



Gait kinematic analysis of the osteoarthritic knee: pre- and post- total knee arthroplasty

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Dedication / Dédicace

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ABSTRACT

Introduction. Patients with knee osteoarthritis tend to modify spatial and temporal parameters during walking to reduce the pain. There are common gait features which are consistently shown to be significantly linked to osteoarthritis severity such as knee adduction moment, knee flexion angle, stiffness and walking speed.

Total knee arthroplasty (TKA) is considered the gold standard treatment for end-stage knee osteoarthritis. Nearly a million of total knee prosthesis are implanted worldwide each year. However, reduced physical function of the knee is partly, but apparently not fully, remedied by surgery.

The purpose of this thesis was to investigate the in vivo, three dimensional knee kinematics during gait at the patients with knee osteoarthritis and the influence of total knee arthroplasty on restoration of normal kinematics.

Material and methods. Thirty patients with medial knee OA and a control group with age-matched subjects were prospectively collected for this study. From the same group 20 patients were re-assessed 11 months after undergoing a total knee arthroplasty with posterior stabilized prosthesis.

All subjects were assessed with a 3D, in vivo, real time device, KneeKGTM, while walking on a treadmill at a self-selected speed. The KneeKG is composed of passive motion sensors fixed on the validated knee harness, an infrared motion capture system (Polaris Spectra camera, Northern Digital Inc.), and a computer equipped with the Knee3DTM software suite (Emovi, Inc.). The whole procedure lasted 20-25 minutes. For each participant, the 4 biomechanical patterns consisting of the 3 knee angles: flexion-extension, abduction-adduction, internal-external tibial rotation and anterior-posterior translation, were calculated.

Results. The patients with knee OA had a reduced extension during stance phase ($p<0.05$) and a reduced flexion during push-off and initial swing phase ($p<0.05$). The adduction angle was consistently greater for the OA patients ($p<0.05$). The 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 the tibial rotation during the midstance phase; OA patients retained a neutral position while control group presented internal tibial rotation.

The patients walked faster after total knee arthroplasty comparing with pre-op assessment ($p<0.05$), but with lower speed comparing with control group. During walking, range of flexion/extension was improved significantly ($p<0.05$) after TKA but it still remained lower than control group. Eventhough there was a visible improvement in the movement in frontal plane after total knee arthroplasty, the difference did not reach the significance. The range of motion in axial plane did not change pre- and post arthroplasty, but remained lower than the matched control group ($p<0.05$). The maximum posterior translation during swing phase was significantly higher at post arthroplasty group comparing with control group, $p<0.05$.

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

Following TKA, patients had better clinical, spatiotemporal and kinametic parameters. They walked longer, faster and with a better range of motion. Despite improvements, the knee kinematics during gait in TKA group differed from healthy control group. They had a lower extension, lower range of axial rotation and an increased tibial posterior translation.

Future research should be focused on comparing different designs of prosthesis pre- and post operatively in a longer follow-up delay.

PLAN

ABSTRACT

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ABBREVIATIONS

AP	Anterior-Posterior
ACL	Anterior Cruciate Ligament
CG	Control Group
CT	Computed Tomography
ER	External Rotation
HKA	Hip-Knee-Ankle Angle
IR	Internal Rotation
LCS	Low Contact Stress
KJC	Knee Joint Center
MRI	Magnetic Resonance Imaging
OA	Osteoarthritis
OP	Operation
PCA	Principal Component Analysis
PCL	Posterior Cruciate Ligament
PCR	Posterior Cruciate Retaining
PS	Posterior Stabilized
RSA	RadioStereometric Analysis
UHMPE	Ultra-high-molecular-weight polyethylene
3D	Three dimensional

CHAPTER 1. INTRODUCTION

1. Introduction

Knee osteoarthritis (OA) represents one of the most prevalent forms of osteoarthritis, with population-based studies estimating severe radiographic disease amongst 1% of 25-34 year olds, and 30% in those aged 75 and above [1]. The high incidence of medial compartment knee OA may be attributable to both anatomical and mechanical factors. Mechanically, functional activities such as gait oblige the medial compartment to bear greater loads than the lateral compartment [2].

Currently, the total knee arthroplasty appears to be the treatment of choice in end stage osteoarthritis in subjects older than 55 years, with severe pain and/or functional problems. Total knee arthroplasty has proven to be a successful and durable solution because the fundamental goal of total knee arthroplasty is to give the patients what they need for their everyday activities: pain relief, a good post-operative range of motion and stability [3, 4]; however, it is still not clear if the restoration of normal knee kinematics is possible.

Quantitative kinematic analysis using advanced motion capture technology has been used as an important tool for thorough understanding of joint function.

This work is focused on the three dimensional, in-vivo kinematics of the osteoarthritic knee before and after total knee arthroplasty during all phases of the gait cycle.

The thesis includes four main parts:

- The first part, the general overview and the literature review of the kinematics of the native, osteoarthritic and replaced knee.
- The second part, the kinematic data during gait of the osteoarthritic knee are compared with the native knee.
- The third part, a comparison of kinematic data of osteoarthritic knee, pre- and post total knee arthroplasty with the control group.
- The last part, a conclusion based on the obtained results is drawn and the perspectives for the next studies is presented.

2. Kinematics Of Native Knee

A. Historical insights

The description of the evolution of spatial position of rigid bodies, without the considerations of the forces involved, is called kinematics. The advancement of the study of locomotion remains dependent on the development of new tools for observation. Over the last several decades, there have been several fundamental advancements that have made a substantial impact on our understanding of the process of knee motion. Flexion and extension are the main movement types of the knee joint and they have been the subject of scientific investigations for over 100 years. Fundamental research on tibiofemoral kinematics was published as early as 1836 by the Weber brothers [5]. For the first time, the kinematics of the knee joint was described as a motion comprising rolling and gliding (Fig. 1). Since that first description, based upon direct visual observation of a cadaveric specimen, several methods have been used to examine the kinematics of the human knee. The axis of knee flexion and extension was derived from the geometry of the femoral condyles, as early as the late 19th century, by analysis of true sagittal plane sections through the femoral condyles. From these sagittal sections, it was clear that the femoral condyles were not circular, but were elongated. The femoral condyles were described as spirals, with the lateral condyle having a greater variation in curvature than the medial condyle. If the femoral condyles were circular, the axis of flexion and extension of the knee would be fixed at its center, like a hinge. The changing curvature of the condyles seen on sagittal sections results in an axis that moves as the knee flexes and extends. This was described as “the instant center of motion” moving along a predictable curved pathway during knee flexion. The instant center of motion (Reuleaux)

model was useful because it linked the shape of the condyles to the motion characteristics of the knee (Fig.2) [6].

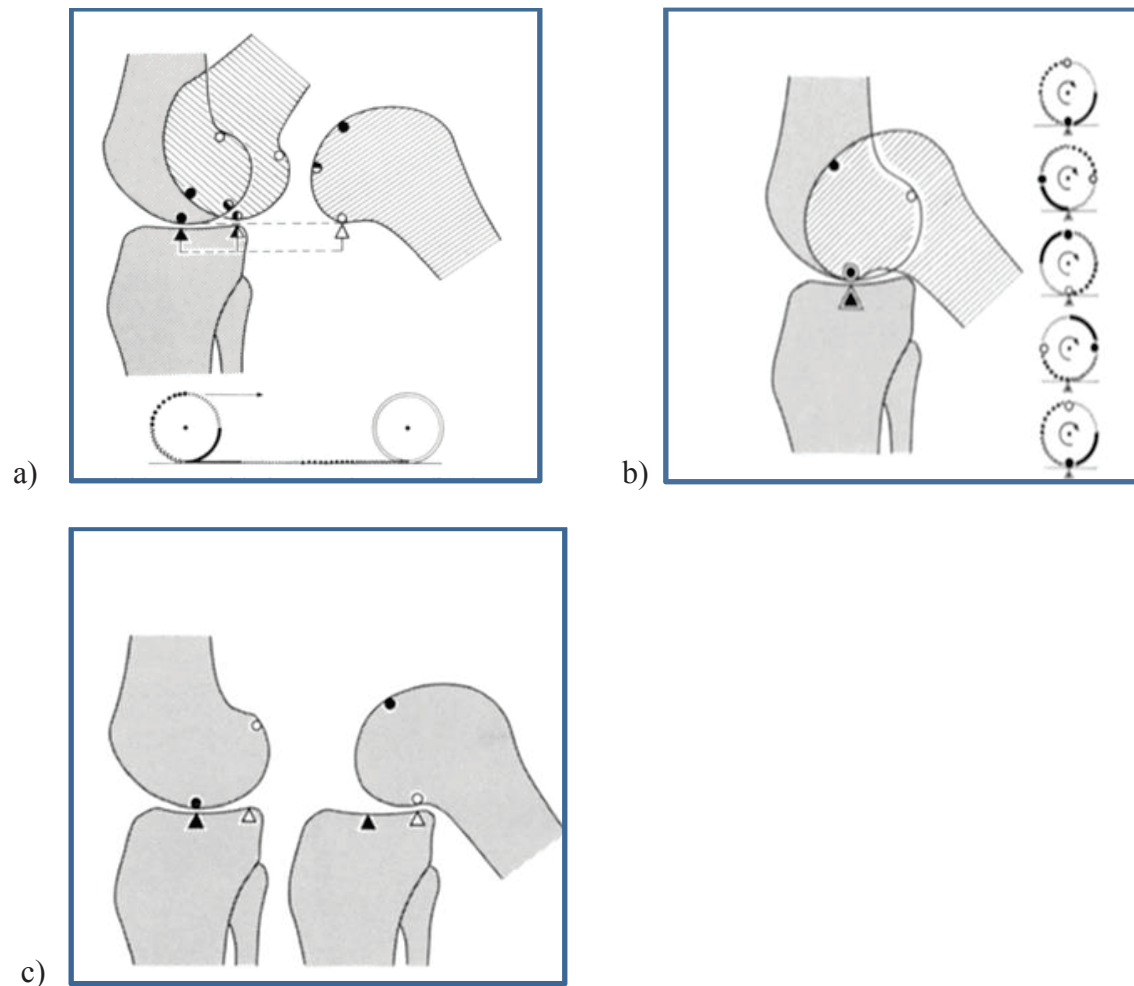


Fig.1. The movement of the femur during flexion/extension. a) the movement of the femur relative to tibia if the movement was a pure rolling, the femur would roll off the tibia; b) the movement of the femur relative to tibia if the movement was a pure gliding, the femur would engage on the posterior rim of tibia; c) the physiologic movement of the femur relative to tibia generated by rolling and gliding [7]

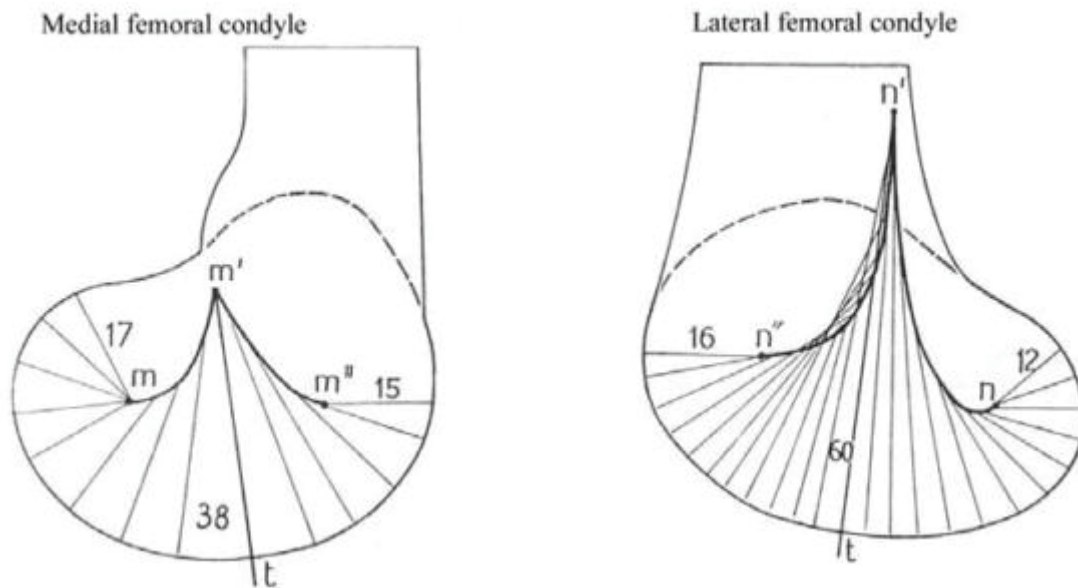


Fig.2. Anatomically sagittal diagram of the medial and lateral femoral condyles. The axis of knee flexion and extension or “instant center” moves as the knee flexes, following a predictable pathway [6]

However, the principal criticism of the instant center theory is that it assumes the flexion and extension movement lies exactly in the sagittal plane. Fick in 1911 reanalyzed the condyle shapes by using 3 rather than 2 dimensions and concluded that the flexion-extension of the knee did not lie in the sagittal plane but was offset by several degrees. This offset orientation of the flexion-extension would result in a single, fixed axis, rather than an instant center. The method of Reuleaux used to map the instant center of motion is correct only if the considered motion lies in a plane. If the plane of movement is offset, then the calculated axis appears to move. However, from Braune and Fisher in 1891, until at least the 1970s, researchers had based their calculations on this assumption.

In the years following, the limitations of this methodology became clear, with the major flaw

being the inability to ascertain the location of the axes of rotation before performing kinematic analyses.

Contemporary movement toward the concept of a fixed flexion-extension axis began in the field of total knee arthroplasty. The dynamic growth in knee arthroplasty in the 1970s required that knee kinematics be understood for prosthetic design.

In 1983, Grood and Suntay [8] presented a joint coordinate system providing a geometric description of the three-dimensional rotational and translational motion between two rigid bodies, applied to the knee joint. With this model, the described joint displacements are dependent of the order in which the rotation components and translations occur.

Hollister [9] essentially described knee motion as pure rotations occurring around two axes: the so called ‘flexion-extension axis’ and the so called ‘longitudinal rotation axis’, with the understanding of the flexion extension axis not being exactly located in the coronal plane and the longitudinal axis not being exactly located in the sagittal plane. As a consequence, these mathematical ‘simple rotations’ meant in reality flexion-extension, varus-valgus and internal-external rotation of the knee joint.

With the improvement of the technology, more advanced tools were possible to be used such as MRI, CT, dual fluoroscopy, dual coil MRI, roentgen stereo photogrammetry, optoelectronic motion capture systems.

These methods have provided a more thorough understanding of the knee kinematics.

B. MRI studies

Hill, Iwaki and coworkers [10] applied MRI scans to 13 unloaded knees and 7 loaded knees for the description of the surface geometry and relative movements of the femur and the tibia. According to this study, during flexion in the unloaded knee, medially, the mean AP position of the femoral condyle did not change from 110° to -5°. Laterally, the femoral condyle rolled

forwards from 110° to 60°, a total of 13 mm, corresponding to 15° of femoral IR (tibial ER) as the knee extended. There was then 1 mm of forward femoral movement, equivalent to 1° of rotation, from 60° to 0°. Finally, the condyle again moved forward 3 mm as the femur internally rotated 4° to 'screw home'. When load is applied it alters tibiofemoral motion in neutral tibial longitudinal rotation. The medial femoral condyle translates forwards about 4 mm between 10° and 45° flexion and the lateral femoral condyle moves backwards further than in the absence of load.

Todo et al. [11] also, analyzed MR images perpendicular to the flexion-extension axis, rather than sagittal images. They concluded that roll back, if present at all, is small, perhaps 2mm, and can be suppressed in either the medial or lateral compartment by the longitudinal rotation of the knee.

In a later paper in 2004, Pinskerova and Freeman [12] reworked these findings and concluded that medially the condyle hardly moves antero-posteriorly from 0° to 120° but the contact area transfers from an anterior pair of tibio-femoral surfaces at 10° to a posterior pair at about 30°. Thus because of the shapes of the bones, the medial contact area moves backwards with flexion to 30° but the condyle does not. Laterally the femoral condyle and the contact area move posteriorly but to a variable extent in the mid-range causing tibial internal rotation to occur with flexion around a medial axis. From 120° to full flexion both condyles roll back onto the posterior horn so that the tibio-femoral joint subluxes.

Nakagawa and co-workers [11] studied unloaded high flexion in 20 Japanese volunteers with an open MRI. Active flexion was measured from 90°-133° and passive flexion from 90° to 162°. They found a mean posterior translation of the medial condyle and lateral condyle of 2 mm and 13 mm respectively from 90-130° of flexion. Pushing the knee further, from 133° to 162° of flexion, caused further posterior translation, medially by 4.5mm and laterally by 15 mm, subluxing the femur behind the tibia.

Johal and co-workers [13] studied the full range of motion of a “loaded squat” (wall supported squat sit) with 10 volunteers, using an interventional MRI. They observed posterior translation of the lateral femoral condyle of 22 mm from hyperextension to 120° of flexion. The medial condyle moved forward 1.7 mm from hyperextension to 30° of flexion and started translating posterior from 90° flexion onwards: 3.6 mm between 90° and 120°, and an additional 8.4 mm from 120° to full flexion. This differential translation on the medial and lateral side leads to an external rotation of 20° of the femur relative to the tibia during flexion.

Beyond 120° of flexion, posterior translation was equal on the medial and lateral side and no further tibiofemoral rotation was occurring.

Although MRI permits imaging of knee motion and has been useful in understanding the motion of the knee transferring the kinematics of these cyclical knee flexion-extension motions to walking or other activities of daily living has not yet been established.

C. RSA Studies

Karrholm et al. [14] confirmed the data published by Hill in an *in vivo* RSA experiment where volunteers performed a step-up activity from 50° to 65° of flexion to full extension. They also found anterior translation of the femoral condyle in early flexion (3 mm from 0° to 50° of flexion). Forced external rotation of the foot suppressed physiologic internal rotation of the tibia with flexion. The main weakness of this study is the variability of the motion that was performed and the fact that they only studied a single motion from moderate flexion to full extension and not a full motion cycle.

On the other hand, this procedure is done while the patient is lying in horizontal position hence it does not allow the 3-D kinematic analysis while performing normal daily activities.

D. Fluoroscopy Studies

Fluoroscopy allows following knee motion in real time *in vivo*, with an obvious drawback of delivering two-dimensional images. A preliminary CT scan allows making a 3D bone model that can convert the 2D fluoroscopic image with a shape-matching technique.

