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Gait knee kinematics of patients with ACL rupture: a 3D assessment before and after the reconstruction

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Abstract

Because of the role of the ACL in knee joint's biomechanics, it is essential to quantify the kinematics of ACL deficient and ACL reconstructed knee. In-vitro models bare the handicap of limited muscle simulation, while static, one-dimensional testing cannot predict the behavior of these groups of patients under realistic loading conditions. Currently, the most widely accepted method for assessing joint movement patterns is gait analysis. Respectively, 3D motion analysis is necessary to provide high reliability movement analysis.

The purpose of the study was in-vivo evaluation of the behavior of the anterior cruciate ligament deficient (ACLD) and anterior cruciate ligament reconstructed (ACLR) knees during walking, using 3D, real-time assessment tool.

The biomechanical data were collected prospectively with KneeKG™ system. In the first study, 30 patients with ACL rupture were compared with 15 healthy subjects as a control group. In the second study, 3D kinematic data were obtained in 15 patients pre- and post-ACL reconstruction, 15 contralateral knees and 15 healthy control subjects. Kinematic data were recorded during treadmill walking at self-selected speed. Flexion/extension, external/internal tibial rotation, adduction/abduction and anterior/posterior tibial translation were compared between groups.

The ACLD patients showed a significant lower extension of the knee joint during stance phase. A significant difference in tibial rotation angle was found in ACLD knees compared to control knees. The patients with ACLD knee rotated the tibia more internally during the mid-stance phase, than control group. There was no significant difference in anterior-posterior translation and adduction–abduction angles.

The ACLR knees showed a significant higher extension of the knee joint during entire stance phase compared to ACLD knees. But the ACLR knees showed still deficit of extension compared to healthy control knees from 46 to 74 % of the gait cycle. In axial plane, there was no significant difference in pre- and post-operative kinematic data. The significant difference was achieved between ACLR knees and healthy control knees, specifically between 28 to 34 % and 44 to 54 % of the gait cycle. There was no significant difference in anterior-posterior translation or in coronal plane between the groups that were compared.

Significant alterations of joint kinematics in the ACLD knee were revealed in this study by manifesting a higher flexion gait strategy and excessive internal tibial rotation during walking that could result in a more rapid cartilage thinning throughout the knee. In the other hand, even though ACLR knees showed some improvements in sagittal plane compared to ACLD knees, in axial plane there still exists difference compared to healthy control knees. These kinematic changes could lead to abnormal loading in knee joint and initiate the process for future chondral degeneration. However, the post-operative kinematic data were collected 10 months after surgery, so a longer follow-up is needed to evaluate if these kinematic changes persist in time, and their effects in joint degeneration.

Keywords: Knee, Anterior cruciate ligament deficient, Reconstruction, Kinematic, 3D assessment, Gait

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ABBREVIATIONS

ACL	Anterior Cruciate Ligament
ACLD	Anterior Cruciate Ligament Deficient
ACLR	Anterior Cruciate Ligament Reconstruction
AM	Anteromedial
BTB	Bone-Tendon-Bone
CG	Control Group
CK	Contralateral Knee
DB	Double Bundle
MRI	Magnetic Resonance Imaging
Nm	Newton meter
NWI	Notch Width Index
OP	Operation
PCL	Posterior Cruciate Ligament
PL	Posterolateral
PT	Patellar Tendon
RSA	RadioStereometric Analysis
SB	Single Bundle
3D	Three dimensional

I. INTRODUCTION

Introduction

Anterior cruciate ligament injury leads to instability of the knee and to biomechanical knee changes. ACL rupture is associated with the development of chondral injuries, meniscal tears, degeneration of articular cartilage, and eventually posttraumatic arthritis.

Currently, patients with ACL injury (especially the young and those with high level activity) generally undergo with ACL reconstruction. This is because surgical techniques of ACL reconstruction have been improved, but there is still room to ameliorate, especially the restoration of normal knee kinematics.

Quantitative kinematic analysis is important for gaining a thorough understanding of normal and pathological knee joint function during human locomotion [1–3]. While significant information can be obtained through the manual clinical examination more precise and objective tools are quite useful [4], particularly in regard to assessment of rotational stability [5].

Although in vitro studies have provided important information about ACL deficient knee, their inability to simulate knee kinematics during daily activities has led to development of new in vivo-3D kinematic tools.

This work is focused on in-vivo evaluation of the kinematics of the knee in patients with ACL rupture, before and after ACL reconstruction during all phases of the gait, using a new 3D, quasi-rigid real time assessment tool (KneeKGTM). The KneeKGTM system was developed with the aim of making kinematical assessment of the behavior of the knee joint during gait a part of clinical care.

This thesis consists of three main parts:

- in the first one, the review of literature of the kinematics of the native knee, ACL deficient and ACL reconstructed knees.

- in the second one, comparison of the ACL deficient knee kinematics with control group during the gait, and

- in the third one, comparison of ACL deficient knee, pre- and post-OP, and the ACLR knee with the contralateral healthy knee and control group.

I.1. Biomechanics in Orthopaedics

The field of biomechanics is a multidisciplinary area of science that involves among other disciplines physicians, surgeons, physical therapists and bioengineers.

Biomechanics is the application of engineering principles to the study of forces and motions of biologic systems. As they relate to sports medicine, biomechanical studies are designed to determine the magnitude and direction of forces and moments of various tissues in and around a diarthrodial joint, as well as to measure the corresponding joint kinematics. This information can then be used by clinicians for the functional assessment of a normal or an injured joint and for planning the appropriate course of treatment [6].

Since most musculoskeletal injuries are caused by imbalance of internal muscle and external environmental forces, resulting in damage to the anatomical biological tissues and structures, biomechanical analysis helps studying these forces and their effects, and establishes the injury mechanism [7]. By understanding the pathomechanics of sport injury, biomechanical studies enhance the development of injury prevention in sports medicine [8].

The new paradigm “Orthopedics Sport Medicine” explains very well the synergy between the orthopedics and biomechanics in the management of sports injuries (Fig 1).

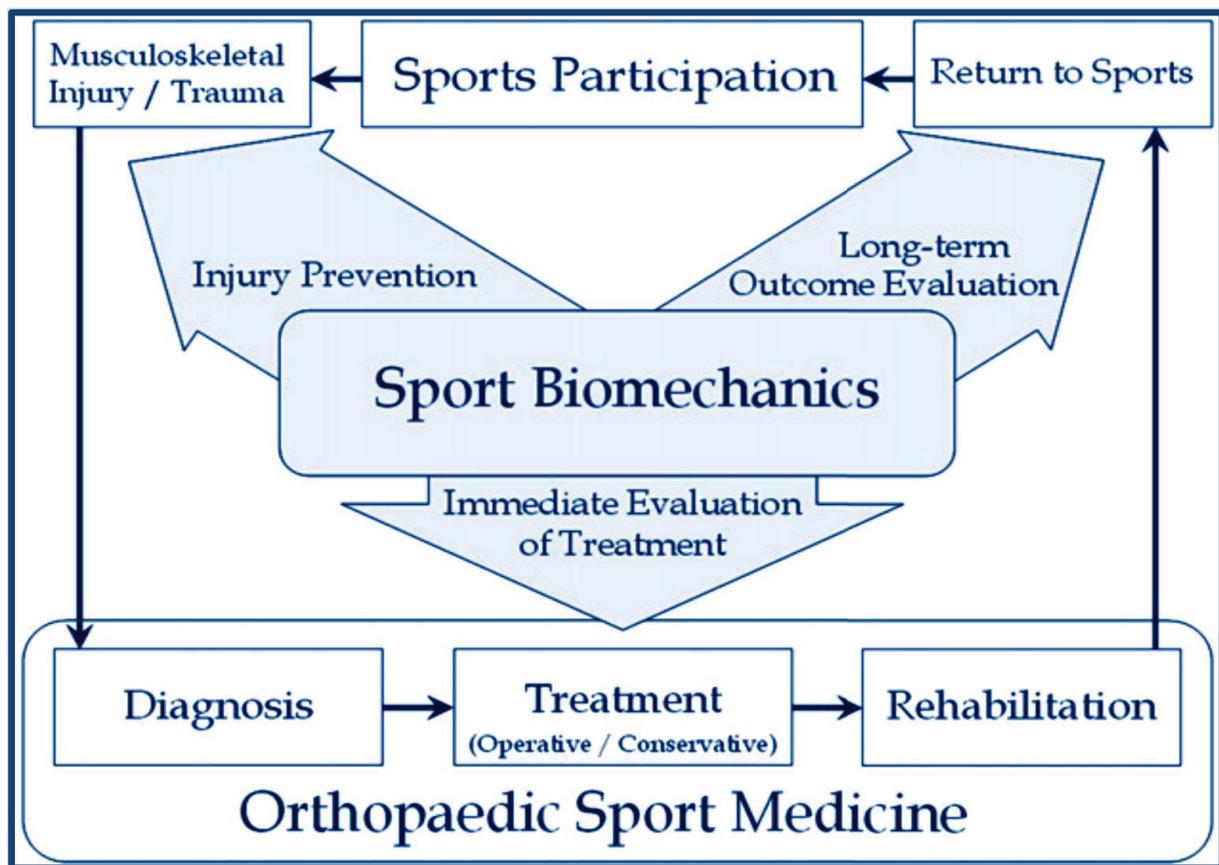


Fig. 1 New paradigm of “Orthopedics sport biomechanics” in preventing and managing sports injury[9]

As it is showed in this figure, biomechanics has three main roles: (1) it helps in the prevention of the musculoskeletal sports-related injuries and trauma, (2) it provides quantitative objective assessment to evaluate the immediate outcome of treatment either operative or conservative, and (3) it acts as an objective tool to monitor the long-term rehabilitation progress, and to indicate if an athlete is adequately recovered to a satisfactory level for returning to sports [9].

By understanding the pathomechanics of sport injury, biomechanical studies enhance the development of injury prevention in sports medicine [8], which is a rapid growing research field in order to promote safety in sports [10].

However, there still can be improvements in the treatment of knee injuries. This improvement can be achieved through a profound understanding of the function of the injured knee during daily living activities. This understanding can be gained through gait analysis.

Gait analysis allows the quantification of gait parameters and provides the objective measures necessary to evaluate dynamic functional levels of patients performing everyday activities. Gait analysis offers a means of studying subclinical differences in a patient's movement and the mechanical demands placed on the knee.

I.1.1. Gait cycle

Gait cycle begins with heel contact of either foot and ends with heel contact of the same foot. Gait is a cyclic phenomenon that can be divided into segments, or phases. Initially, it is divided into two phases: stance phase, where the limb is in contact with the floor, and swing phase, where the limb advanced forward without being in contact with floor. The stance phase begins when the heel of the forward limb makes contact with the ground and ends when the toe of the same limb leaves the floor. While, the swing phase begins with the toe off of this foot and ends with the heel contact of the same foot. The stance phase corresponds approximately to 60 % of the gait cycle and the swing phase to about 40% of the gait cycle [11].

The **stance phase (60% of the gait cycle)** is subdivided as following:

Initial contact: 1-2 %

The moment when the foot touches the floor and the point at which the body's centre of gravity is at its lowest position. The role of the knee is mainly to provide adequate stability in the leg.

Loading phase: 1-10 %

It begins with initial floor contact and continues until the other foot is lifted for swing. The knee is flexed for shock absorption.

Mid-stance: 10%-30%

The foot is flat on the floor and the weight of the body is directly over the supporting limb. This phase is the first half of the single-limb support interval. The main role of the knee is to provide stability of the supporting lower limb.

Terminal stance phase: 30%-50%

This phase completes the single limb support. It begins with heel rise and continues until the other foot strikes the floor. The full extension of the knee during this phase allows the extension of the step.

Pre-swing: 50%-60%

It begins with the initial contact of the opposite limb and ends with the ipsilateral toe-off. A major objective of this phase is to position the limb for swing phase.



Fig. 2 Stance phase of gait cycle

The **swing phase (40% of the gait cycle)** is subdivided as following:

Initial swing phase: 60%-73%

This phase begins with a lift of the foot from the floor and ends when the swinging foot is opposite to the stance foot. In this phase, the limb is advanced by hip flexion and increased knee flexion.

Mid-swing phase: 73%-87%

Begins when the swinging foot is opposite to the stance foot and ends when the swinging limb is forward and tibia vertical. The knee is allowed to extend in response to gravity.

Terminal swing: 87%-100%

The final phase of swing begins with a tibia vertical and ends when the foot strikes the floor. The knee maximally extends.



Fig. 3 Swing phase of gait cycle

I.2. The native knee and ACL

I.2.1. Anatomy of the knee

The knee joint is generally considered to be the largest and the most complex diarthrosis joint of the human body. It enables hinge and rotating movements as the connections between the upper and lower leg. In addition, due to its mobility, it contributes in advancing the human body during movement.

The knee consists of three basic types of structures: 1.Ligaments are passive elastic structures and can be loaded in tension only; 2.Musculotendinous units are active elastic structures and can act only under tension; 3.Bone is essentially non-elastic and serves to take the compressive loads in the joint [12].

The femur, tibia and patella constitute the knee joint and result in two separate articulations: the tibiofemoral and patellofemoral joints.

The bones are connected together by the following structures: the articular capsule, the ligamentum patellae, the oblique popliteal, the tibial collateral, the fibular collateral, the medial and lateral menisci, the transverse ligament, the coronary ligament, the posterior cruciate ligament and anterior cruciate ligament.

The knee joint is stabilized laterally by the tendon of biceps and gastrocnemius muscles, the iliotibial tract, and the fibular collateral ligament (extracapsular ligament). Medially, it is stabilized by the sartorius, gracilis, gastrocnemius (medial head), semitendinosus, semimembranosus and the tibial collateral ligament (extracapsular ligament) (Fig 4).

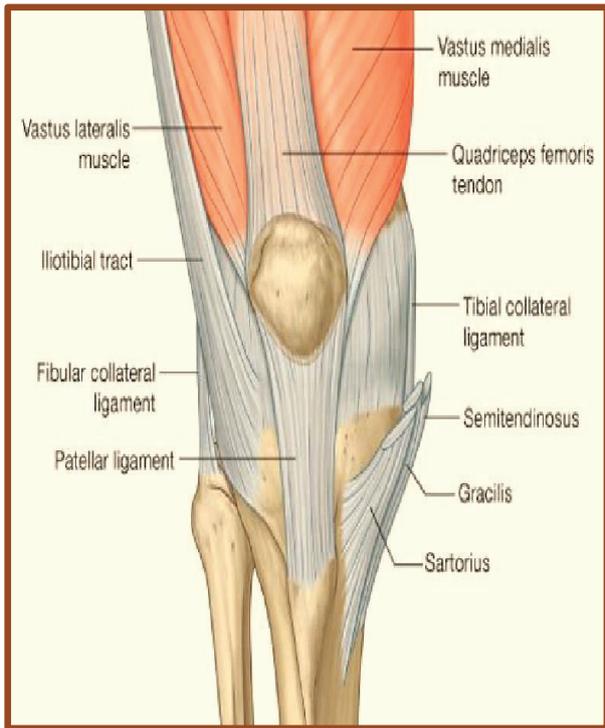


Fig. 4 Musculotendinous units of the knee[13]

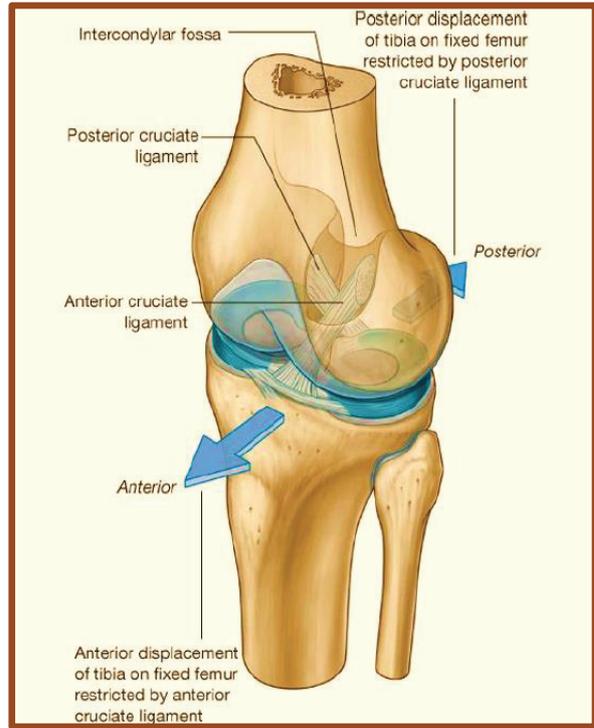


Fig 5 Functional anatomy of cruciate ligaments [13]

I.2.2. Ligamentous stability of the knee

As the knee derives its stability from ligament structures rather than from its osteochondral surface, 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 [14]. The amount that a specific structure contributes to the absorption of deforming forces has been described by the concept of primary and secondary stabilizers of the knee joint [15].

I.2.3. Functional anatomy of the ACL

The ACL is an intraarticular but extrasynovial band of dense connective tissue. It 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 [16]. 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 [17]. The area of origin and insertion of the ACL is reported to average 113 and 136 mm², respectively. The cross-sectional area at mid-substance varies between 36 and 44 mm², while the length of the anterior and posterior aspect of the ligament is reported to vary between 22 and 41 mm [3, 18, 19].

The ACL consists of two bundles, an anteromedial (AM) and a posterolateral (PL) bundle. The anteromedial bundle inserts at a more medial and superior aspect of the lateral femoral condyle, while the posterolateral bundle inserts at a more lateral and distal aspect of the lateral femoral condyle, 36.9 ± 2.9 mm, respectively 20.5 ± 2.5 mm in length. Both bundles are similar in size, with an average width of 5.0 ± 0.7 mm and 5.3 ± 0.7 mm in the mid-substance [20]. Occasionally there is an additional intermediate bundle in between these two bundles [21].

The AM bundle is thought to be more important as a restraint to antero-posterior translation of the knee, while the PL bundle is thought to be a more important restraint to rotational movements about the knee [22]. Amis and Dawkins [21] reported that PL bundle is dominant in resisting anterior tibial translation during extension, whereas during the 90° flexion, it is the AM bundle. Hence, they pointed out a reciprocal relationship between the two bundles. Gabriel et al. [23] have shown that the in situ forces of PL bundle in response to 134 N anterior load are highest in full extension and decreased with increasing flexion. In addition,

their study revealed that PL bundle plays a significant role in the stabilization of the knee against a combined rotatory load.

Dissection by Giuliani et al. [24] found that the primary blood supply to the ACL comes from the middle genicular artery, with additional supply coming from the inferomedial and inferolateral genicular arteries. While, posterior articular branches of the tibial nerve provide innervation to the ACL [25].

Histological studies have shown the presence of mechanoreceptors under the synovial membrane on the surface of the ligament [26, 27]. These special receptors are classified according to their adaptability to excitatory signals. Ruffini corpuscles and Golgi tendon organ like-receptors are slowly adapting mechanoreceptors which exhibit continuous activity in response to changes in motion, position and rotation angle of the joint. On the other hand, the rapidly adapting mechanoreceptors are the most sensitive indicators of changes in tension in the ligament. They identify acceleration and are excited regardless of joint position. Pacinian corpuscles are also rapidly adapting mechanoreceptors of the ACL [18, 26, 27]. In addition, free nerve endings act as nociceptors and may also play a role in vasomotor control [27].

I.2.4. Kinematics of the knee

Biomechanics of the knee describes the function of the knee joint in terms of its mechanical components.

Kinematics, as it relates to the human body, describes the motion of diarthrodial joints as well as locomotion and gait. An understanding of normal joint kinematics is important for comparison purposes during diagnosis of injury and for evaluating the success of treatment protocols [28].

The movement of the knee joint is governed by its ligaments, other supporting soft tissue structures, and the geometric constraints of the articular surfaces. The knee is capable of movement in six degrees of freedom: three rotations and three translations (Fig 6). To follow the recommendations of the International Society of Biomechanics [29] the description of the knee motion can be accomplished by relating movement to three principle axes: the tibial shaft axis, the femur epicondylar axis, and the anterior-posterior axis, which is perpendicular to the two other axes (Fig 7)[30]. Translations along these axes are referred to as proximal-distal or compression-distraction, medial-lateral shift, and anterior-posterior drawer, respectively. Rotations about these axes are referred to as internal-external rotation, flexion-extension, and varus-valgus (or abduction-adduction) rotation, respectively.

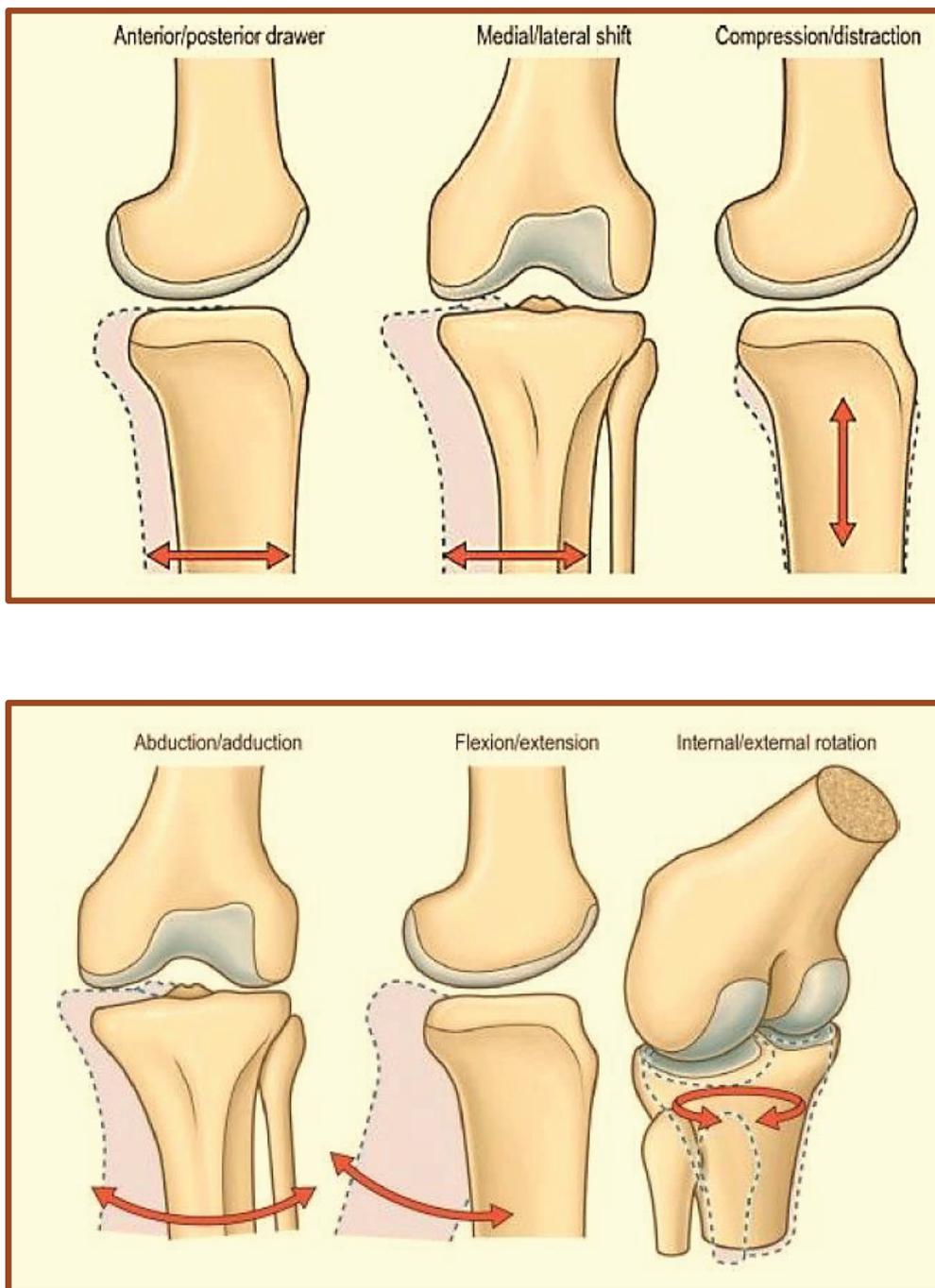


Fig. 6 Knee joint motion in three dimensions, which is described using six independent variables [31]

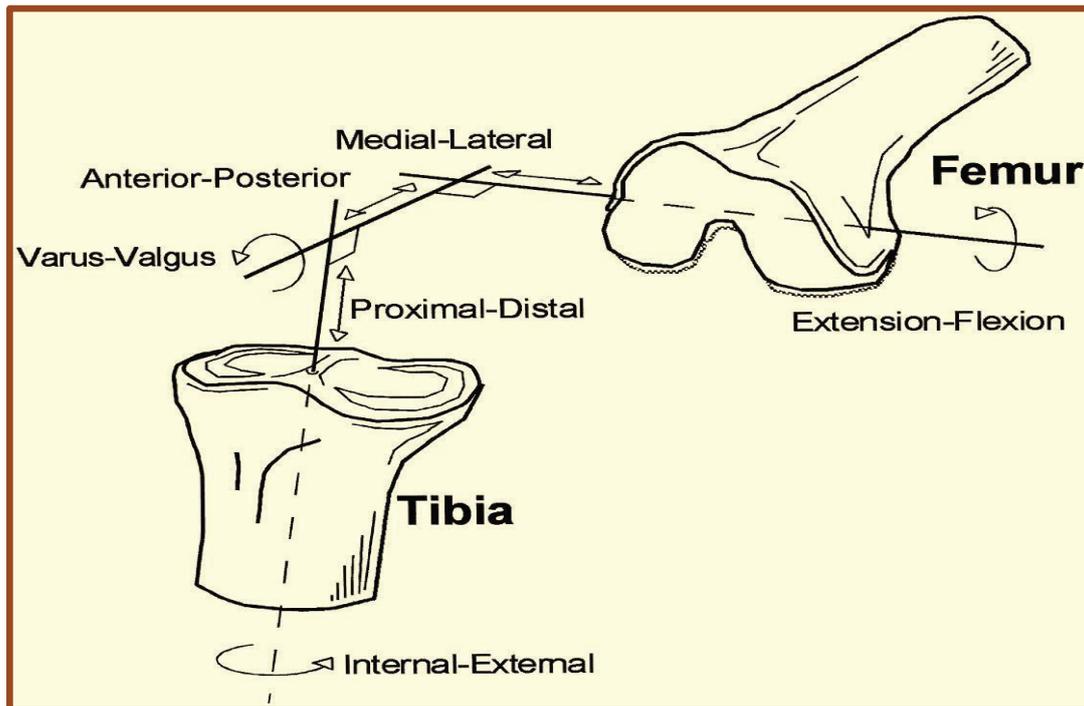
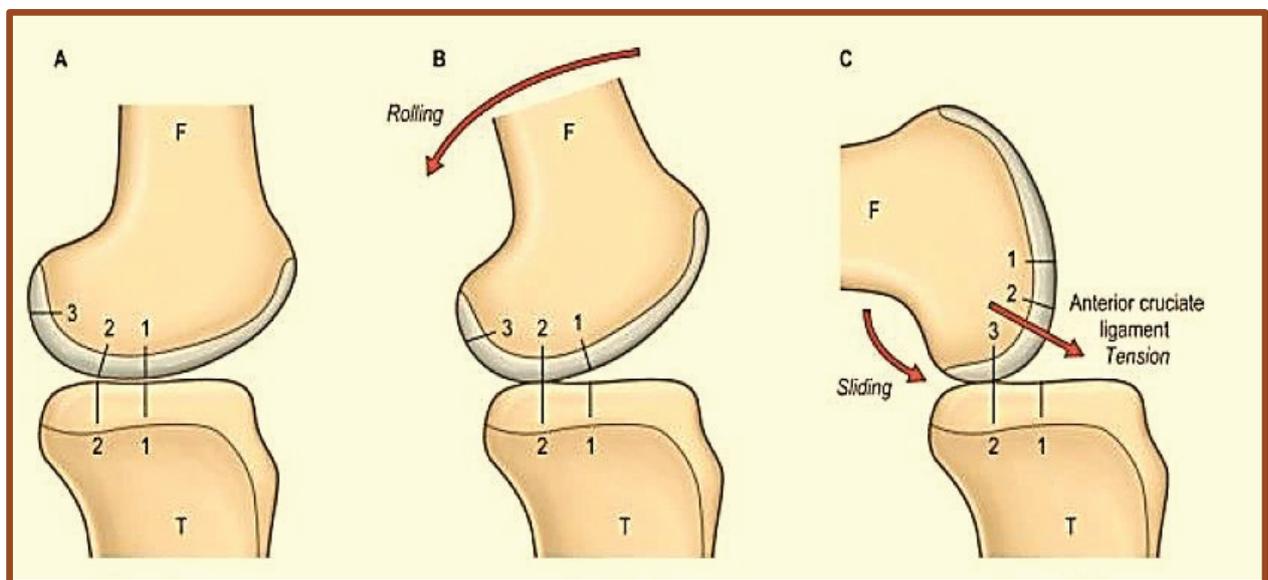


Fig. 7 Schematic diagram illustrating the six degrees of motion of the human knee joint [30]

Description of knee motion implies sagittal plane motion and rotational components of the femorotibial joint. It is best described by the “instantaneous center of motion” theory, which suggests that motion occurs about any instant point that acts as the center of rotation and therefore does not move [1].

