postoperative radiographs have been introduced, hence enabling visual control of bone tunnel placement when using intraoperative flouroscopic imaging [68–72] (Fig. 4). Investigations performed by Klos et al. revealed that the measurement technique described by Amis et al. produced reliable data in both rotated and non-rotated (i.e., full overlap of the femoral condyles) lateral radiographs of the knee [64, 69]. Probably the most popular technique for radiographic bone tunnel measurement is the quadrant method described by Bernard et al. [72]. Although commonly used to identify the anatomic ACL origin, it is reliable only when the projection of the femoral condyles perfectly overlap in lateral radiographs of the knee.

Tibial bone tunnel placement has been reported to be less sensitive with respect to postoperative knee kinematics, however, may cause graft impingement or unphysiologic loading patterns when malplaced [73–75]. The tibial tunnel should be placed within the posterior half of the native ACL footprint, which will allow the anterior fibres

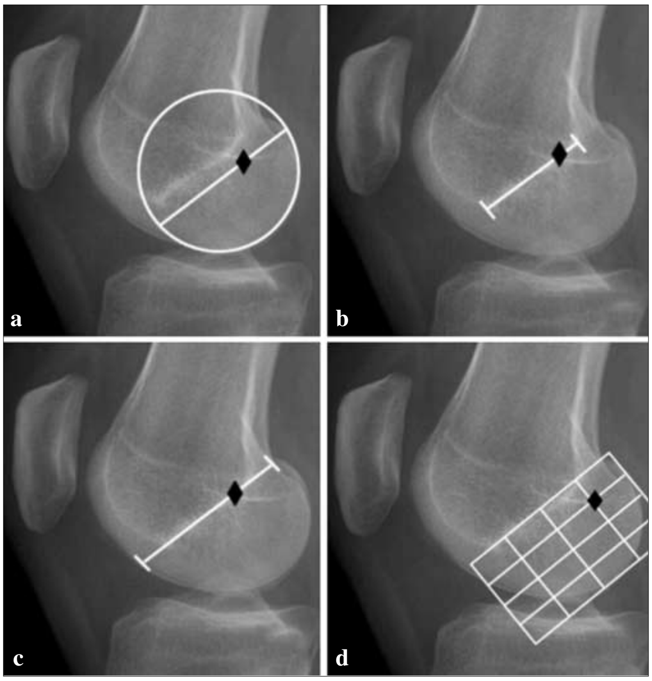

Fig. 4 Radiographic analysis of femoral bone tunnel placement. **a** 60% of the anterioposterior diameter of the lateral femoral condyle ([69]); **b** 80% of the anteroposterior length of Blumensaat's line ([70]); **c** 65% of the anteroposterior cortical depth of the distal femur in line with Blumensaat's line ([71]); **d** anteroinferior corner of the most proximal and posterior quadrant adapted to the height and length of the lateral femoral condyle ([72])

of the graft to run parallel with the intercondylar roof in full extension (Fig. 1) [73]. Anterior placement of the tibial tunnel as well as a vertically oriented intercondylar roof in the sagittal plane will either initially subject the graft to roof impingement or secondarily will cause roof impingement with contraction of the quadriceps muscle. It has been recommended in the literature to leave 6 mm of clearance between the anterior aspect of the graft and the intercondylar roof during extension of the knee. Arthroscopically, the bone tunnel should be drilled at an angle of 40°–50° to the long axis of the tibia and should be placed anteromedially or posterior to the anterior horn of the medial meniscus and slightly anterior to the posterior cruciate ligament insertion [76–79].

In the frontal plane, the position of the femoral tunnel along the intercondylar notch can be described by the angular position of numerals on the face of a clock. In general, a tunnel positioned at 11 o'clock for a right knee (or 1 o'clock for a left knee) has been considered the correct tunnel position [80–82]. However, Markolf et al. emphasised the intercondylar notch to have a three-dimensional configuration, such that variations in graft placement in the frontal plane inevitably resulted in variations in tunnel placement in the sagittal plane [83]. Hence, they demonstrated that the biomechanical consequences of varying femoral tunnel placement in the frontal plane were less critical than varying the anteroposterior position.