Komistek and co-workers [15] used this technology to study gait and deep flexion activities in five volunteers. They found significantly less translation and rotation of the femur during gait than during deep flexion. They confirmed previous findings of less translation on the medial side than on the lateral side but the most important result of their work seems to be the impressive inter-individual variability. For deep flexion, the medial condyle translation ranged from +3 mm to -9 mm versus +1,4 to -30 mm for the lateral condyle. The average tibiofemoral rotation during flexion was 13°.

Bank's group [16] used CT-derived bone models for model registration and added MRI derived articular surfaces for obtaining higher accuracy of the contact areas. They observed the greatest femoral external rotation during the squat activity, but reported no posterior subluxation of either femoral condyle in maximum knee flexion.

E. Optical tracking

In tracking based approaches, an appropriate marker is affixed to the patient limb and a tracking device traces the markers position. It is very useful to get information about knee kinematics during daily activities like walking, running and squatting. There are two approaches for the positioning of markers on the limbs [7]. Invasive approach is skeletal intracortical pins drilled in the limb with markers in surface. Invasive methods are more reproducible and accurate than non-invasive ones when it comes to recording knee kinematics, but they are usually less accessible and less safe, mainly due to risk of infection. For this reason, non-invasive methods with passive markers are widely used. One approach is

to place markers directly onto the skin, usually over bony anatomical landmarks. The other is to fix a set of at least three markers to each limb segment (rigid body), either directly or placed on a rigid structure. Both of these approaches allow representation of the motion of the body segment, but are subject to skin movement artefacts when movements out of the sagittal plane have to be assessed.

KneeKGTM

In order to reduce skin motion artefacts, a new non invasive knee attachment system has been developed by the Imaging and Orthopaedics Laboratory, University of Montreal Hospital Centre. The group developed a harness to be fixed quasi-statically on the thigh and calf, therefore reducing the skin motion artifact. This harness was shown to be accurate in obtaining 3D kinematic data and was validated for gait applications [17, 18]. The potential of the device to assess 3D knee kinematics in a variety of situations led to the commercialization of the device under the name KneeKGTM.

Another good point of this tool is that it allows a quick (20 min) assessment of knee kinematics and patient examination can be performed in a small assessment room with a treadmill. This advantage allows the use of KneeKGTM on the clinical settings.

F. Normal Gait Cycle

Walking and running are the most common human movements and probably the most complex. The individual walking pattern is a personal identity because each of us performs a characteristic and unique way of walking [19].

The gait cycle is defined as the period from heel contact of one foot to the next heel contact of the same foot. This cycle consists of two parts, stance and swing phase. On average, the gait cycle is about one second in duration with 60% in stance and 40% in swing.

It has eight sub-phases which assist in determining overall coordination of the limbs, and assessing functional tasks associated with walking (Fig.3).

Phase 1- Initail contact (Heel strike). The instant when the foot contacts the ground.

The knee is extended and the ankle is neutral.

Phase 2 – Loading. The limb attempts to absorb shock caused by ground reaction forces, stabilize the limb to bear the weight of the body, and continue forward progression.

The knee flexes rapidly, reaching approximately 15° of flexion.

Phases 3 – Midstance. Represents the first half of single support, which occurs from the 10% to 30% periods of the gait cycle. It begins when the contralateral foot leaves the ground and continues as the body weight travels along the length of the foot until it is aligned over the forefoot. The knee extends until it reaches 3° of flexion by midway through terminal stance (Fig.4).

Phase 4- Terminal stance. The second half of the single support from 30 to 50% of the gait cycle and is defined as the time from heel rise until the other limb makes contact with the floor. During this phase, body weight moves ahead of the forefoot.

Phase 5 – Pre swing . From the time of initial contact with the contralateral limb to ipsilateral toe-off. During this period, the stance limb is unloaded and body weight is transferred onto the contralateral limb. Knee flexes from 7° to 40° of flexion.

Phase 6 - Initial swing phase. It begins at the moment the foot leaves the ground and continues until maximum knee flexion occurs, when the swinging extremity is directly under the body and directly opposite to the stance limb. The knee is at 60° of flexion.

Phase 7 – Mid swing. From the time the swing foot is opposite to the stance limb to when the tibia is vertical. Begins from maximum knee flexion (when the swing limb is under the body) until the swing limb passes the stance limb and the tibia becomes in a vertical position.

Phase 8 – Terminal swing. Is the final phase of the swing period from 85% to 100% of the gait cycle. The tibia passes beyond perpendicular, and the knee fully extends in preparation for heel contact. The knee extends rapidly throughout mid- and terminal swing until peak extension is reached just before initial contact. Peak extension can range from 3° of hyperextension to 5° of flexion.

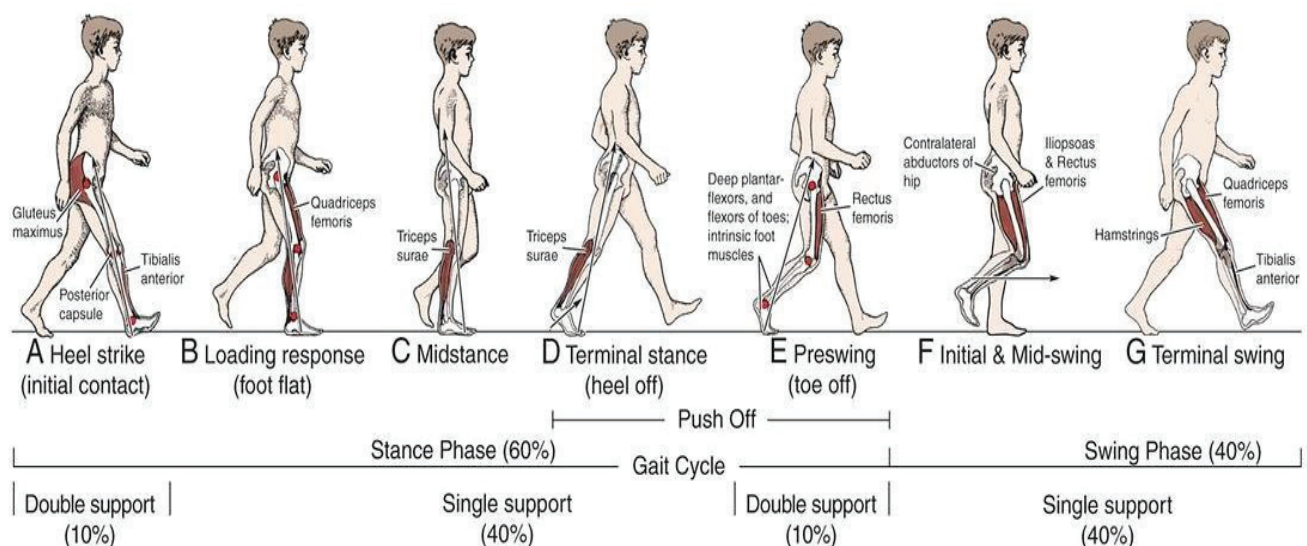


Fig.3. Phases of gait cycle [20]

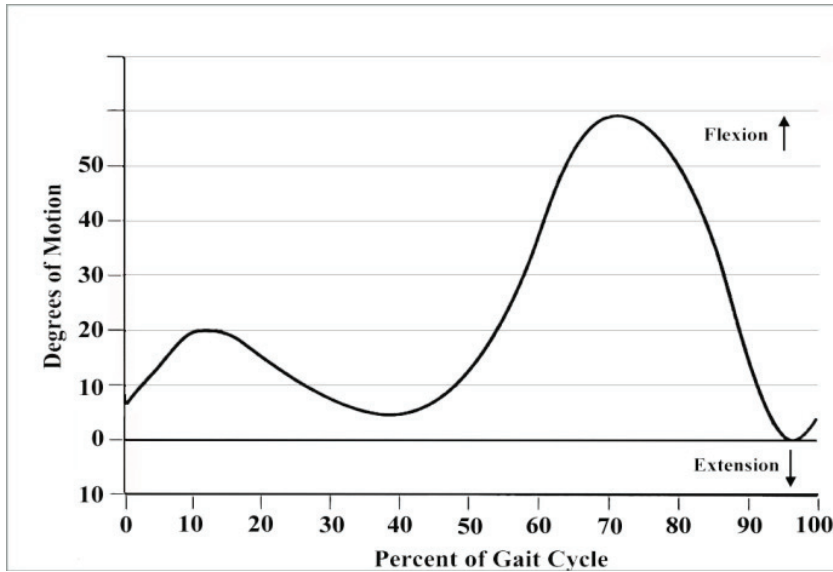


Fig. 4 Healthy knee flexion and extension during gait [21]

Lafortune et al. [22] studied knee kinematics during walking in 5 healthy volunteers using intracortical pins with clusters. They found out that the average pattern of flexion/extension of the tibiofemoral joint during walking was biphasic: a slight flexion followed by an extension during the stance phase and a large flexion also followed by an extension during the swing phase. The average pattern of abduction/adduction of the tibiofemoral joint (rotation around the floating axis) was uniphasic and was limited to 5°. Twice during the stance phase, the tibiofemoral joint rotated internally -as heel strike occurred and again prior to toe-off. During the middle part of the stance phase, the tibiofemoral joint remained close neutral position. From toe-off until heel strike, the tibiofemoral joint rotated externally.

Andriacchi et al. [23] studied the anterior posterior (AP) motion of the knee during walking. At heel strike with the knee at full extension, the tibia is at its maximum anterior position during the gait cycle. The next key event occurs at terminal extension where the tibia is located posteriorly again while the knee is near full extension. Thus, heel strike and terminal extension provide two events where the tibia is located over a range of AP positions relative to the femur.

G. Conclusion

On the basis of the published data, one can conclude the normal kinematic pattern of the human knee consists of internal rotation of the tibia relative to the femur with increasing flexion, following a greater posterior translation of the lateral femoral condyle than of the medial femoral condyle. On the other hand, gait analysis has allowed researchers and clinicians to better understand biomechanical factors of gait in healthy participants and those with lower limb pathology [24]. However, different methodologies seem to discover different kinematic patterns. Moreover, it remains unclear to what extent the natural kinematic patterns can be modified after post-traumatic or degenerative disease of the knee joint and to what extent the prosthetic surgery replaces the functional anatomy.

3. Kinematics Of Osteoarthritic Knee

A. Overview Of Knee Osteoarthritis

Osteoarthritis is a degenerative process that typically affects the synovial joints of the body. WHO estimates that osteoarthritis (OA) affects 9.6% of men and 18% of women older than 60 years of age [25]. Increases in life expectancy and ageing populations are expected to make OA the fourth leading cause of disability by the year 2020 [26]. It leads to social, psychological and economical burdens in patients with substantial financial consequences [27]. In France, the cumulated health costs resulting from OA almost doubled within 10 years (from 1993–2003) [28] .

In recent years, it is shown that not only cartilage, but also the subchondral bone, ligaments, the synovial fluid, and surrounding muscles are involved in the OA process. Although the exact aetiology is still unknown, OA is in general characterized by loss of articular cartilage, osteophyte formation, and subchondral bone sclerosis [29].

Knee OA represents one of the most prevalent forms of osteoarthritis, with population-based studies estimating severe radiographic disease amongst 1% of 25-34 year olds, and 30% in those aged 75 and above [1]. The diagnosis of the knee OA is made at first by clinical examination. Pain, morning stiffness, swelling and crepitus at one patient older than 50 years may be consequences of the knee osteoarthritis. During the examination, it is important to analyze the deformity of the lower limbs and its reductibility, the ligamentary status and the range of motion. Imaging modalities play an important role in the knee OA, since they confirm the diagnosis, determine the involved compartment and evaluate the stage of the disease. Conventional radiography (Fig.5) is still today the technique of reference for evaluating knee OA. The main radiological signs are joint space narrowing corresponding to loss of cartilage; the osteophytes which represent marginal bone reaction to loss of cartilage;

subchondral bone reactions, geodes or bone condensation [30]. There are two main classifications of the structural changes associated with knee OA (Tab.1), Kellgren and Lawrence composed criteria for a 5-point grading scale using radiographic features [31]; and Ahlbäck scaled the knee OA from stage 0 (no OA) to stage V (bone defect/loss >10 mm, often with subluxation and arthritis of the other compartment) [32] .

Kellgren-Lawrence classification	
<i>Grade 0</i>	Normal
<i>Grade 1</i>	Doubtful narrowing of joint space and possible osteophytic lipping
<i>Grade 2</i>	Definite osteophytes and possible narrowing of the joint space
<i>Grade 3</i>	Moderate multiple osteophytes, definite narrowing of joint space and some sclerosis, possible deformity of the bone ends
<i>Grade 4</i>	Large osteophytes, marked narrowing of joint space, severe sclerosis, and definite deformity of bone ends
Ahlbäck classification criteria	
<i>Stage 0</i>	No radiographic sign of arthritis
<i>Stage I</i>	Narrowing of the joint space (JSN) (with or without subchondral sclerosis). JSN is defined by a space inferior to 3 mm, or inferior to the half of the space in the other compartment (or in the homologous compartment of the other knee)
<i>Stage II</i>	Obliteration of the joint space
<i>Stage III</i>	Bone defect/loss <5 mm
<i>Stage IV</i>	Bone defect/loss between 5 and 10 mm
<i>Stage V</i>	Bone defect/loss >10 mm, often with subluxation and arthritis of the other compartment

Tab.1. The radiographic classifications of osteoarthritis

According to Petersson et al., Ahlbäck and Kellgren and Lawrence classifications show a good correlation [33]. Moreover, Ahlbäck classification seems easy to apply and suitable for the assessment of medial compartment arthritis of the knee; thus, it seems particularly useful for the orthopaedic treatment of knee disorders [34].

Whilst the symptoms and diagnosis of the disease have been clearly defined in medical research, the underlying causes are not yet fully understood. It is thought that biomechanical factors play a key role in OA aetiology. Furthering the understanding of these factors could lead to better treatments and help reduce prevalence through preventative measures.

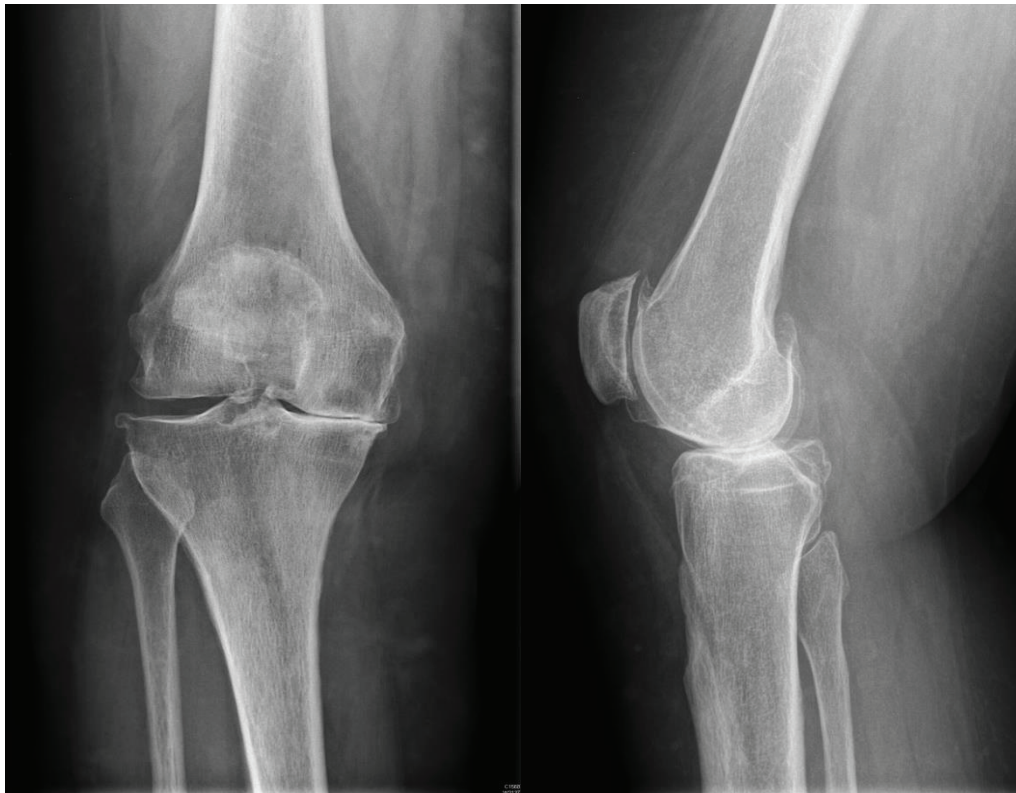


Fig. 5. The radiography of an osteoarthritic knee

B. Kinematics Of Osteoarthritic Knee – Literature Review

Over the past 20 years, there have been numerous investigations into the effects of OA on knee kinematics and also into what biomechanical factors may be causing the initiation of the disease. Kinematic data allow physicians to obtain and process accurate objective measurements of sophisticated movement such as human walking.

Hamai et al. [35] studied knee kinematics in 12 subjects with medial OA during three activities using dynamic imaging and model-image registration with CT-derived models. They concluded that knees with medial OA differed from normal knees. First, they displayed a femoral internal rotation bias of about 8° compared to normal knees: femoral external rotations of 4° and 15° during squatting (20° and 100° flexion, respectively) were less in knees with medial OA than the 12° and 24° observed in normal knees. Second, the natural screw-home movement was not observed in knees with medial OA, perhaps because they did not reach full extension in squatting or stair climbing. Finally, femoral condylar contact in knees with medial OA did not exhibit significant AP translation between 30° and 80° for either squatting or stair climbing.

Saari et al. [36] studied the kinematics of the knee during weight-bearing active extension in 14 patients with medial osteoarthrosis (OA) and in 10 controls using dynamic radiostereometry. They found out that between 50° and 20° of extension, the OA knees showed decreased internal tibial rotation corresponding to less posterior displacement of the lateral femoral flexion facet center.

Matsui et al. [37] evaluated the rotational deformity at 150 osteoarthritic knees compared with 31 control knees using CT scans. Results of the this study indicate that the varus deformity in OA of the knee is associated with significant rotational deformity. The tibia tended to locate in an externally rotated position in the knees with severe varus deformity.

Siston et al. [38] performed a study where a surgical navigation system measured normal passive kinematics from 7 embalmed cadaver lower extremities and in vivo intraoperative passive kinematics on 17 patients undergoing primary total knee arthroplasty. Osteoarthritic knees displayed a decreased screw-home motion and abnormal varus/valgus rotations between 10^0 and 90^0 of knee flexion when compared to normal knees. The anterior–posterior motion of the femur in osteoarthritic knees was not different than in normal knees.