Relative surface motion between the tibia and the femur during flexion and extension occurring about the center pathway appears to be that of gliding and rolling, the ratio of which is determined by the geometry of the articulating joint surface [32]. It changes from flexion to extension, with rolling of the femorotibial joint predominating near extension and gliding predominating as the knee is flexed [14](Fig 8). The articular surface of the medial condyle is shorter and wider compared to the lateral condyle, which results in a rotatory movement around the longitudinal axis of the knee during knee flexion and extension. The medial tibial plateau is slightly concave and deepened by the medial meniscus [33]. Together with the saddle - shape of the lateral tibial plateau, this results in the localisation of the axis of longitudinal rotation of the knee through the medial side of the knee [2]. This, again, results in

a greater posterior displacement of the lateral femoral condyle than the medial femoral condyle during knee flexion, which in turn induces an internal rotation of the tibia. 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 [34].



Extension:	Early flexion:	Deeper flexion:
Central contact	Posterior rolling.	ACL prevents femur rolling
femur onto tibia	Contact now moves	back further, so now it slides
	back to F2 onto T2	on the tibia. F3 moves onto T2

Fig. 8 Knee joint kinematics during gait: rolling and gliding [31]

I.2.5. Biomechanics of ACL

The ACL has a role in providing mechanical stability to the knee joint and also plays an integral role in knee joint sensorimotor control and proprioception [35].

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 mid-substance and the ligament-to-bone insertion site. Basically, they can be characterized 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 [36]. In vivo studies have shown that the maximum tension of ACL varies between 1725 and 2195 Newton [37]. In addition, its biomechanical properties depend on its orientation, so it is anisotropic tissue.

But we have to mention that the biomechanical properties of ligaments and tendons depend on several biologic factors. For example, differences in specimen age, species, skeletal maturation, anatomic location, as well as exercise or immobilization can affect the properties of these tissues [28].

The parameters describing the mechanical properties of the ligament substance include tangent modulus, ultimate tensile strength, ultimate strain, and strain energy density [38]. Further, the tangent modulus, tensile strength, and strain energy density of the AM bundle in the human ACL are larger than that for the PL bundle [39]. This fact that different bundles have different properties suggest that each bundle contributes to knee joint stability differently, which may have important ramifications on their replacements [28].

The ACL was found to be loaded under both internal and external tibial rotation between full extension and 45° of flexion. An internal tibial torque of 10 Nm produce a force on the order

of 100 N in the ACL at 20⁰ of flexion, while an external tibial torque of 10 Nm produced a smaller force on the order of 50 N [40].

In terms of arthrokinematics, the ACL primarily prevents anterior tibial translation which is clinically assessed with the Lachman test, a reliable non-invasive diagnostic test for the ACL rupture [41]. In addition, the ACL plays a crucial role in limiting axial rotation [42].

I.3. ACL rupture and reconstruction

I.3.1. Incidence of ACL injury

Anterior cruciate ligament injury is one of the most common injuries in sport. Incidence rate for ACL tears are difficult to assess because some injuries remain undiagnosed. In the United States, anterior cruciate ligaments injuries total between 100 000 and 200 000 yearly, making this the most common injury [43]. It was estimated that 175.000 ACL reconstructions were performed in the year 2000 in the United States [44]. In France, 35501 ACL reconstructions were performed in 2006, against 32333 in 2005 [45].

While the rate for noncontact injuries ranges from 70-84% [46, 47], the incidence of noncontact injuries in female is 2 to 8 times greater than in males [46, 48, 49]. A meta-analysis of ACL rupture rates has revealed the incidence to be highest in the team sports of female collegiate basketball and soccer, roughly 0.30 per 1000 exposures. In comparison, male collegiate counterparts in the same sports have an incidence of roughly 0.10 per 1000 exposures [49].

I.3.2. Mechanism of injury

As a step toward more effective injury screening and prevention, the maneuvers during which most ACL injuries occur, have been identified and examined. Most non-contact ACL injuries are reported to arise from a sudden deceleration while either running or changing direction or landing from a jump [50]. Other described mechanisms of ACL tears include knee hyperextension and hyperflexion [51, 52].

High quadriceps activation in combination with a small knee flexion angle during a landing is thus viewed as a worst-case scenario for sagittal plane-based ACL injury. An extended hip

joint, coupling with the knee joint posture during landings, has also been linked to greater anterior tibial shear forces during a landing task [53, 54].

Notch stenosis may contribute to an increase in rates of ACL injury [55, 56]. The notch width index (NWI) has been used as a measure of notch stenosis [55]. The NWI is the ratio of the width of the intercondylar notch to the width of the distal femur at the level of the popliteal groove. In Souryal's study [55], the mean NWI for normal knees was 0.2338, for acute ACL injured knees, it was 0.2248, and for those with bilateral ACL injuries, it was 0.1961. Notch stenosis is not felt to be a factor in gender differences in ACL injury [19].

I.3.3. Impact of ACL rupture

The main symptom of patients with ACL rupture is an unstable knee joint - "giving way" symptom. Patients complain that the knee gives away particularly if they are not running in a straight line. In addition, this instability affects the knee function during daily activities and sports. It has been proposed that decline in a proprioception follows disruptions of the ACL, impairing the ability of muscles around the knee to respond appropriately to applied loads. This may make the knee more susceptible to episodes of instability and predisposes it to further injury [57]. Each time the knee gives way, there is subluxation, shearing and compression, which can cause tears of menisci, further stretching of the capsular ligaments, and damage to the articular surface. It is interesting to note that amount of intraarticular damage correlates extremely well with the number of episodes of giving way. In fact, over 50% of patients with ACL deficiency have radiological signs of knee osteoarthritis 10 years after injury [58].

It has been widely accepted that ACL deficient subjects alter in the knee function after injury. A small percentage of ACL-deficient subjects are able to return to function at a high level after injury without complaint of instability (*copers*) [59–61]. The other group of ACL-

deficient subjects are able to avoid evoking episodes of instability by mitigating their activity levels (*adapters*) [62]. And, the subjects of the last group are unable to return to their premorbid level of sports play or activity because of repeated episodes of giving way (*noncopers*) [60]. Therefore, this last group of patients requires an ACL reconstruction to restore joint stability.

I.3.4. Diagnosis of ACL rupture

The history of injury is the first step toward diagnosis of ACL rupture. Key points of the history suggesting ACL tear include a noncontact mechanism of injury, the identification of an audible popping sound, the early occurrence of swelling as a result of bleeding (hemarthrosis) from the rupture of the vascular ACL, and an inability to continue to participate in the game or practice after the injury.

During the physical examination, Lachman test, pivot shift test and drawer test are useful in the assessment of an ACL tear. The next step in diagnosis of the ACL tear is magnetic resonance imaging (MRI), which according to a systematic review, had a sensitivity of 86%, a specificity of 95%, and an accuracy of 93%, as confirmed by arthroscopy [63].

Lachman test

In the Lachman test, the knee is placed in a position of 20° to 30° of flexion and is slightly externally rotated to relax the pull of quadriceps and iliotibial band. The femur is firmly stabilized with the examiner's outer hand and anteriorly direct force is applied to the proximal calf with the examiner's other hand on the tibia. Crawford R et al. [64] have revealed that sensitivity and specificity of the Lachman test for an ACL tear were 85% and 94 %, respectively.

Anterior drawer test

The anterior drawer test is performed with the knee at 90° of flexion. The examiner sits on the patient's feet and grasps the proximal part of tibia and pulls it forward. Excessive translation of the tibia anteriorly compared to contralateral side indicates that the ACL is likely torn. The sensitivity and specificity of this test in evaluation of chronic ACL rupture is 92% and 91%, respectively, while in the acute phase these values go down to 49% and 58% [64].

Pivot shift test

In this test, the patient is in a supine position and relaxed. While the limb is held in external rotation and the knee in full extension, the lateral tibia is subluxed anteriorly in relation to the femur. While external rotation of the limb is maintained, the hip is brought into abduction to relax the iliotibial band, and an axial and valgus load is applied to the knee as it slowly flexed. This test simultaneously evaluates both rotation and translation of the tibial plateau relative to the femoral condyle.

All variations of pivot shift test are based on anterior subluxation of both lateral and medial tibial plateaus of the tibia in extension and very early flexion (lateral side subluxing more than the medial). With further flexion (at 20° to 40°), the posterior pull of the iliotibial tract reduces the lateral tibia. Usually, the clinician grades the relocation event subjectively (absent pivot, 0; a pivot with a smooth glide, 1+; a pivot with an abrupt shift, or jump, 2+; or momentary locking in a subluxed position before reduction, 3+) [65].

For the pivot shift test, while the specificity is high (98%), the sensitivity is low (24%) [66].

I.3.5. Imaging evaluation

Plain radiography should be the first imaging study ordered when any knee with a suspected acute ACL injury is evaluated. Ligament injury may be found and associated fractures may be identified. ACL injury has three main radiographic signs: avulsion of the intercondylar tubercle, anterior displacement of the tibia with respect to the femur, and Segond fracture, results from an avulsion of the lateral capsule from the tibia at location posterior to the Gerdy tubercle and superior and anterior to the fibular head. However, these radiographic signs are frequently absent in patients with ACL injuries. X ray films of chronic ACL-deficient knee may demonstrate a 'lateral notch sign', which is an exaggeration of the normal indentation of the sulcus terminalison the lateral femoral condyle [67].

MRI has been established as a highly useful method of accurately evaluating the ACL in adult patients. According to various studies, the accuracy to detect a torn ACL varies between 93% and 97%. Although the ACL can be visualized on CT, its visibility is impaired in the presence of haemarthrosis and MRI is also the best in evaluation of associated meniscal tears, bone bruises, chondral injuries, and complex ligament injury patterns. Rubin DA et al. [68] have revealed that MRI's sensitivity is significantly decreased if other major ligamentous injuries are present in the knee.

I.3.6. Treatment of ACL rupture

The treatment of ACL rupture depends mainly on: age, functional disability, functional requirements and association with other structures injuries.

The ‘potential copers’, patients without concomitant injuries, patients with comorbid medical conditions (serious cardiac, renal, or hepatic disease) or who no longer wish to participate in strenuous physical activities, could be candidates for conservative treatment. For these patients who opt for conservative treatment, physical therapy with an experienced physical therapist is recommended. However, without surgical repair, the knee generally remains unstable and prone to further injuries [69].

Because of the frequent failure of the nonsurgical treatment of the ACL injuries, surgery remains the treatment of choice in almost all athletes who want to remain active. Over the years, surgical techniques for reconstruction of deficient ACL have evolved and changed. Nowadays, there are two main surgical approaches for ACL reconstruction: single- and double-bundle reconstruction. There are studies that have shown that double bundle reconstruction restores knee stability closer to normal than does single-bundle ACL reconstruction [70]. Claes S et al.[71] have found that both ‘anatomic’ single- and double-bundle reconstruction adequately restore tibial rotational excursion in a human, in vivo kinematic model. As well, a meta-analysis performed by Meredith et al. [72] failed to prove the superiority of the double-bundle method.

There are a number of choices available to the orthopedic surgeon in determining which graft is best for a patient who will undergo ACL reconstructive surgery. The patellar bone-tendon-bone (BTB) graft has been the gold standard graft choice for ACL reconstruction since it was popularized in the mid-1980’s. The hamstring tendon graft, quadriceps tendon graft, allografts and synthetic (prosthetic) grafts are the other alternatives.

1.3.7. The operation technique in our center

The operation technique has previously been published [73–75].

The patient is placed in supine position. A distal leg holder is positioned at the end of table, which enables the flexion of the knee at 90 degrees and, the second one is positioned lateral of thigh to maintain the knee in this position. The tourniquet is inflated to 300 mmHg and is placed at the top of the thigh (Fig 9).



Fig. 9 Installation of patient

For harvesting of the bone-patellar tendon-bone autograft, a longitudinal incision at the medial border of the patellar tendon starting from the distal pole of patella to a point approximately 2 cm distally to the tibial tuberosity is performed. Then a 10 mm wide central strip of the patellar tendon is prepared including a small block from the patella and a longer,

wedge shaped bone block from the tibial tuberosity. Both of them are armed with strong resorbable sutures (Fig 10).

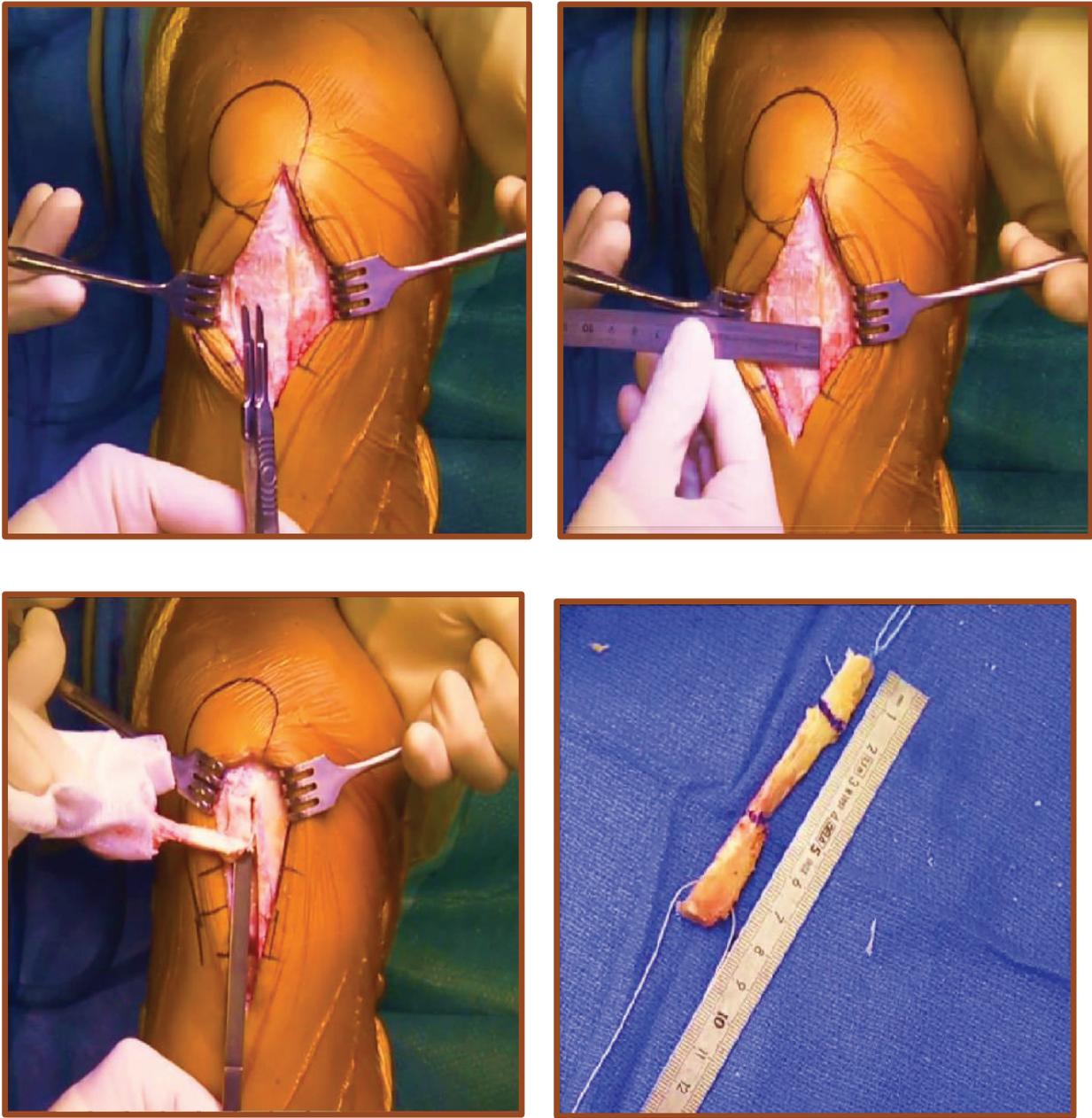


Fig 10. Preparation of the graft. Form and dimensions of the graft.

Arthroscopic surgery - Femoral tunnel preparation

The intercondylar notch, the tibial and the femoral ACL-attachment area are cleaned using a shaver blade. It is of crucial importance to visualize the posterior attachment area of the ACL at the deep posterior femoral condyle, back at the bone-cartilage transition.

Guided by a special outside-in aiming device, a short second skin incision is established. The aiming device is placed in the center of the native ACL attachment resulting in an extra-articular starting point on the distal lateral femoral metaphysis. The guide wire is then drilled from outside into the joint from proximal to distal. The guide pin position is verified via the medial portal and then overdrilled using a 10 mm cannulated drill bit (Fig 11).

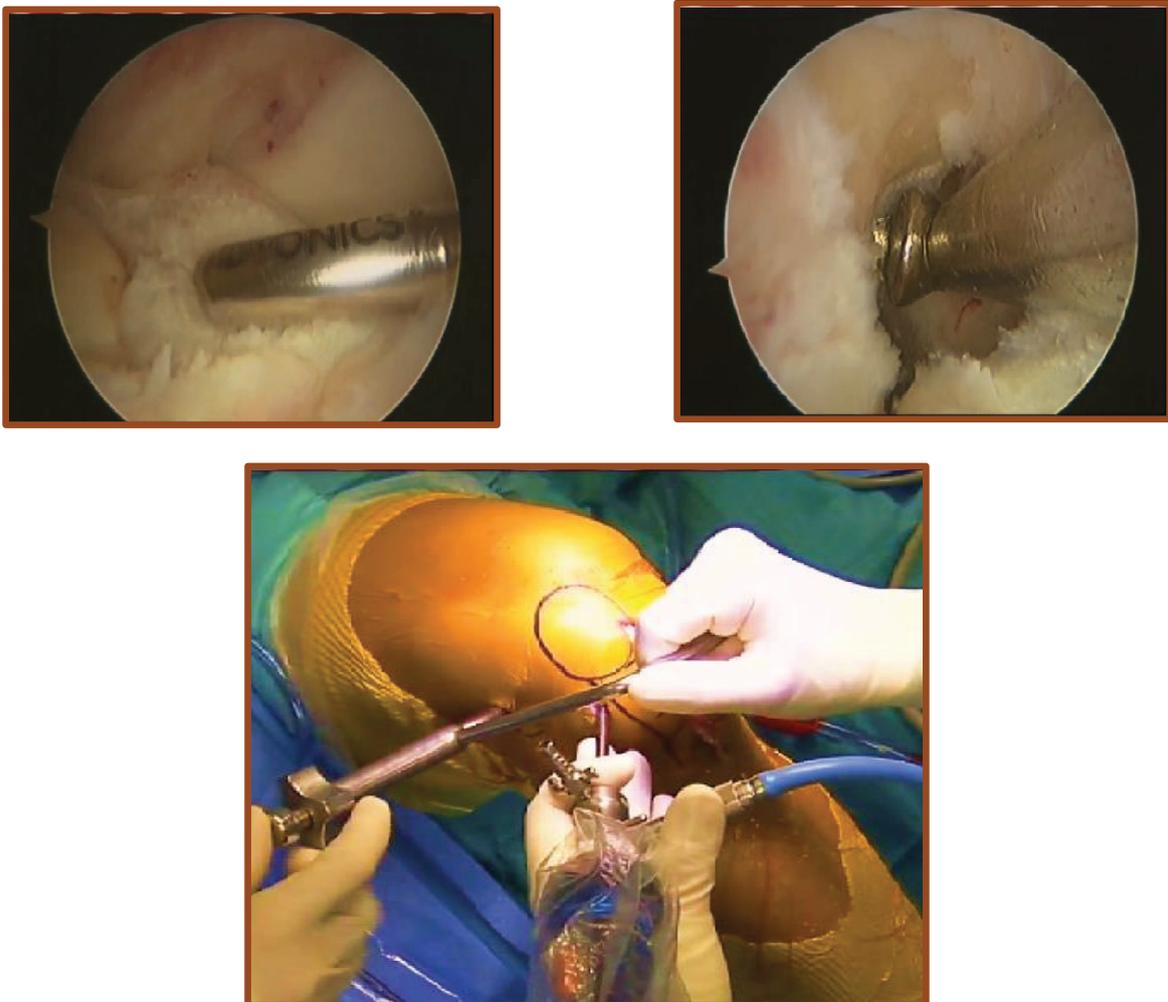


Fig 11. Femoral tunnel preparation

Tibial tunnel preparation

For the tibial tunnel preparation, a tibial aiming device is placed between tibial spine close to the shallow border of PCL, behind the anterior corn of medial meniscus and medial to the anterior corn of the lateral meniscus. Then, a guide wire is introduced under arthroscopic control through aiming device. The tunnel is expanded firstly to a diameter of 9 mm. The tibial tunnel aperture is debrided from soft tissues with a shaver blade (Fig 12).

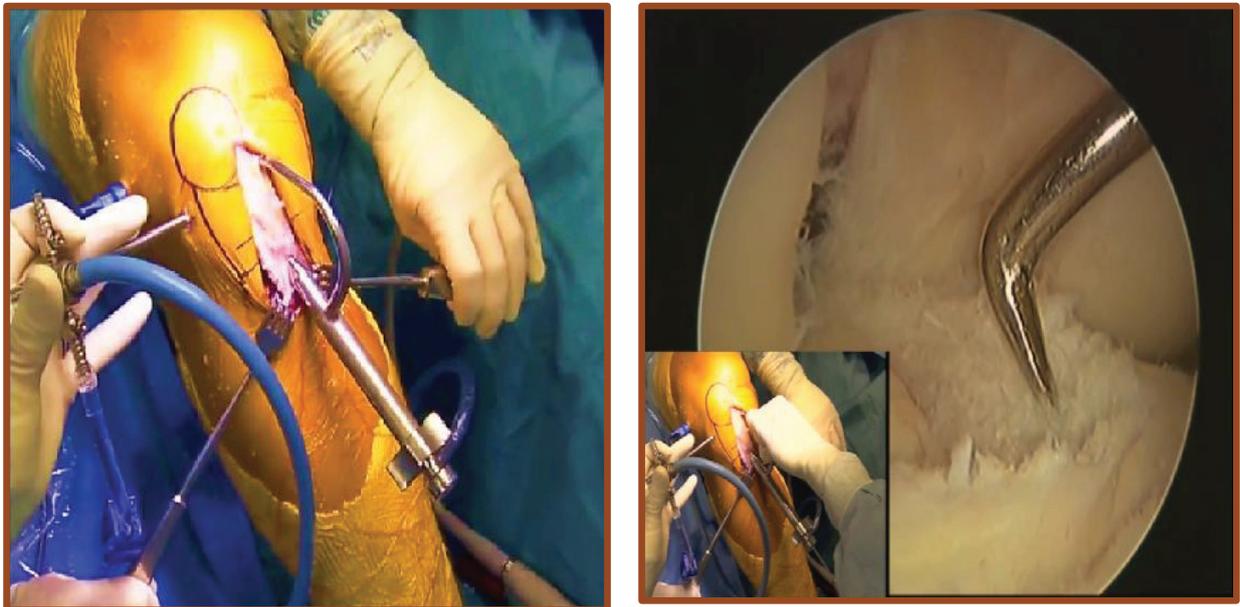


Fig 12. Tibial tunnel preparation

Anterograde graft introduction and graft fixation

Shuttle sutures are introduced through the femoral tunnel and pulled out through the tibial tunnel. The graft is then inserted into the femoral tunnel. Here it is possible to determine the rotational position of the bone block in the femoral tunnel. The goal is the tendon's backside to face anteriorly. Under constant pull distally the bone block is then tapped into the femoral tunnel until a press-fit anchorage is achieved, ideally flush at the femoral aperture (Fig 13).

The graft is preconditioned in 10 cycles, and the isometry of the graft is checked. The graft is then fixed in 20° of flexion using an interference screw (Fig 14).