In order to obey the complex structure of the intact ACL, several authors have claimed ACL reconstruction to be more anatomical when both the anteromedial and the posterolateral bundle were replaced independently [84–87]. Yagi et al. measured the *in situ* ACL force as well as knee kinematics with the ACL intact, sectioned, reconstructed with one bundle and reconstructed with two bundles using a robotic universal force-moment sensor testing system [84]. They demonstrated that under combined anterior, internal tibial and valgus torque, knee stability using a double-bundle technique was superior to a single-bundle reconstruction technique. There is agreement among ACL surgeons that double-bundle ACL reconstruction is a demanding procedure that currently may only be performed in experienced trauma centres.

Adequate graft (pre-)tension

The tension applied to the graft before final graft fixation significantly influences the kinematics of the knee joint and the long-term ability of a graft to stabilise the knee joint. A low initial graft tension will not provide adequate joint stability, while excessive initial graft tension will restrain range of motion and is susceptible of early graft failure. Yoshia et al. studied the effect of initial graft tension on knee stability using an *in vivo* animal model [88]. They demonstrated that anteroposterior knee stability did not significantly differ between grafts fixed at 1 N and 39 N three months after surgery, however, knee joints with higher initial graft tension showed evidence of degenerative cartilage lesions. In a goat model, Abramowitch et al. reported similar results with the biomechanical characteristics of an ACL substitute not differing significantly when comparing grafts fixed with high (35 N) and low (5 N) initial tension six weeks after surgery, but differing significantly when compared to an uninjured control group [89]. In a prospective randomised trial, Kim et al. failed to prove that forces of either 8, 12 or 15 kg applied to the graft during fixation resulted in significant differences in anterior knee laxity one year after surgery [90].

In accordance, *in vitro* studies on the course of graft tension have shown that both the patellar tendon-bone graft and the hamstring tendon graft lose their initial tension when being cyclically loaded [91, 92]. There is a lack of scientifically based data on the clinical impact of initial graft tension as follow-up studies so far have failed to prove that variations of graft tension resulted in clinical symptoms after ACL reconstruction. At present, the amount of tension that should be applied to a graft prior to fixation has yet to be precisely defined. Excessive tension as well as loose fixation should therefore be controlled intraoperatively by testing range of motion and anterior knee stability under arthroscopic visualisation.

Although the influence of the viscoelastic behaviour of ACL replacements so far has not been entirely characterised, the viscoelastic creep or relaxation may contribute to a decrease in graft tension over time. Consequently, several authors have emphasised that grafts should be cyclically preconditioned prior to implantation in order to decrease the viscoelastic elongation behaviour during rehabilitation [93, 94]. The subject of graft preconditioning remains controversial as the preconditioning protocols described in the literature are not commonly applicable to the surgical procedure and may not be as beneficial as suggested by *ex vivo* biomechanical studies [93–95].

Anatomical graft fixation

Most fixation devices currently used rely on linkage material between the graft substance and the bone, thus influencing graft healing and altering the initial biomechanical properties of a graft material. Generally, fixation devices distant to the joint line (i.e., buttons, staples, washers or post-screws) fail to reconstruct the complex nature of the native tibial or femoral ACL insertion close to the joint surface. As a consequence, the strain that is induced in a substitute during cyclic loading is significantly larger when compared to the intact ACL [48]. This allows for longitudinal ('bungee-effect') and transverse ('windshield-wiper-effect') graft motion within the bone tunnel, which in turn may lead to bone tunnel dilation, may impair healing of the graft to the bone tunnel and may complicate revision surgery due to loss of bone material.

Furthermore, it should be considered that the linear stiffness of grafts fixated distant to the joint line is less than placing the graft close to the entrance of the bone tunnel. Although Magen et al. demonstrated that the stiffness of a graft complex was influenced more by the method of fixation than the length of the graft, it has been previously reported that keeping the length of a substitute as short as possible may increase graft stiffness and knee stability throughout the range of motion [96–99]. In accordance, studies have shown that linear graft stiffness is higher when fixation systems that are placed close to the tunnel entrance are used [48]. Investigations on the overall stability of porcine knees after ACL reconstruction revealed that fixating the graft proximally on the tibial side resulted in significantly more anterior knee stability than central or distal tibial graft fixation [97].