In 2001, Kaufman et al. [39] performed a large study on 139 adults diagnosed with early stage knee OA and compared their gait characteristics against those of 20 healthy control. They found that, for level walking, the OA sufferers had 6° less peak knee flexion than healthy controls. OA subjects also showed significantly lower knee extensor moments which were conjectured to be a method of minimising pain in the joint. Cadence was also significantly reduced during walking for the OA subjects. The stair ascent and descent exercises did not show any differences in the range of joint motion, however speed and knee extensor moment was found to be significantly smaller for OA sufferers for both activities.

Briem and Snyder-Mackler in 2009 [40], looked at the inter-limb differences in 32 patients with moderate medial knee OA. Asymmetry was seen between affected and unaffected knees for flexion and adduction. Knee on the involved side had a significantly smaller flexion and greater adduction angle than on the uninvolved side during weight acceptance.

Mundermann, Dyrby and Andriacchi in 2005 [41] performed a gait analysis of 42 patients with bilateral medial knee osteoarthritis and they observed that all patients with knee OA made initial contact with the ground with the knee in a more extended position than that of the control subjects.

Nagano et al. (2012) [42] in their kinematic gait study of 45 patients with different stages of knee OA and 13 healthy subjects, also found out similar results. The flexion angle at the time of foot contact was significantly less in patients with severe and moderate osteoarthritis than

in normal subjects. The abduction angle at the 50% stance phase was significantly less in patients with severe osteoarthritis than in normal subjects. The excursion of axial tibial rotation was significantly less in patients with early osteoarthritis than in normal subjects.

On the other hand, Heiden, Lloyd and Ackland [43] revealed different results in their study of gait parameters of 54 patients with knee OA. Gait differences in the knee osteoarthritis patients were greater knee flexion at heel strike and during early stance along with reductions in the peak external knee extension moment in late stance[43].

Childs et al. [44] compared a group of 24 knee OA sufferers against a group of 24 healthy controls. Gait recording was performed using an electromagnetic system, a force plate and a surface EMG system with subjects performing both walking and step descent tasks. During the walking task, they also found that OA sufferers had a higher knee angle at heelstrike and also had a lower knee flexion range of motion in the loading response phase of stance .

Following on from the work of Astephen in 2004 [45], Deluzio and Astephen in 2007 [46] used a group of 50 end-stage knee OA patients to look at the gait waveform data of three variables; knee flexion angle, flexion moment and knee adduction moment. As with the previous investigation, a force plate and optoelectronic system was used to collect gait data and a control group of 63 healthy subjects was also used. The authors then used principle component analysis (PCA) to compare the two groups. OA patients knees were less flexed throughout the gait cycle than the controls, and also they had less range of motion in the joint. OA subjects were also shown to have a smaller range of flexion moment during gait and a lower flexion moment during the first half of the stance phase. A lower adduction moment in early stance was also shown in the OA subjects.

Astephen et al. [47] in 2011 investigated the associations between joint biomechanics and neuromuscular control for moderate OA, looking at the differences between radiographic changes and pain severity. Data were collected on a group of 40 OA patients (with a range of

severities) using an optoelectronic system, force plate and EMG system. Radiographic OA severity was found to be correlated with knee adduction moment during stance and maximum knee flexion angle over the whole cycle with higher knee adduction moments and lower maximum flexion angles associated with more severe OA.

In 2007, Landry et al. [48] used the PCA technique previously developed [55] to look at the effect of walking speed on OA. An optoelectronic system and force plate was used to collect data, and 41 patients with radiographic grade 1-3 on the KL scale were compared against 43 asymptomatic patients. Two gait speeds were analysed: self-selected and 150% of self-selected speed. They found that the OA patients had similar stride characteristics and joint kinematics to the control group. This does not agree with the majority of the literature.

Lewek, Rudolph and Snyder-Mackler in 2004 [49] studied control of frontal plane knee laxity during gait in patients with medial compartment knee osteoarthritis. Twelve subjects with genu varum and medial compartment knee osteoarthritis (OA group) and 12 age-matched uninjured subjects underwent stress radiography to determine the presence and magnitude of frontal plane laxity. All subjects also went through gait analysis with surface electromyography of the medial and lateral quadriceps, hamstrings, and gastrocnemius to calculate knee joint kinematics and kinetics and co-contraction levels during gait. The OA group showed significantly greater knee instability, medial joint laxity, greater medial quadriceps - medial gastrocnemius (VMMG) co-contraction and greater knee adduction moments than the control group. Also, the OA group had a knee flexion significantly less than excursion of the healthy control subjects.

More recently, in 2013, Baert et al. [50] assessed the gait adaptations of the knee with early and established OA in comparison with a control group. Fourteen female patients with early knee OA, 12 female patients with established knee OA and 14 female control subjects participated in the study. The gait parameters were acquired using 3D LED motion analysis

system. None of the kinematic variables were significantly different between the early OA and control group. Early OA patients showed significantly less knee adduction in stance phase and more maximal knee extension in late stance than established OA patients. In stance phase, established OA patients showed significantly more knee adduction and less late stance maximal knee extension than controls.

Conclusion: This review of the existing literature on the links between osteoarthritis and gait kinematics has highlighted several areas of interest. There are common gait features which are consistently shown to be significantly linked to osteoarthritis severity such as knee adduction moment, knee flexion angle, stiffness and walking speed.

Most previous studies of changes in OA patients during gait were focused on spatiotemporal parameters [51, 52]; demonstrating that knee OA patients walked slower, with a reduced stride length, and a lower single-limb support compared to controls. Some studies have examined kinematic alterations during phase-specific gait cycle and reported a decreased knee excursion during flexion, decreased peak flexion during stance phase and increased knee flexion at heel strike [41, 47]. Although the kinematic changes in sagittal plane have been enlightened, there are still contradictions. On the other hand there are only few studies that observed the changes in frontal and axial planes and anterior-posterior (AP) translation which still remain unclear.

Authors	Subjects	Method	Activity Studied
Saari T et al 2005	14 patients medial knee OA 10 control group	Dynamic radiostereometry	Active extension
Matsui et al. 2005	150 medial OA knees 31 control knees	CT scans	Passive Flexion-Extension
Siston et al 2006	17 patients OA 7 cadaver normal knees	a surgical navigation system	Passive flexion- extension
Hamai S et al 2009	12 subjects medial knee OA	CT scans	kneeling, squatting, stair climbing
Kaufman et al. 2001	139 patients knee OA 20 control group	Reflective markers, six video cameras	-walking -stair ascent -stair descent
Briem et Snyder-Mackler 2009	32 patients with medial knee OA -interlimb differences	Optoelectric motion analysis	-gait, stance phase
Mündermann, Dyrby and Andriacchi 2005	42 patients with medial knee OA 42 control group	Reflective markers	-gait, stance phase
Heiden , Lloyd, Ackland 2009	54 patients knee OA 30 control group	Reflective cluster markers	-gait, stance phase
Deluzio, Astephen 2007	50 patients knee OA 63 control group	Optoelectric system	-gait
Nagano et al. 2012	45 patients knee OA 13 control group	Reflective cluster markers	-gait, foot contact and 50% of stance phase

Childs et al. 2003	24 patients knee OA 24 control group	Electromagnetic motion analysis system	-gait, stance phase
Landry et al. 2007	41 patients knee OA 43 control group	optoelectronic system	-gait
Lewek, Rudolph and Snyder- Mackler 2004	12 patients medial knee OA 12 control group	-stress radiography -surface electro- myography	-gait, stance phase
Baert et al. 2013	14 female early knee OA (Early OA) 12 female established knee OA (Estab. OA) 14 female control group (CG)	3D motion analysis (LED)	-gait

Tab.2 Material and methods characteristics of the studies included in the review

Author	Flexion/ Extension	Internal/Extrenal Rotation	Adduction/ Abduction	Anterior/Posteri or Translation
Saari 2005	N.S	↓ tibial IR	/	↓
Matsui 2005	/	↑ tibial ER	/	/
Siston 2006	/	↓ screw-home mechanism	/	N.S
Kaufman 2001	↓ peak motion	/	/	/
Briem et Snyder- Mackler 2009	↓ flexion	/	↑ adduction	/
Mündermann, Dyrby and Andriacchi 2005	↓ flexion at heel strike	/	/	/
Heiden , Lloyd, Ackland 2009	↑ flexion at heel strike	/	/	/
Deluzio, Astephen 2007	↓ flexion	/	/	/
Nagano et al. 2012	↓ flexion	↓ tibial ER at heel strike	↓ abduction	/
Childs et al. 2003	↑ flexion at heel strike	/	/	/
Landry et al. 2007	N.S	/	/	/
Lewek,Rudolph and Snyder-Mackler 2004	↓	/	/	/
Baert et al. 2013	↓ extension at stance	/	↑ adduction	/

Tab. 3. Reported results of the studies comparing knee kinematics within OA groups or with a control group.

4. Kinematics Of Replaced Knee

A. Overview Of Knee Replacement

Nearly a million of total knee prostheses are implanted worldwide each year. Approximately 600,000 knee replacements are performed every year in the United States of America [53], more than 70,000 in France [54].

Attempts to replace the knee joint with an arthroplasty have been made for at least 140 years. In 1860, Verneuil suggested “the interposition of tissues between resected bone to prevent fusion”. Jules Emelie Péan, one of the leading surgeons in Paris, described in 1894 the first attempt with an articular prosthesis of metal of humerus. In 1938, Venable and Struck presented an important work when they significantly improved the quality of vitallium and certain steel alloys. This raised the success rate for all metal implants and meant a significant step forward in the development of implant surgery.

It was not until the late 60s that the first series of replacements of the knee with metal implants emerged and the early 70s saw the rise of prostheses of the knee.

In 1971, Gunston [55] importantly recognized that the knee does not rotate on a single axis like a hinge but rather the femoral condyles roll and glide on the tibia with multiple instant centres of rotation. His polycentric knee replacement had early success with its improved kinematics over hinged implants but failed because of inadequate fixation of the prosthesis to bone.

The Total Condylar prosthesis was designed by Insall at the Hospital for Special Surgery in 1973. This prosthesis concentrated on mechanics with the intention to create a knee replacement with kinematic characteristics as similar as possible to the normal knee.

Many studies are published since that time and the prostheses have continued to evolve.

Nowadays, there is a wide diversity of total knee prosthesis designs, developed with specific properties and with a specific patient group in mind and therefore each one has its own theoretical advantages and disadvantages. However, there are some main implant variations.

A set of knee-prosthesis components consists of: an anatomically shaped distal ***femoral component*** made of Cobalt-Chrome alloy which is a very hard and durable material, allowing it to withstand the massive loads and cycles a knee endures on a daily basis; a ***tibial component*** that is usually composed of two main pieces, the tibial tray made of Titanium or Cobalt-Chrome and the tibial bearing component made of ultra high molecular weight polyethylene (UHMWPE); and the optional ***patellar component*** is also made of plastic (UHMWPE).

The medial collateral ligament and lateral collateral ligament are critical in holding the joint in place and producing joint motion. Implantation of prosthesis typically requires removal of the anterior cruciate ligament and, depending on the prosthesis design, may also involve removal of the posterior cruciate ligament. The posterior cruciate ligament (PCL) is an important structure that stabilizes the knee joint. The cruciate retaining prosthesis enables the surgeon to preserve the ligament. It has a small groove that helps the ligament continue to provide flexion. When PCL is removed, the posterior stabilized prosthesis is used. These implants employ a “cam and post” system that substitutes for the posterior ligament and provides support on the posterior part of the knee (Fig.6).

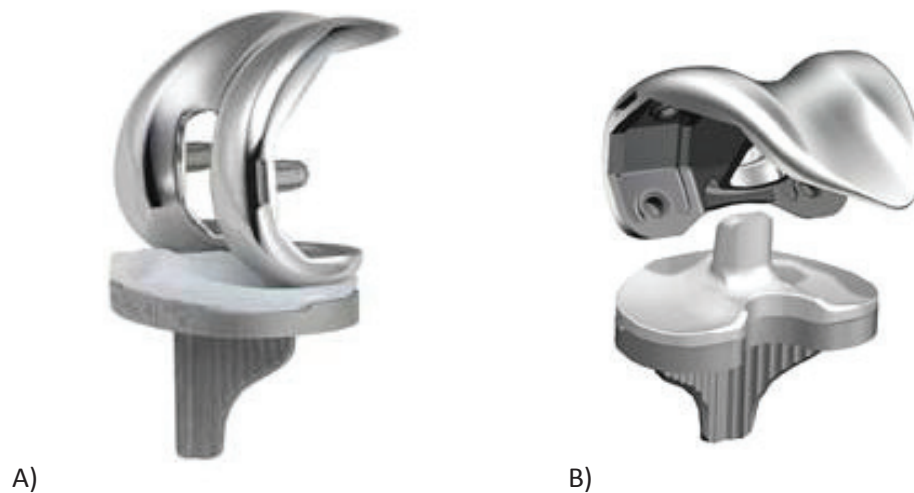


Fig.6. A) PCLR prosthesis and B) PCLS prosthesis, note the post and cam mechanism of the PCLS implant, which prevents the posterior translation of the tibia.

For fixation, most predominantly knee prostheses are cemented to the bone with polymethylmethacrylate but some other types of cementless prosthesis, instead of cement, have a special surface that encourages bone to grow into the implant for fixation.

Current TKR devices can be subdivided into two groups based on different fundamental design principles: fixed-bearing knees, in which the UHMWPE insert is snapped or press fitted into the tibial tray, and mobile-bearing designs which facilitate movement of the insert relative to the tray (Fig.7).

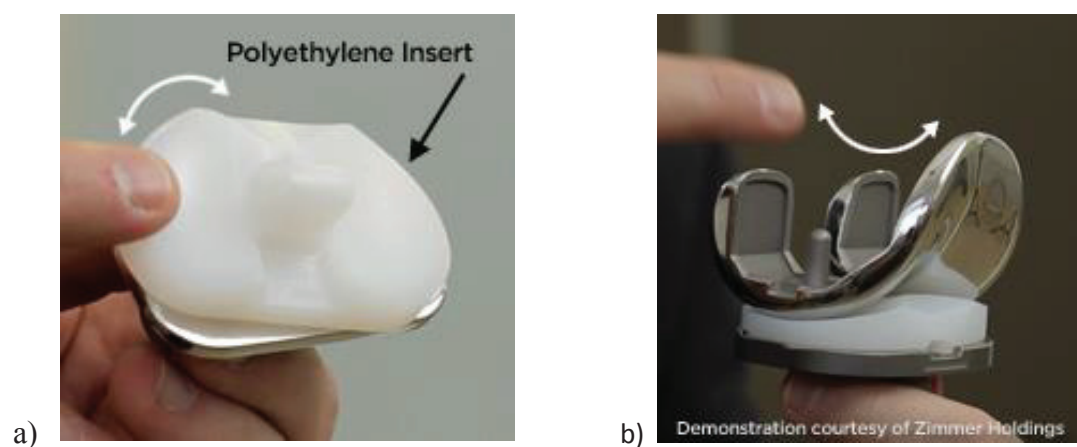


Fig.7. a) Mobile bearing and b) Fixed bearing

HLS KneeTec Prosthesis (Tornier), the prosthesis used in our study, is a posterior stabilized total knee replacement (Fig.8) with rotatory platform without conservation of cruciate ligaments. Its distinctive feature is the system of posterior stabilization: a third condyle in the midline of the intercondylar notch that permits progressive contact with the tibial cam from 35° flexion, thus ensuring stability very early on in flexion. This causes roll-back of the femoral condyles that optimizes quadriceps function, reduces load on the extensor system and improves flexion. Finally, the third condyle constitutes an additional point of tibiofemoral contact, which permits better load distribution on the polyethylene [56].

The trochlear groove is convex-dome with a constant radius along its trajectory, giving better congruence on engagement and therefore stability throughout all movement of the knee.

The tibial axis of rotation of the mobile platform is located posterior to the midline to restore a movement close to anatomic rotation. An anterior circular rail allows for 30° range of axial rotation.



Fig.8 The HLS KneeTec Prosthesis

B. Kinematics of Replaced Knee

There is a long-standing controversy on which type of total knee prosthesis provides better kinematics and clinical outcome. Several studies have analyzed the kinematics of total knee prosthesis, comparing different designs or with a control group during passive movements.

Dennis et al., in 1998 [57], have studied the effect of implant designs in range of motion after total knee arthroplasty. The range of motion of twenty patients with posterior cruciate retaining (PCR) prosthesis; 20 patients with posterior substituting (PS) prosthesis and 20 control group were obtained using video fluoroscopy. The normal knee group exhibited superior flexion over either TKA subgroup, whether measured under passive non-weightbearing or active weight-bearing conditions. Maximal mean postoperative flexion for PCR and PS TKA groups was similar when evaluated under passive non-weightbearing conditions. When measured under weight-bearing conditions, patients implanted with PS TKA exhibited significantly greater mean range of motion than those with PCR TKA.

Siston et al., in 2006 [38], measured intraoperatively using surgical navigation system passive kinematics of 17 patients pre- and post- TKA and compared it with that of 7 normal cadaver knees. Throughout the range of flexion, no systematic relationship of varus/valgus rotation angle with flexion was present in any knee following TKA, resulting in a motion pattern significantly different from normal knees. The screw-home motion following TKA was significantly less than the screw-home motion in the OA knees. AP femoral translation was significantly different in the knees following TKA compared to normal knees. Following TKA, the femur translated anteriorly on the tibia until approximately 60° of flexion before beginning a posterior translation.

Yoshiya et al., in 2005 [58], realized a study where in vivo comparison of flexion kinematics for posterior cruciate- retaining (PCR) and posterior stabilized (PS) total knee arthroplasty (TKA) was performed. In the PCR TKA, an anterior femoral translation from 30° to 60° of

flexion was observed in the weight-bearing condition. In contrast, flexion kinematics for the PS TKA was characterized by the maintenance of a constant contact position under weight-bearing conditions and posterior femoral rollback in passive flexion.