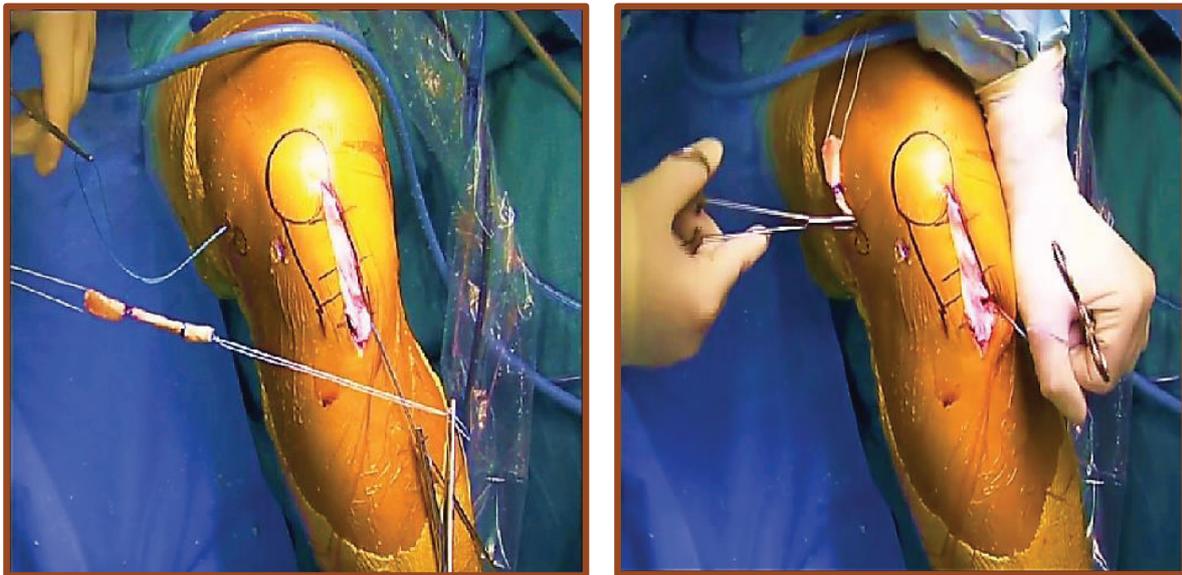


Fig. 13 Anterograde graft introduction

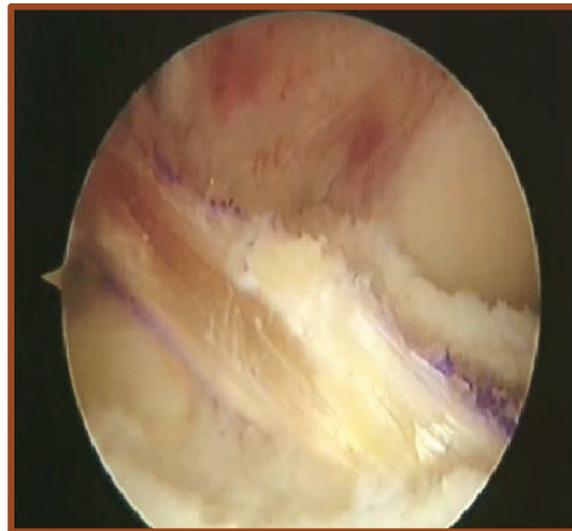
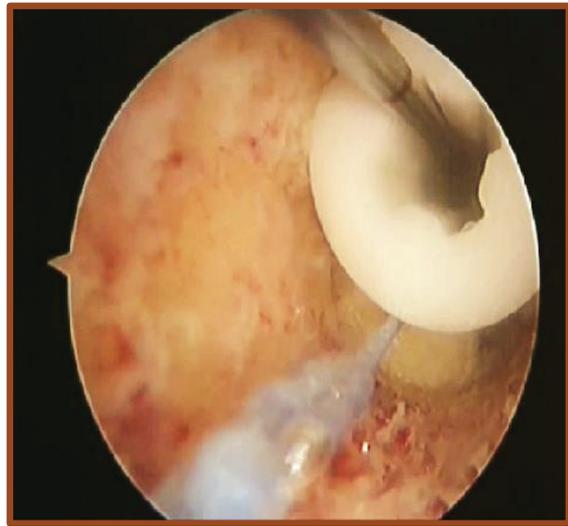
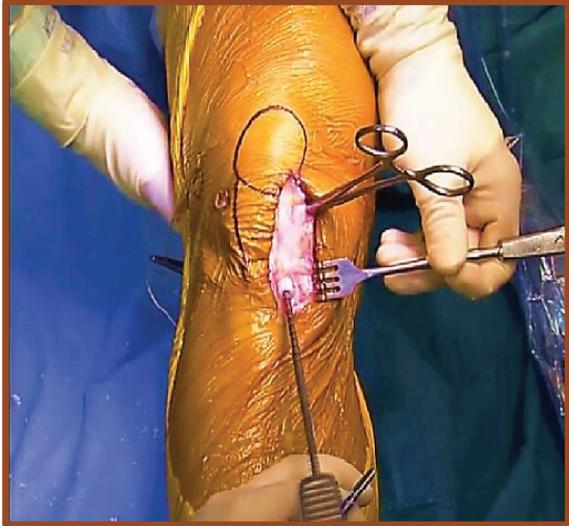


Fig. 14 Tibial graft fixation. The graft in place

II. Review of literature

Several studies (in vitro and in vivo) have evaluated the knee kinematics in patients with ACL deficient pre- and post-operation by comparing the results with healthy contralateral knee or control group. These studies used different systems to assess the knee kinematics: dynamic radiographs, dynamic MRI, static measurement, navigation, dynamic radio-stereometry, stereo dynamic fluoroscopy, accelerometers and opto-electronic systems.

Because of pathologic knee laxity resulting from ACL injury is a complex and 3D motion, for better understanding the pathology and efficacy of the treatment, it is necessary to assess the knee angular patterns in sagittal, axial, frontal plane and the antero-posterior translation.

II.1. Kinematic evaluation of the anterior cruciate ligament deficient knees and anterior cruciate ligament reconstructed knees

Numerous studies have provided information on the alterations of the kinematics at the ACLD and ACLR knees [76–85]. Some of the in vitro cadaveric studies [86, 87] have provided some insight into the kinematical behavior of the anterior cruciate ligament deficient knees under controlled conditions but they were unable to accurately simulate the effects of weight bearing and muscles contraction on the joint kinematics.

The most commonly used method for assessing dynamic movement uses skin-marker-based motion capture system, but some studies reported that soft tissue artifact biased the measurements, especially in knee joint translation and internal/external rotation angle [88, 89]. According to Cappozzo et al. [90], the motion of the marker with respect to the underlying bone, as a result of skin movement, ranges from a few millimeters up to 40 mm. In order to overcome the skin-to-bone movements, the reflective markers have been fixed to pins

driven into the skeletal bones [91], but this method is invasive, which inevitably limit the patient population, and will never be used in clinical setting.

Dynamic MRI can be used to study static or quasi-static knee joint motion [92]. During the last few years, a method combining cine-PC and MRI has been developed [93]. The investigation is performed with the patient lying down and actively moving the knee between full extension and 30 degrees of flexion, but without weight bearing.

Radiostereometry (RSA) is another very accurate method that has been developed to measure knee kinematics during active motion and in vivo. RSA is able to measure motion down to 0.1 mm and 0.1-0.3 degrees and can be used for three-dimensional (3D) recordings [94]. On the other side, there are some disadvantages to using RSA. It is an invasive method and each patient and investigation will take a long time and is extremely labour intensive. A specially designed laboratory is needed. Even if the radiation is lower than that in an ordinary radiographic examination, the examinations add to the total radiation burden. And, despite being accurate and reliable, RSA will never been used in everyday clinical work.

II.2. KneeKGTM and its role in knee kinematic evaluation

As mentioned, obstacles like skin movement artifact, non-weight bearing, and invasive methods make acquiring the movement accurately at knee level not simple.

Therefore, the KneeKGTM attachment system and KneeKGTM axis definition procedure were developed with the objective of providing high reliability movement analysis.

The development of the KneeKGTM began in 1992 at the Imaging and Orthopaedics Research Laboratory in Montreal, Canada, to find out more thoroughly the impact of tunnel positioning on ACL graft elongation, torsion, and bending. After reviewing the available scientific literature, they came to the conclusion that there is a need for an assessment device to accurately quantify knee biomechanics in 3D [5]. Because skin motion artifact is a major factor in precision of measurements of knee joint motion, the group developed a harness to be fixed quasi-rigid on the thigh and calf, therefore reducing the skin motion artifact [95]. In a comparative study of three non-invasive attachment systems, where the authors utilized a low-dose biplanar x-ray system to evaluate the displacement of these different attachment systems, the KneeKGTM attachment system had the most stable attachment [96].

The precise quantitative rotational data provided by KneeKGTM make it suitable tool for evaluation:

- Of risk factors for ACL injury
- To predict certain in vivo ligament bending and torsion deformations
- To evaluate if the knee is biomechanically ready to resume contact sports when orthopedic surgeons release the patient from ACL reconstruction
- To illustrate the importance of 3D biomechanical evaluation in ACL injuries.

II.3. Flexion-Extension movement patterns in anterior cruciate ligament deficient knees and anterior cruciate ligament reconstructed knees

ACL deficiency and its effects on knee joint movement patterns have been investigated extensively regarding flexion-extension using gait analysis [76, 97–101].

As the primary restraint in antero-posterior translation, the rupture of the ACL would cause instability in the knee. As consequence, the ACL deficient patients use stronger contraction of the hamstrings to pull tibia posteriorly or they walk with weaker contraction of the quadriceps to avoid pulling the tibia anteriorly.

It was found that ACLD knees showed higher flexion than contralateral knees ($39.2 \pm 13.1^\circ$ respectively $34.8 \pm 10.4^\circ$) at the end of the stance phase [97], which means that patients use higher flexion strategy during gait. Some of the other studies [98, 99] showed also higher flexion from mid to late stance phase, or throughout the stance phase [100].

On the contrary, other reports have shown that ACLD knees tended to flex less throughout the mid- and terminal stance phase [76, 101].

Besides ACL deficiency, ACL reconstruction has also been investigated regarding its effects on knee flexion-extension movement patterns. It is very important to note that time from surgery may play a very important role in this context.

Knoll et al. [102] studied the knee kinematics in ACL reconstructed knee 3 weeks and 4 months post-operatively. They revealed that after 3 weeks from reconstruction, the ACL reconstructed knee flexed more during stance phase and less during swing phase. They noted the same results after 4 months post-operatively, but no significant differences were found 8 months after reconstruction.

Devita et al. [103] examined the gait of ACL-reconstructed individuals 3 weeks and 6 months post-operatively. They have found that ACL reconstructed subjects walked with approximately 10° higher flexion at knee joint 3 weeks after surgery, while the kinematics were not different between ACLR group and healthy subjects 6 months post-operatively. In addition, Bulgheroni et al. [104] examined ACL-reconstructed individuals 2 years post-operatively and reported no significant differences in the flexion-extension patterns.

There are studies that exhibited significantly less knee joint flexion [77, 105] after ACL reconstruction. Deneweth et al. [77] noted that ACL reconstructed knees exhibited significantly less flexion, 4 months post-operatively. Delahunt et al. [105] also noted that ACLR knee exhibited significantly less peak knee joint flexion, 4.4 years after surgery. It is important to note that subjects in these studies were investigated during the performance of a diagonal jump landing task [105] and during 3 single-legged, forward hop landing trials [77].

Thus, it seems that given time, ACL reconstructed individuals can eventually regain normal knee gait patterns regarding knee flexion-extension.

Study	Subjects	Methods	Activity	Results
Chen et al [97]	10-ACLD 10-CK*	Dual fluoroscopic imaging system	Walking	ACLD higher flexion than CK
Beard et al [98]	18-ACLD 18-CK	3D VICON gait analysis system	Walking	ACLD higher flexion than CK
Roberts et al [99]	18-ACLD 18-CK	Video-based motion analysis system	Walking	ACLD higher flexion than CK
Berchuck et al [101]	16-ACLD 10-CG^	Optoelectronic system	Walking	ACLD less flexion than CG
Georgoulis et al [76]	13-ACLD 10-CG	3D optoelectronic gait analysis system	Walking	ACLD less flexion than CG

Knoll et al [102]	25-ACLD 25ACLR 51-CG	3D ultrasound based system	Walking	3 weeks, 4 months postop-ACLD higher flexion than CG 8 months postop- No significant difference
Devita et al [103]	8-ACLD 22-CG	Video-based motion analysis system	Walking	3 weeks postop- ACLD higher flexion than CG 6 months postop-No significant difference
Bulgheroni et al [104]	15-ACLR 5-CG	3D optoelectronic gait analysis system	Walking	2 years postop-No significant difference
Delahunt et al [105]	13-ACLR 16-CG	3 CODA mpx1 units integrated with 4 AMTI walkway embedded force-plates	Diagonal jump landing task	4.4 years postop- ACLR less flexion than CG
Deneweth et al [77]	9-ACLR 9-CK	3D-High-speed biplane radiography	3 single-legged, forward hop landing trials	4 months postop- ACLR less flexion than CK

Table 1 - Literature review of flexion-extension movement patterns. *CK=Contralateral Knee, ^CG= Control Group

II.4. Internal-external rotation movement patterns in anterior cruciate ligament deficient knees and anterior cruciate ligament reconstructed knees

Even though the knee flexion-extension movement patterns have been extensively studied in both ACL deficient and reconstructed subjects, less is known regarding axial plane movements of the tibia with respect to the femur.

Some studies [78, 79] have found that ACL deficiency caused a statistical difference in internal tibial rotation when compared to healthy, contralateral knees, during walking respectively during stair ascent and descent. They showed also that ACLR knees exhibited some improvements in joint kinematics, but not being fully restored to a normal level. There is still significant difference between ACLR knees and ACL intact knees.

Claes et al. [71] have studied rotational stability in the ACLD knees, single-bundle ACL reconstruction and double-bundle reconstruction. They have found that there were significant differences between ACLD knees and control group (greater tibial rotation was seen in ACLD knees), while there is no significant difference between single-bundle and double-bundle ACL reconstruction when compared to contralateral knees and healthy control group. That means that both single- and double-bundle ACL reconstruction adequately restore tibial rotational excursion.

Stergiou et al. [106], in their systematic review, explored movements in the axial plane in ACL deficient and reconstructed subjects during different situations. In their first study [76], they found that ACLD group exhibited significantly increased tibial rotation range of motion during the initial swing phase of the gait cycle comparing to the ACLR group and control group, and no significant difference were found between ACLR group and control group. They used bone-patellar tendon-bone (BPTB) autograft for ACL reconstruction. In the second work, they wanted to test the ACLR group in higher demanding activities [107]. They

examined ACLR group (BPTP autograft) and control group during descending stairs and subsequent pivoting. The evaluation was done at an average time, 12 months from surgery. The tibial rotation range of motion during the pivoting period was found to be significantly larger in the ACLR group compared with contralateral knee and with the healthy control group. In the next study, they wanted to identify if tibial rotation remains excessive after a longer time period (2 years) following the reconstruction, during high demanding activities [108]. They found that tibial rotation remains significantly excessive even 2 years after reconstruction. Finally, they verified if the changing of autograft would be the solution of this excessive tibial rotation. In their last study [109], quadrupled hamstring tendon was used as autograft, and kinematic evaluation was done during high demanding activities. Even in this study, tibial rotation was found to be significantly larger in ACLR knees when compared to contralateral knee and healthy control group.

Study	Subjects	Methods	Activity	Results
Claes et al (2011) [71]	20-ACLD 8-ACLR-SB* 8-ACLR-DB^ 10-CG	3D optoelectronic gait analysis system-reflective skin markers	- Walking - Descending a 25 cm high platform - Descending this platform followed by subsequent pivoting	1.Higher internal tibial rotation in ACLD 2.No significant difference between ACLR-SB and DB
Gao et Zheng (2010) [78]	14-ACLD 14-ACLR 14-CG	3D optoelectronic gait analysis system-reflective skin markers	- Walking	1.Higher internal rotation ACLD vs CG 2.Higher internal rotation ACLR vs CG
Gao et al (2012) [79]	12-ACLD 12-ACLR 12-CG	3D optoelectronic gait analysis system-reflective skin markers	- Stair ascent and descent	1.Higher internal rotation ACLD vs CG 2.Higher internal rotation ACLR vs CG

Georgoulis et al (2003) [76]	13-ACLD 21-ACLR 10-CG	3D optoelectronic gait analysis system-reflective skin markers	- Walking	1. Higher internal rotation ACLD vs CG 2. No significant difference between ACLR vs CG
Ristanis et al (2003) [107]	20-ACLR	3D optoelectronic gait analysis system-reflective skin markers	- Descending stairs and subsequent pivoting	Higher tibial rotation range of motion- ACLR vs CG
Ristanis et al (2006) [108]	9-ACLR 10-CG	3D optoelectronic gait analysis system-reflective skin markers	- Descending stairs and subsequent pivoting - Jumping from platform and subsequent pivoting	Higher tibial rotation range of motion- ACLR vs CG 2 years post-OP
Chouliaras V et al (2007) [109]	11-ACLR – PT ^a 11-ACLR- HT ^b 11-CG	3D optoelectronic gait analysis system-reflective skin markers	- Descending stairs and subsequent pivoting	Higher tibial rotation range of motion in both ACLR groups vs CG

**Table 2 - Literature review of internal-external rotation. *SB=Single bundle, ^DB= Double bundle,
PT^a=Patellar tendon, HT^b=Hamstring tendon.**

II.5. Adduction-abduction movement patterns in anterior cruciate ligament deficient knees and anterior cruciate ligament reconstructed knees

The adduction-abduction movement patterns of the knee in ACLD and ACLR subjects have been presented in several studies [76–83].

Even though these studies used different activities in analyzing knee kinematics, such as: walking, running, 3 single legged-forward hop landing, stair ascent-descent, step-up exercise; most of these studies have shown greater tibial adduction when compared to contralateral knee or healthy control group. Studies that have analyzed the frontal movement pre- and post-operatively revealed that ACLR knees exhibited less abnormality, but the adduction-abduction movements were not fully restored to a normal level.

Some of the studies [78, 79] which analyzed ACLD, ACLR and intact knees have shown that the kinematics of the ACLR knees were more similar to those of the ACLD knees when compared to the ACL intact knees. This suggest that the ACLR knees had been ‘under corrected’ by surgery procedure.

Webster et al. [82] analyzed alterations kinematics following hamstring and patellar tendon ACLR surgery. They found that the hamstring group had significantly reduced adduction compared to both the patellar tendon and control groups at all-time points during stance. But, there were no difference between the patellar tendon and control groups. There were also no significant differences when the reconstructed knee was compared to the contralateral knee for either patient group.

In the other hand, Zhang et al. [83], showed that ACLD patients and control subject abducted-adducted their knees similarly, except at the heel contact where ACLD knees abducted more than the normal subjects did.

Study	Subjects	Methods	Activity	Results
Georgoulis et al (2003) [76]	13-ACLD 21-ACLR 10-CG	3D optoelectronic gait analysis system- reflective skin markers	Walking	No significant difference
Deneweth et al (2010) [77]	9-ACLR 9-CK	High-speed biplane radiography	3 single legged, forward hop landing trials	No significant difference
Gao et Zheng (2010) [78]	14-ACLD 14-ACLR 14-CG	3D optoelectronic gait analysis system- reflective skin markers	Walking	ACLD more adduction compared to CG; ACLR no significant difference compared to CG
Gao et al (2012) [79]	12-ACLD 12-ACLR 12-CG	3D optoelectronic gait analysis system- reflective skin markers	Stair ascent and descent	ACLD and ACLR more adduction compared to CG
Kozánek et al (2011) [80]	30-ACLD 30-CK	Combination of MRI, dual fluoroscopy and advanced computer modeling	Step-up exercise	No significant difference
Tashman et al (2004) [81]	6-ACLR 6-CK	High-speed biplane radiographic system	Downhill running	ACLR more adduction compared to CK
Webster et al (2011) [82]	18-ACLR (patellar tendon graft) 18-ACLR (hamstring graft) 18-CG	3D optoelectronic gait analysis system- reflective skin markers	Walking	Hamstring group reduced adduction compared to both other groups. No difference between patellar and control group.

Table 3 - Literature review of adduction-abduction movement.

II.6. Antero-posterior translation patterns in anterior cruciate ligament deficient knees and anterior cruciate ligament reconstructed knees

Because of its function as the primary restraint against anterior tibial translation, ACL disruption inevitably causes alterations in the sagittal plane. Various studies describe different adaption strategies, like quadriceps avoidance gait [76, 101] or higher knee flexion [78, 85, 98–100], to prevent anterior translation of the tibia during walking. It was thought that reduced quadriceps contraction would reduce the anterior shear force applied to the tibia during the stance phase of gait, while increased angle flexion of the knee during the stance phase would increase activity of the hamstring muscle to improve joint stability, as agonist of the ACL. Waite et al. [85] have found no significant difference in A-P translation between ACLD knees and control group. Meanwhile, there are studies that have shown statistically significant difference in the A-P translation of the ACLD knees when compared to control group during walking [97]. Hoshino et al. and Tashman et al. [81, 110] have shown that the anterior tibial translation is significantly reduced after ACL reconstruction, respectively have found no significant difference between ACLR knees and healthy contralateral knees. Otherwise, Papannagari et al. [111] have presented increasing of anterior tibial translation in ACL reconstructed knees during a single-legged weight bearing lunge in full extension and 15° of flexion, when compared to contralateral intact knee.

Study	Subjects	Methods	Activity	Results
Gao et Zheng (2010) [78]	14-ACLD 14-ACLR 14-CG	3D optoelectronic gait analysis system- reflective skin markers	Walking	No significant difference
Chen et al [97]	10-ACLD 10-CK	Dual fluoroscopic imaging system	Walking	More tibial anterior translation in ACLD
Hoshino et al [110]	7-ACLD 7-ACLR	Dynamic stereo X-ray system	Downhill running	Significant reduction of tibial anterior translation
Keays et al [84]	8-ACLD 8-CG	Qualisys 3D-Motion Analysis System	Seated knee extension with 3 kg weight and a unilateral wall squat	No significant difference
Papannagari et al [111]	7-ACLD 7-ACLR 7-CK	Dual-orthogonal fluoroscopic system	Single-legged weight bearing lunge	Increasing of tibial anterior translation in ACLR
Tashman et al (2004) [81]	6-ACLR 6-CK	High-speed biplane radiographic system	Downhill running	No significant difference
Waite et al [85]	15-ACLD 15-CK	3D optoelectronic gait analysis system- reflective skin markers	Running and cutting	No significant difference

Table 4 Literature review of antero-posterior translation.

Gait changes of the
ACL-deficient knee
3D kinematic
assessment

III. Gait changes of the ACL -deficient knee 3D kinematic assessment

III.1. Introduction

Knowledge of in vivo movement of the knee is important for understanding normal function as well as addressing clinical problems, including instability and function after anterior cruciate ligament (ACL) injury. There are numerous studies that have provided information on biomechanical changes in the anterior cruciate ligament-deficient (ACLD) knees [100, 101, 112, 113]. In vitro studies using cadavers [87, 114] have provided some insight into the kinematical behaviour of the ACLD knees under controlled conditions. However, these studies are unable to accurately simulate the effects of weight bearing and muscles contraction on the joint kinematics.

The most commonly used method for assessing dynamic movement is the skin marker-based motion capture system. Nevertheless, studies using skin marker-based motion capture system have reported soft tissue artefacts, especially in knee joint translation and internal/external rotation angles [88, 89]. According to Cappozzo et al. [90], the marker displacement with respect to the underlying bone, as a result of skin movement, ranges from few millimetres up to 40 mm. On the other hand, markers that are attached to pins drilled into the bone are potentially more accurate, but are invasive.

Despite that there are several devices currently available to assess the knee kinematics [83, 115–119], the 3D biomechanical changes caused by ACL injury and possible compensations adopted by the patients following the injury are still not clearly understood. Thus, establishing an objective evaluation of the kinematics of the knee in a clinically feasible way is critical in

extensive evaluation of the ACL function and a valuable feedback for further progress of ACL treatment.

The purpose of the study was the *in vivo* evaluation of the ACL-deficient knee behaviour during all phases of gait, using a new 3D, quasi-rigid, real-time assessment tool. It was hypothesized that ACLD knees would exhibit altered joint kinematics. In order to test the hypothesis, we examined 3D knee joint kinematics (flexion–extension; internal– external rotation; adduction–abduction; anterior–posterior tibial translation) during walking at self-selected speed, in two subject groups: ACLD knees and healthy controls with bilateral ACL-intact knees.

III.2. Materials and methods

This study was prospectively conducted from January 2011 to June 2011, in the facilities of the biomechanical laboratory at our clinical center. Patients who were planned to be operated for ACL reconstruction were selected for kinematic analysis. The ACL rupture was diagnosed by clinical examination, MRI and confirmed during the surgery by arthroscopy. Patients with unilateral ACL rupture and healthy contralateral knees (that had never suffered of any kind of orthopaedic or neurological condition) were included in the study. Patients with meniscal injury where partial meniscectomy or repair was feasible and patients with grade I or II medial collateral injury were also part of the study. All patients who participated in this study were non-copers (unable to return to their premorbid level of sports play or activity). On the other hand, patients who had pain, subtotal or total meniscectomy, concomitant PCL injury, knee joint movement restriction, full thickness cartilage defect $>1 \text{ cm}^2$ and had previous history of any surgery in both knees were excluded from the study. Thirty patients were eligible for inclusion in this study; 40 patients were excluded because they did not meet the inclusion criteria.

A control group consisting of 15 participants of similar age, who had no history of musculoskeletal injury or surgery in the lower extremities and exhibited no measurable ligamentous instability on clinical examination (pivot shift, Lachman and drawer test), was selected (Table 5).

Parameters	ACLD group	Control group	p value
Age (years)	29.8±8.9	29.3±9.7	0.53 ^a
Height (cm)	172±7.8	173.1±10.2	0.71
Weight (kg)	72.2±12.1	70.9±14.6	0.74
BMI (Kg/m ²)	24.4±3.9	23.2±2.5	0.30
Time from injury (month)	5.7±5.3	-	-

Table 5 - Participant characteristics. a. Mann-Whitney test

III.2.1. Data collection

Clinical assessment was performed for all subjects by two fellow clinicians experienced in orthopaedic surgery. Static knee stability was evaluated with the manual Lachman test, drawer test and pivot shift test. The International Knee Documentation Committee (IKDC) objective evaluation was also acquired to assess clinical outcomes (Table 6).

<i>Range of Motion</i>	
Flexion (mean±SD)	137.7° ±4.3
Extension (mean±SD)	0.5°±1.5
Recurvatum (mean±SD)	1.9° ±3.2
<i>Lachman test with Telos device</i>	
Mean	8.75mm±3.78
5-10 mm (nr. of patients)	22
10-15mm (nr. of patients)	5
>15 mm (nr. of patients)	3
<i>IKDC (nr. of patients)</i>	
Grade A	-
Grade B	-
Grade C	23
Grade D	7

Table 6 - Clinical characteristics of ACLD patients

Imaging investigations were done in our radiological department and included four series of radiographs: anteroposterior; lateral at 30° of flexion; axial views at 30° of flexion; and radiological Lachman test using a Telos device (150 N pressure was applied 6 cm below the hollow of the knee) (Fig 15).



Fig. 15 Lachman test using TELOS device.

Biomechanical data of walking were collected using The KneeKG™ System. The KneeKG™ System is composed of passive motion sensors fixed on a validated knee harness [95], an infrared motion capture system (Polaris Spectra camera, Northern Digital Inc.) and a computer equipped with the Knee3D™ software suite (Emovi, Inc.). The system measures and analyses the 3D positioning and movements of patient's knee (Fig 16) [5].

To reduce the skin motion artefact, the group developed a harness that is fixed quasi-statically on the thigh and the calf (Fig.17a) [120]. This harness was shown to be accurate in obtaining 3D kinematic data that could be used to evaluate ACL and ACL graft deformation in vivo [95, 120].

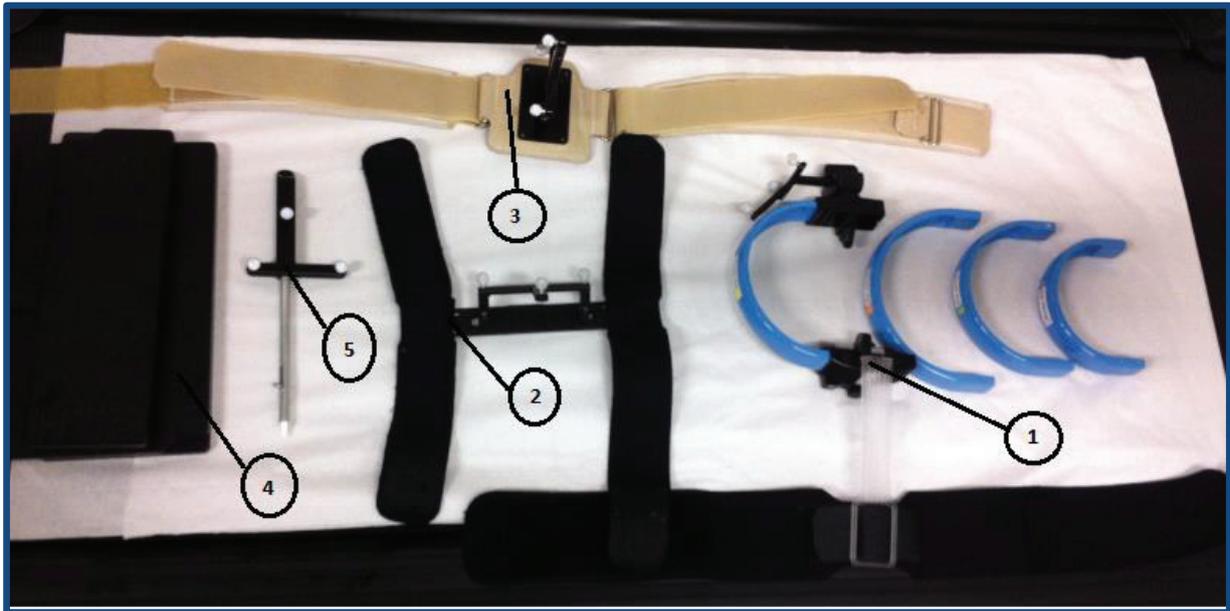


Fig. 16 The KneeKG™ system and its parts. 1. Femoral harness (4 interchangeable arches), 2. Tibial harness, 3. Sacroiliac belt, 4. Feet position guide, 5. Pointer, 6. Computer, 7. Cart, 8. Treadmill, 9. Video camera, 10. Reference body.