Sufficient initial graft fixation strength

The importance of secure graft fixation has dramatically increased as current rehabilitation protocols emphasise early weight bearing after ACL reconstruction and as the fixation site is known to be the weakest link during the early postoperative period [26]. Graft fixation to bone

Table 2 Biomechanical data on graft material and fixation devices currently used in ACL reconstruction

	Fixation technique	Ultimate failure load [N]	Stiffness [N/mm]	Reference
Intact ACL		2160±157	242±28	[33]
Quadrupled hamstring tendon graft		4140±n.n.	807±n.n.	[26]
Tibial	Interference screw	776±155	226±56	[96]
	Suture/post	830±187	60±14	[96]
	Washer (20 mm)	930±323	126±28	[96]
Femoral	Interference screw (b)	507±93	58±14	[91]
	Interference screw (b)	621±139	76±20	[104]
	Interference screw (t)	419±77	40±11	[99]
	Interference screw (t)	774±154	80±15	[104]
	Cross-pin	737±140		[108]
	Endobutton	864±164		[108]
	Transfix	746±119		[108]
Patellar tendon-bone graft		2376±151		[94]
Tibial	Interference screw (b)	718±219	46±5	[104]
Femoral	Interference screw (b)	707±169	115±26	[106]
	Interference screw (b)	702±168	190±78	[107]
	Interference screw (t)	681±146	107±25	[106]
	Press-fit	571±109	125±29	[106]
	Cross-pin	639±156	226±63	[107]

should furthermore consider that the bone mineral density and the angle of force application significantly differ between the femoral and the tibial bone. In accordance with the surgical procedure of drilling the femoral tunnel with the knee flexed between 90° and 120°, studies on the line of force transmission have shown that the femoral graft fixation strength increases as the angle between the axis of the bone tunnel and the axis of the ligament increases during extension of the knee [100–103].

On the tibial side, the line of force on the graft is directly in line with the tibial bone tunnel. Investigations on the biomechanical properties of different tibial graft fixation techniques for patellar tendon-bone grafts showed that interference screw fixation provided superior ultimate strengths of 293–758 N when compared to techniques using sutures and staples [99]. In contrast, tibial fixation of quadrupled hamstring tendon grafts using washerplates or sutures is reported to provide superior fixation strength when compared to metal or biodegradable interference screws (Table 2). Studies on the femoral fixation of patellar tendon-bone grafts demonstrated that interference screws, extracortical buttons and transverse fixation systems provided similar fixation strengths ranging from 418 to 640 N [104–109]. Options for femoral soft tissue graft fixation have likely been shown not to significantly differ with respect to initial fixation strength and stiffness and to generally resist those forces arising during rehabilitation. Fixation not only needs to withstand continuous loading cycles but also should facilitate biologic graft healing and should allow return to a histologic transition zone from ligament to bone. Therefore, efforts have focused on the reduction of ligament fibre movement within the bone tunnel and the elimination of linkage materials that may impair healing of the graft–tunnel interface.

Conclusions

Reviewing the literature on biomechanical aspects of ACL reconstruction, it may be concluded that all autologous graft materials as well as fixation techniques and fixation materials currently promoted and used in ACL surgery provide sufficient initial fixation strength during the early postoperative period. Although being the weakest link until graft healing to bone is completed, studies on the elongation behaviour of an ACL substitute suggest that rather a loss of viscoelastic properties, resulting from either excessive graft pretension or malplacement of the bone tunnels, may account for postoperative knee laxity and limited clinical success in some cases. Anatomically, as well as functionally, any graft material generally will fail to mimic the complex nature of the native ACL and therefore will inevitably alter the kinematics of the knee joint. Femorotibial joint motion, which is characterised by a well balanced ratio of rolling and gliding, especially reacts to changes in range of motion as well as changes in motion patterns of the joint surfaces after ACL injury, most likely resulting in degeneration of secondary joint stabilisers. When reconstruction of the ACL is performed, no functional restitio ad integrum may be expected, as an anatomically placed single bundle ACL substitute will reconstruct only part of the fibre mass of the intact ACL.

Whether or not double-bundle ACL reconstructions functionally imitated the native ACL more closely and consequently restored the kinematics of the knee joint more accurately, currently remains subject to debate. So far, clinical follow-up studies have shown that the physiologic anterior knee stability cannot be completely restored after ACL reconstruction. However, under observance of the biomechanical factors that significantly influence the kinematics of the knee joint, encouraging mid- to long-term clinical and functional outcomes after ACL reconstruction have been reported.

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