In 2008, Cates et al. [59] also analyzed the kinematics from full extension to maximum flexion for 30 subjects (15 PS, 15 PCR) using fluoroscopy. Ranges of motion were not statistically different between the 2 implant designs. In comparing the PCR group to the PS group at each flexion angle (ie, at 0°, 30°, 60°, 90°, and 120°), no significant differences were found in axial rotation angles. The PCR knees had a significantly larger mean axial rotation angle than the PS knees at maximum flexion.

Other studies analyzed the kinematics of the replaced knee during daily activities as walking, stair climbing, knee bending.

Haas et al., in 2002 [60], investigated in vivo kinematics of 10 subjects with either posterior stabilised (PS) or posterior cruciate substituting (PCS) mobile bearing TKAs during gait and during a knee bend from 0° to 90° flexion. This study showed that the kinematic patterns for subjects having either a PS or PCS mobile bearing TKA were similar during gait but subjects having a PS TKA experienced more posterior femoral rollback of the lateral condyle during the deep-knee bend.

In 2011, Hatfield et al. [61] investigated 3D kinematic and kinetic gait patterns of 42 patients with severe knee osteoarthritis, collected 1 week prior and 1 year post-TKA. Overall and midstance knee adduction moment magnitude decreased while knee flexion angle magnitude increased during swing. Increases in the early stance knee flexion moment and late stance knee extension moment were found, indicating improved impact attenuation and function. A decrease in the early stance knee external rotation moment indicated alteration in the typical rotation mechanism.

Moro-oka et al., in 2007 [62], compared knee kinematics in 9 patients with bi-cruciate preserving total knee arthroplasty (ACL/ PCL knees) and 5 patients with posterior cruciate ligament (PCL) preserving total knee arthroplasty. They studied treadmill gait, stair stepping, and maximum flexion activities using lateral fluoroscopy and shape matching. The ACL/PCL knees showed greater knee flexion, greater tibial internal rotation and greater posterior condylar translation than the PCL knees all through the gait cycle. For the stair activity, posterior translations of the lateral condyle were significantly greater in the ACL/PCL knees from 30° to 70° flexion.

Argenson et al., in 2006 [63], studied in vivo kinematics of the femorotibial joint during a deep knee bend using fluoroscopy for 20 subjects having a TKA designed for deep flexion. The average weight-bearing range of motion was 125°. On average, subjects in this study experienced 5.4° more internal rotation of normal axial rotation.

In 2007, van der Linden et al. [64] evaluated the knee kinematics in functional activities seven years after TKA. Nineteen patients with knee osteoarthritis were assessed using electro-goniometry before surgery, 18–24 months and seven years after total knee surgery. Maximum knee angle during the swing phase while walking on a level surface increased at 18–24 months after surgery but decreased again seven years after surgery. Patients used a significantly greater range of motion of the knee during ascending and descending a flight of stairs, seven years after surgery compared to 18–24 months after surgery.

Recently, in 2012, Joglekar et al. [65] investigated the gait of 18 subjects with either a PS or PCR TKA and sacrificed PCL and compared with the normal contralateral knee using a passive reflective arrays and an optoelectric system. The data for the operated knees for stair descent demonstrated significant differences in maximum overall knee flexion angle; maximum stance phase knee flexion angle; maximum swing phase knee flexion angle;

average swing phase knee flexion angle with the PS knees showing higher flexion angles than the PCR knees.

One of the few studies that included coronal and transverse planes is that of McClelland et al., in 2011 [66], who compared the knee kinematics of 40 patients following TKR and 40 unimpaired controls during comfortable and fast walking speeds using three dimensional motion analysis.

In the sagittal plane, the TKR group walked with less knee flexion during stance and swing phases. The TKR group also had less knee flexion at initial contact during fast speed walking but the reduced knee flexion at initial contact during comfortable speed walking did not reach significance. There were no differences in the coronal plane kinematics between groups. In the transverse plane, the TKR group walked with significantly less internal rotation and significantly more external rotation compared to the control group.

Conclusion. Evidence from multiple studies included in this review indicates that knee kinematics after total knee arthroplasty differ from normal healthy controls. TKA patients walked with less total range of knee motion than their control counterparts.

The range of flexion during the loading phase of stance was also reduced compared to controls. Peak knee flexion during weight acceptance and knee flexion excursion are less in the operated knee after TKA than in healthy controls.

Substantial differences were also identified in the characteristics of the knee replacement prostheses designs used in the various studies. There is ongoing discussion in the literature about the biomechanical effects of either retaining or resecting the PCL in TKA.

However, there is a lack in literature in analyzing the knee kinematics during gait in frontal, transverse and axial plane. Also most of the studies compare designs of the prosthesis or the TKA group with the healthy control subjects but there are only few studies that investigated the pre- and post-TKA knee kinematics.

Authors	Subjects	Methods	Activity studied	Prosthesis design
Dennis et al. 1998	40 patients with TKA 20 control group	Video fluoroscopy	Passive flexion-extension	PCR vs PS prosthesis
Siston et al. 2006	17 patients with TKA 7 cadaver knees	Surgical navigation system	Passive flexion-extension	PS Prosthesis
Yoyisha et al. 2005	20 patients with bilateral TKA	Fluoroscopy	Passive and weight-bearing Flexion-extension	PCR prosthesis in one knee PC prosthesis on the other knee
Cates et al. 2008	30 patients with TKA	Fluoroscopy	Flexion-extension	PS prosthesis PCR prosthesis
Hatfield et al. 2011	42 patients pre- and post -TKA	Electromyography, infrared reflective markers	Gait	PS prosthesis
Kitagawa et al. 2010	10 subjects pre- and post TKA	Fluoroscopy	Flexion/Extension	PCR- prosthesis
Saari et al. 2005	39 patients with TKA 18 control group	Skin reflective markers	Gait	PCR and PS prosthesis
Van de Linden et al. 2007	19 patients pre – and post TKA (after 18 months and 7 years)	electrogoniometry	Gait, level walking, ascend and descend	LCS prosthesis

Yue et al. 2011	11 patients pre- and post TKA 22 control group	Fluoroscopy	Flexion/Extens ion	PCR prosthesis
Joglekar et al. 2012	18 patients with TKA Control: contralateral knee	Reflective markers	Gait: stair ascend, descend	PS prosthesis PCR prosthesis with sacrificing PCL
McClelland et al. 2011	40 patients with TKA 40 control group	Reflective markers motion analysis	Gait	PS prosthesis

Tab.4. Material and methods characteristics of the studies included in the review

Author	Flexion/ Extension	Internal/Extrenal Rotation	Adduction/ Abduction	Anterior/Posteri or Translation
TKA vs normal healthy knees				
Denniss et al. 1998	↓ flexion	-	-	-
Siston et al. 2006	-	↓ screw-home mechanism	↑ varus	↑ femoral anterior translation
McClelland et al. 2011	↓ flexion	↓ tibial IR ↑ tibial ER	-	-
Saari et al. 2005	↓ extension	-	↓ abduction ↑ adduction	-
Yue et al. 2011	-	↓ tibial IR	N.S	↓ posterior femoral translation
TKA vs pre – OP				
Hatfield et al. 2011	↑ flexion	-	↓ adduction	-
Van de Linden et al. 2007	↑ flexion	-	-	-
Yue et al. 2011	-	↓ tibial IR	N.S	↓ posterior femoral translation
Kitagawa et al. 2010	-	N.S		N.S
PCR vs PS prosthesis				
Yoyisha et al. 2005	-	-	-	↑ anterior shift
Cates et al. 2008	N.S	↑ range	-	N.S
Joglekar et al. 2012	↓ flexion	-	-	-

Tab.5. Reported results of the studies comparing knee kinematics pre- and post-TKA, within TKA groups or with a control group.

CHAPTER 2. GAIT KNEE KINEMATIC ALTERATIONS IN MEDIAL OSTEOARTHRITIS: 3D ASSESSEMENT

The study described in this chapter has been presented at EFORT 2014 and published.

D Bytyqi , B Shabani, S Lustig, L Cheze, N Karahoda Gjyrgjeala, P Neyret (2014). Gait Knee Kinematic Alterations in Medial Osteoarthritis: 3D assessment. *International Orthopaedics*, **38**(6):1191-8, DOI 10.1007/s00264-014-2312-3.

A. Introduction

Knee function can be quantified with either patient-based scales (questionnaires) or performance based measures. Seeking to improve knee function evaluation, numerous studies have analysed movement of intact, pathological and treated knees. Quantitative kinematic analysis has been used as an important tool for thorough understanding of joint function [67]. Kinematics of osteoarthritic (OA) knees has been evaluated using surgical navigation systems, magnetic resonance imaging (MRI), and computed tomography (CT) [35, 38, 68] but these techniques could not be used to study weight-bearing activities, and their results may be affected by the lack of weight contribution.

With advances in motion capture technology, three dimensional knee motion during weight-bearing is now available. Gait, as the most common activity of daily living, has been analyzed to clarify the biomechanical characteristics of OA knees.

Most previous studies about changes in OA patients during gait were focused on spatiotemporal parameters [51, 69]; demonstrating that knee OA patients walked slower, with a reduced stride length, and a lower single-limb support compared to controls. Some studies

have examined kinematic alterations during phase-specific gait cycle and reported a decreased knee excursion during flexion, decreased peak flexion during stance phase and increased knee flexion at heel strike [24, 41, 70]. Although the kinematic changes in sagittal plane have been enlightened, the 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 [40, 42] but, both studies were concentrated only in stance phase. Because the AP translation is small in magnitude and can be affected by the choice of assessment system, only few studies have analyzed this parameter [36, 38]. The quasi-static fixation of our assessment system on the bones allows us to be one of the first studies to examine the anterior-posterior translation during walking [18, 71].

The objective of this study was to use three dimensional motion analyses (KneeKGTM) to identify the changes in the kinematic variables of patients with osteoarthritic knee during a complete gait cycle and to correlate them with clinical characteristics.

We hypothesised that the OA knee patients exhibit an altered knee kinematics in sagittal, frontal and axial plane.

B. Material and Methods

Participants

This prospective study was performed between February and April 2011 in the biomechanical laboratory at our Clinical Center. Thirty patients (18 females, 12 males; mean age of 65.7 years) with varus malalignment and medial compartment knee OA were recruited into the study. They had been scheduled for total knee arthroplasty. Diagnosis was based on clinical history, a physical examination, and from radiographic evaluation that included comparative anteroposterior knee view with monopodal support, bilateral posteroanterior view at 45° knee flexion in weight-bearing, comparative lateral view at 30° knee flexion, axial view at 30° knee flexion, stress valgus and varus X-ray using a Telos system [72]. 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 center of the femoral head to the midpoint of the tibial eminential spine and another line from this midpoint to the center of the tibial plafond. The medial angle between the lines is the HKA angle (varus < 180°, valgus > 180°). Subjects were classified in terms of OA disease severity using the Ahlbäck's radiographic grading system. The integrity of the anterior cruciate ligament (ACL) was assessed intraoperatively. ACL was present in 21 knees, attenuated in five knees and ruptured in 4 knees. Patients were included in the study if they were able to walk along 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 subjects with varus malalignment were selected (Tab. 6). The asymptomatic subjects were evaluated by a trained orthopedic surgeon and were excluded if they had orthopedic (joint fracture, joint laxity, OA, lower leg discrepancy and arthritis) or neurological problems that could affect their gait pattern.

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

Tab.6. Demographic characteristics of the study groups

In vivo kinematic evaluation

Patients' knee motions were recorded using KneeKGTM system, a new innovative tool that enables a 3D analysis of the knee kinematics in a weight-bearing, dynamic condition (Fig. 9). The knee marker attachment system is designed to reduce skin motion artifacts [71]. Several studies have assessed the accuracy and the reproducibility of the device, and validated it [18, 49, 71, 73].

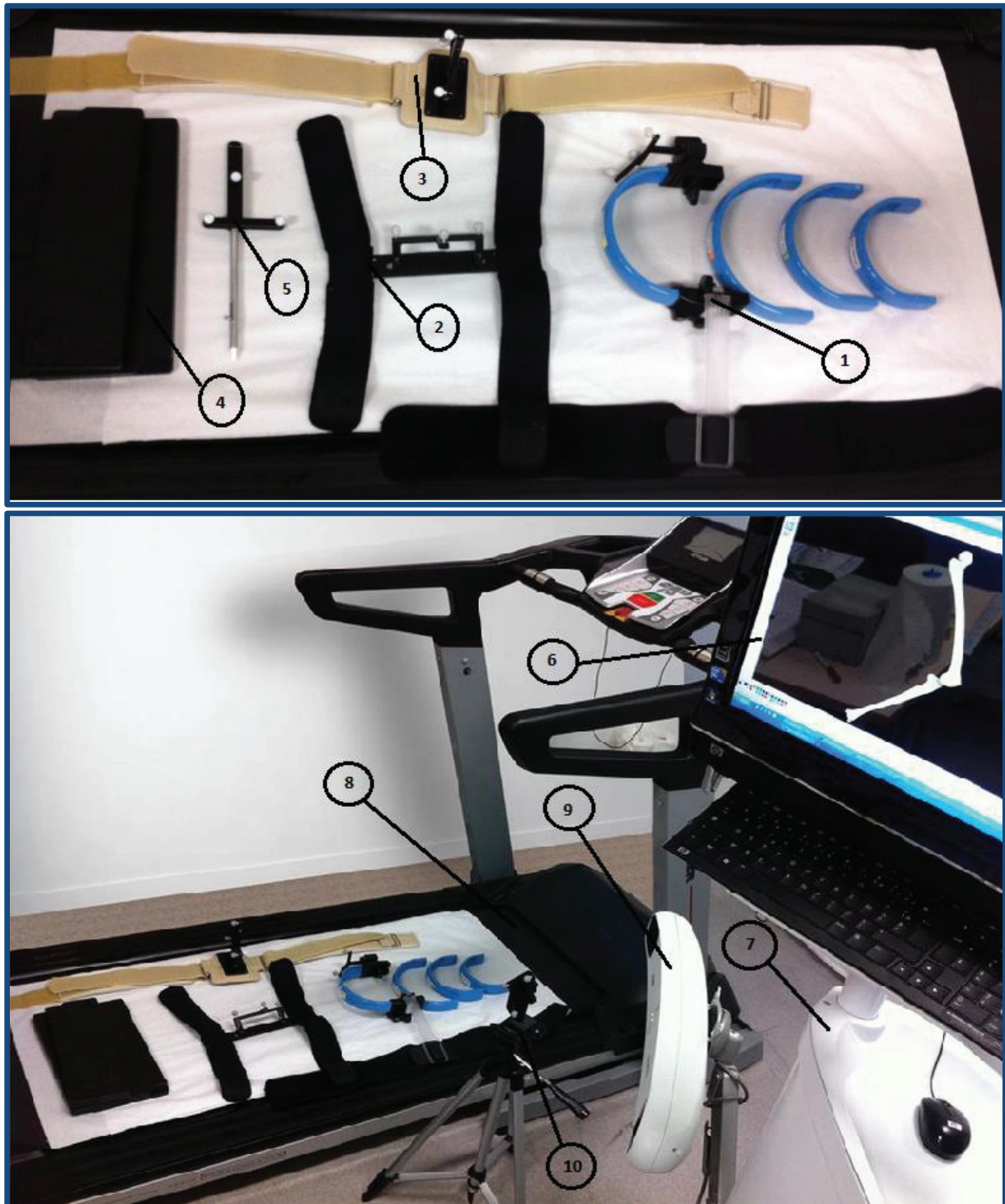


Fig.9 The KneeKG™ system and its parts. 1. Femoral harness (4 interchangeable arches), 2. Tibial harness, 3. Sacroiliac belt, 4. Feet position guide, 5. Pointer, 6. Computer, 7. Cart, 8. Treadmill, 9. Video camera, 10. Reference body.

The mean inter-observer repeatability value ranged between 0.4 degrees and 0.8 degrees for rotation angles and between 0.8 and 2.2 mm for translation [18, 73]. This clinical tool enables an accurate and objective assessment of the 3D function of the knee joint.

The kinematic analysis of the knee during gait was done walking on the treadmill at comfortable speed chosen by subjects themselves. Firstly, because treadmill walking can be unfamiliar, a treadmill walking habituation period of 10 min was initiated prior to data collection. After the installation of the femoral, tibial and sacral trackers, the calibration of the device was done.



Fig .10. The placement of the femoral and tibial harnesses

The first step of the procedure was localizing anatomical landmarks with the retroreflective localizer. The second step of the procedure was defining the center of the hip, knee and ankle joints. The hip joint center (HJC) was defined using an optimization computation method during a leg circumduction movement (Fig 11).



Fig. 11 Calibration: Defining of the Hip Joint Center

The knee joint center (KJC) was then determined to be the projection of the femoral epicondyle's midpoint on the mean helical knee flexion-extension axis (again computed during flexion-extension movement) (Fig 12).

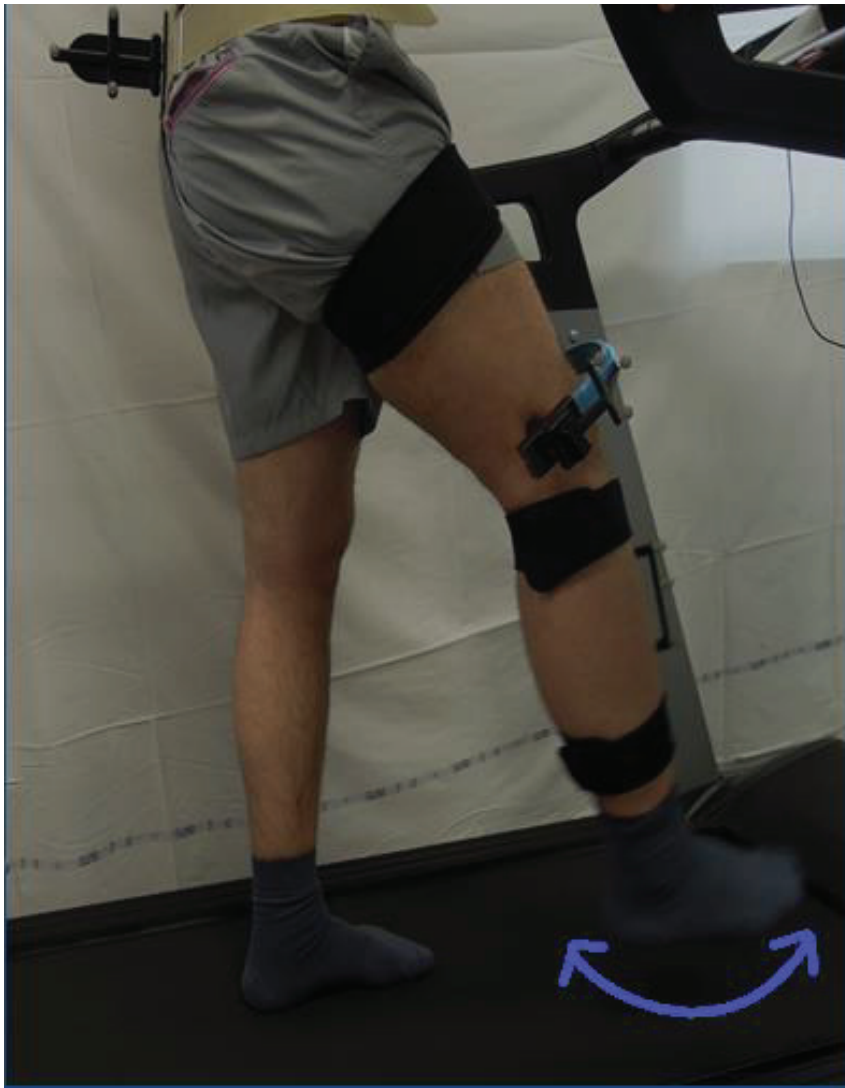


Fig.12 Defining of the Knee Joint flexion-extension axis

The ankle's joint center (AJC) was defined as the midpoint between the malleoli. Lastly, the neutral transverse rotation was set when the knee was determined to be at 0° of flexion during a slight flexion-hyperextension movement (Fig 13).