The inter-observer ICC (Intra-class Correlation Coefficient) for flexion/extension is 0.94, for adduction/abduction is 0.92 and for internal/external rotation is 0.89. The standard error of measurement (SEM) for flexion/extension is 0.5° , for adduction/abduction is 0.4° and for internal/ external rotation is 0.7° . The intra-observer ICC is 0.92 for flexion/extension, 0.94 for adduction/abduction and 0.88 for internal/external rotation. The SEM is 0.7° for flexion/extension, 0.5° for adduction/abduction and 0.8° for internal/external rotation [121]. The mean repeatability value ranged between 0.41° and 0.81° for rotation angles and between 0.8 and 2.2 mm for translation [122].

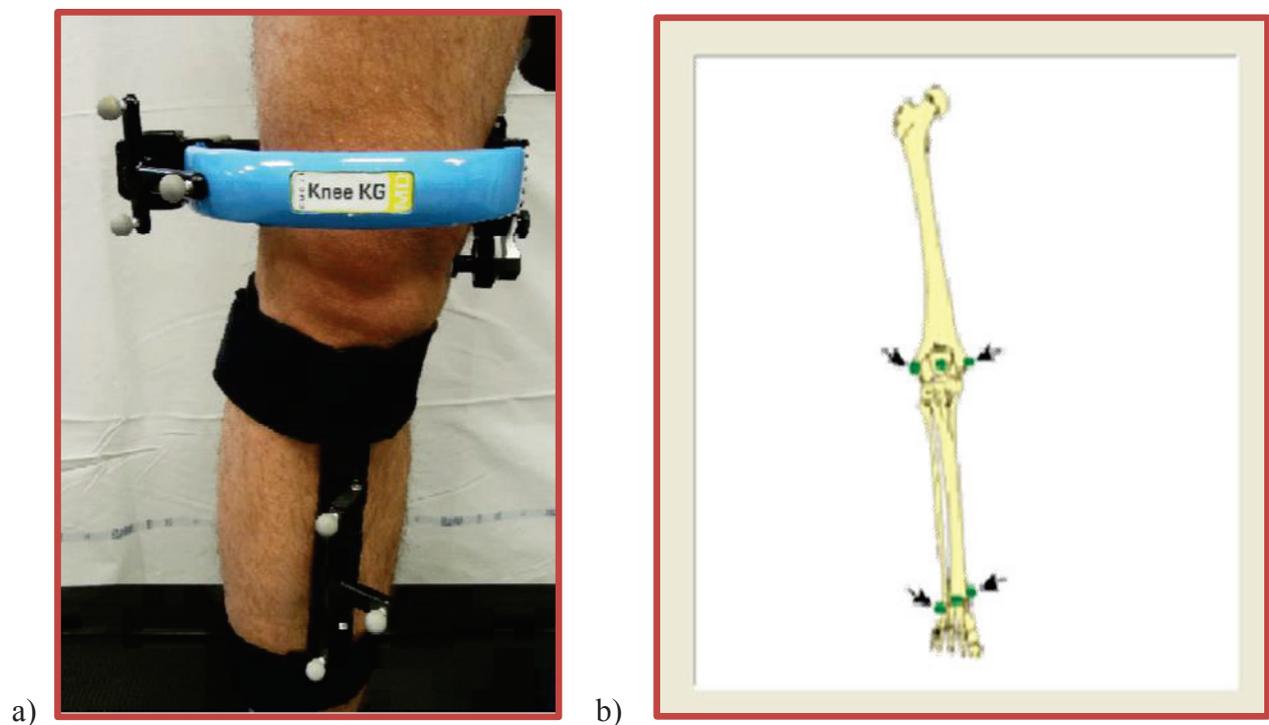


Fig. 17 a) Anterior view of a right knee fitted KneeKG tracker system. b) Identification of four anatomical landmarks.

After the installation of femoral, tibial and sacral trackers, the calibration procedure was done as described by Hagemester et al. [122]. This procedure includes two main parts: defining the joint centers and defining the joint system of axes based on predetermined postures.

The calibration begins with the identification of four anatomical sites: the medial malleolus, the lateral malleolus, the medial condyle and the lateral condyle (Fig 17b).

The 3D position of the femoral head was defined using a functional method (Fig 18). While the subject was performing a circumduction movement of the leg, the Knee3D™ recorded the motion of the sensors for a period of 5 s. The Knee3D™ then calculated the optimal point defining the center of the femoral head.

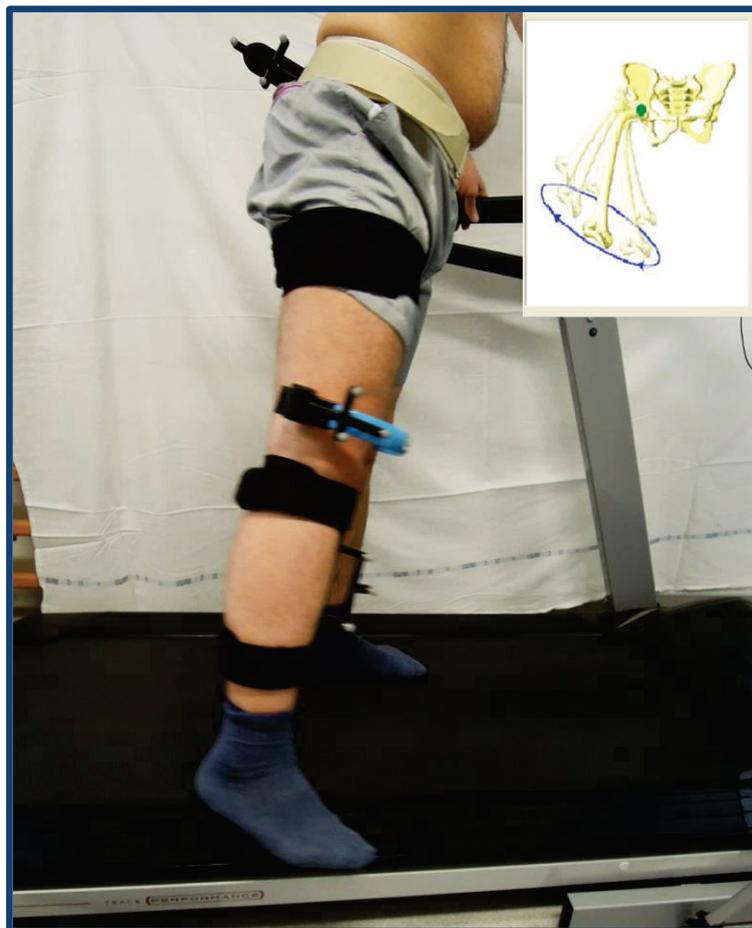


Fig. 18 Hip Joint Center definition

The next phase was defining the center of the knee in terms of 3D position (Fig 19). The subject has to put the leg in complete extension and then perform repetitive leg flexion/extension for a period of 10 s. Once the movement has been recorded, the Knee3D™

calculated a medio-lateral, middle axis for that movement. Based on this axis, the Knee3D™ then defined the knee center from the 3D positions of the medial and lateral condyles measured in the previous steps. The mid-point of both condyles was projected on this axis, thereby defining the knee center.

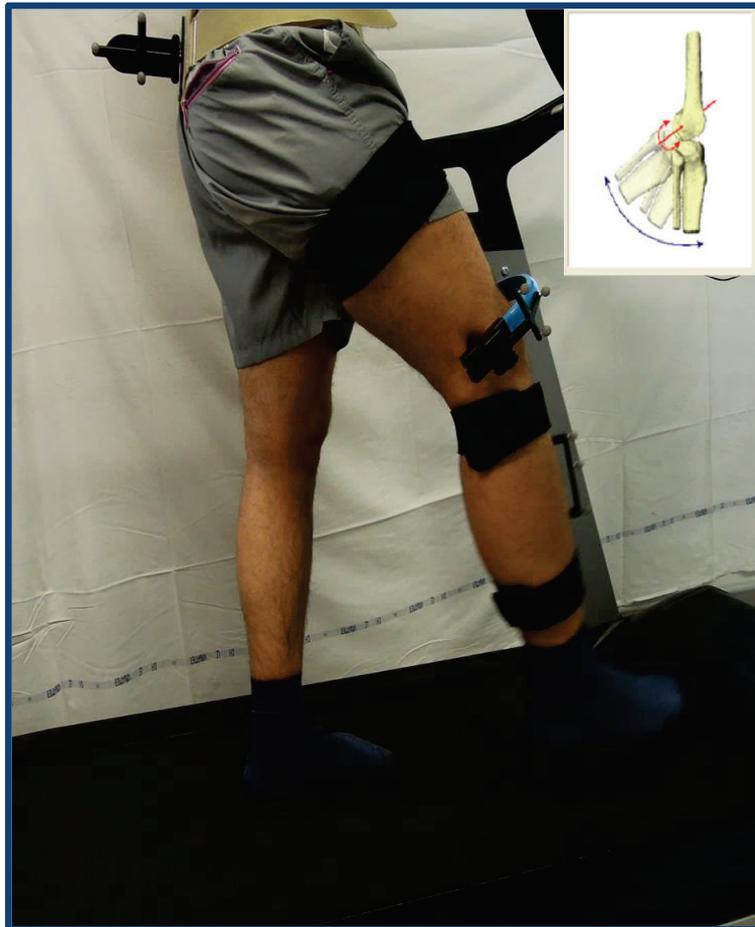


Fig. 19 Knee joint center definition

The final phase of calibration was the set of the neutral transverse rotation when the knee was determined to be at 0° of flexion during a slight flexion–hyperextension movement (Fig 20).

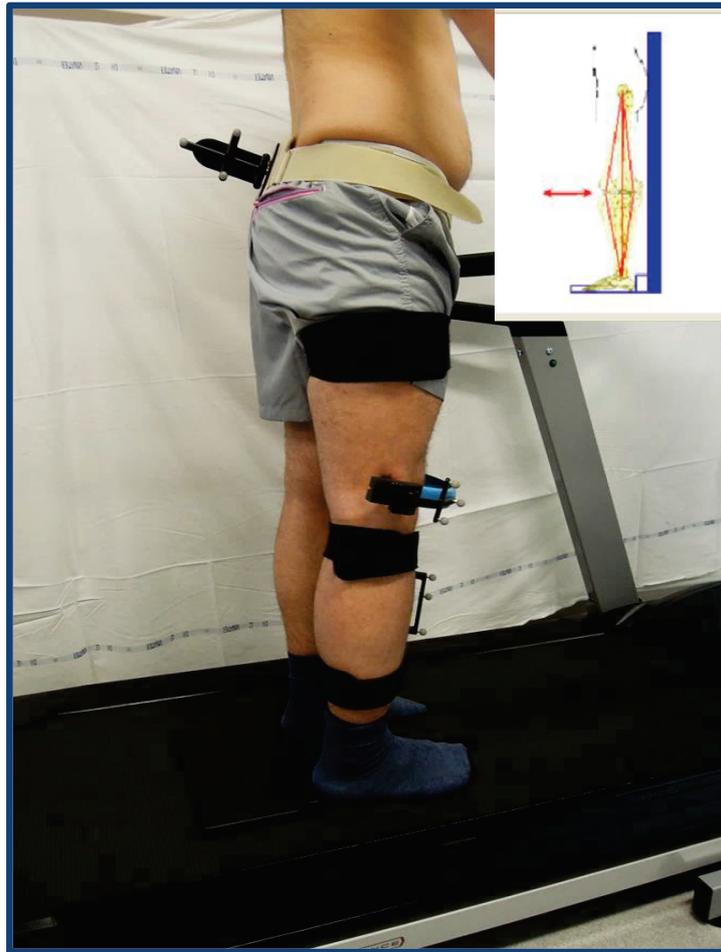


Fig. 20 Final step of axis definition: posture with knee in full extension

After calibration, kinematic data were collected during treadmill walking at a self-selected comfortable speed. To avoid the effect of footwear on lower limb biomechanics, all subjects were asked to walk barefoot. Before starting the trials collection, all patients walked 10 min to get used to walking on the treadmill.

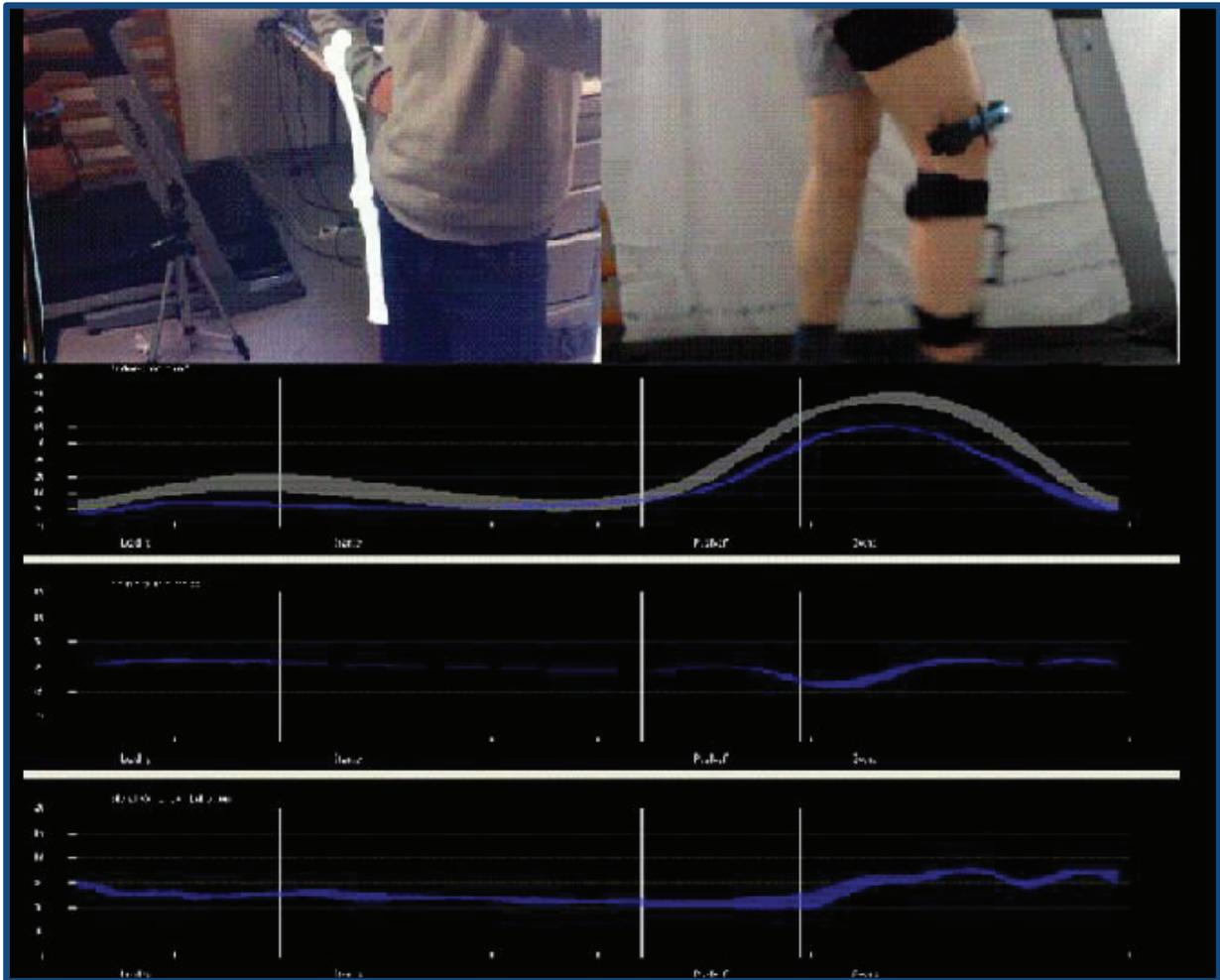


Fig. 21 Positions and orientations of the virtual models are set by the control unit in real time, allowing the user to see virtual bones in movement in accordance with patient's real bone movement.

Once calibration and measurements have been performed, the KneeKG™ computed the knee kinematical parameters throughout the 45-s recording. A database containing, for each participant, the 4 biomechanical patterns consisting of the three knee angles (flexion–extension, abduction–adduction and internal–external tibia rotation) and anterior-posterior tibial translation was created in Microsoft Excel 2010. In addition, graphical form and a report are also available (Fig.21,22).

	A	B	C	D	E	F	G	H	I	J	K	L	M	N	O	P	
8	Angle type	Data type	v1	v2	v3	v4	v5	v6	v7	v8	v9	v10	v11	v12	v13	v14	v15
9	Flex-ext	Avg	-3,26	-2,77	-2,29	-2,07	-1,62	-2,24	-2,68	-2,98	-3,15	-3,04	-2,95	-2,88	-3,05	-3,72	
10	Flex-ext	SD	1,22	1,06	1,27	1,05	1,03	1,13	0,78	0,61	1,18	1,38	1,57	1,47	1,58	1,52	
11	Flex-ext	Avg + 1SD	-2,04	-1,71	-1,03	-1,02	-0,59	-1,11	-1,89	-2,37	-1,97	-1,66	-1,39	-1,41	-1,47	-2,2	
12	Flex-ext	Avg - 1SD	-4,48	-3,82	-3,56	-3,12	-2,66	-3,37	-3,46	-3,6	-4,32	-4,42	-4,52	-4,35	-4,62	-5,23	
13		Init. Angle	-3,26														
14		Loading	-6,25														
15		Minimum	-12,58														
16		Maximum	57,26														
17		Amplitude	67														
18																	
19	Add-abd	Avg	-1,01	-1,2	-1,53	-1,75	-2,05	-2,32	-2,74	-3,01	-3,13	-3,08	-3,14	-3,18	-3,03	-2,9	
20	Add-abd	SD	0,31	0,42	0,47	0,54	0,57	0,62	0,54	0,46	0,39	0,35	0,42	0,43	0,29	0,24	
21	Add-abd	Avg + 1SD	-0,7	-0,78	-1,06	-1,21	-1,47	-1,7	-2,2	-2,55	-2,75	-2,73	-2,72	-2,75	-2,74	-2,66	
22	Add-abd	Avg - 1SD	-1,33	-1,62	-2	-2,3	-2,62	-2,94	-3,27	-3,47	-3,52	-3,42	-3,57	-3,6	-3,32	-3,13	
23		Init. Angle	-1,01														
24		Loading	-2,63														
25		Minimum	-10,22														
26		Maximum	-0,59														
27		Amplitude	9														
28																	
29	Rot-int-ext	Avg	-0,93	-0,61	-0,42	-0,02	0,91	1,48	2,61	3,17	3,16	3,01	3,15	3,61	4,45	4,93	
30	Rot-int-ext	SD	0,54	0,64	0,8	0,91	1,44	1,37	0,92	0,56	0,68	0,7	0,51	0,72	0,73	0,66	
31	Rot-int-ext	Avg + 1SD	-0,38	0,04	0,38	0,89	2,35	2,85	3,53	3,73	3,84	3,71	3,67	4,34	5,18	5,59	

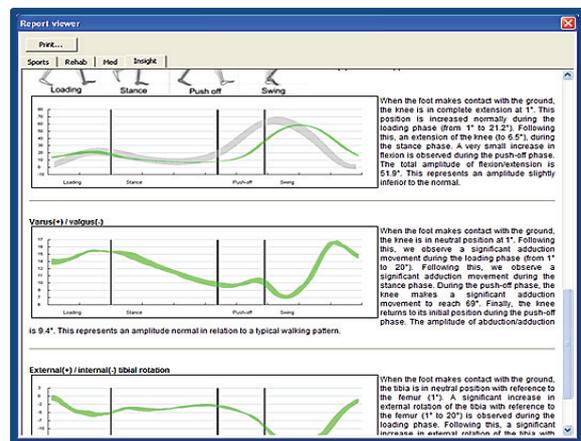
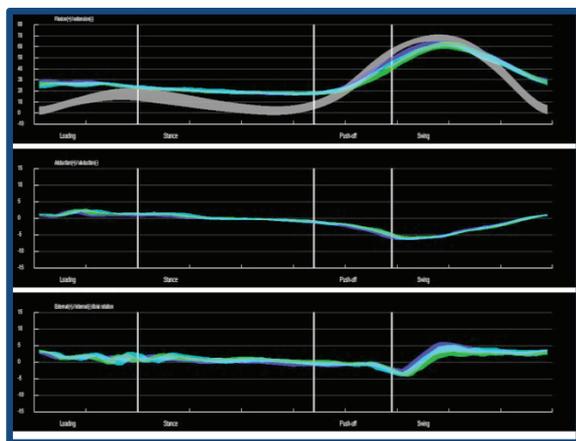


Fig. 22 Forms of results after the trial is finished

III.2.2. Statistical analysis

Sample size calculation ($\alpha < 0.05$; power 80 %; difference in means 1) showed that a minimum of 34 subjects were needed for this study. Participant characteristics (such as age, height, weight, BMI and gait speeds) were tested to determine whether parametric assumptions were met using the Levene test. Mann–Whitney test and two-tailed independent test were used for nonparametric and parametric variables, respectively. ANOVA test was utilized to compare kinematic parameters of ACL-deficient group and control group. All

statistical analyses were done using SPSS v 21 (SPSS Inc, Chicago, Illinois, USA), and significance level was set at 0.05.

III.3. Results

None of the participants' characteristics were statistically different between groups, except walking speed, where the ACLD group walked with lower speed than the control group. Table 7 summarizes the spatiotemporal and main kinematic data of the ACLD and control group.

Kinematic parameters	ACLD	Control Group	p value
Flexion-Extension- <i>SP</i> (mean)	13.2°±2.1	7.3° ±2.7	p<0.05
Range of Flexion-Extension	45.9°±7.3	54.2°±4.5	p<0.001
Internal-external rotation- <i>MSP</i> (mean)	-1.4°±0.2	0.2°±0.3	p<0.05
Range of internal-external rotation	9.0°±2.3	10.3°±2.5	p>0.05
A-P translation (mean)	-1.9mm±1.3	-2.2mm±1.5	p>0.05
Range of A-P translation	8.9mm±14.3	7.5mm±2.6	p>0.05
Adduction-Abduction (mean)	-0.4°±0.9	-0.9°±1.1	p>0.05
Range of adduction-abduction	5.8°±2.7	7.8°±3.2	p>0.05
Speed (Km/h)	2.2±0.4	2.5±0.4	p<0.05

Table 7 - Kinematic parameters of ACLD knees and control knees. SP=Stance phase, MSP=Mid-stance phase

III.3.1. Sagittal plane

The ACL-deficient patients showed a significant lower extension ($13.2^{\circ} \pm 2.1^{\circ}$) of the knee joint during entire stance phase compared to the control group ($7.3^{\circ} \pm 2.7^{\circ}$) (Fig. 23). More specifically, a statistically significant difference ($p < 0.05$) was identified between 1 and 61 % of the gait cycle under self-selected speed walking. Significant difference ($p < 0.001$) was

identified also in the range of flexion–extension between the two groups, where the ACLD group showed less range of flexion–extension than control group ($45.9^\circ \pm 7.3^\circ$ and $54.2^\circ \pm 4.5^\circ$, respectively).

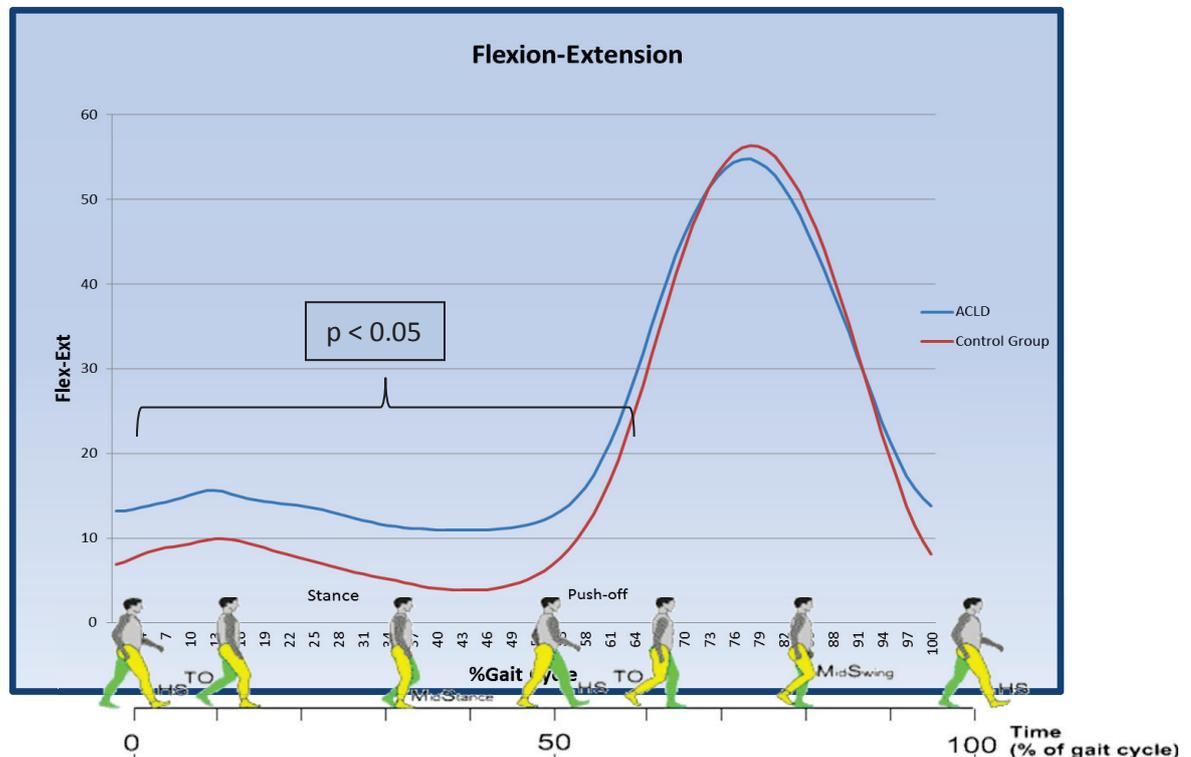


Fig. 23 Tibiofemoral kinematics of ACLD knees and healthy control knees-Sagittal plane

III.3.2. Axial plane

Even though in ACL-deficient group, tibia rotated more internally during entire stance phase and at the beginning of the swing phase, a significant difference ($p < 0.05$) was found only during the end of mid-stance phase, between 26 and 34 % of the gait cycle ($-1.4^\circ \pm 0.2^\circ$ and $0.2^\circ \pm 0.3^\circ$ for ACLD and control group, respectively), (Fig. 24).