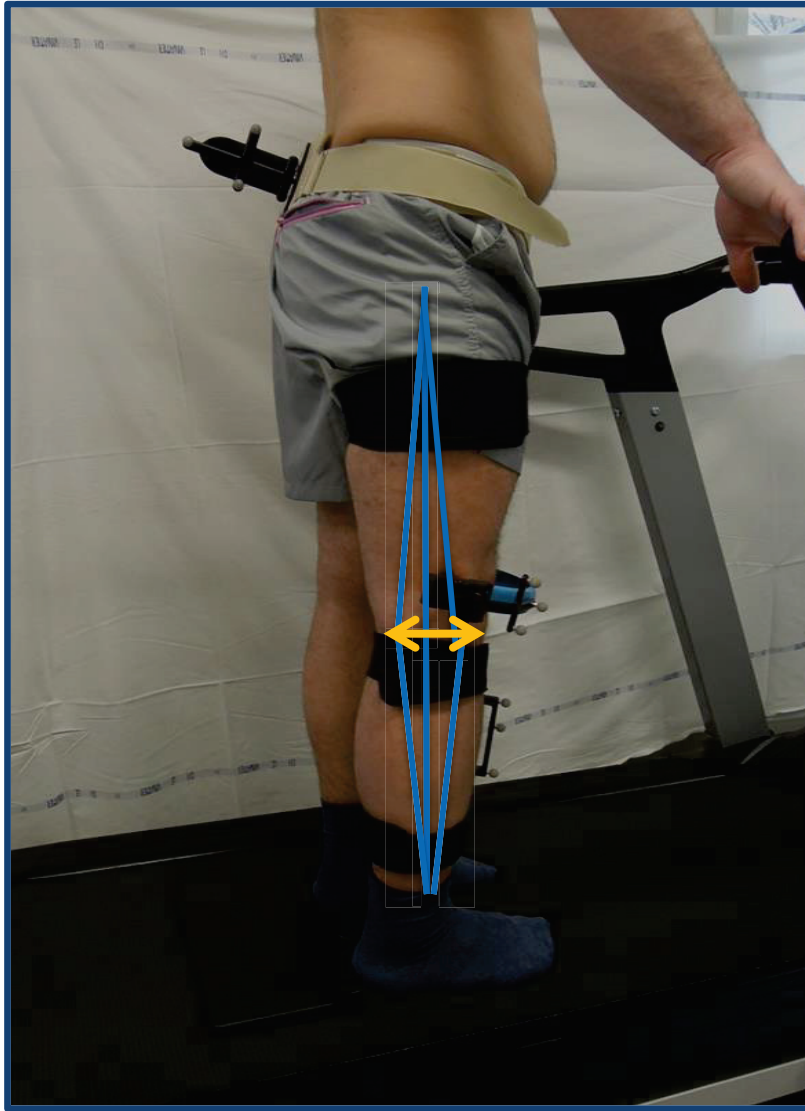


Fig.13 Defining the neutral transverse rotation

A sagittal plane is formed from the successive positions of the vector joining HJC to AJC during the movement; longitudinal axis for femur is defined from HJC to KJC; longitudinal axis for tibia is defined from KJC to AJC; projections of the longitudinal axes onto sagittal plane are calculated; absolute angle between projections is calculated. The neutral posture is defined when this absolute angle is zero. At this posture, anterior-posterior axes are defined lying in sagittal plane and perpendicular to longitudinal axes, respectively for femur and tibia. Medio-lateral axes are then defined perpendicular to the other two axes, for the femur and for the tibia. Finally, the origin of the axes is fixed at KJC.

Once the installation and calibration of the system have been finalized, the subject was asked to walk at the pre-selected speed and the 3D displacements of the reflective markers were recorded for 45 s by the operator. Once calibration and measurement had been performed, the Knee3D suite computed the various angle values on anatomical axes between the tibia and the femur and provided a chart that contains the subject's curves.

The whole procedure lasted 20-25 minutes; after which the trackers were taken off and the patient was dismissed.

A database in Microsoft Excel 2010 was created containing, for each participant, the 4 biomechanical patterns consisting of, resumed on 100% of the gait cycle, the 3 knee angles: flexion-extension, abduction-adduction, internal-external tibial rotation and anterior-posterior translation.

Using these data, we analysed the movements in sagittal, frontal, axial and transverse plane during walking and their correlations with clinical data.

Statistical Analysis

The assumptions of normality and equality of variance were assessed in the data using Shapiro-Wilk and Levene's tests respectively. If the data met the parametric assumptions, t-test was used, if not, non-parametric Mann-Whitney test was used, to compare differences in age, height, weight, BMI. ANOVA and PostHoc (Tuckey) tests were used to compare the kinematic characteristics. Pearson's correlations (r) were used to examine relationships between clinical and kinematic gait parameters of knee OA patients. The statistical difference was set as $p < 0.05$. All the data were analyzed using SPSS v21 (SPSS Inc., Chicago).

C. Results

The OA group walked with a lower speed comparing with the control group (1.2 km/h; 2.1 km/h , respectively, $p<0.05$). The perimeter of walking for most patients (14 out of 30) of the OA group was around 1 km. Their mean range of motion was 5.6° of extension and 110.8° of flexion (Tab. 7). The table 8 summarizes the spatiotemporal and kinematic data for OA and control groups.

<i>Clinical characteristics of knee OA patients</i>		
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 osteoarthritis (number of patients)	Stage I	-
	Stage II	8
	Stage III	16
	Stage IV	6
Range of motion (n=30) (mean ± SD)	Extension	5.6 ⁰ (12.6)
	Flexion	110.8 ⁰ (18.5)

Tab.7 The clinical characteristics of the patients with medial knee osteoarthritis

Kinematic characteristics (mean±SD)	KneeOA group	Control group	Significance
Speed	1.2 km/h (0.3)	2.1 km/h (0.2)	$p < 0.05^*$
Flexion angle at initial contact	19 ⁰ (7.7)	17.4 ⁰ (12.5)	$p > 0.05$
Maximum flexion during stance	22.4 ⁰ (8.1)	28.1 ⁰ (7.9)	$p < 0.05^*$
Maximum extension during stance	7.6 ⁰ (4.1)	2.2 ⁰ (4)	$p < 0.05^*$
Maximum flexion during swing	48.2 ⁰ (6.3)	54.4 ⁰ (5.3)	$p < 0.05^*$
Maximum extension during swing	14.9 ⁰ (6.5)	12.5 ⁰ (6.7)	$p > 0.05$
Range of Flexion-Extension	40.6 ⁰ (6.1)	52.2 ⁰ (5.3)	$p < 0.05^*$
Adduction(+)/Abduction(-) angle at initial contact	5.7 ⁰ (7.3)	-0.4 ⁰ (2.8)	$p < 0.05^*$
Range of adduction-abduction	7.7 ⁰ (5)	5.5 ⁰ (1.6)	$p > 0.05$
Internal(-)/External(+) Rotation at initial contact	0.3 ⁰ (3.6)	-0.1 ⁰ (2.4)	$p > 0.05$
Range of internal-external rotation	7.6 ⁰ (3.1)	9.3 ⁰ (2.4)	$p < 0.05^*$
Anterior (+)/Posterior (-) Translation of tibia at initial contact	-2.9mm (5.4)	0.4mm (2.2)	$p < 0.05^*$
Range of Anterior-Posterior Translation	7.9mm (4.1)	9.3mm (4.4)	$p > 0.05$

Tab. 8 The spatiotemporal and kinematic data of the knee OA and control group

*statistically significant

Flexion-Extension

The range of motion in sagittal plane was significantly lower for OA group ($40.6^\circ \pm 6.1$) compared to control group ($52.2^\circ \pm 5.3$), $p < 0.05$. The maximum flexion during stance was significantly lower for OA group than control group ($22.4^\circ \pm 8.09$ and $28.1^\circ \pm 7.97$, respectively, $p < 0.05$). During the gait cycle we observed a significant difference during terminal stance which corresponds to maximum extension of the knee ($p < 0.05$; 8.5° and 4.4° , OA patients and control group, respectively). During initial swing phase, the OA patients had a reduced flexion ($p < 0.05$; 41.9° and 49.4° , OA patients and control group, respectively) (Fig.14).

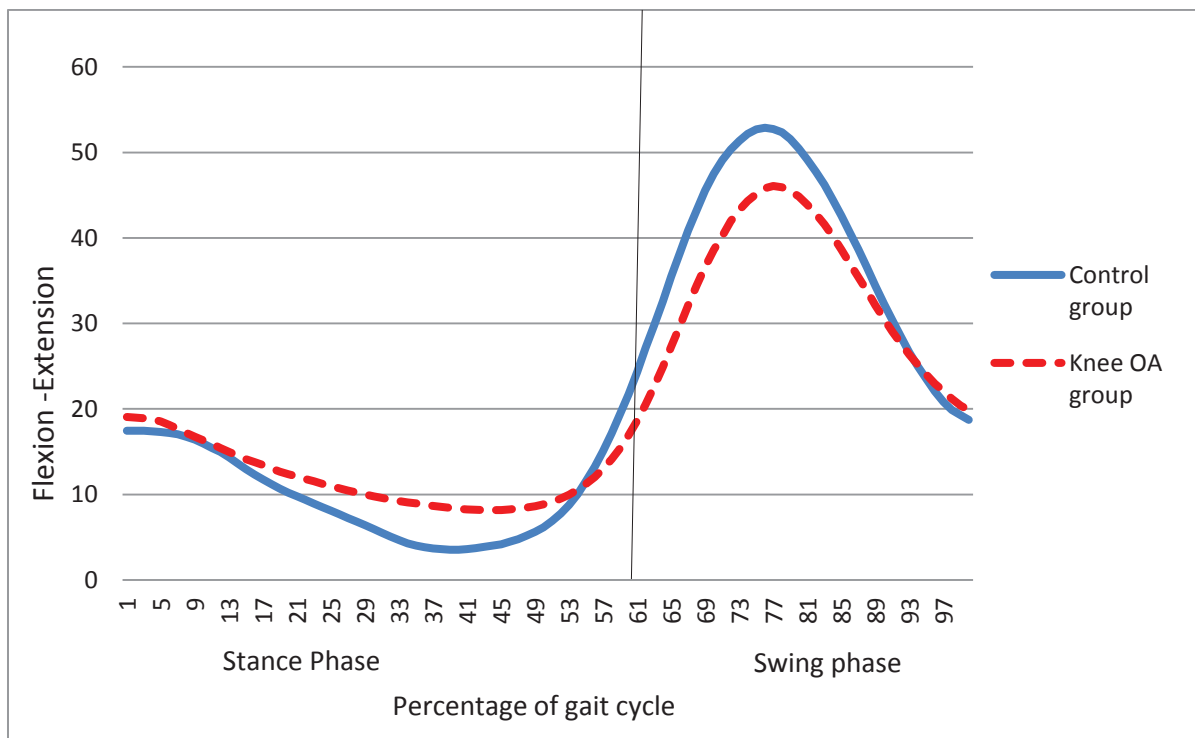


Fig 14. Flexion-Extension during gait cycle. Positive values represent flexion.

Adduction-Abduction

At initial contact, the OA patients showed an adduction angle of $5.7^\circ \pm 7.3$, while control group stayed in a relatively neutral position, $-0.4^\circ \pm 2.8$, $p < 0.05$. This difference was

observed during the whole gait cycle (Fig.15). 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.

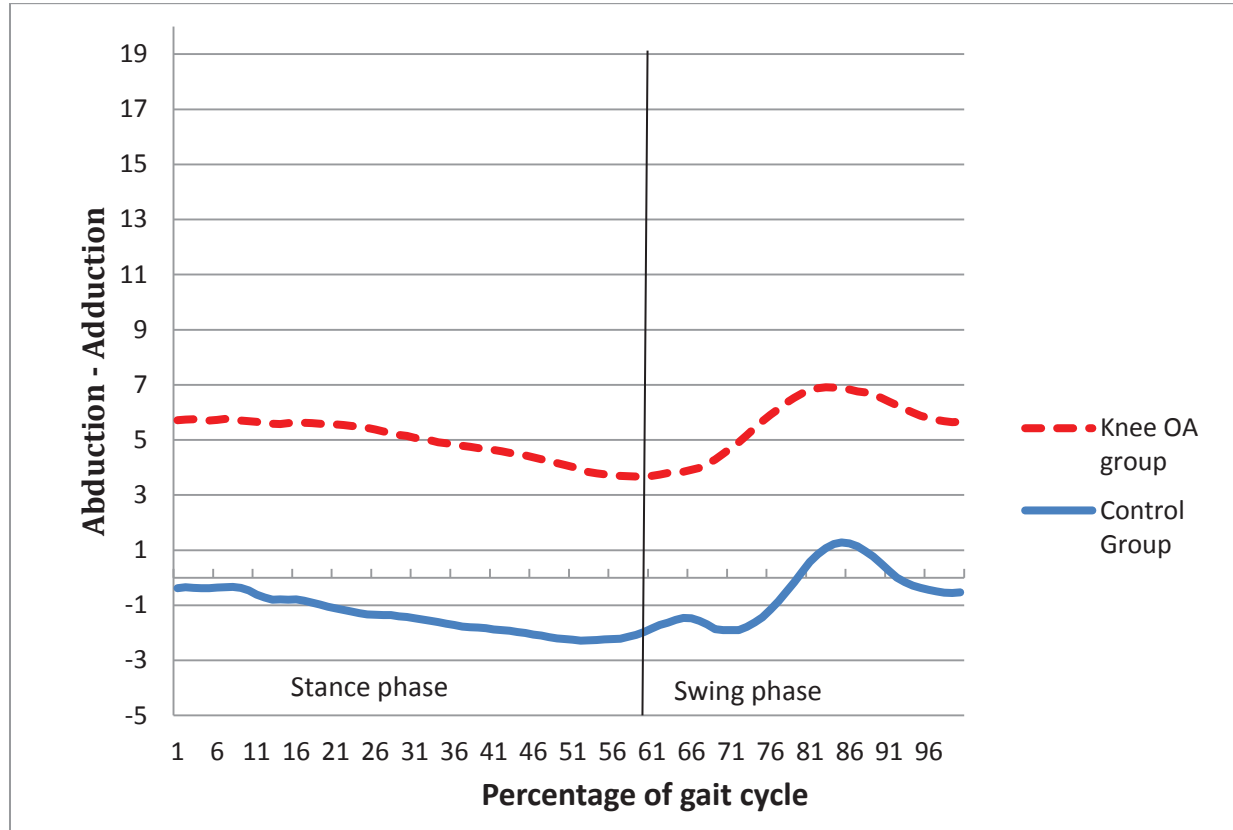


Fig 15. Adduction-Abduction during gait. Positive values mean adduction.

Internal-External Rotation

The range of motion of internal-external rotation of OA patients was significantly lower than that of the control group ($7.6^\circ \pm 3.1$; $9.3^\circ \pm 2.4$, respectively, $p < 0.05$). In terms of internal-external rotation during the gait cycle, the OA group tended to behave in contrast with control group. During the stance phase, the OA group remained in relatively neutral position (mean $-0.5^\circ \pm 0.4$) while the control group showed internal rotation (mean $-2^\circ \pm 0.7$), $p < 0.05$. On the other hand, during swing phase, while the control group started to rotate externally with the peak at 86% of gait cycle, at midswing, (2.1°), the OA group firstly rotated internally, then

restored the neutral position again with the peak at 93% of gait, terminal swing phase (0.5°) (Fig.16).