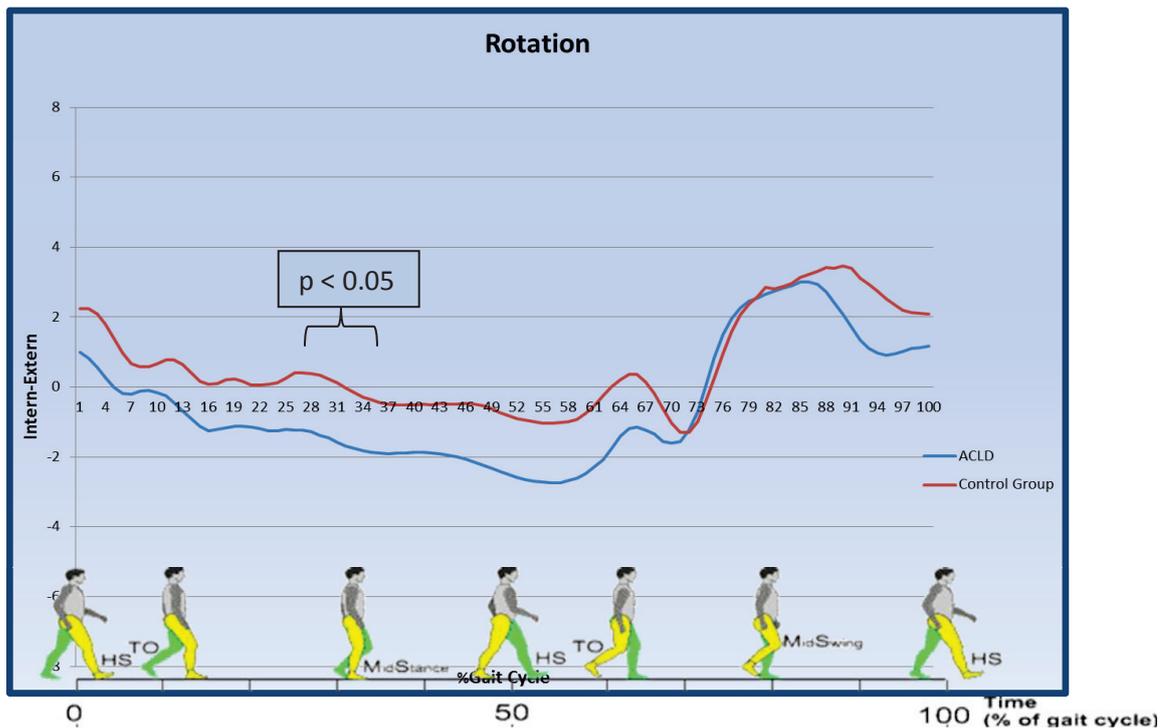


Fig. 24 Tibiofemoral kinematics of ACLD knees and healthy control knees-Axial plane

III.3.3. Anterior-posterior translation and coronal plane

Although in ACL-deficient patients during the entire gait cycle tibia was in anterior and adduction position compared to the control group, there were no statistically significant differences between the two groups (Fig 25, 26). Likewise, the range of A-P translation and adduction–abduction did not show any significant difference, but in contrast with flexion–extension and internal–external rotation, the range of A-P translation was higher at the ACLD group compared to control group (Table 7).

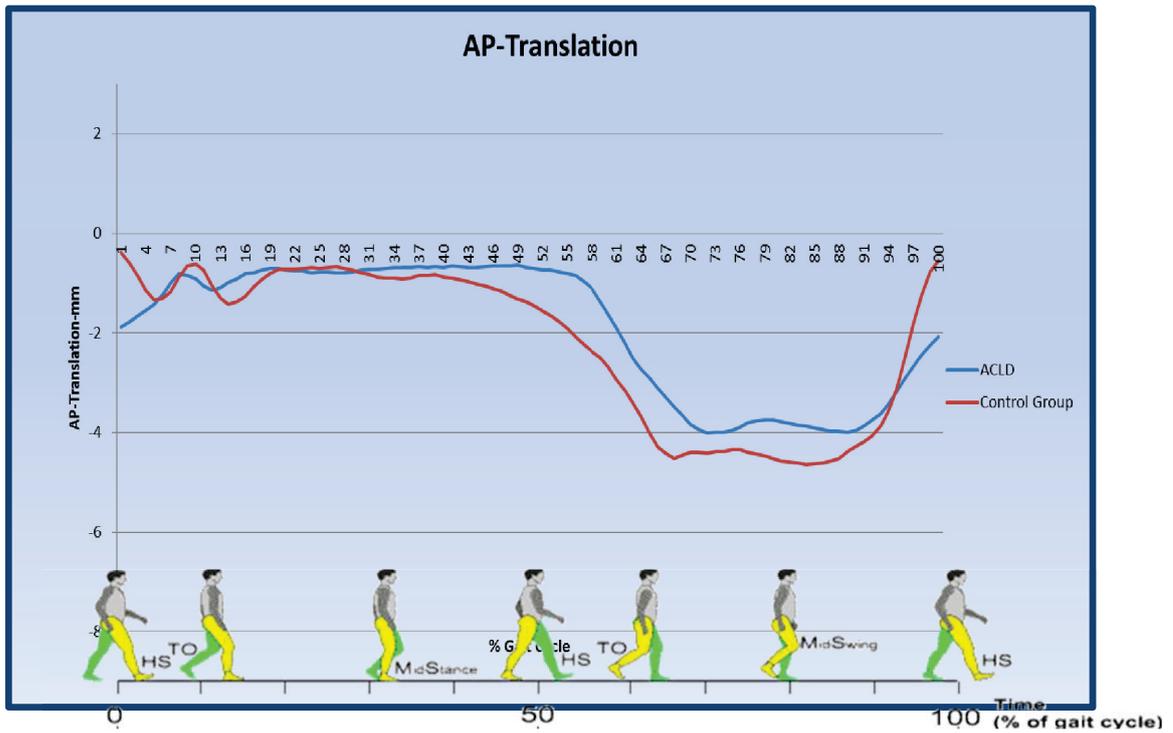


Fig. 25 Tibiofemoral kinematics of ACLD knees and healthy control knees-Anterior-posterior translation

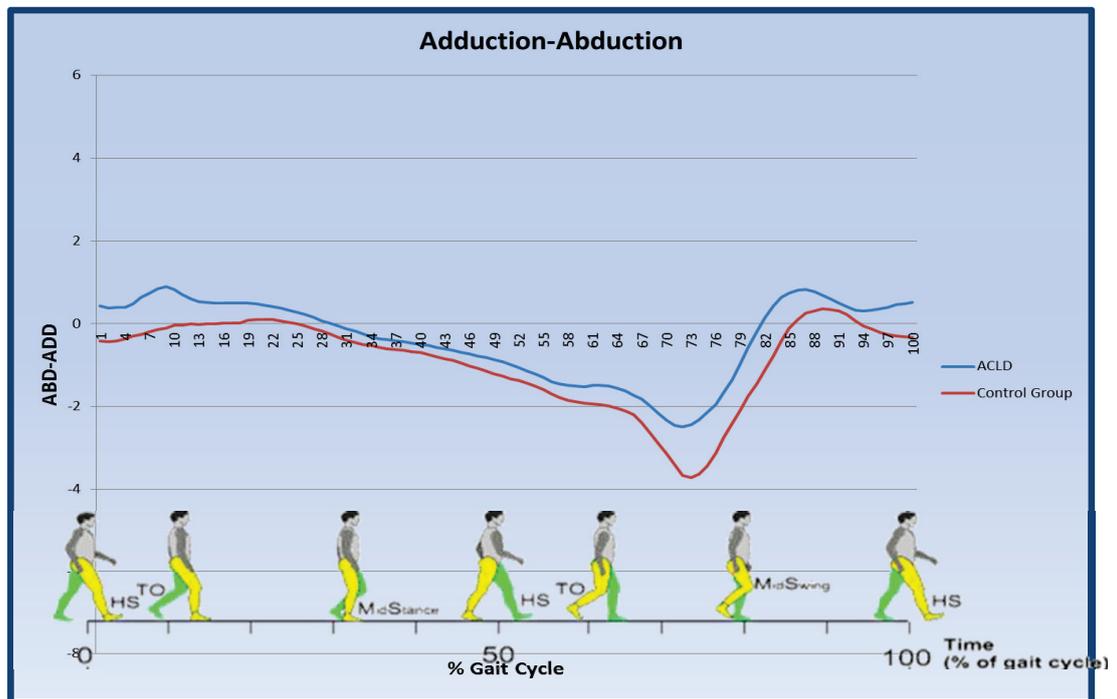


Fig. 26 Tibiofemoral kinematics of ACLD knees and healthy control knees-Coronal plane

III.4. Discussion

The most important finding of the present study was that ACLD knees adapted higher flexion strategy of walking to avoid excessive A-P tibial translation, and they also exhibited more internal tibial rotation.

Statistically significant differences were found in sagittal plane knee kinematics during the gait cycle, where the ACLD knees showed higher flexion of the knee joint throughout stance phase and lower knee range of motion. One major function of the ACL is to limit anterior tibial translation when the knee is close to extension, so the ACLD patients appeared to limit knee extension in order to reduce the functional absence of ACL. Gao and Zheng [78] explained that ACLD knees tend to stay in flexion before the knee reaches maximum extension in order to accomplish a less abrupt weight shift.

The finding of the extension deficit was consistent with the study of Hurd and Snyder-Mackler [123] who explained that this higher angle observed at ACLD patients could potentially be a result of higher hamstring activation. However, there are studies that show less flexion throughout mid-stance phase of gait [101, 124]. Because of the variability in kinematic patterns that exist between the group with ACLD knees compared to normal knees [125], the number of patients in the last two studies [101, 124] (16 and 18 ACLD knees, respectively) might be the cause of the difference between our results and theirs.

Even though the movements in sagittal plane have been studied extensively, less is known regarding the axial and coronal plane. In the present study, in axial plane, although in ACL-deficient group tibia rotated more internally during entire stance phase and at the beginning of the swing phase, a significant difference was noted only during the end of mid-stance phase. This significant increase of internal tibial rotation happens during the phase when the knee is in near maximum extension. ACL, as a primary restraint in anterior-posterior translation and

as a secondary restraint in rotation stability of the knee, elongates maximally during mid-stance phase [126, 127], thus its rupture could change the kinematics of this part of the gait cycle. While Georgoulis et al. [76] have reported that tibia rotated more internally during the swing phase, Fuentes et al. [100] have found less internal rotation knee joint moment during terminal stance phase. This discrepancy could arise from the difference in the time since injury, (the period between the occurrence of injury and the time of kinematic evaluation) (Table 1). The average time of injury in our study was $5.7 \text{ months} \pm 5.3$, compared to Fuentes et al. [14] study where the average time of injury was 22 months (from 5 to 78 months). Additionally, in Andriacchi and Durby work [112] where less external rotation in ACLD knees was found, the time of injury was 127 ± 142 months (from 3 to 400 months). It has been reported that the time since injury can influence the development of gait alterations following ACL injury [128]; thus, we may say that ACLD patients could adapt different gait compensation strategies. The increase in internal tibial rotation when the knee is in its most extended position could also be because the hamstring muscle may not be able to prevent abnormal rotation [129]. Ciccoti et al. [130] reported a significant increase in rectus femoris muscle activity among the patients with ACL rupture in extension moment. Thus, the increased activity of the rectus femoris muscle could probably result in increased internal tibial rotation. This axial position can result in increased stress in the cartilage and accelerate the loss of cartilage relative to the intact knee, especially in medial compartment [131]. In addition, even though in this study there was no statistically significant difference in coronal plane, the tibia of ACLD knees remained slightly in adduction, which could increase the probability of cartilage wear in medial compartment. Seon et al. [132] reported that in ACLD knees, the medial compartment of the joint is more vulnerable to cartilage degeneration and osteoarthritis development. This is in agreement with the kinematic abnormalities that we have observed in this study.

Even though in ACLD knees tibia remained slightly more anterior for the most part of gait cycle, there was no significant difference in knee joint translation compared to control group. While it is well known that ACLD could lead to excessive anterior tibial translation during the passive knee laxity test [133, 134], the excessive knee laxity in ACLD knees could be less prominent during active movement. Non-loaded extension exercises in a seated and supine position have showed increase of anterior-posterior translation at different knee angles [135, 136]. Nevertheless, compressive loading of the knee, such as weight bearing, has been shown to reduce A-P laxity and stiffen the tibiofemoral joint in comparison with the non-weightbearing condition [137]. Andriacchi et al. [112] have observed a significant difference in anteroposterior translation only during the terminal swing phase. However, there are studies that have shown that the amount of passive laxity was unrelated to gait patterns [120, 138]. This compensation strategy could be either because of a stronger co-contraction of muscles to stabilize the joint or through anterior tibial subluxation, where forward motion of the tibia is limited by other surrounding passive structures [133]. Several studies have suggested that hamstring muscles can compensate for instability of the ACLD knee during functional activity [130, 139].

The increase in hamstring muscle contraction produces an additional knee flexion moment that must be balanced by a corresponding increase in a quadriceps muscle force. This increase of co-contraction of antagonist muscles would increase the loading in regions that may not be adapted to such loading conditions [140]. Furthermore, this compensatory strategy to aid in the knee joint laxity provides a possible explanation for post-traumatic cartilage degeneration and early OA development in the knee.

This study has some limitations. First, the artefacts from soft tissue movements could be considered as a limitation of this study. Nevertheless, Sati and Larouche [95] have shown that the harnesses that are fixed quasi-statically on the knee and on the tibia reduce the skin

motion artefacts. Additionally, this harness was shown to be accurate in assessing 3D kinematic data that could be used to evaluate ACL and ACL graft deformation in vivo [95, 120]. Second, we have not compared the data from the injured knee against data from contralateral knee. Instead, we have opted for a control group of healthy subjects. Thus, variations in individual walking could have potentially affected our results. However, Branch et al. [4] observed that the contralateral knees in ACLD patients demonstrated greater rotational laxity compared to healthy subjects.

The findings in this study provide clinically relevant information regarding the effects of ACL deficiency on gait parameters. The ACLD knees adapted functionally to protect excessive anterior–posterior translation but failed to avoid rotational instability. These kinematic alterations may result in changes of load distribution and cartilage thinning [140]. Thus, these results may be taken into consideration for planning the ACL reconstruction and rehabilitation. We consider that further work is needed in the future in order to assess whether current ACL reconstruction techniques restore the normal knee kinematics for daily activities.

III.5. Conclusion

In this study using 3D, in vivo motion analysis, significant alterations in knee joint kinematics in ACLD patients were identified. These alterations were manifested as a higher flexion gait strategy and excessive internal tibial rotation during walking. However, no significant difference was shown in anterior-posterior translation. The increased flexion angle was accompanied by a restoration of anterior–posterior stability, but it did not restore normal rotation stability. These kinematic changes could lead to abnormal tibiofemoral cartilage contact during daily activities, thus may represent a biomechanical mechanism of joint degeneration after ACL injury.

Publication

Gait changes of the ACL-deficient knee 3D kinematic assessment

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Gait changes of the ACL-deficient knee 3D kinematic assessment

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Abstract

Purpose Static, one-dimensional testing cannot predict the behaviour of the anterior cruciate ligament (ACL)-deficient knee under realistic loading conditions. Currently, the most widely accepted method for assessing joint movement patterns is gait analysis. The purpose of the study was in vivo evaluation of the behaviour of the anterior cruciate ligament-deficient (ACL D) knees during walking, using 3D, real-time assessment tool.

Methods Biomechanical data were collected prospectively on 30 patients with ACL rupture and 15 healthy subjects as a control group, with KneeKg™ System. Kinematic data were recorded in vivo during treadmill walking at self-selected speed. Flexion/extension, abduction/adduction, anterior/posterior tibial translation and external/internal tibial rotation were compared between groups.

Results The ACL D patients showed a significant lower extension of the knee joint during stance phase ($p < 0.05$; $13.2^\circ \pm 2.1^\circ$ and $7.3^\circ \pm 2.7^\circ$, for ACL D and control group, respectively). A significant difference in tibial rotation angle was found in ACL D knees compared to control knees

($p < 0.05$). The patients with ACL D rotated the tibia more internally ($-1.4^\circ \pm 0.2^\circ$) during the mid-stance phase, than control group ($0.2^\circ \pm 0.3^\circ$). There was no significant difference in anteroposterior translation and adduction–abduction angles.

Conclusion Significant alterations of joint kinematics in the ACL D knee were revealed in this study by manifesting a higher flexion gait strategy and excessive internal tibial rotation during walking that could result in a more rapid cartilage thinning throughout the knee. The preoperative data obtained in this study will be useful to understand the post-ACL reconstruction kinematic behaviour of the knee.

Clinical relevance The findings in this study indicate that ACL D knee may adapt functionally to prevent excessive anterior–posterior translation but they fail to avoid rotational instability.

Keywords Knee · Anterior cruciate ligament deficient · Kinematic · 3D assessment

Introduction

Knowledge of in vivo movement of the knee is important for understanding normal function as well as addressing clinical problems, including instability and function after anterior cruciate ligament (ACL) injury. There are numerous studies that have provided information on biomechanical changes in the anterior cruciate ligament-deficient (ACL D) knees [2, 7, 15, 29]. In vitro studies using cadavers [28, 34] have provided some insight into the kinematical behaviour of the ACL D knees under controlled conditions. However, these studies are unable to accurately simulate the effects of weight bearing and muscles contraction on the joint kinematics.

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The most commonly used method for assessing dynamic movement is the skin marker-based motion capture system. Nevertheless, studies using skin marker-based motion capture system have reported soft tissue artefacts, especially in knee joint translation and internal/external rotation angles [6, 44]. According to Cappozzo et al. [10], the marker displacement with respect to the underlying bone, as a result of skin movement, ranges from few millimetres up to 40 mm. On the other hand, markers that are attached to pins drilled into the bone are potentially more accurate, but are invasive.

Despite that there are several devices currently available to assess the knee kinematics [21, 30, 31, 35, 37, 40], the 3D biomechanical changes caused by ACL injury and possible compensations adopted by the patients following the injury are still not clearly understood. Thus, establishing an objective evaluation of the kinematics of the knee in a clinically feasible way is critical in extensive evaluation of the ACL function and a valuable feedback for further progress of ACL treatment.

The purpose of the study was the *in vivo* evaluation of the ACL-deficient knee behaviour during all phases of gait, using a new 3D, quasi-static real-time assessment tool. It was hypothesized that ACLD knees would exhibit altered joint kinematics. In order to test the hypothesis, we examined 3D knee joint kinematics (flexion–extension; internal–external rotation; adduction–abduction; anterior–posterior tibial translation) during walking at self-selected speed, in two subject groups: ACLD knees and healthy controls with bilateral ACL-intact knees.

Materials and methods

This study was prospectively conducted from January 2011 to June 2011, in the facilities of the biomechanical laboratory at our clinical centre. Patients who were planned to be operated for ACL reconstruction were selected for kinematic analysis. The ACL rupture was diagnosed by clinical examination, MRI and confirmed during the surgery by arthroscopy. Patients with unilateral ACL rupture and healthy contralateral knees (that had never suffered of any kind of orthopaedic or neurological condition) were included in the study. Patients with meniscal injury where partial meniscectomy or repair was feasible, and patients with grade I or II medial collateral injury were also part of the study. All patients who participated in this study were non-copers (unable to return to their premorbid level of sports play or activity). On the other hand, patients who had pain, subtotal or total meniscectomy, concomitant PCL injury, knee joint movement restriction, full thickness cartilage defect >1 cm² and had previous history of any surgery in both knees were excluded from the study. Thirty patients

Table 1 Participant characteristics

Parameters	ACLD group	Control group	<i>p</i> value
Age (years)	29.8 ± 8.9	29.3 ± 9.7	n.s. ^a
Height (cm)	172 ± 7.8	173.1 ± 10.2	n.s.
Weight (kg)	72.2 ± 12.1	70.9 ± 14.6	n.s.
BMI (Kg/m ²)	24.4 ± 3.9	23.2 ± 2.5	n.s.
Time from injury (month)	5.7 ± 5.3	–	–

^a Mann–Whitney test

Table 2 Clinical characteristics of ACLD patients

Range of Motion	
Flexion (mean ± SD)	137.7° ± 4.3°
Extension (mean ± SD)	0.5° ± 1.5°
Recurvatum (mean ± SD)	1.9° ± 3.2°
Lachman test with Telos device	
Mean	8.75 mm ± 3.7
5–10 mm (nr. of patients)	22
10–15 mm (nr. of patients)	5
>15 mm (nr. of patients)	3
IKDC (nr. of patients)	
Grade A	–
Grade B	–
Grade C	23
Grade D	7

were eligible for inclusion in this study; 40 patients were excluded because they did not meet the inclusion criteria.

A control group consisting of 15 participants of similar age, who had no history of musculoskeletal injury or surgery in the lower extremities and exhibited no measurable ligamentous instability on clinical examination (pivot shift, Lachman and drawer test), was selected (Table 1).

Data collection

Clinical assessment was performed for all subjects by two fellow clinicians experienced in orthopaedic surgery. Static knee stability was evaluated with the manual Lachman test, drawer test and pivot shift test. The International Knee Documentation Committee (IKDC) objective evaluation was also acquired to assess clinical outcomes (Table 2).

Imaging investigations were done in our radiological department and included four series of radiographs: anteroposterior; lateral at 30° of flexion; axial views at 30° of flexion; and radiological Lachman test using a Telos device (150 N pressure was applied 6 cm below the hollow of the knee).

Biomechanical data of walking were collected using The KneeKg™ System. The KneeKg™ System is composed of

passive motion sensors fixed on a validated knee harness [38], an infrared motion capture system (Polaris Spectra camera, Northern Digital Inc.) and a computer equipped with the Knee3D™ software suite (Emovi, Inc.). The system measures and analyses the 3D positioning and movements of patient's knee [41].

To reduce the skin motion artefact, the group developed a harness that is fixed quasi-statically on the thigh and the calf [39]. This harness was shown to be accurate in obtaining 3D kinematic data that could be used to evaluate ACL and ACL graft deformation in vivo [38, 39].

The inter-observer ICC (*Intra-class Correlation Coefficient*) for flexion/extension is 0.94, for adduction/abduction is 0.92 and for internal/external rotation is 0.89. The standard error of measurement (SEM) for flexion/extension is 0.5°, for adduction/abduction is 0.4° and for internal/external rotation is 0.7°. The intra-observer ICC is 0.92 for flexion/extension, 0.94 for adduction/abduction and 0.88 for internal/external rotation. The SEM is 0.7° for flexion/extension, 0.5° for adduction/abduction and 0.8° for internal/external rotation [27]. The mean repeatability value ranged between 0.41° and 0.81° for rotation angles and between 0.8 and 2.2 mm for translation [19].

After the installation of femoral, tibial and sacral trackers, the calibration procedure was done as described by Hagemester et al. [19]. This procedure includes two main parts: defining the joint centres and defining the joint system of axes based on predetermined postures.

The first phase of calibration was the identification of four anatomical sites: the medial malleolus, the lateral malleolus, the medial condyle and the lateral condyle.

In the second phase, the 3D position of the femoral head was defined using a functional method. While the subject was performing a circumduction movement of the leg, the Knee3D™ recorded the motion of the sensors for a period of 5 s. The Knee3D™ then calculated the optimal point defining the centre of the femoral head.

The next phase was defining the centre of the knee in terms of 3D position. The subject has to put the leg in complete extension and then perform repetitive leg flexion/extension for a period of 10 s. Once the movement has been recorded, the Knee3D™ calculated a medio-lateral, middle axis for that movement. Based on this axis, the Knee3D™ then defined the knee centre from the 3D positions of the medial and lateral condyles measured in the previous steps. The mid-point of both condyles was projected on this axis, thereby defining the knee centre. The final phase of calibration was the set of the neutral transverse rotation when the knee was determined to be at 0° of flexion during a slight flexion-hyperextension movement.

After calibration, kinematic data were collected during treadmill walking at a self-selected comfortable speed. To avoid the effect of footwear on lower limb biomechanics,

all subjects were asked to walk barefoot. Before starting the trials collection, all patients walked 10 min to get used to walking on the treadmill.

Once calibration and measurements have been performed, the KneeKG™ computed the knee kinematical parameters throughout the 45-s recording. A database containing, for each participant, the 4 biomechanical patterns consisting of the three knee angles (flexion–extension, abduction–adduction and internal–external tibia rotation) and anteroposterior tibial translation was created in Microsoft Excel 2010.

Statistical analysis

Sample size calculation ($\alpha < 0.05$; power 80 %; difference in means 1) showed that minimum of 34 subjects were needed for this study. Participant characteristics (such as age, height, weight, BMI and gait speeds) were tested to determine whether parametric assumptions were met using the Levene test. Mann–Whitney test and two-tailed independent test were used for nonparametric and parametric variables, respectively. ANOVA test was utilized to compare kinematic parameters of ACL-deficient group and control group. All statistical analyses were done using SPSS v 21 (SPSS Inc, Chicago, Illinois, USA), and significance level was set at 0.05.

Results

None of the participants' characteristics were statistically different between groups, except walking speed, where the ACLD group walked with lower speed than the control group. Table 3 summarizes the spatiotemporal and main kinematic data of the ACLD and control group.

Sagittal plane

The ACL-deficient patients showed a significant lower extension ($13.2^\circ \pm 2.1^\circ$) of the knee joint during entire stance phase compared to the control group ($7.3^\circ \pm 2.7^\circ$) (Fig. 1a). More specifically, a statistically significant difference ($p < 0.05$) was identified between 1 and 61 % of the gait cycle under self-selected speed walking. Significant difference ($p < 0.001$) was identified also in the range of flexion–extension between the two groups, where the ACLD group showed less range of flexion–extension than control group ($45.9^\circ \pm 7.3^\circ$ and $54.2^\circ \pm 4.5^\circ$, respectively).

Axial plane

Even though in ACL-deficient group, tibia rotated more internally during entire stance phase and at the beginning of the swing phase, a significant difference ($p < 0.05$) was found only during the end of mid-stance phase, between 26

Table 3 Kinematic parameters of ACLD knees and control knees

Kinematic parameters	ACLD	Control group	p value
Flexion–extension-SP (mean)	13.2° ± 2.1°	7.3° ± 2.7°	p < 0.05
Range of flexion–extension	45.9° ± 7.3°	54.2° ± 4.5°	p < 0.001
Internal–external rotation-MSP (mean)	-1.4° ± 0.2°	0.2° ± 0.3°	p < 0.05
Range of internal–external rotation	9.0° ± 2.3°	10.3° ± 2.5°	n.s
A-P translation (mean)	-1.9 mm ± 1.3	-2.2 mm ± 1.5	n.s
Range of A-P translation	8.9 mm ± 14.3	7.5 mm ± 2.6	n.s
Adduction–abduction (mean)	-0.4° ± 0.9°	-0.9° ± 1.1°	n.s
Range of adduction–abduction	5.8° ± 2.7°	7.8° ± 3.2°	n.s
Speed (Km/h)	2.2 ± 0.4	2.5 ± 0.4	p < 0.05

SP Stance phase, MSP Mid-stance phase

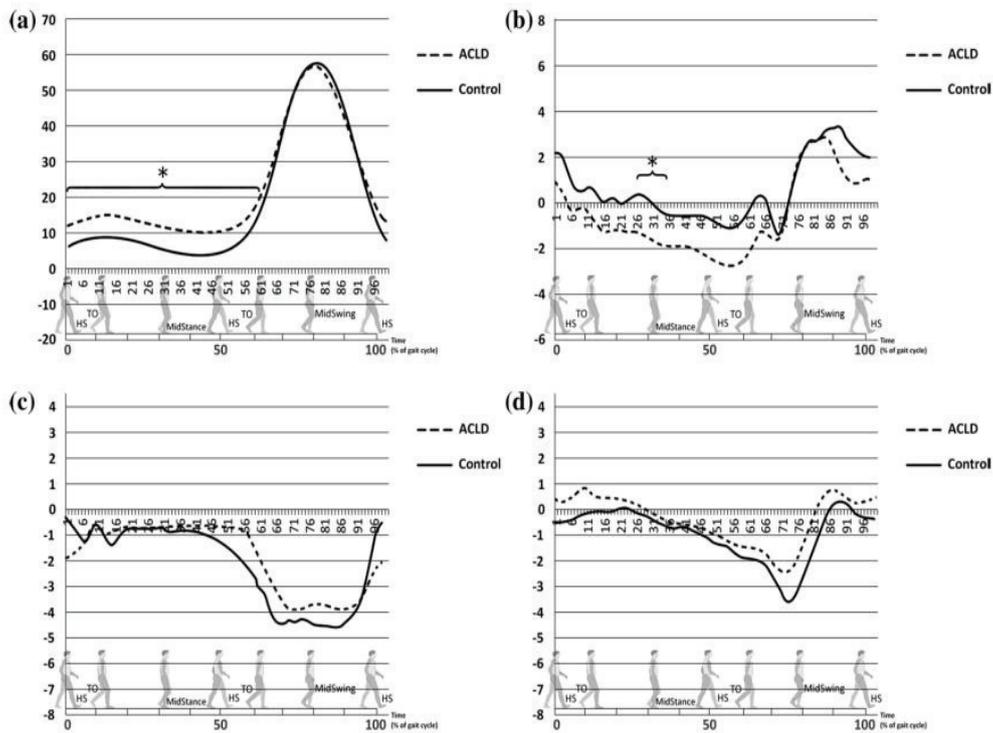


Fig. 1 Tibiofemoral kinematics of ACLD knees and healthy control knees. **a** Sagittal plane. **b** Axial plane. **c** Anteroposterior tibial translation. **d** Coronal plane; (*) Indicates significant difference between ACLD knees and control knees

and 34 % of the gait cycle ($-1.4 \pm 0.2^\circ$ and $0.2^\circ \pm 0.3^\circ$ for ACLD and control group, respectively), (Fig. 1b).