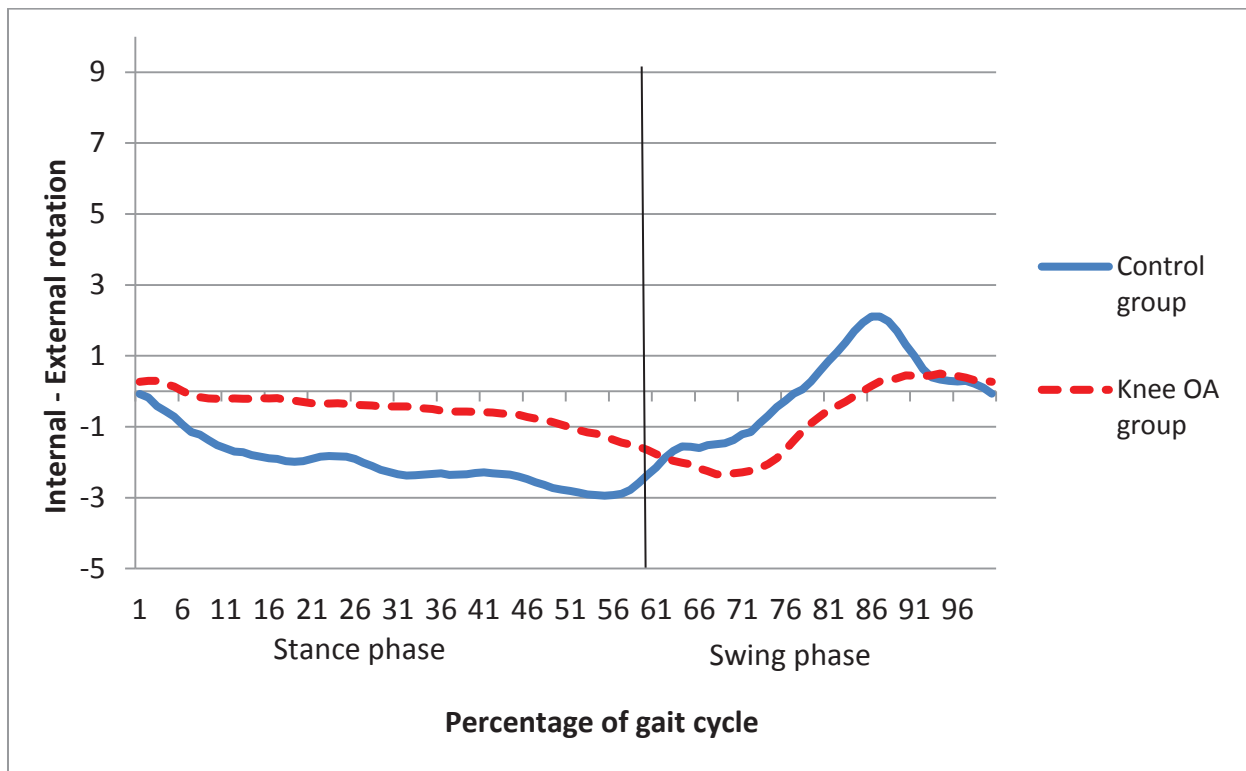


Fig. 16. External(+)/ Internal(-) rotation during gait.

Anterior-Posterior Translation

At initial contact, the OA group were significantly more posterior translated ($-2.9\text{mm} \pm 5.4$) compared to control group ($0.4\text{mm} \pm 2.21$), $p < 0.05$. The tibia, at OA group, stayed in a more posterior position during the whole gait cycle but the significance was reached only at loading and midstance phase (Fig.17).

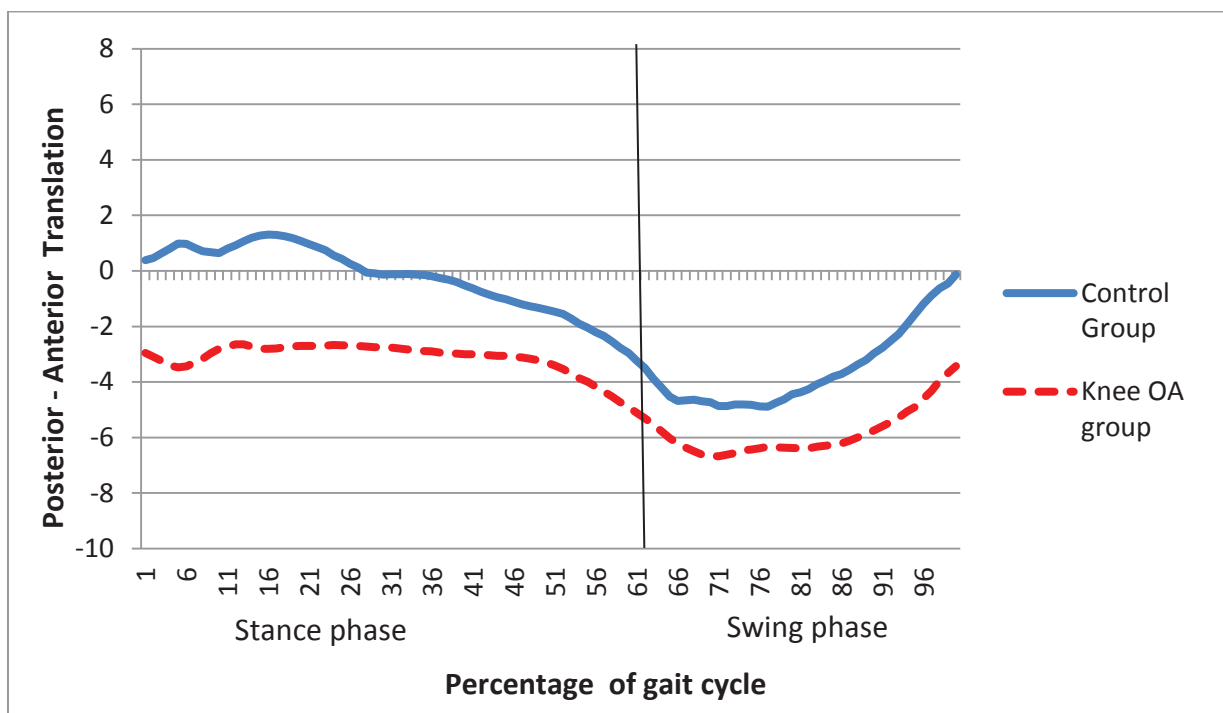


Fig 17. Antero-posterior tibial translation (mm). Positive values mean anterior translation of tibia relative to femur.

There was no significant difference in the kinematics between intact/attenuated and ruptured ACL groups of osteoarthritic knees (Tab.9).

Kinematic data (mean±SD)	ACL-intact	ACL- attenuated	ACL- ruptured	Significance
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	8.2mm (4.4)	5.69mm (2.8)	9.03mm (3.4)	>0.05

Tab. 9 The ACL status of the knee OA

D. Discussion

The major finding of this study was that the OA knee group showed an overall stiffening gait strategy by exhibiting reduction of motion not only in sagittal but also in axial plane.

Subjects with OA showed a reduction in flexion-extension range. Reductions in flexion excursion have been found in most of the studies that examined the patients with OA knees [42–44, 74]. Heiden et al. [43] found no difference in flexion-extension excursion, but they analyzed only one phase of gait cycle, from heel strike to midstance, and included all-stage OA knee patients. Our OA knee group showed a significant decreased maximum flexion angle during both stance and swing phase, which is in agreement with previous studies [42, 70]. The knee OA group in this study along with Schmitt and Rudolph study showed less extension during the single-limb support stance phase (34-51%) than control group. This lack of extension might be because of greater levels of flexor activation found in the study of Heiden et al. [43] and quadriceps weakness [70, 75] observed at the severe OA knees.

Eventhough the range of adduction-abduction was greater at the knee OA group, it did not reach the statistical significance. On the other hand, there was a positive correlation between varus deformity and adduction-abduction laxity. The results of this study revealed that knee OA patients showed a significant increased knee adduction angle throughout the gait cycle. These data are consistent with the findings of previous studies [36, 42, 68]. This frontal instability associated with an increased adduction angle may increase the further degradation of the cartilage. Furthermore, Lewek, Rudolph and Snyder-Mackle [49] showed that the excessive frontal laxity was observed only on the medial side of the joint and was accompanied by greater medial muscle co-contraction as an effort to control this laxity.

In this 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. [42], who observed a decreased excursion of axial tibial rotation at the OA group. There was no

significant difference at the initial contact rotational angle, but the knee OA group differed from normal knees during the whole gait cycle. Firstly, during stance phase, the knee OA group displayed a relatively neutral position while control group rotated internally, reaching a significant difference at the midstance. This difference at this gait phase which corresponds to maximal knee extension, may come because of the lack of full extension of the OA knees. Hamai et al. [35] evaluated the rotational angles using CT and reported a femoral internal rotation bias compared to control group. Saari et al. [36] using dynamic RSA, found a decreased internal rotation for knee OA group. Secondly, during swing phase, the knee OA group maintained a neutral position at the terminal swing phase while control group rotated externally. To our knowledge, this is the first study that provides information for external-internal rotation during swing phase. These data show that the “screw-home” mechanism may alter at the osteoarthritic knee. Nagao et al. [75] measured the rotational angle at osteoarthritic knees with ultrasound and reported that the rotation of screw-home decreases with progression of the osteoarthritis and the knee joint then moved more like a simple hinge joint.

There is no consensus in literature concerning AP translation in OA knees. In this study, the knee OA group displayed a significantly increased posterior translation from initial contact to midstance. During swing phase, the tibia of OA knees still remained more posteriorly but in a less degree. Siston et al. did not find differences in AP translation between OA and control group [38], but in their study the kinematics is evaluated with passive motion and the control group consisted in cadaveric knees, whereas we assessed dynamic motions in vivo. Hamai et al. [35] observed a less posterior femoral translation at the patients with medial knee OA. Saari et al. [36] concluded that the increased posterior displacement of knees with arthrosis tended to deviate from normal knees in the same way as previously observed in knees with total knee arthroplasty. There was no significant difference in AP translation between

intact/attenuated/ruptured ACL in OA knees. However, the sample sizes of the ruptured and attenuated ACL groups (4 and 5, respectively) were small to draw 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, the OA patients were heavier and walked with a slower speed than control group. This BMI and speed difference is consistent with literature for the similar age-groups with and without OA [46, 76]. Some studies compared gait trials between groups by selecting the speed usually near 1 m/s. Walking speed effect on the gait measures, particularly in sagittal plane, has been studied by previous authors [24, 76]. While the “normal” group showed significant difference between 1 m/s and self-selected speeds for all variables, the severe OA group did not show any differences between the two speeds. We did not select the speed because we wanted the subjects to walk as in their natural, daily activities. Another point is that the gait speed is linked with the presence of knee OA and it is difficult to separate the confounding effect [70]. Secondly, the artifacts from soft tissue could have affected the results. Nevertheless, Sati and Larouche [17] showed that the skin motion artifacts are reduced with the harness used because it is fixed quasi-statically in thigh and calf. Lustig et al. [18] concluded that this evaluation system provides an objective assessment of the precise biomechanical behavior of the knee.

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

E. Conclusion

This study investigated the in vivo kinematics of the osteoarthritic knee during gait on a treadmill. Knee OA group showed an altered “screw-home” mechanism by decreased excursion in sagittal and axial tibial rotation and a posterior translation of the tibia. On the other hand, we observed an adduction angle during whole gait cycle and an increased frontal laxity with the increase of the varus malalignment.

Analysing post-arthroplasty knee function would be of interest to understand if the changes described in this study could predict post-arthroplasty knee kinematics and if the kinematics of the knee after the arthroplasty comes back to normal.

CHAPTER 3. DOES THE TOTAL KNEE ARTHROPLASTY IMPROVE THE GAIT?

IN VIVO KNEE KINEMATICS ANALYSIS

The study described in this chapter has been submitted for publication.

D Bytyqi , B Shabani, S Lustig, L Cheze, N Karahoda Gjyrgjeala, P Neyret (2014). Does the total knee arthroplasty improve the gait? In vivo knee kinematics analysis. *Journal of Biomechanics*

A. Introduction

Approximately 600,000 knee replacements are performed every year in the United States of America [53], more than 70,000 in France [54]. Nowadays, there is a wide diversity of total knee prosthesis designs, developed with specific properties and with a specific patient group in mind and therefore each one has its own theoretical advantages and disadvantages.

It is well documented that total knee arthroplasty for patients with osteoarthritis of the knee results in a decrease of pain and an increase in physical function and quality of life [27, 77].

However, the long-term performance of total knee replacement is dependent on the kinematics of knee joint. Retrieval studies have shown that the wear of total knee replacements is highly variable and this is attributable to the diverse kinematic and stress conditions that occur in vivo.

Many *in vitro* studies [78–80] compared knee kinematics before and after TKA, however, the knees were not at the end-stage of OA and they lack the influence of muscles and weight-bearing.

Most *in vivo* studies assessed the kinematics of either OA knees or TKA knees. A few intraoperative studies compared passive knee flexion kinematics before and after TKA using surgical navigation systems [38].

Three-dimensional gait analysis provides comprehensive joint kinematic changes during walking that enhance our understanding of altered joint function and loading with pathologic conditions and treatment options such as TKA.

However, it remains unclear how the gait kinematics of advanced OA knees change as a result of TKA. Because knee motion following TKA depends on the preoperative conditions, understanding the kinematics of OA knees, and how a total knee arthroplasty changes those kinematics, could lead to improved implant design and surgical techniques.

In our study, we compared *in vivo* gait analysis of knee kinematics before and after TKA in patients with medial compartment knee OA, and compared the data to the kinematics of a control group of healthy subjects using a 3D, *in vivo* assessment device, KneeKgTM.

The goal was to determine how medial compartment OA affects kinematics during a weight-bearing, daily activity as walking and if contemporary PCS-TKA can restore the kinematics towards normal.

B. Material and Methods

Participants

This prospective study was performed on 20 patients (14 female, 6 male) who underwent total knee arthroplasty for knee osteoarthritis (OA) between February to April 2011. Patients were included in the study if they were able to walk along without a gait aid and were excluded if they had any neuromuscular disease, cardiovascular disorders, or other lower limb surgeries that would affect their gait or put them at risk while participating. Given that, 21.2% of TKR patients will undergo knee replacement on the contralateral knee within 5 years, patients with bilateral knee replacement were included, provided that the most recent procedure was undertaken at least 10 months prior to testing [66, 81].

The follow-up examination was done at least 10 months post total knee replacement.

A control group of 12 subjects at the similar age, with no history of musculoskeletal disorders on the lower limbs was selected (Tab.10).

	Knee OA group (mean ±SD)	Control group (mean ±SD)	Significance
Age	67 yrs (6.9)	61.7 yrs (3.1)	$p < 0.05^*$
Weight	81.8 kg (14.2)	73.5 kg (9)	$p > 0.05$
Height	168.6 cm (6.5)	166.2 cm (5.7)	$p > 0.05$
BMI	29.08 kg/m ² (4.1)	26.5 kg/m ² (1.8)	$p > 0.05$
Gender	Female:14	Female:8	
	Male: 6	Male:4	

Tab.10 Demographic characteristics of the study groups.

Implant and Surgical Technique

All the surgeries were done by three surgeons of one team. All the patients recieved posterior-stabilized tricompartmental TKA (KneeTec prosthesis, Tornier, Saint Ismier, France). A description of the original design has been published in a previous study [56]. The femoral part of this prosthesis has a third condyle that provides replacement of the central pivot and after 35° of flexion it lays in the mating projection on the polyethylene insert to create a tripodal support. The tibial part has a delta keel to avoid interfering with cortical bone and to increase the contact surface. A rotatory version of the insert was used which allows rotation of 15°. There exist ten sizes from which three most common sizes are available in narrow version also.

The medial parapatellar arthrothomy was performed in all cases [82]. All bone cuts were made using a oscillating saw over guide pins. The cut in tibia was done at 90° in sagittal plane and 9 mm for medial femorotibial osteoarthritis in reference with the intact part of the plateau. The intramedullary rod guarantees a good balance in the sagittal plane while extramedullary rod refines the varus-valgus in relative to the first intermetatarsal space. The insert part of the prosthesis provides a slope of 4°.

The posterior femoral cut was done in 7° valgus, relative to centromedullary guide, in the case of the medial femorotibial osteoarthritis. A patellar resurfacing arthroplasty with a dome patella was performed in all cases. All components were cemented. (Fig.18).

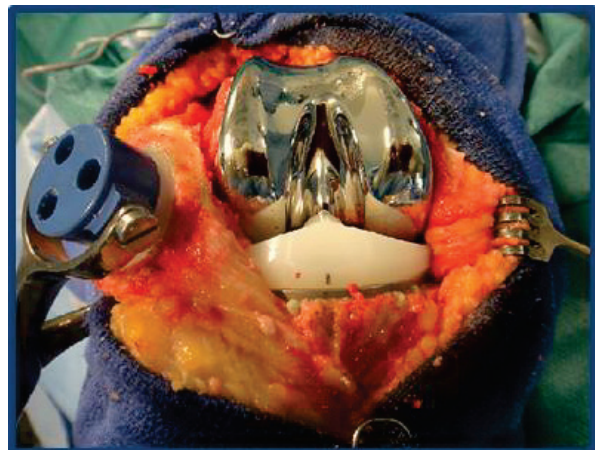
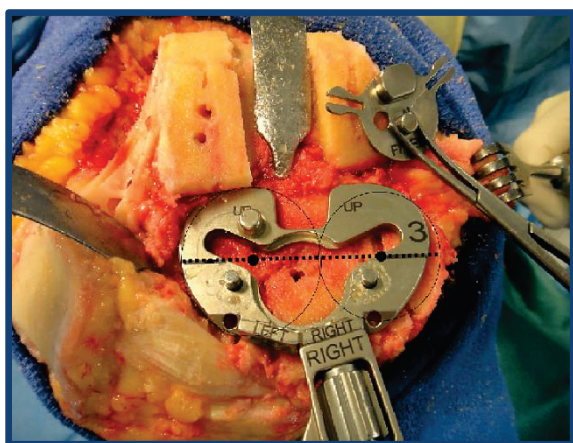
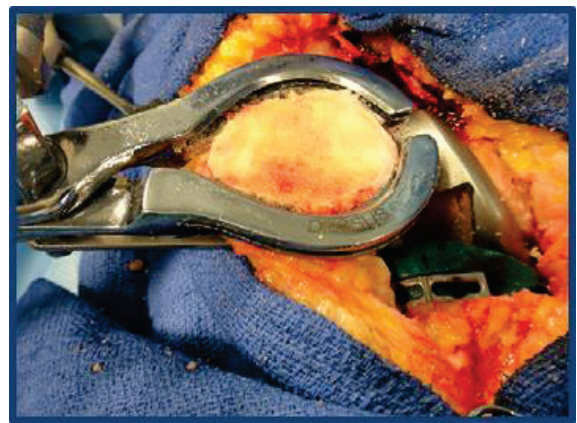
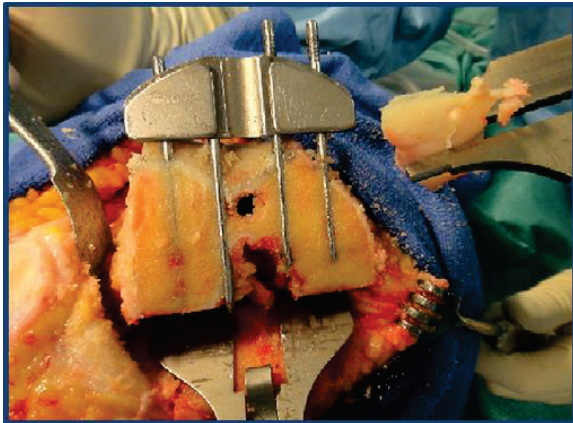
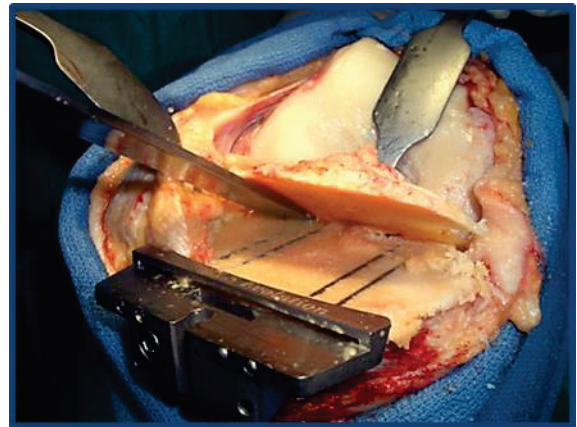
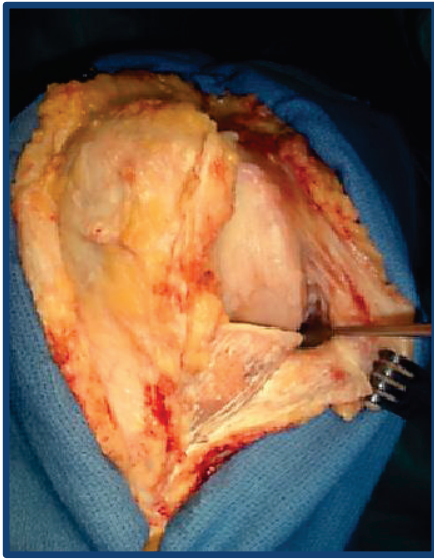