Anteroposterior translation and coronal plane

Although in ACL-deficient patients during the entire gait cycle tibia was in anterior and adduction position compared to the control group, there were no statistically significant differences between the two groups (Fig. 1c,d). Likewise, the range of A-P translation and adduction–abduction did

not show any significant difference, but in contrast with flexion–extension and internal–external rotation, the range of A-P translation was higher at the ACLD group compared to control group (Table 3).

Discussion

The most important finding of the present study was that ACLD knees adapted higher flexion strategy of walking to

avoid excessive A-P tibial translation, and they also exhibited more internal tibial rotation.

Statistically significant differences were found in sagittal plane knee kinematics during the gait cycle, where the ACLD knees showed higher flexion of the knee joint throughout stance phase and lower knee range of motion. One major function of the ACL is to limit anterior tibial translation when the knee is close to extension, so the ACLD patients appeared to limit knee extension in order to reduce the functional absence of ACL. Gao and Zheng [16] explained that ACLD knees tend to stay in flexion before the knee reaches maximum extension in order to accomplish a less abrupt weight shift.

The finding of the extension deficit was consistent with the study of Hurd and Snyder-Mackler [20] who explained that this higher angle observed at ACLD patients could potentially be a result of higher hamstring activation. Furthermore, there are studies that show less flexion throughout mid-stance phase of gait [7, 23]. Because of the variability in kinematic patterns that exist between the group with ACLD knees compared to normal knees [13], the number of patients in the last two studies [7, 23] (16 and 18 ACLD knees, respectively) might be the cause of the difference between our results and theirs.

Even though the movements in sagittal plane have been studied extensively, less is known regarding the axial and coronal plane. In the present study, in axial plane, although in ACL-deficient group tibia rotated more internally during entire stance phase and at the beginning of the swing phase, a significant difference was noted only during the end of mid-stance phase. This significant increase of internal tibial rotation happens during the phase when the knee is in near maximum extension. ACL, as a primary restraint in anteroposterior translation and as a secondary restraint in rotation stability of the knee, elongates maximally during mid-stance phase [45, 48], thus its rupture could change the kinematics of this part of the gait cycle. While Georgoulis et al. [17] have reported that tibia rotated more internally during the swing phase, Fuentes et al. [14] have found less internal rotation knee joint moment during terminal stance phase. This discrepancy could arise from the difference in the time since injury, (the period between the occurrence of injury and the time of kinematic evaluation) (Table 1). The average time of injury in our study was 5.7 months \pm 5.3, compared to Fuentes et al. [14] study where the average time of injury was 22 months (from 5 to 78 months). Additionally, in Andriacchi and Durby work [2] where less external rotation in ACLD knees was found, the time of injury was 127 \pm 142 months (from 3 to 400 months). It has been reported that the time since injury can influence the development of gait alternations following ACL injury [47]; thus, we may say that ACLD patients could adapt different gait compensation strategies. The increase in internal

tibial rotation when the knee is in its most extended position could also be because the hamstring muscle may not be able to prevent abnormal rotation [12]. Ciccoti et al. [11] reported a significant increase in rectus femoris muscle activity among the patients with ACL rupture in extension moment. Thus, the increased activity of the rectus femoris muscle could probably result in increased internal tibial rotation. This axial position can result in increased stress in the cartilage and accelerate the loss of cartilage relative to the intact knee, especially in medial compartment [1]. In addition, even though in this study there was no statistically significant difference in coronal plane, the tibia of ACLD knees remained slightly in adduction, which could increase the probability of cartilage wear in medial compartment. Seon et al. [42] reported that in ACLD knees, the medial compartment of the joint is more vulnerable to cartilage degeneration and osteoarthritis development. This is in agreement with the kinematic abnormalities that we have observed in this study.

Even though in ACLD knees tibia remained slightly more anterior for the most part of gait cycle, there was no significant difference in knee joint translation compared to control group. While it is well known that ACLD could lead to excessive anterior tibial translation during the passive knee laxity test [5, 8], the excessive knee laxity in ACLD knees could be less prominent during active movement. Non-loaded extension exercises in a seated and supine position have showed increase of anteroposterior translation at different knee angles [24, 26, 32]. Nevertheless, compressive loading of the knee, such as weight bearing, has been shown to reduce A-P laxity and stiffen the tibiofemoral joint in comparison with the non-weightbearing condition [33, 46]. Andriacchi et al. [2] have observed a significant difference in anteroposterior translation only during the terminal swing phase. However, there are studies that have shown that the amount of passive laxity was unrelated to gait patterns [18, 36, 38]. This compensation strategy could be either because of a stronger co-contraction of muscles to stabilize the joint or through anterior tibial subluxation, where forward motion of the tibia is limited by other surrounding passive structures [8]. Several studies have suggested that hamstring muscles can compensate for instability of the ACLD knee during functional activity [11, 22, 43].

The increase in hamstring muscle contraction produces an additional knee flexion moment that must be balanced by a corresponding increase in a quadriceps muscle force [33]. This increase of co-contraction of antagonist muscles would increase the loading in regions that may not be adapted to such loading conditions [3, 4, 25]. Furthermore, this compensatory strategy to aid in the knee joint laxity provides a possible explanation for post-traumatic cartilage degeneration and early OA development in the knee.

This study has some limitations. First, the artefacts from soft tissue movements could be considered as a limitation of this study. Nevertheless, Sati and Larouche [38] have shown that the harnesses that are fixed quasi-statically on the knee and on the tibia reduce the skin motion artefacts. Additionally, this harness was shown to be accurate in assessing 3D kinematic data that could be used to evaluate ACL and ACL graft deformation in vivo [38, 39]. Second, we have not compared the data from the injured knee against data from contralateral knee. Instead, we have opted for a control group of healthy subjects. Thus, variations in individual walking could have potentially affected our results. However, Branch et al. [9] observed that the contralateral knees in ACLD patients demonstrated greater rotational laxity compared to healthy subjects.

The findings in this study provide clinically relevant information regarding the effects of ACL deficiency on gait parameters. The ACLD knees adapted functionally to protect excessive anterior–posterior translation but failed to avoid rotational instability. These kinematic alterations may result in changes of load distribution and cartilage thinning [14]. Thus, these results may be taken into consideration for planning the ACL reconstruction and rehabilitation. We consider that further work is needed in the future in order to assess whether current ACL reconstruction techniques restore the normal knee kinematics for daily activities.

Conclusion

In this study using 3D, in vivo motion analysis, significant alterations in knee joint kinematics in ACLD patients were identified. These alterations were manifested as a higher flexion gait strategy and excessive internal tibial rotation during walking. However, no significant difference was shown in anteroposterior translation. The increased flexion angle was accompanied by a restoration of anterior–posterior stability, but it did not restore normal rotation stability. These kinematic changes could lead to abnormal tibiofemoral cartilage contact during daily activities, thus may represent a biomechanical mechanism of joint degeneration after ACL injury.

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Presentation

**In vivo kinematic alterations of the ACL-deficient knee -
3D assessment**

Abstract:

Introduction:

Following anterior cruciate ligament (ACL) rupture, the knee joint kinematics alters including anterior tibial displacement, and internal tibial rotation. Static, one-dimensional testing cannot predict the behavior of the ACL-deficient knee under realistic loading conditions. Currently, the most widely accepted method for assessing joint movement patterns is gait analysis. Thus, the objective of this study was the in vivo evaluation of behavior of the ACLD knee during walking, using a 3D, real time assessment tool.

Materials and methods:

Biomechanical data were collected prospectively on 30 patients with ACL rupture and 15 healthy subjects as a control group, with KneeKGTM System. This system is composed of passive motion sensors fixed on the validated knee harness, an infrared motion capture system (Polaris Spectra camera, Northern Dig. Inc.), and a computer equipped with the Knee3DTM software suite (Emovi, Inc.). Kinematic data were recorded in vivo during treadmill walking at self-selected speed. Flexion/extension, abduction/adduction, antero/posterior tibial translation and external/internal tibial rotation were calculated. Statistical analyze was performed to determine differences between ACLD and control group.

Results:

The ACLD patients showed a significant lower extension of the knee joint during stance phase, ($p < 0.05$; $13.16^{\circ} \pm 2.08$ and $7.33^{\circ} \pm 2.73$, for ACLD and control group respectively). A significant difference in tibial rotation angle was found in ACLD knees compared with control knees, ($p < 0.05$). The patients with ACLD rotated the tibia more internally ($-1.4^{\circ} \pm 0.22$) during the midstance phase, than control group ($0.15^{\circ} \pm 0.26$). There was no significant difference in antero-posterior translation and adduction-abduction angles.

Discussion:

The ACLD patients in our study showed a limited knee extension during the stance phase that appeared to be the adaptation strategy to avoid A-P translation during maximum extension and degrade the functional need for ACL. With a more internally rotated tibia position observed in ACLD knees in our study during midstance phase, the axial position alters and this could result in the changes of contact points and in a more rapid cartilage thinning throughout the knee, especially in the medial compartment.

Even though at ACLD knees tibia remained slightly more anterior most of the gait cycle there was no significant difference. This may be explained by active contraction of the muscles that help to increase the mechanical stability of the knee and thereby reduce massive translation of the femoral condyles relative to the tibial plateau.

Conclusion: Our study revealed significant alterations of joint kinematics in the ACLD knee by manifesting a higher flexion gait strategy and excessive internal tibial rotation during walking. The preoperative data obtained in this study will be useful to understand the post-ACL reconstruction kinematic behavior of the knee, a study that is ongoing in our department.

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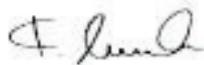
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In vivo kinematic alterations of the ACL deficient knee - 3D assessment

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Alterations Of The 3D Kinematics Of The ACL Deficient Knee - In Vivo Assessment With KneeKG System

Shabani B, Bytyqi D, Lustig S, Cheze L, Bytyqi C, Neyret P

Introduction

Following anterior cruciate ligament (ACL) rupture, the knee joint kinematics alters including anterior tibial displacement, and internal tibial rotation. Static, one-dimensional testing cannot predict the behavior of the ACL-deficient knee under realistic loading conditions. Currently the most widely accepted method for assessing joint movement patterns is gait analysis. Thus, the objective of this study was the in vivo evaluation of behavior of the ACLD knee during walking, using a 3D, real time assessment tool.

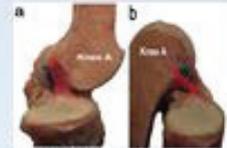


Fig.1 Anatomy of ACL

Materials and methods:

Biomechanical data were collected prospectively on 30 patients with ACL rupture and 15 healthy subjects as a control group, with KneeKG™ system. This system is composed of passive motion sensors fixed on a validated knee harness, an infrared motion capture system (Polaris Spectra camera, Northern Dig. Inc.), and a computer equipped with the Knee3D™ software suite (Emovi, Inc.). Kinematic data were recorded in vivo during treadmill walking at self-selected speed. Flexion/extension, abduction/adduction, antero-posterior tibial translation and external/internal tibial rotation were calculated. Statistical analysis was performed to determine differences between ACLD and control group.



Fig.2 KneeKG system

Parameter	ACLD group	Control group	p-value
Subjects	30	15	-
Age (years)	39.3±8.9	39.3±8.7	0.99
Height (cm)	173.5±10.3	173.5±10.3	0.71
Weight (kg)	72.3±10.1	72.3±10.0	0.74
BMI (kg/m ²)	24.4±2.9	24.4±2.9	0.99
Speed (km/h)	3.2±0.4	3.2±0.4	0.91
Time Treadmilling (min:00)	3:29±0:22	-	-

Table 1. Participants' characteristics



Fig.3 KneeKG examination



Fig.4 Anterior view of a right knee fitted KneeKG tracking system

Results:

Kinematic parameter	ACLD	Control group	p-value
Peak flexion (°)	113.9±2.28	73.9±2.73	p<0.01
Internal rotation at mid-stance (°)	-0.1±0.22	0.1±0.28	p<0.01
A-P tibial motion	-0.1±0.21	0.1±0.28	p<0.01
Abduction/Adduction	0.3±0.17	-0.1±0.17	p<0.01
Range of Knee Flexion	18.9±0.28	11.1±0.32	p<0.01
Range of Internal rotation	0.1±0.28	0.2±0.18	p<0.01
Range of A-P translation	0.1±0.28	0.2±0.18	p<0.01
Range of abduction/adduction	0.3±0.17	-0.1±0.17	p<0.01

Table 2. Kinematic parameters of ACLD knee and control knee. ST-stance phase, MSF-midstance phase

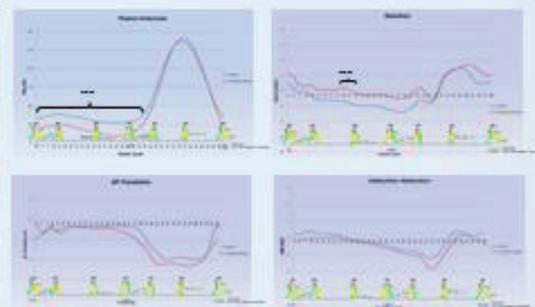


Fig.5 Tibiofemoral kinematics of ACLD knee and healthy control knee

Discussion:

The ACLD patients in our study showed a limited knee extension during the stance phase that appeared to be the adaptation strategy to avoid A-P translation during maximum extension and degrade the functional need for ACL. With a more internally rotated tibia position observed in ACLD knees in our study during midstance phase, the axial position alters and this could result in changes of contact points and in a more rapid cartilage thinning throughout the knee, especially in the medial compartment. Even though in ACLD knees tibia remained slightly more anterior, during most of the gait cycle there was no significant difference. This may be explained by active contraction of the muscles that help to increase the mechanical stability of the knee and thereby reduce massive translation of the femoral condyles relative to the tibial plateau.

Conclusion:

Our study revealed significant alterations of joint kinematics in the ACLD knee by manifesting a higher flexion gait strategy and excessive internal tibial rotation during walking. The preoperative data obtained in this study will be useful to understand the post-ACL reconstruction kinematic behavior of the knee, a study that is ongoing in our department.

Gait knee
kinematics after
ACL reconstruction
– 3D assessment

IV. Gait knee kinematics after ACL reconstruction – 3D assessment

IV.1. Introduction

The alterations in biomechanical features of the knee during walking following ACL deficient and ACL reconstruction have been evaluated in different studies. While many studies about ACL deficient patients have demonstrated functional adaptations to protect the knee joint, an increasing number of patients undergo ACL reconstruction surgery in order to return to their desired level of activity.

Many studies have reported good clinical outcomes following ACLR. However, long-term patient follow-up studies have reported a high incidence of degenerative changes [141], abnormal knee laxity [142], the need for revision surgery [143], anterior knee pain [144]. The precise mechanism contributing to these postoperative complications are unknown.

Abnormal knee kinematics has been thought to be one of the possible reasons for long-term development of degenerative changes after ACLR. It is therefore interesting to study the kinematics associated with this type of surgery. But, even though there are several devices currently available to assess the knee joint kinematics [83, 115, 116] the 3D biomechanical changes caused by ACL injury and the effect of ACLR in kinematics of the knee are still not clearly understood. Thus, establishing an objective evaluation of the kinematics of the knee in a clinically feasible way is critical in extensive evaluation of the ACL function and a valuable feedback for ACLR.

The purpose of this study was to compare 3D kinematic patterns between individuals having undergone ACL reconstruction with healthy contralateral knee and control group.

IV.2. Material and methods

This prospective study was conducted in periods from January 2011 to January 2014, in the facilities of the biomechanical laboratory at our Clinical Center. Patients who were planned to be operated for ACL reconstruction were selected for kinematic analysis. The ACL rupture was diagnosed by clinical examination, MRI and confirmed in the time of the surgery by arthroscopy. The patients with unilateral ACL rupture and healthy contralateral knees (that had never suffered of any kind of orthopaedic or neurological condition) were included in the study. Patients with meniscal injury where partial meniscectomy or repair is feasible, grade I or II medial collateral injury were also part of the study. On the other hand patients who had subtotal or total meniscectomy, concomitant PCL injury, with knee joint movement restriction, full thickness cartilage defect $>1 \text{ cm}^2$, with previous history of any surgery in both knees, were excluded from the study.

Of the 30 patients who were included in the first study, we were able to obtain follow-up evaluations after the reconstruction on 15 patients. ACL reconstruction was done in the same center by three surgeons of the same team and with the same technique.

The post-OP examination was done in average time 10.23 ± 1.4 months from ACL reconstruction. The kinematic analyzes were done in ipsilateral knees and in contralateral healthy knees. These kinematic data were compared with kinematic data of 15 participants as control group, who had no history of musculoskeletal injury or surgery in the lower extremities (Table 8).

Demographic	ACL-Group	Control Group	p - value
Age (years)	30±9.8	29.3±9.7	n.s
Height (cm)	171.6±9.3	173.1±10.1	n.s
Weight (kg)	70.8±13.7	70.9±14.6	n.s
BMI (kg/m ²)	23.9±3.6	23.2±2.5	n.s
Female:Male	7:8	7:8	-
Right:Left	6:9	7:8	-
Time from injury (months)	4.7±4.3	-	-
Time surgery to examination (months)*	10.23±1.4	-	-

Table 8 – Participants’ characteristics

IV.2.1. Data collection and operation technique

Clinical assessment was performed for all subjects by two fellow clinicians experienced in orthopaedic surgery. Static knee stability was evaluated with the manual Lachman test, drawer test and pivot shift test. The International Knee Documentation Committee (IKDC) objective evaluation was also acquired to assess clinical outcomes (Table 9).

The operation technique which consist of: ‘Double incision iso-anatomical ACL reconstruction’, using a patellar tendon auto graft has previously been presented in general part. The in-vivo, 3D kinematic data were collected during walking at self-selected comfortable speed. We used the KneeKGTM system, which has precisely been presented in the first study.

<i>Clinical characteristics</i>	ACL D	ACL R	p value
Range of motion			
Flexion (mean ± SD)	136.6 ⁰ ±4.88 ⁰	138.33 ⁰ ±4.88 ⁰	n. s
Extension (mean ± SD)	0.53 ⁰ ±1.45 ⁰	0 ⁰	n. s
Recurvatum (mean ± SD)	2.33 ⁰ ±3.6 ⁰	0.66 ⁰ ±1.5 ⁰	n. s ^a
IKDC (nr. of patients)			
Grade A	-	15	
Grade B	-	-	
Grade C	11	-	
Grade D	4	-	

Table 9 - Clinical characteristics of ACLD and ACLR knees. ^a Mann Whitney test

IV.2.2. Statistical analysis

Participant characteristics (such as age, height, weight, BMI, gait speeds and range of motion) were tested to determine whether parametric assumptions were met using the Levene test. Mann–Whitney test was used for non-parametric variables, while ANOVA and paired t-test were used for parametric variables.

Paired t-test was utilized to compare kinematic parameters of ACLD and ACLR group, and ACLR with contralateral group, while ANOVA test was used to compare ACLR with control group. All statistical analyses were done using SPSS v 21 (SPSS Inc, Chicago, Illinois, USA), and significance level was set at 0.05.

IV.3. Results

None of the participants' characteristics were statistically different between groups, except walking speed, where the participants after surgery walked with higher speed compared to the speed before having surgery. Table 10 and 11 summarizes the main kinematic data of the ACLD, ACLR, healthy contralateral and control knees.

	ACLR vs ACLD <i>Stance Phase</i>	ACLR vs Contralateral <i>Stance Phase</i>	ACLR vs Control <i>Stance Phase</i>
Sagittal plane	10.03 ^o ±4.96 vs 14.22 ^o ±6.37 p < 0.05	10.03 ^o ±4.96 vs 8.42 ^o ±5.57 p > 0.05	29.24 ^o ±18.06 vs 23.29 ^o ±18.35 * p < 0.05
Axial plane	-1.68 ^o ±2.67 vs -1.35 ^o ±1.97 p > 0.05	1.35 ^o ±1.97 vs -1.35 ^o ±2.97 p > 0.05	-1.53 ^o ±0.21 vs -0.07 ^o ±0.25 ^ -2.78 ^o ±0.27 vs -0.86 ^o ±0.21 @ p < 0.05
AP translation	-0.29mm±2.31 vs 0.5mm±2.52 p > 0.05	-0.29mm±2.31 vs 0.38mm±2.31 p > 0.05	-0.29mm±2.31 vs -1.2mm±1.5 p > 0.05
Coronal plane	-1.12 ^o ±3.42 vs -0.55 ^o ±2.78 p > 0.05	-1.12 ^o ±3.42 vs -1.1 ^o ±3.21 p > 0.05	-1.12 ^o ±3.42 vs -0.59 ^o ±3.08 p > 0.05
Speed # (Km/h)	2.46 ± 0.21 vs 2.1 ± 0.38 p < 0.05	-	2.46 ± 0.21 vs 2.51 ± 0.4 p > 0.05

Table 10 - Kinematic parameters of ACLR, ACLD, healthy contralateral and control knees during stance phase

* Terminal stance, Pre-swing, Initial swing phase; ^ Mid-stance phase; @ Terminal stance phase

Speed throughout all gait cycle

	ACLR vs ACLD <i>Swing Phase</i>	ACLR vs Contralateral <i>Swing Phase</i>	ACLR vs Control <i>Swing Phase</i>
Sagittal plane	38.81 ^o ±3.87 vs 37.59 ^o ±6.08 p > 0.05	38.81 ^o ±3.87 vs 38.3 ^o ±5.1 ^o p > 0.05	29.24 ^o ±18.06 vs 23.29 ^o ±18.35 * p < 0.05
Axial plane	1.76 ^o ±3.28 vs 0.82 ^o ±3.62 p > 0.05	1.76 ^o ±3.28 vs 2.16 ^o ±4.76 p > 0.05	1.76 ^o ±3.28 vs 1.26 ^o ±3.1 p > 0.05
AP translation	-3.69mm±2.76 vs -2.49mm±3.6 p > 0.05	-3.69mm±2.76 vs -3.54mm±2.45 p > 0.05	-3.69mm±2.76 vs -4.39mm±1.66 p > 0.05
Coronal plane	-1.53 ^o ±4.97 vs -0.34 ^o ±3.56 p > 0.05	-1.53 ^o ±4.97 vs 2.05 ^o ±4.76 p > 0.05	-1.53 ^o ±4.97 vs -1.42 ^o ±3.32 p > 0.05

Table 11 - Kinematic parameters of ACLR, ACLD, healthy contralateral and control knees during swing phase.

* Terminal stance, Pre-swing, Initial swing phase

IV.3.1. Sagittal plane

The ACL reconstructed knees showed a significant higher extension (10.03^o±4.96^o) of the knee joint during entire stance phase compared to the ACL deficient knees (14.22^o±6.37^o) while there is no significant difference during swing phase (38.81^o±3.87; respectively 37.59^o±6.08). No statistical difference was detected between reconstructed (10.03^o±4.96^o) and intact contralateral knees either in stance (8.42^o±5.57^o) or swing phase (38.81^o±3.87, respectively 38.3^o±5.1^o) (Fig 27).

In the other hand there was found statistically significant difference between reconstructed (29.24^o±18.06^o) and healthy control (23.29^o±18.35^o) knees during terminal stance, pre-swing and initial swing phase, where the reconstructed knee showed lower extension. More specifically, the difference was identified between 46 and 74 % of the gait cycle (Fig 27).

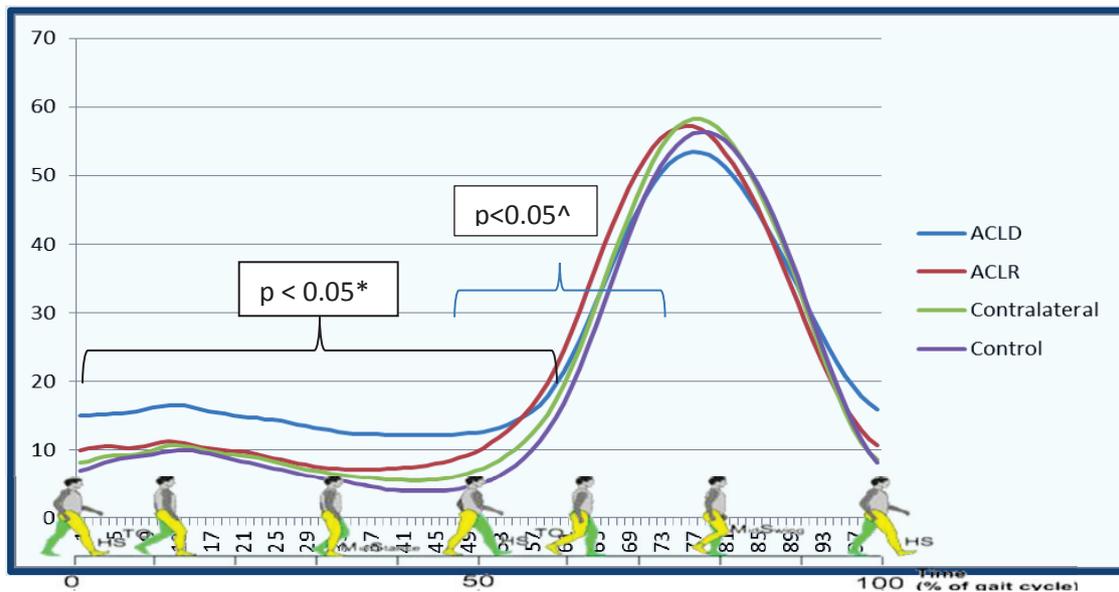


Fig. 27 Tibiofemoral kinematics of ACLD, ACLR, Contralateral and control knees -Sagittal plane.

*ACLR vs ACLD; ^ACLR vs Control

IV.3.2. Axial plane

There were no statically significant differences between pre- and postoperative kinematic data, neither in stance ($-1.68^0 \pm 2.67$; $-1.35^0 \pm 1.97$) nor in swing phase ($0.82^0 \pm 3.62$; $1.76^0 \pm 3.28$). The significant difference is not achieved also between the ACL reconstructed and intact contralateral knees in none of the phases of the gait cycle (stance: $-1.35^0 \pm 1.97$; $-1.35^0 \pm 2.97$; swing: $1.76^0 \pm 3.28$; $2.16^0 \pm 4.76$). Even though the tibia of ACL reconstructed knees rotated more internally from mid-stance to initial swing phase, compared to healthy control knees, the significant difference is achieved only from 28 to 34 % and 44 to 54 % of the gait cycle (mid-stance and terminal stance phase) ($-1.53^0 \pm 0.21$: $-0.07^0 \pm 0.25$, respectively $-2.78^0 \pm 0.27$: $-0.86^0 \pm 0.21$) (Fig 28).