Fig. 18 The surgical technique of our center of implantation of total knee arthroplasty [82]

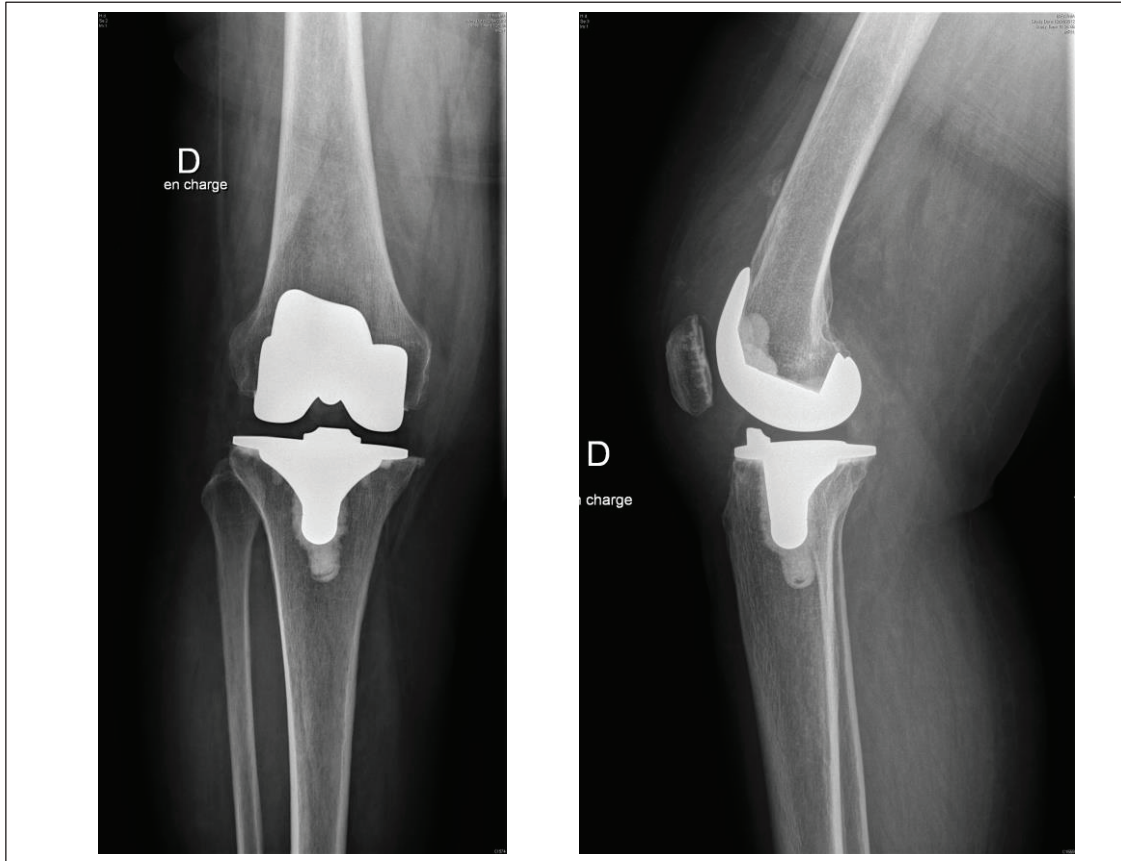


Fig. 19 The radiograph of the knee after TKA

Clinical and Imaging Examination

The clinical knee examination of each patient consisted on defining the morphotype, the range of motion and frontal and sagittal laxity. Also, for each patient, a series of X-rays was done: the entire lower extremities in frontal, lateral and axial planes (Fig.19).

In Vivo Kinematic Evaluation

The *in vivo*, 3D kinematic data were collected during walking at self-selected comfortable speed. We used the KneeKGTM system, which has precisely been prescribed in the first study.

Statistical Analysis

Participant characteristics (such as age, height, weight, BMI, gait speed 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 between pre- and post - TKA groups, while ANOVA test was used to compare TKA group 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.

C. Results

The patients walked faster after total knee arthroplasty comparing with pre-op assesement (1.9 km/h \pm 0.5 and 1.3 km/h \pm 0.5, respectively, p <0.05), but with lower speed comparing with control group (2.1 km/h \pm 0.2, p<0.05) (Tab.11) .

The overall alignment was significantly improved, 4.75° \pm 2.9, preoperatively to 1.7 \pm 1.5 postoperatively, p<0.05).

<i>Clinical characteristics of knee OA patients, pre- and post – TKA</i>				
Clinical parameter		Pre – TKA	Post - TKA	P – value
Perimeter of walking in daily life (number of patients)	< 0.5 km	1	0	
	0.5-1 km	3	1	
	~1 km	13	4	
	>1 km	3	15	
HKA Alignment (mean \pm SD)		4.75° \pm 2.9	1.7° \pm 1.5	p<0.05
Range of Motion (mean \pm SD)	Recurvatum	0.5° \pm 1.5	1.5° \pm 2.3	p<0.05
	Extension	6° \pm 7.4	0.7° \pm 1.6	p<0.05
	Flexion	111.2° \pm 16.6	120° \pm 10.6	p<0.05
Speed (mean \pm SD)		1.3 km/h \pm 0.5	1.9 km/h \pm 0.5	p<0.05
Time from surgery (mean \pm SD)		-	10.7 months \pm 1.08	-

Tab.11 The clinical characteristics of knee OA patients, pre- and post - TKA

Flexion Extension

The passive range of motion in sagittal plane was significantly improved after total knee arthroplasty. The mean extension preoperatively was $6.1^{\circ} \pm 7.4$, while postoperatively it was $0.75^{\circ} \pm 1.6$, $p < 0.05$. The mean flexion before arthroplasty was $111.2^{\circ} \pm 16.6$, and $120^{\circ} \pm 10.6$ after arthroplasty, $p < 0.05$. During walking, range of flexion/extension was also improved significantly (pre-OP: $39.9^{\circ} \pm 5.5$, and post-OP: $44.8^{\circ} \pm 5.1$, $p < 0.05$) but it still remained lower than control group ($52.2^{\circ} \pm 5.35$, $p < 0.05$). The maximum flexion during swing phase was significantly higher at arthroplasty group comparing with pre-OP assesement ($51.7^{\circ} \pm 5.4$ and $48.4^{\circ} \pm 5.4$, respectively, $p < 0.05$). There was no significant difference between arthroplasty group and control group. On the other hand, there was no difference in the arthroplasty group before and after operation at the max extension during stance phase. But there was a difference between this TKA and control group (pre-OP $8.4^{\circ} \pm 3$; post OP $6.9^{\circ} \pm 5.5$ and control group $2.2^{\circ} \pm 3.9$, $p < 0.05$).

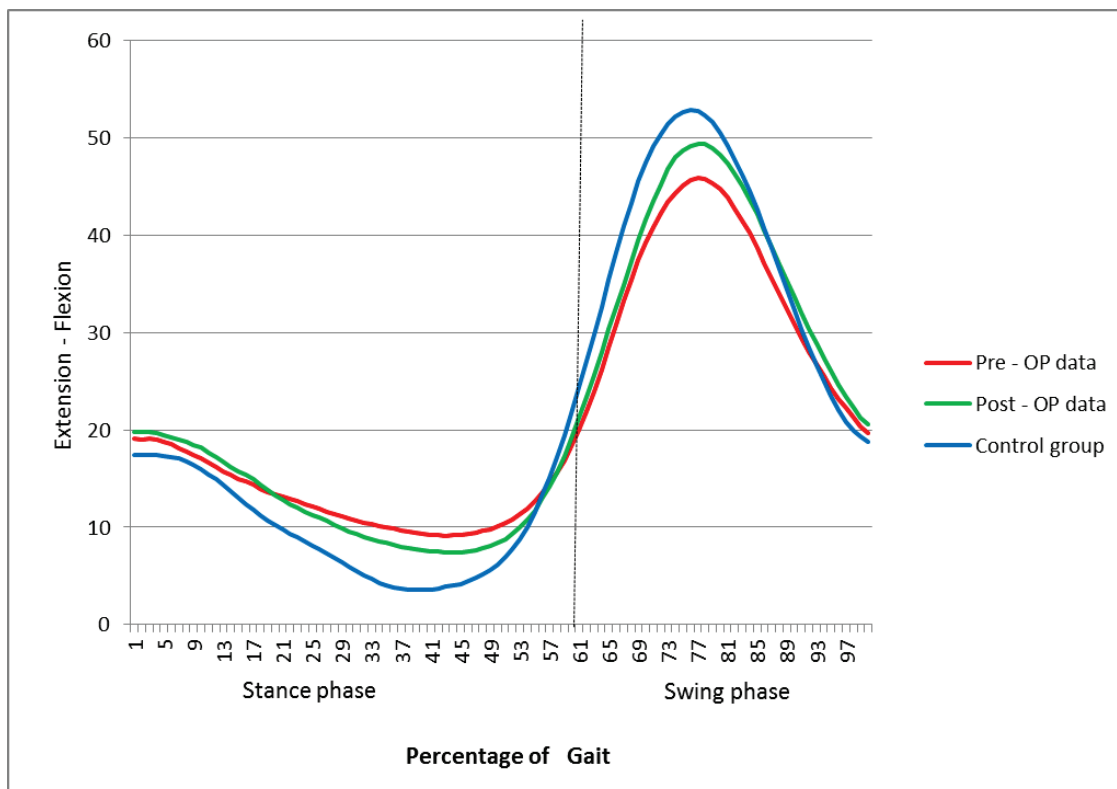


Fig.20 Flexion-Extension (degrees) during gait cycle; positive values represent flexion

Adduction-Abduction

Eventhough there was a visible improvement in the movement in frontal plane after total knee arthroplasty, the difference did not reach the significance. This may be a result of high variability in the pre-OP data.

Despite amelioration, the post – arthroplasty group showed a significant difference with control group during the initial and mid swing phase of gait cycle (post-OP: $2.15^{\circ} \pm 0.44$ and control group $-1.49^{\circ} \pm 0.49$).

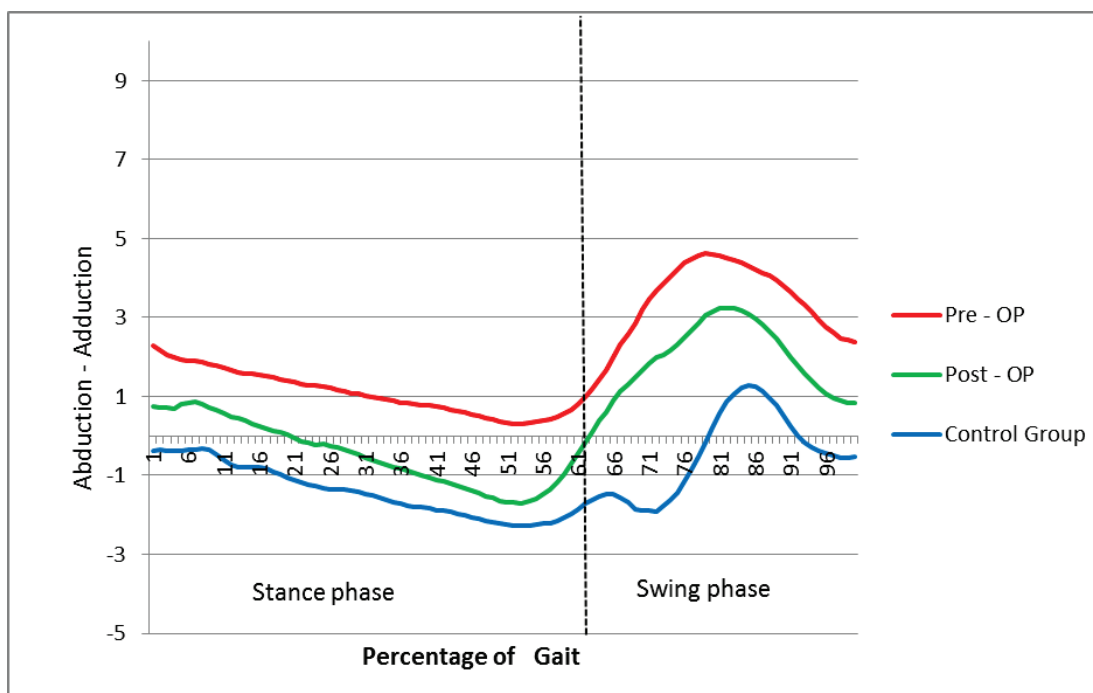


Fig. 21 . Adduction-Abduction (degrees) during gait. Positive values mean adduction.

Internal/External Rotation

The range of motion in axial plane did not change pre- and post arthroplasty, but remained lower than the matched control group, (pre-OP $7^{\circ} \pm 2.2$; post-OP $6.7^{\circ} \pm 2.4$; control group $9.3^{\circ} \pm 2.4$, $p < 0.05$).

During the midstance phase, there was a significant difference between TKA and control group, but there was no difference pre- and post – arthroplasty. While control group rotated

internally ($-2.33^{\circ} \pm 0.02$) during midstance, TKA group stayed in a relatively neutral position (pre-OP: $-0.12^{\circ} \pm 0.02$; post-OP: $-0.19^{\circ} \pm 0.03$), $p < 0.05$.

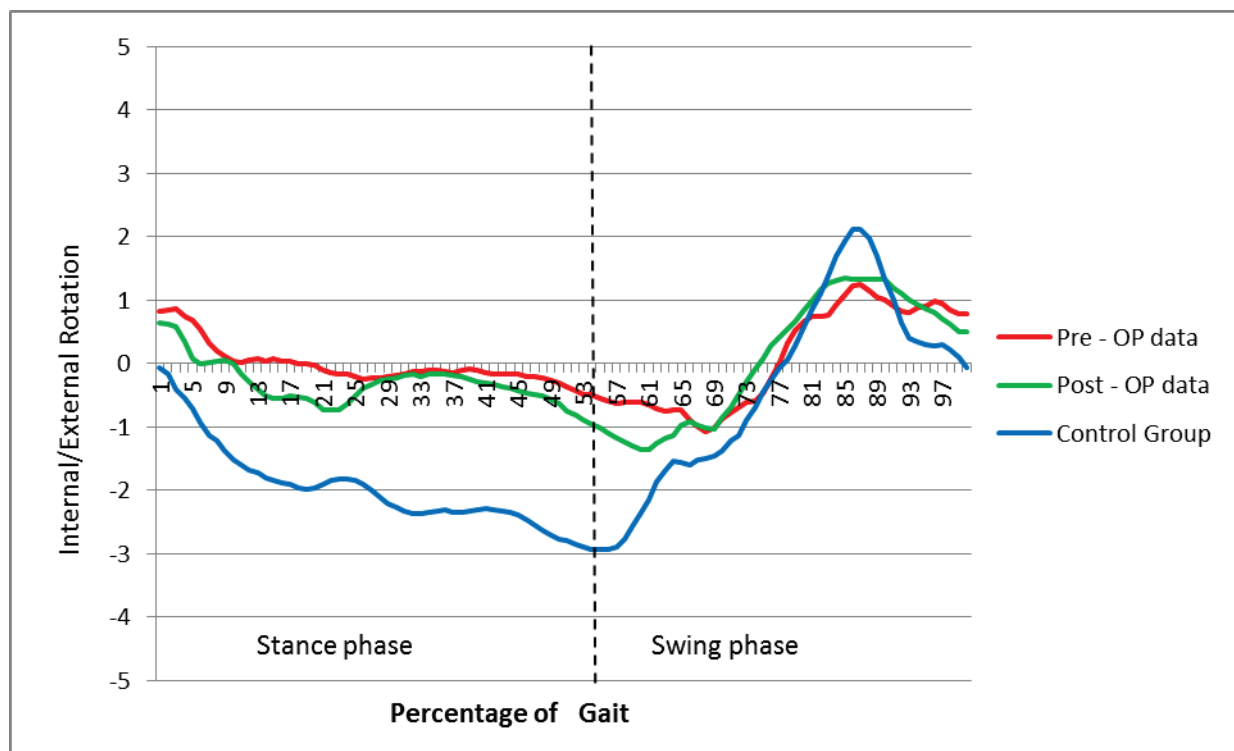


Fig.22 Internal/External tibial Rotation (degrees) during gait. Positive values mean external rotation

Anterior/Posterior Translation

There was a significant difference in anterior/posterior translation between TKA and control group. During loading and midstance, the tibia in control group was slightly anteriorly translated ($1.07^{\circ} \pm 2.74$) while in TKA group the tibia was posteriorly translated (pre OP: $-1.67^{\circ} \pm 2.81$; post Op: $-2.7^{\circ} \pm 1.92$), $p < 0.05$. There was no difference between pre- and post – arthroplasty, however after arthroplasty the tibia was more in posterior translation comparing with pre-OP data.