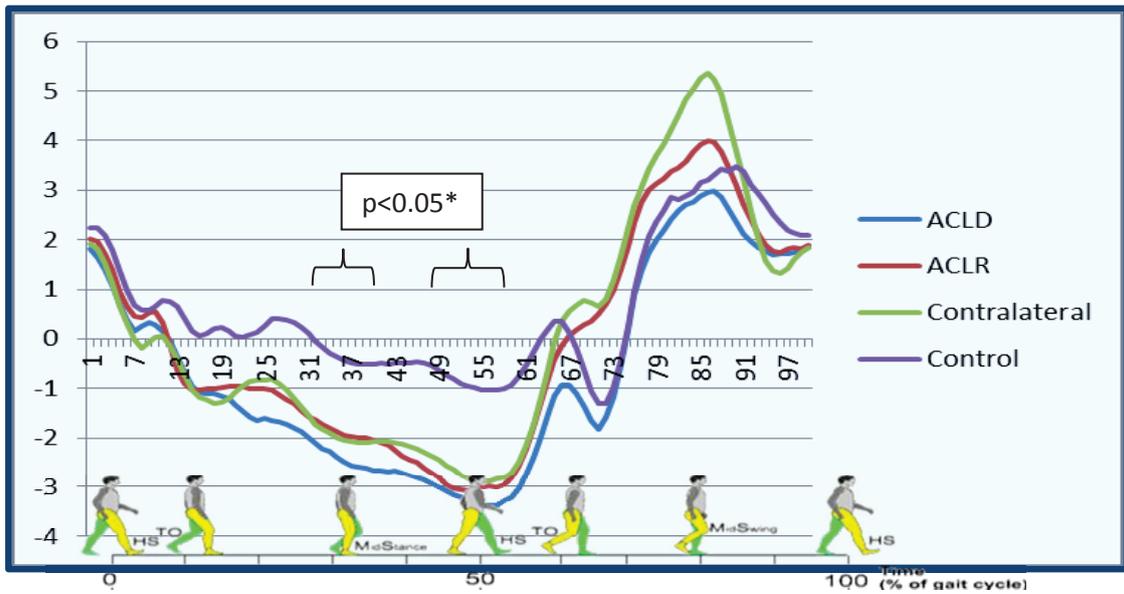


Fig. 28 Tibiofemoral kinematics of ACLD, ACLR, Contralateral and control knees -Axial plane.

*ACLR vs Control

IV.3.3. Antero-posterior translation and coronal plane

Even though in ACLD knees during entire gait cycle tibia was in anterior position compared to ACLR knees, statistically there were no significant differences between the two groups. However, there were no significant differences between ACLR knees when compared to intact contralateral knees and healthy control knees (Fig 29).

Although the ACLR and ACLD knees remain in adducted position in initial and mid-swing phase compared to intact contralateral and healthy control knees, in coronal plane there were no statistically significant differences between all groups that are compared (Fig 30).

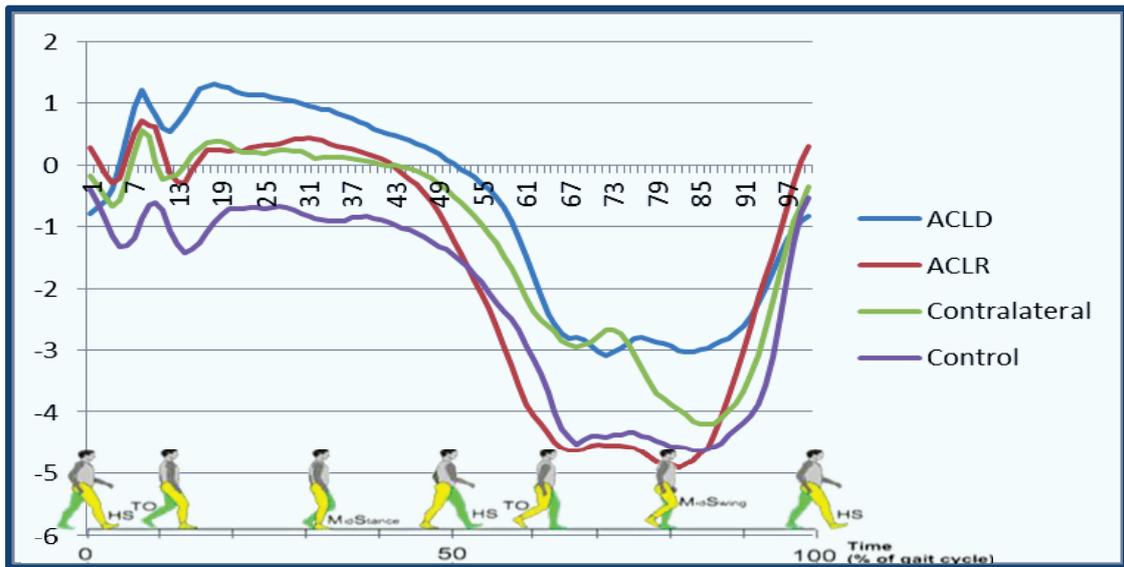


Fig. 29 Tibiofemoral kinematics of ACLD, ACLR, Contralateral and control knees –Anterior-Posterior translation.

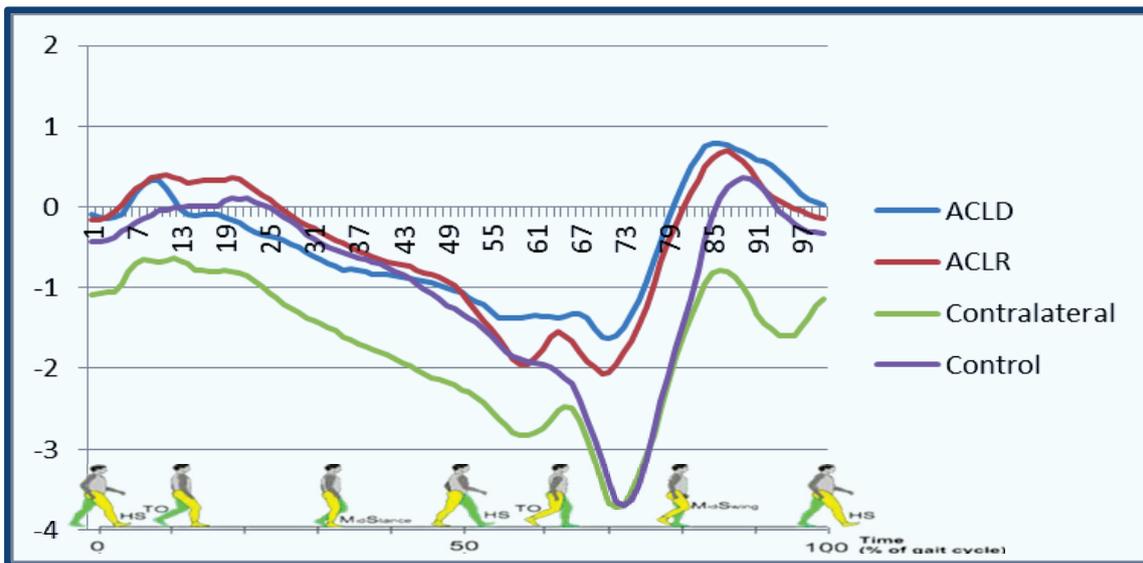


Fig. 30 Tibiofemoral kinematics of ACLD, ACLR, Contralateral and control knees – Coronal plane.

IV.4. Discussion

The most important finding of the present study was that ACLR knees improved significantly extension compared to ACLD knees, but not compared to control group. There were no differences in AP translation, while ACLR knees showed more internal tibial rotation.

Statistically significant differences were found in sagittal plane, where the ACLR knees showed more extension during stance phase compared to ACLD knees, while there were no significant differences compared to intact contralateral knees. However, during terminal stance and initial swing phase ACLR knees showed significantly less extension than healthy control knees. So, after reconstructive surgery the extension has been improved in comparison with ACLD knees, but not fully restored compared to healthy control group. The findings of the extension deficit were consistent with the study of Gao et al. [78]. They showed that ACLR knees exhibited less extension during stance phase and during second period of swing phase.

Seeing that in ACLR knees passive ROM were fully restored, this deficit in extension could be due to quadriceps strength weakness. Freiwald et al. [145] have found that the maximal isokinetic quadriceps ratio was 81% of that of the normal knee 16 months after surgery, while in our study this time is 10 months. Arciero et al. [146] reported that patients regained 98.5 % of thigh girth and 97 % of quadriceps muscle strength at an average follow-up of 31 months. In addition, quadriceps strength weakness has been noticed after harvesting the BPTB autograft and hamstring muscle weakness after harvesting HST autograft [147]. These alterations in muscle performance could be neural or mechanical in origin. Specifically, the lack of proprioceptive activity deriving from the ruptured ligament or graft harvest site may alter neural control of the muscles around the knee [139, 148]. But, several investigations

have suggested that ACL-reconstructed patients will return to pre-injury gait status over time [146, 149].

Kinematic alterations were also identified in axial plane. There were no statistical differences in kinematic data before and after surgery, neither between ACLR knees and intact contralateral knees. The differences were observed between ACLR knees and healthy control group. The significant differences were achieved during mid-stance and terminal stance phase, when the leg was under full body weight. The finding of greater internal tibial rotation in ACLR knees was consistent with other studies [78, 150]. They have found that the ACLR knees exhibited more internal tibial rotation during midstance phase, or throughout the whole gait cycle, respectively. This may be explained by the material properties of the graft, which was different from the native ACL [151]. Handl et al. [151] showed that BTPB graft showed more stiffness compared to original ACL. In addition, the native ACL has two functional bundles, and the attachment site area is much larger than the insertion site of single bundle. The decreased attachment area and posteriorly shifted insertion may affect the graft's ability to constrain the internal tibial twisting [150]. Butler et al. [39] evaluated the strain distribution within the ACL, a spatial variation in strain was measured along the length of the ACL, with the greatest strain found at the insertion site.

As we see above, while there were significant differences between ACLR knees and healthy control group in both planes (sagittal, axial), there were no significant difference between ACLR knees and intact contralateral knees. It has been shown that there are biomechanical adaptations in the intact contralateral knee. The same phenomenon was observed also in other studies [147, 152]. Decler et al. [152] proposed that this alteration is a compensatory mechanism in order to maintain some degree of symmetry between the two legs.

Even though in term of AP translation there was improvement throughout gait cycle in ACLR knees, there were no significant differences between all groups that were compared. As well, the significant difference was not achieved even in comparison between ACLD knees and control group. But here it is important to note that while the ACLD knees walked with less extension to prevent anterior translation of tibia, the ACLR knees have improved significantly extension compare to pre-OP, and there was not any significant difference compared to control group. There are studies that have presented similar results [78, 153]. Gao et Zheng [78] have found no significant statistical difference of knee joint translations between ACLD, ACLR and ACL intact knees. They explain that these results may come from a combination of two aspects: the relatively low inter-group difference and relatively high intra-group variability.

In the first part of swing phase, the ACLR and ACLD knees remain in adducted position even though there were no statistically significant differences. Wang et al.[154] have not found any significant difference between ACLR knees and controls. They thought that this does not necessarily reflect no changes in compartmental loading postoperatively; the medial/lateral load sharing can be further evaluated by characterizing knee kinetics, specifically abduction/adduction moments. Furthermore, Schipplein et Andriacchi [155] claimed that co-contraction of antagonistic muscle action and/or pretension in the passive soft tissue was needed for dynamic joint stability during walking. So, further studies should concern also in kinetic parameters of gait cycle to complement kinematic studies.

There is support in the literature that kinematic abnormalities in ACLR knees are associated with the OA development and progression. If the kinematic changes are sufficient to shift cyclic loading during ambulation to region that cannot adapt to a change in the local mechanical environment, then normal homeostasis is disrupted in a manner that can initiate a degenerative pathway. The knee joint is particularly sensitive to kinematic changes, since

there is a larger range of translational motion at the knee than in other joints, and the movement is dependent on stable ligaments, healthy menisci, and coordinated muscular function [157]. Thus, maintaining consistent patterns of gait within an envelope of healthy homeostasis between external ambulatory mechanics and cartilage metabolism is a necessary condition to sustain cartilage health [157, 158].

This study has some limitations. First, the fact that patients included in this study were operated by three surgeons could be considered as a limitation of this study. Nevertheless, the three surgeons were part of the same team and they used the same technique. Reynaud [156] in his medical thesis has elaborated the position of femoral tunnel in our center. In addition, there was no significant difference between surgeons' outcomes.

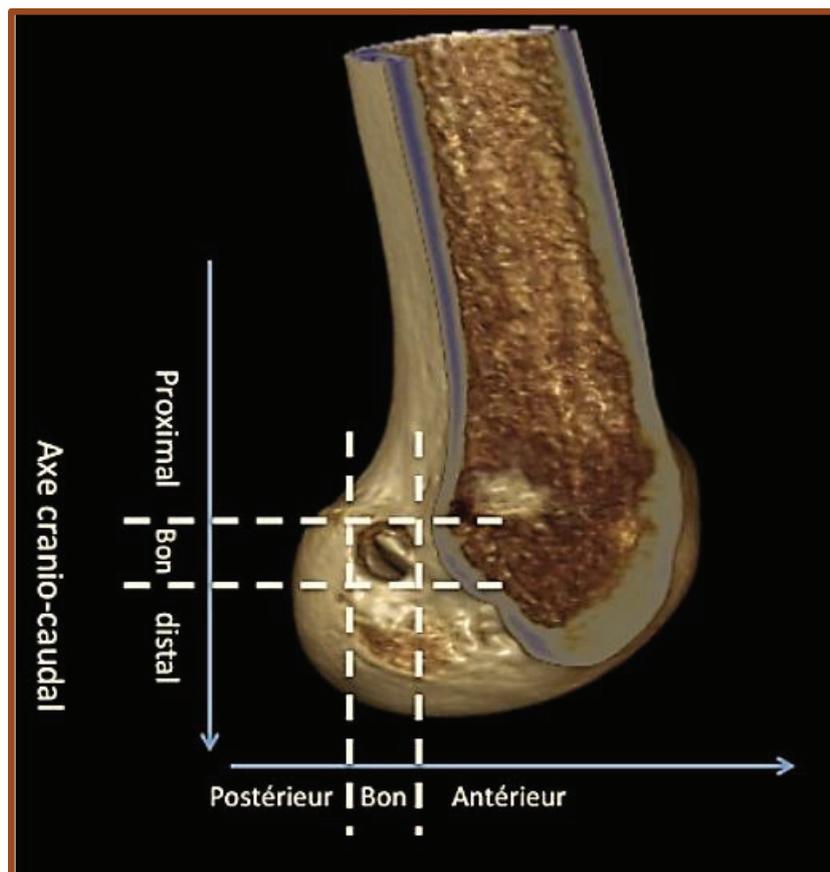


Fig. 31 Analysis of femoral tunnel position (Anatomic position) [156]

Another possible limitation is the small number of patients that formed four groups that we compared (ACLD, ACLR, ACLI and control group). However, it is consistent with other gait analysis studies of the ACLD population. The fact that there was a prospective follow up made it difficult to have a more important number of patients.

IV.5. Conclusion

In vivo – 3D motion analysis in this study revealed that ACLR knees improve significantly extension compared to ACLD knees, but there were still difference compared to healthy control group. In the axial plane, tibia remains in internal position significantly compared to healthy control group, while there were no any significant difference in antero-posterior translation and in coronal plane. These kinematic changes could lead to abnormal loading in knee joint and initiate the process for future chondral degeneration. However, the postoperative kinematic data were collected 10 months after surgery, so a longer follow-up is needed to evaluate if these kinematic changes persist in time, and their effects in joint degeneration.

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Title: **Gait Knee Kinematics After ACL Reconstruction – 3D Assessment**

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Keywords: ACL, Reconstruction, Kinematics, Gait

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CONCLUSIONS and PERSPECTIVES

This thesis has investigated the influence of ACL rupture and ACL reconstruction on knee joint kinematics during walking with the aim of providing a better understanding of the mechanism adaptation of ACLD knees and the effect of surgery in these patients. As a step towards achieving the goal, we used KneeKG system for kinematic analysis as a 3D, real time, non-invasive tool.

It has been shown that in sagittal plane, patients with ACL rupture manifested higher flexion strategy. This study revealed that ACLD knees adapted functionally to protect excessive anterior-posterior translation but failed to avoid rotational instability. However, while this study is performed during walking; it would be of high interest to investigate kinematic of ACLD knees during high demanding activities, as walking downhill, pivoting maneuvers, running. Further information could be gathered from electromyography (EMG) measurements to determine relative muscle activity and the onset patterns of muscle activation.

A reconstruction of the ACL affects the knee joint kinematics. We observed that in sagittal plane ACLR knees improved significantly the extension deficit compared to ACLD knees, but not in such level as control healthy knees. These results coincide with previous investigations that have noted that ACLR knees have altered post-surgical lower extremity locomotive strategies. Studies with long-term follow-up are essential to investigate if these alterations persist in time.

No significant difference was found in axial plane before and after reconstruction. The tibia of ACLR knees remains significantly in internal rotation during mid-stance and terminal stance phase compared to healthy control knees. While the rotational instability after ACL

reconstruction remains an issue, we propose that maybe extra-articular lateral reinforcement could be potential solution. Lording et al. [159] in their study have noticed that lateral extra-articular reinforcement in conjunction with intra-articular reconstruction may be an important option in the control of rotational laxity of the knee. So, further studies should be focused in assessing kinematics of this type of surgery.

There were no significant differences in anterior-posterior translation and in the coronal plane between the groups that are compared. Here, studies utilizing EMG would be helpful in determining differences in co-contraction of antagonist muscles (hamstring and quadriceps) and better characterizing the intra-articular loading in an ACLR during gait. This would better explain the role of biomechanics in initiating and development of premature OA in ACLR knees.

In our second study, we have noticed that kinematic adaptations occurred in contralateral intact knee. In sagittal and in axial planes, while we found significant difference between ACLR knees and healthy control knees, this difference was not present when comparing with contralateral intact knees. This alteration could be as compensatory mechanism in order to maintain some degree of symmetry between the two legs. It would be interesting to study and compare the kinematics of contralateral knees pre- and postoperatively, and to identify if this 'laxity' could be a risk factor or not.

Résumé

Introduction

La rupture du ligament croisé antérieure (LCA) conduit à une instabilité et à des modifications biomécaniques du genou. Elle est associée au développement de lésions méniscales et à une atteinte dégénérative du cartilage articulaire pouvant évoluer jusqu'à l'arthrose.

Actuellement, les patients présentant une lésion du LCA (en particulier les jeunes et ceux souhaitant poursuivre une activité sportive soutenue) bénéficient généralement une reconstruction du LCA. Ainsi même si les techniques chirurgicales de reconstruction du LCA sont actuellement éprouvées, il y a encore place pour les améliorer, notamment pour perfectionner la restauration cinématique du genou normale.

L'analyse cinématique quantitative est un outil important pour acquérir une compréhension approfondie de la fonction articulaire du genou normal et pathologique au cours de la locomotion humaine. Même si des informations importantes peuvent être obtenues par l'examen clinique manuel, les outils plus précis et objectifs sont très utiles, en particulier en ce qui concerne l'évaluation de la stabilité rotatoire.

De même pour les études in vitro qui fournissent des informations importantes sur les patients avec rupture de LCA, leur incapacité à simuler la cinématique du genou lors des activités quotidiennes a conduit au développement de nouveaux outils cinématiques in vivo-3D.

Ce travail est axé sur l'évaluation in vivo de la cinématique du genou chez les patients avec rupture du LCA, avant et après la reconstruction du LCA au cours de toutes les phases de la

marche, en utilisant un nouvel outil d'évaluation 3D, quasi-rigide, en temps réel (KneeKG™). Le système KneeKG™ été développé avec pour objectif l'évaluation en pratique clinique du comportement cinématique de l'articulation du genou lors de la marche.

Les données quantitatives précises de rotation fournies par le KneeKG™ en font un outil adapté:

- pour l'évaluation des facteurs de risque de lésions de LCA,
- pour prévoir certaines déformations in vivo en flexion et en torsion du ligament,
- pour évaluer si le genou est biomécaniquement prêt à reprendre les sports de contact à distance d'une reconstruction chirurgicale du LCA,
- pour illustrer l'importance de l'évaluation biomécanique 3D des lésions ACL.

Matériel et Méthodes

Cette étude prospective a été réalisée durant la période de Janvier 2011 à Janvier 2014, dans les aménagements du laboratoire biomécanique à notre centre clinique. Les patients qui étaient programmés pour une opération de reconstruction du LCA ont été sélectionnés pour l'analyse cinématique. La rupture du LCA a été diagnostiqué par l'examen clinique, l'IRM et confirmé durant le 1er temps de la chirurgie par arthroscopie. Les patients avec rupture unilatérale du LCA et un genou controlatéral sain (qui n'avaient jamais souffert d'aucune sorte de problème orthopédique ou neurologique) ont été inclus dans l'étude. Les patients présentant une lésion méniscale pour laquelle une ménisectomie partielle ou une suture était possible, des lésions grade I ou II du ligament collatéral médial étaient également inclus dans l'étude. Par contre les patients qui présentaient une ménisectomie subtotale ou totale, une

lésion concomitante du PCL, avec restriction de mouvements de l'articulation du genou, un défaut cartilagineux grade IV de cartilage >1 cm², avec des antécédents d'une chirurgie dans les deux genoux, ont été exclus de l'étude.

Dans la première étude, 30 patients (LCAD) ont pu être inclus dans cette étude; 40 patients ont été exclus parce qu'ils ne répondaient pas aux critères d'inclusion. Un groupe de contrôle composé de 15 participants du même âge, sans antécédents de lésions musculo-squelettiques ou chirurgie des membres inférieurs et qui ne présentait aucune instabilité ligamentaire mesurable sur l'examen clinique (pivot shift, Lachman et test du tiroir), a été sélectionné.

Dans la deuxième étude, sur les 30 patients qui ont été inclus dans la première étude, nous avons pu obtenir des évaluations de suivi après la reconstruction sur 15 patients. L'examen post-opératoire a été fait avec un délai moyen de $10,23 \pm 1,4$ mois à compter de la reconstruction du LCA. Les analyses cinématiques ont été réalisées dans les genoux ipsilatéraux et genoux sains controlatéraux. Ces données cinématiques ont été comparées avec les données cinématiques de 15 participants du groupe de contrôle, qui n'ont pas eu d'antécédents de lésions musculo-squelettiques ou la chirurgie dans les extrémités inférieures.

La technique opératoire consiste en une reconstruction du LCA par technique Kenneth-Jones Out-In . Préparation de tunnel fémoral - Guidé par un particulier viseur fémoral out-in , une deuxième courte incision de la peau était établie. Le viseur était placé dans le centre de l'attachement du LCA natif résultant en un point de départ extra-articulaire de la métaphyse fémorale latérale distale. La broche de guidage était ensuite forée de dehors en dedans dans l'articulation. Tout ce processus était contrôlé par arthroscopie par la voie d'abord médiale. Préparation de tunnel tibial - Le viseur tibial était placé entre les épines tibiales au niveau du pied du LCA. Ensuite, une broche guide était introduite sous contrôle arthroscopique à travers le viseur. Le tunnel tibial était ensuite foré de dehors en dedans. Après la préparation des

tunnels, le greffe était insérée dans le tunnel fémoral et tractée jusqu'à obtenir une fixation en press-fit. L'isométrie était vérifiée et une fixation tibiale avec une vis d'interférence était réalisée.

Les données biomécaniques de marche ont été collectées en utilisant le système KneeKG™. Le système KneeKG™ est composé de détecteurs de mouvement passifs fixés sur un harnais validé du genou, un système de capture de mouvement infrarouge (Polaris Spectra camera, Northern Digital Inc.), et un ordinateur équipé du logiciel Knee3D™ (Emovi, Inc.). Le système mesure et analyse la position 3D et le mouvement du genou du patient.

Afin de réduire les artefacts de mouvement de la peau, le groupe a développé un harnais fixé de façon quasi-rigide sur la cuisse et du mollet. Ce harnais a été démontré pour être précis dans l'obtention de données 3D cinématiques qui pourraient être utilisés pour évaluer les déformation du LCA et des greffes de LCA in vivo.

Au début, la procédure de calibration est effectuée. Cette procédure inclut deux parties principales: la définition des centres des articulations et définissant le système joint d'axes basés sur des postures prédéterminées.

Après calibration, les données cinématiques étaient collectées pendant la marche sur tapis roulant à une vitesse confortable, auto-sélectionnée, et pour éviter l'effet de la chaussure sur la biomécanique des membres inférieurs, tous les sujets marchaient pieds nus. Avant de commencer les essais de tous les patients marchaient 10 minutes pour s'habituer à marcher sur un tapis roulant.

Une fois la calibration et les mesures effectués, le KneeKG™ calcule les paramètres cinématiques du genou tout au long de l'enregistrement. Une base de données contenant, pour chaque participant, les 4 modèles biomécaniques composé des trois angles du genou (flexion-

extension, abduction-adduction et de rotation du tibia interne-externe) et la translation tibiale antéro-postérieure a été créé dans Microsoft Excel 2010.

Résultats

Dans la première étude, il a été montré que dans le plan sagittal les patients avec rupture de LCA manifestent une extension de l'articulation du genou inférieure au cours de toute la phase d'appui. Alors que dans le plan axial, bien que le tibia reste en rotation interne pendant toute la phase d'appui, une différence significative a été observée seulement entre 26 et 34% du cycle de marche (au milieu de la phase d'appui). Il n'y avait aucune différence statistiquement significative entre les deux groupes en translation antéro-postérieure et dans le plan coronal.

Dans la deuxième étude, dans le plan sagittal, les genoux après reconstruction du LCA ont montré une extension de l'articulation du genou significativement supérieure pendant toute la phase d'appui par rapport à avant l'opération. Aucune différence significative n'a été trouvée entre les genoux après reconstruction et les genoux controlatéraux intacts. En revanche, il a été trouvé une différence statistiquement significative entre genou reconstruit et genou contrôle sain (entre 46 et 74% du cycle de marche), où les genoux reconstruits ont montré une extension inférieure. Dans le plan axial, il n'y avait pas de différences statistiquement significatives entre les données cinématiques pré et postopératoires. Cependant, la différence significative a été obtenue de 28 à 34 % et 44 à 54% du cycle de marche, où les genoux reconstruits pivotaient plus à l'intérieur.

Il n'y avait aucune différence statistiquement significative entre les groupes comparés dans la translation antéro-postérieure et dans le plan coronal.

Conclusion

Dans la première étude, utilisant la 3D, l'analyse du mouvement in vivo, des modifications cinématiques significatives ont été identifiées dans les genoux avec une rupture du LCA. Ces modifications se sont manifestées sous la forme d'une stratégie plus en flexion de la marche et un rotation tibiale interne excessive pendant la marche. Cependant, aucune différence significative n'a été retrouvée en translation antéro-postérieure. L'augmentation de l'angle de flexion a été accompagnée par une restauration de la stabilité antéro-postérieur, mais il n'a pas restauré la stabilité en terme de rotation. Ces changements cinématiques pourraient mener à des contacts fémoro-tibiaux anormaux du cartilage pendant les activités quotidiennes, représentant potentiellement un mécanisme biomécanique de l'usure des articulations après une lésion du LCA.

Dans la seconde étude, l'analyse 3D in vivo du mouvement a révélé que les genoux avec LCA reconstruits améliorent significativement leur extension par rapport aux genoux avec LCA déficient, mais il y avait encore une différence par rapport au groupe témoin sain. Dans le plan axial, le tibia reste en position interne significative par rapport à un groupe témoin, tandis qu'il n'y avait pas de différences significatives en translation antéro-postérieur et dans le plan coronal. Ces changements cinématiques pourraient conduire à une charge anormale dans l'articulation du genou et initier le processus d'atteinte dégénérative cartilagineuse à venir.

Toutefois, les données cinématiques post-opératoires ont été collectées 10 mois après la chirurgie, ainsi un suivi plus long serait nécessaire pour évaluer si ces changements cinématiques persistent dans le temps, et leurs effets sur l'articulation.

Mots-clés : Ligament croisé antérieur, Reconstruction, Genou, Analyse de la marche, Cinématique, 3D

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APPENDIX

Publication in co-authorship

**Gait knee kinematic alterations in medial osteoarthritis:
3D assessment**

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Gait knee kinematic alterations in medial osteoarthritis: three dimensional assessment

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Abstract

Purpose Although kinematic changes in the sagittal plane of the osteoarthritic knee (OA) have been elucidated, very few studies have analysed changes in the frontal and horizontal planes. Therefore, the aim of this study was to investigate in vivo 3D knee kinematics during walking in patients with knee OA.

Methods Thirty patients with medial knee OA and a control group of similarly aged individuals were prospectively collected for this study. All participants were assessed with KneeKG™ system while walking on a treadmill at a self-selected speed. In each trial, we calculated the angular displacement of flexion/extension, abduction/adduction and external/internal tibial rotation. Statistical analysis was performed to determine differences between the knee OA group and the control group.

Results Patients with knee OA had reduced extension during the stance phase ($p < 0.05$; 8.5° and 4.4° , OA and control group, respectively) and reduced flexion during pushoff and

initial swing phase ($p < 0.05$; 41.9° and 49.4° , respectively). Adduction angle was consistently greater for OA patients ($p < 0.05$; 3.4° and -0.9° , respectively). Frontal laxity for OA patients was positively correlated with varus deformity ($r = 0.42$, $p < 0.05$). There was a significant difference ($p < 0.05$) in tibial rotation during the midstance phase; OA patients retained a neutral position (-0.4°), while the control group presented internal tibial rotation (-2.2°).

Conclusion Weight-bearing kinematics in medial OA knees differs from that of normal knees. The knee OA group showed an altered “screw-home” mechanism by decreased excursion in sagittal and axial tibial rotation and posterior tibial translation.

Keywords Knee osteoarthritis · Kinematics · Gait · In vivo assessment

Introduction

Knee function can be quantified with either patient-based scales (questionnaires) or performance-based measures. Seeking to improve knee-function evaluation, numerous studies analysed movement of intact, pathological and treated knees. Quantitative kinematic analysis is an important tool for a thorough understanding of joint function [1]. Kinematics of osteoarthritic (OA) knees has been evaluated using surgical navigation systems, magnetic resonance imaging (MRI) and computed tomography (CT) [2–4], but these techniques cannot be used to study weight-bearing activities, and their results may be therefore be affected. With advances in sophisticated motion-capture technology, 3D knee motion during weight-bearing is now available. Gait, as the most common activity of daily living, has been analysed to clarify the biomechanical characteristics of OA knees.

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Most previous studies of knee kinematic changes in OA patients during gait focused on spatiotemporal parameters [5, 6]; demonstrating that knee OA patients walk slower, with reduced stride length and lower single-limb support compared with controls. Some studies examined kinematic alterations during phase-specific gait cycle and reported decreased knee excursion during flexion, decreased peak flexion during stance and increased knee flexion at heel strike [7–9]. Although kinematic changes in the sagittal plane were found, changes in frontal and axial planes and anterior–posterior (AP) translation remain unclear. Some studies reported an increase in knee adduction angle at initial contact and midstance and a smaller external tibial rotation angle at initial contact [10, 11]; however, both those studies concentrated only the stance phase. Because AP translation is small and can be affected by the choice of assessment system, few studies have analysed movements in this plane [2, 12]. The quasistatistical fixation of our assessment system is the first to examine anterior–posterior translation during walking [13, 14]. The objective of this study was to use 3D motion analyses (KneeKGTM) to identify changes in kinematic variables of patients with OA of the knee during a complete gait cycle and to correlate them with clinical characteristics. We hypothesised that patients with an OA knee would exhibit altered knee kinematics in the sagittal, frontal and axial planes.

Materials and methods

Participants

This prospective study was performed between February and April 2011 in the biomechanical laboratory at our clinical centre. Thirty patients (18 women, 12 men; mean age 65.7 years) with varus malalignment and medial-compartment knee OA were recruited into the study. They had been scheduled for total knee arthroplasty (TKA). Diagnosis was based on clinical history, physical examination and radiographic evaluation that included comparative anteroposterior knee view with monopodal support, bilateral posteroanterior view at 45° flexion in weight-bearing, comparative lateral view at 30° flexion, axial view at 30° flexion and stress valgus and varus X-ray using a Telos system [15]. Malalignment was confirmed by measuring mechanical axis of the leg, hip–knee–ankle (HKA) angle, from bilateral weight-bearing anteroposterior long-leg films. A line is drawn from the centre of the femoral head to the midpoint of the tibial eminential spine and another from this midpoint to the centre of the tibial plafond. The medial angle between the lines is the HKA angle (varus < 180°, valgus > 180°). Study participants were classified in terms of OA disease severity using Ahlbäck's radiographic grading system. The integrity of the anterior cruciate ligament (ACL) was assessed intra-operatively. The ACL was present in 21 knees, attenuated in five and ruptured in four

(Table 1). Patients were studied if they could walk without a gait aid and were excluded if they had any neuromuscular disease, cardiovascular disorders or lower-limb surgeries that would affect their gait or put them at risk while participating.

A control group of 12 similar-aged individuals with varus malalignment was selected (Table 2). The asymptomatic individuals were evaluated by a trained orthopaedic surgeon and were excluded if they had orthopaedic (joint fracture, joint laxity, OA, lower-leg discrepancy and arthritis) or neurological problems that could affect their gait pattern.

In vivo kinematic evaluation

Patients' knee motions were recorded using KneeKGTM system [14, 16], a noninvasive system for 3D analysis of the knee, in a weight-bearing and dynamic condition. It consists of an exoskeleton (Fig. 1) (femoral arch with lateral sensors; tibial sensor attachment and sacral belt); infrared camera (Polaris Spectra camera, Northern Digital Inc.) and a computer equipped with Knee3DTM software suite (Emovi, Inc.). It measures and analyses the 3D position and movement of markers embodied in a harness that is placed on the knee of patients whose movement function is to be assessed. The knee-marker attachment system is designed to reduce skin-motion artifacts [13]. Several studies have assessed its accuracy and reproducibility and validated it [13, 14, 17, 18]. Mean interobserver repeatability value ranged between 0.4° and 0.8° for rotation angles and between 0.8 and 2.2 mm for translation [14, 17]. This clinical tool enables an accurate and objective assessment of the 3D function of the knee joint.

Kinematic analysis of the knee during gait was done with the study participant walking on the treadmill at a comfortable speed chosen by each participant. Firstly, because treadmill

Table 1 Clinical characteristics of the knee osteoarthritis (OA) group

Clinical characteristics of knee OA patients		
Perimeter of walking in daily life (number of patients)	< 0.5 km	4
	0.5–1 km	8
	~1 km	14
	> 1 km	4
Ahlbäck's classification of knee OA (number of patients)	Stage I	–
	Stage II	8
	Stage III	16
	Stage IV	6
Status of ACL (number of patients)	Present	21
	Attenuated	5
	Ruptured	4
Range of motion (<i>n</i> =30) (mean ± SD)	Extension	5.6° (12.6)
	Flexion	110.8° (18.5)

ACL anterior cruciate ligament, SD standard deviation

Table 2 Demographic characteristics of the study groups

	Knee OA group (mean \pm SD)	Control group (mean \pm SD)	Significance (<i>p</i>)
Age	65.73 years (10.0)	61.67 years (3.1)	>0.5
Weight	81.8 kg (14.2)	73.5 kg (9)	<0.05*
Height	167.3 cm (7.2)	166.2 cm (5.7)	>0.5
BMI	29.08 kg/m ² (4.1)	26.5 kg/m ² (1.8)	<0.05*
Side	Right: 14 Left: 16	Right: 6 Left: 6	
Gender	Female: 18 Male: 12	Female: 8 Male: 4	

BMI body mass index, *SD* standard deviation

walking can be unfamiliar, a habituation period of ten minutes was initiated prior to data collection. After installation of the femoral, tibial and sacral trackers, the device was calibrated. The first step of the procedure was localising anatomical landmarks with the electromagnetic localiser (Fig. 2a). The second



Fig. 1 KneeKGTM trackers installed on the knee

step was defining the centre of the hip, knee and ankle joints. The hip-joint centre (HJC) was defined using an optimisation computation method during leg circumduction (Fig. 2b). The knee-joint centre (KJC) was then determined to be the projection of the femoral epicondyle midpoint on the mean helical knee flexion–extension axis (again computed during flexion–extension) (Fig. 2c). The ankle-joint centre (AJC) was defined as the midpoint between the malleoli. Lastly, neutral transverse rotation was set when the knee was determined to be at 0° of flexion during a slight flexion–hyperextension movement (Fig. 2d). A sagittal plane forms from HJC to AJC vector joining; femoral longitudinal axis is defined from HJC to KJC; tibial longitudinal axis is defined from KJC to AJC; projections of longitudinal axes onto the sagittal plane are calculated, as are absolute angle between projections. With the patient in this posture, anterior–posterior axes are defined lying in the sagittal plane perpendicular to longitudinal axes, respectively, for femur and tibia. Mediolateral axes are defined for femur and for tibia perpendicular to the other two axes. Finally, the origin of the axes is fixed at KJC.

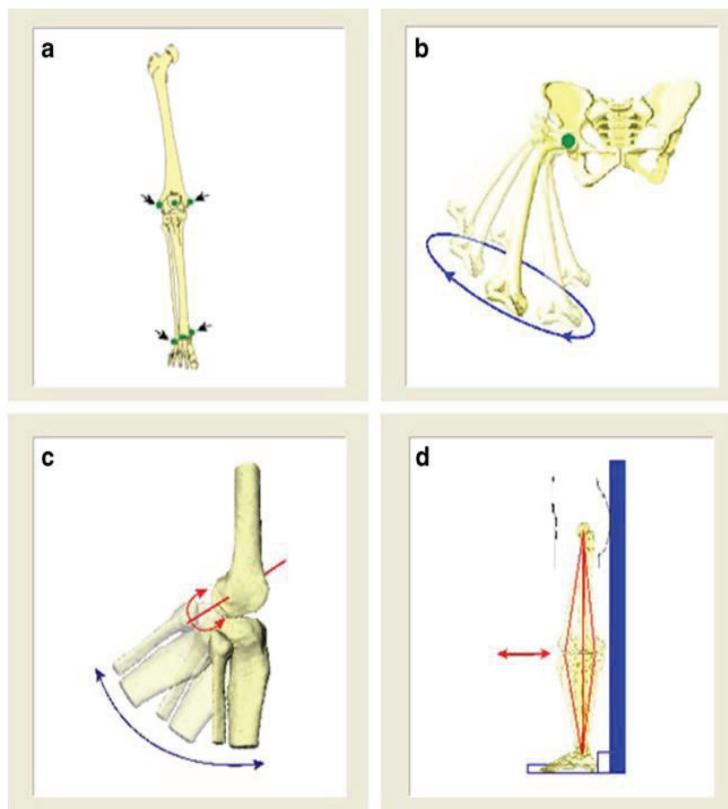
Once system installation and calibration was finalised, the study participant was asked to walk at the preselected speed, and 3D displacements of the reflective markers were recorded by the operator. Once calibration and measurement is performed, the Knee3D suite computes various angle values on anatomical axes between tibia and femur and provides a chart that contains the study individual's curves. The entire procedure lasts 20–25 minutes. After which the trackers were taken off and the individual was dismissed. A database in Microsoft Excel 2010 was created for each participant that contained the four biomechanical patterns consisting of the three knee angles: flexion–extension, abduction–adduction, internal–external tibial rotation (Fig. 3) and anterior–posterior translation.

Using these data, we analysed movement in sagittal, frontal, axial and transverse planes during walking and their correlations with clinical data.

Statistical analysis

The assumptions of normality and equality of variance were assessed using Shapiro-Wilk and Levene's tests, respectively.

Fig. 2 Calibration of measurements: **a** localising anatomical landmarks; **b** defining hip centre; **c** determining knee-joint centre; **d** setting neutral transverse rotation



If data met parametric assumptions, the *t* test was used; if not, nonparametric Mann–Whitney test was used to compare differences in age, height, weight and body mass index (BMI). Analysis of variance (ANOVA) and post hoc (Tukey) tests were used to compare kinematic characteristics. Pearson's correlations (*r*) were used to examine relationships between clinical and kinematic gait parameters. The statistical difference was set as $p < 0.05$. All data were analysed using SPSS v21 (SPSS Inc., Chicago, IL, USA).

Results

Table 3 summarises spatiotemporal and kinematic data for OA and control groups.

Flexion–extension

The range of motion in the sagittal plane was significantly lower for the OA ($40.6^\circ \pm 6.1$) than the control ($52.2^\circ \pm 5.3$; $p < 0.05$) group. Maximum flexion during stance was significantly lower for the OA than the control ($22.4^\circ \pm 8.09$ and $28.1^\circ \pm 7.97$, respectively, $p <$

0.05). group During the gait cycle, we observed a significant difference during terminal stance, which corresponds to maximum knee extension ($p < 0.05$; 8.5° and 4.4° , OA and control group, respectively). During the initial swing phase, OA patients had reduced flexion ($p < 0.05$; 41.9° and 49.4° , OA and control groups, respectively) (Fig. 4).

Adduction–abduction

At initial contact, OA patients showed an adduction angle of $5.7^\circ \pm 7.3$, whereas the control group stayed in a relatively neutral position ($-0.4^\circ \pm 2.8$, $p < 0.05$). This difference was observed during the entire gait cycle (Fig. 5). There was a negative correlation between HKA and range of adduction–abduction ($r = -0.42$, $p < 0.05$), which means that a higher varus deformity is associated with a higher frontal laxity.

Internal–external rotation

Range of motion of internal–external rotation of OA patients was significantly lower than that of the control group ($7.6^\circ \pm$

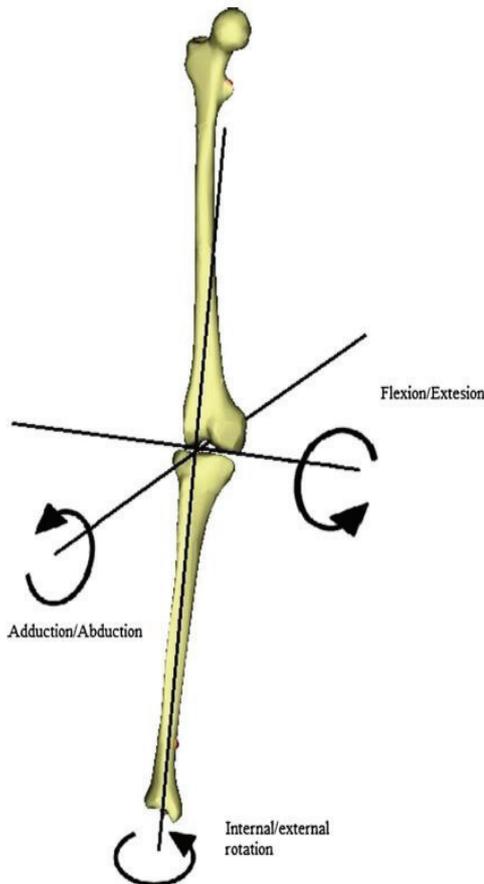


Fig. 3 Knee rotation axes

3.1; $9.3^{\circ} \pm 2.4$, respectively; $p < 0.05$). In terms of internal–external rotation during the gait cycle, the OA group tended to behave in contrast with the control group. During the stance phase, the OA group remained in a relatively neutral position (mean $-0.5^{\circ} \pm 0.4$), whereas the control group showed internal rotation (mean $-2^{\circ} \pm 0.7$), $p < 0.05$. On the other hand, during the swing phase, the control group started to rotate externally, with the peak at 86 % of the gait cycle at midswing (2.1°); the OA group firstly rotated internally then restored the neutral position, with the peak at 93 % of the gait cycle at terminal swing phase (0.5°) (Fig. 6).

Anterior–posterior translation

At initial contact, the OA group were translated significantly more posteriorly ($-2.9 \text{ mm} \pm 5.4$) compared with the control group ($0.4 \text{ mm} \pm 2.21$); $p < 0.05$. In the OA group, the tibia stayed in a more posterior position during the entire whole gait cycle, but significance was reached at loading and midstance phases (Fig. 7).

There was no significant difference in kinematics of OA knees between intact/attenuated and ruptured ACL groups (Table 4).

Discussion

The major finding of this study was that the OA knee group showed an overall stiffening gait strategy by exhibiting reduced motion not only in the sagittal but also in the axial plane. Patients with OA showed a reduction in flexion–extension range. Reductions in flexion excursion have been found in most studies that examined patients with OA knees [10, 19–21]. Heiden

Table 3 Kinematic data of study participants

Kinematic characteristics (mean ± SD)	Knee OA group	Control group	Significance (p)
Speed	1.2 km/h (0.3)	2.1 km/h (0.2)	<0.05*
Flexion angle at initial contact	19° (7.7)	17.4° (12.5)	>0.05
Maximum flexion during stance	22.4° (8.1)	28.1° (7.9)	<0.05*
Maximum extension during stance	7.6° (4.1)	2.2° (4)	<0.05*
Maximum flexion during swing	48.2° (6.3)	54.4° (5.3)	<0.05*
Maximum extension during swing	14.9° (6.5)	12.5° (6.7)	>0.05
Range of flexion–extension	40.6° (6.1)	52.2° (5.3)	<0.05*
Adduction (+)/abduction (–) angle at initial contact	5.7° (7.3)	-0.4° (2.8)	<0.05*
Range of adduction–abduction	7.7° (5)	5.5° (1.6)	>0.05
Internal(-)/external(+) Rotation at initial contact	0.3° (3.6)	-0.1° (2.4)	>0.05
Range of internal–external rotation	7.6° (3.1)	9.3° (2.4)	<0.05*
Anterior (+)/posterior (–) translation of tibia at initial contact	-2.9 mm (5.4)	0.4 mm (2.2)	<0.05*
Range of anterior–posterior translation	7.9 mm (4.1)	9.3 mm (4.4)	>0.05

SD standard deviation, OA osteoarthritis

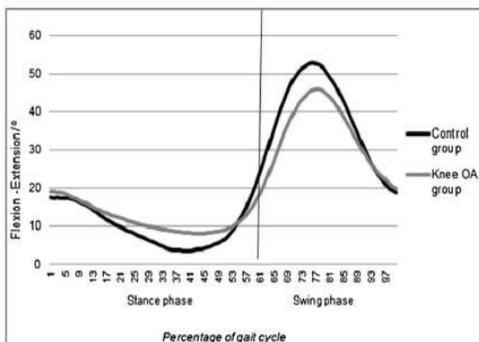


Fig. 4 Knee flexion (+)/extension (-) angle during gait cycle

et al. found no difference in flexion–extension excursion, but they analysed only one phase of the gait cycle, from heel strike to midstance, and included all-stage OA knee patients [22]. Our OA knee group showed a significantly decreased maximum flexion angle during both stance and swing phases, which is in agreement with previous studies [7, 10]. The knee OA group in this study, as in the Schmitt and Rudolph study, showed less extension during the single-limb support-stance phase (34–51 %) than the control group. This lack of extension might be because of greater levels of flexor activation found in the study by Heiden et al. [22] and quadriceps weakness [7, 23] observed at severe OA knees.

Even though the range of adduction–abduction was greater in the OA group, it did not reach statistical significance. On the other hand, there was a positive correlation between varus deformity and adduction–abduction laxity. Results of this study revealed that knee OA patients showed a significantly increased

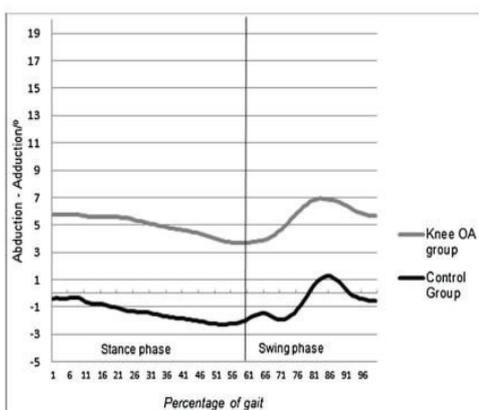


Fig. 5 Knee adduction (+)/abduction (-) angle during gait cycle

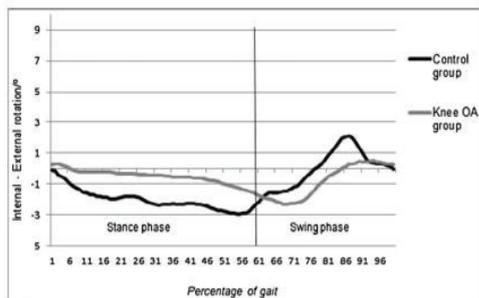


Fig. 6 Knee external (+)/internal (-) rotation angle during gait

knee adduction angle throughout the gait cycle. These data are consistent with the findings of previous studies [4, 10, 12]. This frontal instability associated with increased adduction angle may increase further degradation of the cartilage. Furthermore, Lewek, Rudolph and Snyder-Mackle showed that excessive frontal laxity was observed only on the medial side of the joint and was accompanied by greater medial muscle cocontraction as an effort to control this laxity [19].

In our study, the range of internal–external rotation in the knee OA group was significantly smaller than that of the control group. The same result was observed by Nagano et al. [10], who reported decreased excursion of axial tibial rotation in the OA group. There was no significant difference at the initial contact rotational angle, but OA knees differed from normal knees during the entire gait cycle. Firstly, during the stance phase, the knee OA group displayed a relatively neutral position, whereas the control group rotated internally, reaching a significant difference at midstance. This difference at this gait phase, which corresponds to maximal knee extension, may be due to the lack of full extension of OA knees. Hamai et al. evaluated rotational angles using CT and reported a femoral internal rotation bias

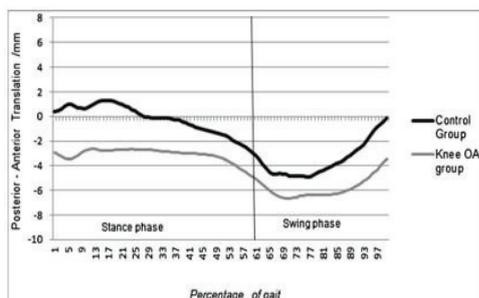


Fig. 7 Tibia anterior (+)/posterior (-) translation (mm) with respect to femur

Table 4 Kinematic data of the knee OA group based on ACL status. There was no significant difference in either plan

Kinematic data (mean \pm SD)	ACL intact	ACL attenuated	ACL ruptured	Significance (<i>p</i>)
Range of flexion/extension	40.9° (6.9)	39.8° (4.4)	40° (2.4)	>0.05
Range of adduction/abduction	7.7° (4.2)	9.4° (8.7)	4.5° (1.2)	>0.05
Range of internal/external rotation	7.1° (2.6)	8.6° (5.2)	8.4° (2.3)	>0.05
Range of anterior/posterior translation	8.2 mm (4.4)	5.69 mm (2.8)	9.03 mm (3.4)	>0.05

OA osteoarthritis, ACL anterior cruciate ligament, SD standard deviation

compared with the control group [3]. Saari et al., using dynamic radiostereometric analysis (RSA), found a decreased internal rotation in the knee OA group. Secondly, during the swing phase, the knee OA group maintained a neutral position at the terminal swing phase, whereas the control group rotated externally. To our knowledge, this is the first study to provide information for external–internal rotation during the swing phase. These data show that the “screw-home” mechanism may be altered in the OA knee. Nagao et al. used ultrasound to measure the rotational angle in OA knees and reported that the rotation of screw-home decreases with progression of knee OA then moved more like a simple hinge joint [24].

There is no consensus in the literature concerning anterior–posterior translation in OA knees. In this study, the knee OA group displayed a significantly increased posterior translation from initial contact to midstance. During the swing phase, the tibia in patients with OA knees remained more posteriorly but to a lesser degree. Siston et al. found no differences in anterior–posterior translation between OA and control groups [2], but in their study, kinematics were evaluated with passive motion, and the control group comprised cadaveric knees, whereas we assessed dynamic motions in vivo. Hamai et al. observed a less posterior femoral translation in patients with medial knee OA [3]. Saari et al. concluded that the increased posterior displacement of knees with arthrosis tended to deviate from normal knees in the same way as previously observed in TKA [12]. There was no significant difference in anterior–posterior translation between intact/attenuated/ruptured ACL in OA knees. However, sample sizes of ruptured and attenuated ACL groups (four and five patients, respectively) were small for drawing frank conclusions. Further investigations with sufficient sample sizes are needed to compare the kinematics in OA knees with different ACL status.

This study has some limitations. First, OA patients were heavier and walked with a slower speed than those in the control group. BMI and speed difference is

consistent with the literature for similar age groups with and without OA [25, 26]. Some studies compared gait trials between groups by selecting a speed usually near 1 m/s. The effect of walking speed on gait measures, particularly in the sagittal plane, has previously been studied [9, 27]. Whereas the “normal” group showed significant difference between 1 m/s and self-selected speeds for all variables, the severe OA group showed no differences between the two speeds. We did not select the speed for patients to walk because we wanted the individual to walk as they do in their normal daily activities. Another point is that gait speed is linked with the presence of knee OA, and it is difficult to separate the confounding effect [7]. Secondly, artifacts from soft tissue could have affected results. Nevertheless, Sati and Larouche showed that skin-motion artifacts are reduced with this harness because it is fixed quasistatically in the thigh and calf [18]. Lustig et al. concluded that this evaluation system provides an objective assessment of the precise biomechanical behavior of the knee [14].

The material presented here provides insight into the profound alterations in knee kinematics during walking as a result of OA progression. This information could be taken into consideration in the development of new methods of OA treatment, either conservative or in the innovative designs of knee prostheses.

Conclusion

This study investigated in vivo kinematics of the OA knee during gait on a treadmill. The knee OA group showed an altered screw-home mechanism by decreased excursion in sagittal and axial tibial rotation, and tibial translation posteriorly. On the other hand, we observed an adduction angle during the entire gait cycle and an increased frontal laxity with increased varus malalignment. Analysing postarthroplasty knee function would be of interest to understand whether the changes described in this study could predict postarthroplasty knee kinematics and whether knee kinematics after arthroplasty return to normal.

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Résumé

La rupture du ligament croisé antérieure (LCA) conduit à une instabilité et à des modifications biomécaniques du genou. Actuellement, les patients présentant une lésion du LCA bénéficient généralement une reconstruction du LCA.

L'analyse cinématique quantitative est un outil important pour acquérir une compréhension approfondie de la fonction articulaire du genou normal et pathologique au cours de la locomotion humaine.

Ce travail est axé sur l'évaluation in vivo de la cinématique du genou chez les patients avec rupture du LCA, avant et après la reconstruction du LCA au cours de toutes les phases de la marche, en utilisant un nouvel outil d'évaluation 3D, quasi-rigide, en temps réel (KneeKGTM).

Dans la *première étude*, des modifications cinématiques significatives ont été identifiées dans les genoux avec une rupture du LCA. Ces changements cinématiques pourraient mener à des contacts fémoro-tibiaux anormaux du cartilage pendant les activités quotidiennes, représentant potentiellement un mécanisme biomécanique de l'usure des articulations après une lésion du LCA.

Dans la *seconde étude*, l'analyse 3D in vivo du mouvement a révélé que les genoux avec LCA reconstruits améliorent significativement leur extension par rapport aux genoux avec LCA déficient, mais il y avait encore une différence par rapport au groupe témoin sain. Dans le plan axial, le tibia reste en position interne significative par rapport à un groupe témoin, tandis qu'il n'y avait pas de différences significatives en translation antéro-postérieur et dans le plan coronal. Ces changements cinématiques pourraient conduire à une charge anormale dans l'articulation du genou et initier le processus d'atteinte dégénérative cartilagineuse à venir.

Toutefois, les données cinématiques post-opératoires ont été collectées 10 mois après la chirurgie, ainsi un suivi plus long serait nécessaire pour évaluer si ces changements cinématiques persistent dans le temps, et leurs effets sur l'articulation.

Mots-clés : Ligament croisé antérieur, Reconstruction, Genou, Analyse de la marche, Cinématique

DISCIPLINE :

Biomécanique

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