Also, the maximum posterior translation during swing phase was significantly higher at post arthroplasty group ($-9.53^{\circ} \pm 2.25$) comparing with control group ($-5.74^{\circ} \pm 3.02$), $p < 0.05$. On the

other hand, there was no difference between pre- and post –OP data, and between pre-OP and control group.

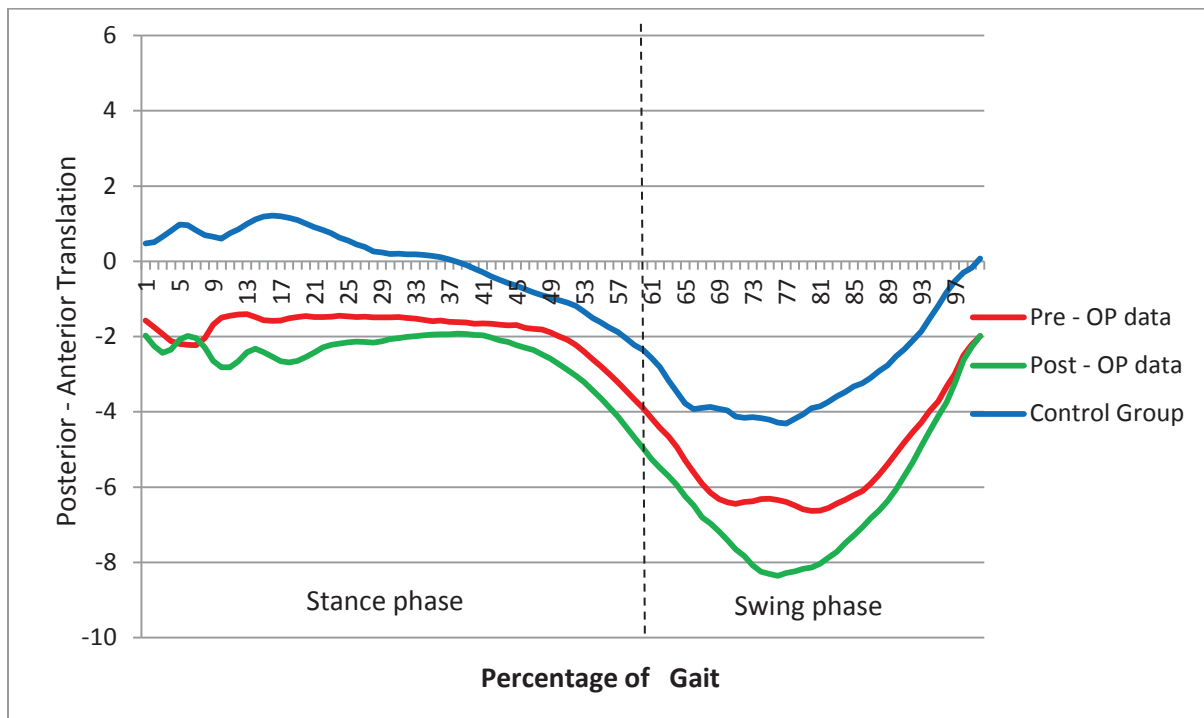


Fig. 23 Anterior/ Posterior tibial translation (mm). Positive values mean anterior translation of tibia with respect to femur.

D Discussion

The major finding in our study is that the total knee arthroplasty improves the kinematics during gait but not to the level of the control healthy group.

Outcome measures such as walking perimeter and walking velocity improved post-TKA similar to reports from previous studies [61, 83]. However, as most reports, our study also concludes that, after TKA, patients walk with an altered gait pattern in relation to movements about the knee and at a reduced velocity compared with healthy controls.

In a review of 11 gait analysis studies, McClelland et al. [84] found that similar values were reported by all for the self-selected walking velocity of TKA patients (0.8–1.1 m/s) and that these values were lower than normal when compared with their respective controls.

In the sagittal plane, the TKR group walked with reduced knee extension during stance but no significant difference at maximum knee flexion during swing phase of gait, compared with control group, but an improved knee flexion compared with pre-surgery. This finding is consistent with other studies [61] and suggests that after total knee arthroplasty the flexion during gait is more efficiently recovered while extension is more difficult to regain.

The incidence of flexion deformity following TKR may be as high as 17% [85], therefore it is reasonable to expect that following surgery TKR patients may be unable to extend their knees as much as healthy individuals without knee problems.

Previous studies of motion analysis have reported no significant differences in the knee extension during stance phase between TKR patients and controls, however these findings may be limited by small sample sizes in these studies [86, 87]. Stance phase knee extension is required primarily to provide stability for power generation during propulsion [88]. A lack of knee extension in patients may partially explain why walking can be difficult following TKR. In addition to the presence of a flexion deformity following TKR, a lack of extension may also be related to issues of muscular control. There is some evidence that hamstrings activity

during stance is prolonged in TKR patients, which may prevent full knee extension from being achieved.

A stiff knee attitude which may serve to protect the quadriceps as a feature of total knee replacement gait, pre- and post-surgery was also recently confirmed by Mandeville et al. [83]. There are only a few articles that have reported data in coronal plane.

Similar to Alnahde, Zeni and Snyder Mackler report [89], our findings revealed that the adduction angle of the post TKA and control knees demonstrated a biphasic pattern of both adduction and abduction, while the pre-TKA data showed that knees remained in adduction throughout the gait cycle. Even though in our study there was no significant difference between pre- and post surgery data, pre-operatively the knee tended to be in a more adduction position.

Knee adduction angle during stance has been shown to be related to dynamic loads during gait with higher adduction angles associated with higher dynamic loads [90]. On the other hand, Hatfield et al. [61], using principal component analysis, demonstrated that at 1-year post-TKA there was a decrease of the adduction moment that implies an overall decrease in medial compartment loading during gait.

During the initial and mid swing phase, the post-TKA group showed a significant increased adduction angle compared with control group. Our findings support the work of Leffler et al. [91] that demonstrated that patients with bicompartimental prosthesis had a systematic shift towards increased varus/decreased valgus, mainly during swing phase.

Meanwhile, Mc Clelland et al. [66] and Saari et al. [87] were one of the few authors who compared the coronal plane knee angles of patients with TKA to controls, but they found no significant difference for any patient group.

In our study, osteoarthritic knees, pre- and post-arthroplasty, walked with a smaller range of internal/external tibial rotation. They also walked with less internal tibial rotation than

controls. The internal-external rotation in knees following TKA was not restored to normal. These findings suggest that TKA patients walk with a knee that is offset into less internal rotation than controls. These results are consistent with findings of other studies. Dennis et al. [92] found that TKA group demonstrated reduced average rotational values and a reduced incidence of normal axial rotation patterns when compared with the normal knee during gait. Siston et al. [38] also, found less screw-home motion in knees following TKA than in the normal knees.

A possible cause is that the prosthetic geometries of TKAs are inconsistent with the morphology of intact knee. The internal tibial rotation in early flexion (the screw-home motion) is attributed to the function of the ACL, asymmetry between the medial and lateral femoral condyles and the asymmetry of the tibial plateau. The lateral tibial plateau of intact knee is of convex shape but most contemporary tibial inserts are symmetrically concave in design. Again, the medial condyle of intact knee is more distal than the lateral condyle. However, most TKAs employ a consistent femoral condylar height to match the symmetrical insert articulation. Therefore, ACL resection and loss of medial-lateral asymmetry following PCS-TKAs may change tibial rotation [93].

However, in our study, different from other studies [93], there was no difference between the pre- and post – TKA group and post – TKA did not display any paradoxal movement.

In our study, osteoarthritic knees displayed a more posterior tibial translation during the loading and midstance compared to control group. There was no significant difference between pre- and post TKA data. During the swing phase, the post TKA group showed an even more posterior tibial translation compared to pre-TKA data and control group.

Similar results were found in other studies. Uvehammer, Kärrholm and Brandsson [94] compared kinematics of concave versus posterior-stabilised tibial joint surface prosthesis and found out that the mean femoral anterior displacement was more important in the TKA

groups than in the normal knees. Dennis et al. [92] also reported a lack of posterior femoral rollback at TKA. Similar patterns were observed in both PCL-retaining and PCL-substituting TKA subjects. This has been attributed to the fact that the cam-and-post mechanism of most PCL-substituting TKA designs does not engage during lesser flexion activities such as gait.

The longitudinal pre- and post-surgery data presented by Smith et al. [95] indicated that pre-surgery gait patterns were retained up to 18 months after surgery. Therefore, just because individuals are no longer suffering from pain at the knee and have the ability to move through a sufficient range of motion, it does not necessarily follow that they will spontaneously modify their gait to a more normal pattern [96].

Although these findings include some error associated with soft tissue movement, the differences found between groups may provide some direction for measurement of knee kinematics in these patients using other methods.

There are clear limitations to our study. The sample size is small, but is consistent with other gait analysis studies of the TKA population. The fact that there was a prospective follow up made it difficult to have a more important number of patients. Another limitation could be the short follow-up period of 11 months post operatively considered. Van der Linden et al. [64] reports that over the years after total knee arthroplasty further improvement in knee function towards more normal values is possible, even when the overall function of the patient decreases.

D. Conclusion

In conclusion, this study is one of the few that compares the gait 3D-kinematic parameters in knee osteoarthritic patients before and after the total knee arthroplasty and a healthy age-matched control group. Following TKA, patients had better clinical, spatiotemporal and kinematic parameters. They walked longer, faster and with a better range of motion. Despite improvements, the knee kinematics during gait in TKA group differed from healthy control group. They had a lower extension, lower range of axial rotation and an increased tibial posterior translation. Future research should be focused on comparing different designs of prosthesis pre- and post operatively in a longer follow-up delay.

CHAPTER 4. FINAL CONCLUSIONS AND FUTURE PERSPECTIVES

The most important part of this work was dedicated to the investigation of the kinematics of the native knee joint, osteoarthritic knee and arthroplastic knee during gait.

In the first part of the study, we compared the kinematics of **osteoarthritic knee with an age-matched control group** using a 3D, real time, non-invasive assessment tool. The results revealed that knee OA group showed an altered “screw-home” mechanism by decreased excursion in sagittal and axial tibial rotation and a posterior translation of the tibia. On the other hand, we observed an adduction angle during whole gait cycle and an increased frontal laxity with the increase of the varus malalignment.

In the **second part of the study**, the goal was to analyze the knee kinematics of the same patients **after total knee arthroplasty** with a posterior stabilized prosthesis. Following TKA, patients walked longer, faster and with a better range of motion. But despite improvements, the knee kinematics during gait in TKA group displayed a lower extension, lower range of axial rotation and an increased tibial posterior translation compared with healthy control group.

The long-term performance of total knee replacement is dependent on the kinematics of knee joint. It should also be noted that the total knee prosthesis is designed for use under normal gait conditions. Abnormal gait patterns after TKA may accelerate damage to and deterioration of the prosthesis itself and increase the likelihood of revision surgery in the future.

In the future, complementary studies should be done to enlighten further the biomechanics of degenerative and arthroplastic knees. A complex study that analyze both kinematics and kinetics would probably gives us more informations about the knee function.

With the improvement of the sensor tracking smart technology, the analysis of these parameters in daily basis has become more feasible. Such research would provide the information of the kinematics in everyday activities and a more realistic, factual correlation of clinical and biomechanical parameters could be estimated.

Keeping in mind that the expectations of the TKA recievers are getting higher, the prospective research should be focused on improvement of the designs and surgical techniques of total knee prosthesis to meet patients satisfaction.

Another area for future work would be to further analyze the kinematics and kinetics of the knees during initiation and progression of the osteoarthritis. This could be done by investigating periodically the native healthy, young knees and to analyze the evolution of their biomechanics during the years. The results may reveal the time and the factors that initiate knee osteoarthritis and could lead to a better prevention.

RÉSUMÉ

Analyse cinématique de la marche chez des patients souffrant d'arthrose du genou; Pré - et post - arthroplastie totale du genou

Introduction. L'arthrose est un processus dégénératif qui affecte typiquement les articulations synoviales du corps. L'augmentation de l'espérance de vie et donc du nombre de personnes âgées devraient faire de l'arthrose la quatrième cause d'invalidité en 2020.

L'arthrose du genou représente une des formes les plus répandues de l'arthrose, des études estiment la présence de la maladie à un stade radiographique sévère chez 1% des 25-34 ans, et 30% dans la population de 75 ans et plus. Les patients atteints d'arthrose ont tendance à modifier les paramètres spatiaux et temporels pendant la marche pour réduire la douleur. Il existe des caractéristiques de la marche significativement liées à l'arthrose du genou, comme une augmentation du moment d'adduction et de la raideur articulaire, une diminution de l'angle de flexion du genou et de la vitesse de marche.

L'arthroplastie totale du genou (PTG) est considérée comme le traitement de référence pour l'arthrose du genou en phase terminale. Près d'un million de prothèses totales de genou sont implantées dans le monde entier chaque année. Environ 130 000 arthroplasties du genou sont effectuées chaque année aux États-Unis d'Amérique, et plus de 60 000 en France.

Cependant, la réduction de la fonction du genou ne semble que partiellement corrigée par la chirurgie.

Le but de cette thèse était d'étudier, in vivo, la cinématique en 3D du genou lors de la marche sur des patients souffrant d'arthrose du genou et de quantifier l'apport de l'arthroplastie totale du genou (PTG) sur la restauration d'une cinématique normale.

Matériel et méthodes. Cette étude prospective a été réalisée durant la période de Janvier 2011 à Janvier 2014, dans le laboratoire de biomécanique aménagé au sein de notre centre clinique. Afin de répondre aux objectifs de recherche, des patients atteints d'arthrose médiale du genou qui étaient programmés pour une opération d'arthroplastie ont été sélectionnés pour l'analyse cinématique.

Trente patients ont été inclus dans la première étude. Un groupe de contrôle composé de 12 participants du même âge, sans antécédents de lésions musculo-squelettiques ou chirurgie des membres inférieurs et qui ne présentaient aucune instabilité ligamentaire mesurable sur l'examen clinique a été sélectionné.

Sur les 30 patients qui ont été inclus dans cette première étude, nous avons obtenu des évaluations de suivi après l'arthroplastie totale du genou sur 20 patients, avec un délai moyen de 11 mois. La prothèse qui a été utilisée était une prothèse totale de genou postéro-stabilisée.

L'analyse cinématique tridimensionnelle du genou a été réalisée en utilisant le système KneeKGTM. Le système KneeKGTM a été développé dans l'objectif d'obtenir une évaluation quantifiée précise du comportement cinématique de l'articulation du genou. Il est composé de détecteurs de mouvement passifs fixés sur un harnais (la précision du suivi de ces détecteurs par rapport à l'os sous-jacent ayant été validée), un système de capture de mouvement infrarouge (caméra Polaris Spectra, Northern Digital Inc.), un tapis roulant et un ordinateur équipé de la suite logicielle Knee3DTM (Emovi, Inc.). L'efficacité du système KneeKG a déjà été démontrée dans le cadre de différentes études cliniques.

Après calibrage, les données cinématiques étaient collectées durant une tâche de marche à vitesse confortable (sélectionnée par le patient) sur tapis roulant. L'ensemble de la procédure durait 20-25 minutes. Pour chaque participant, 4 paramètres biomécaniques constitués par les 3 angles du genou: flexion-extension, d'abduction-adduction, rotation tibiale interne-externe et de translation antéro-postérieure, ont été calculés.

Pour les variables non paramétriques, le test de Mann-Whitney a été utilisé avec un niveau de signification de 5% afin de vérifier les différences entre les groupes. ANOVA test a été menée avec l'ensemble alpha à 0,05 pour toutes les variables paramétriques. Toutes les analyses statistiques ont été effectuées en utilisant SPSS v21 (SPSS Inc, Chicago, Illinois, USA).

Résultats. Le groupe avec arthrose de genou marchait avec une vitesse inférieure comparativement au groupe témoin ($p < 0,05$). Le périmètre de marche pour la plupart des patients du groupe avec arthrose était d'environ 1 km. L'amplitude de mouvement dans le plan sagittal était significativement plus faible pour le groupe avec arthrose par rapport au groupe témoin ($p < 0,05$). Les patients atteints d'arthrose du genou présentaient une extension réduite au cours de la phase d'appui ($p < 0,05$) et une flexion réduite pendant le « push-off » et en début de la phase oscillante ($p < 0,05$). L'angle d'adduction était constamment plus grand pour les patients atteints d'arthrose ($p < 0,05$). La laxité frontale chez les patients atteints d'arthrose était positivement corrélée avec l'angle de varus ($r = 0,42$, $p < 0,05$). L'amplitude des mouvements de rotation interne-externe des patients arthrosiques était significativement inférieure à celle du groupe témoin ($p < 0,05$). On notait une différence significative ($p < 0,05$) dans la rotation du tibia en milieu de phase d'appui : les patients atteints d'arthrose conservaient une position neutre tandis que le groupe contrôle présentait une rotation tibiale interne. Au contact initial, le tibia du groupe arthrose était significativement en translation

postérieure par rapport au fémur, en comparaison au groupe témoin ($p < 0,05$). Le tibia du groupe avec arthrose restait dans une position plus postérieure tout au long du cycle de marche.

Les patients marchaient plus vite après arthroplastie totale du genou comparativement avec l'évaluation pré-opératoire ($p < 0,05$), mais à une vitesse inférieure comparativement au groupe témoin. Pendant la marche, l'amplitude de flexion / extension a été améliorée significativement ($p < 0,05$) après arthroplastie totale de genou, mais celle-ci restait encore inférieure à celle du groupe contrôle. La flexion maximale pendant la phase oscillante était significativement plus élevée dans le groupe arthroplastie, en comparaison avec l'évaluation pré-OP ($p < 0,05$). Il n'y avait pas de différence significative pour ce paramètre entre le groupe arthroplastie et le groupe témoin. D'autre part, il n'y avait pas de différence dans le groupe avec arthrose du genou avant et après l'arthroplastie pour l'extension maximum au cours de la phase d'appui. Même si une amélioration était visible dans le mouvement dans le plan frontal après arthroplastie totale du genou, la différence n'a pas atteint le degré de significativité. L'amplitude de mouvement dans le plan axial n'a pas changé avant et après arthroplastie, mais est restée inférieure à celle du groupe témoin ($p < 0,05$). La translation postérieure maximale pendant la phase envol était significativement plus élevée dans le groupe après arthroplastie comparativement au groupe témoin ($p < 0,05$). D'autre part, il n'y avait pas de différence entre les données pré- et post-OP.

Conclusion. Cette analyse de la marche a révélé que la cinématique de genou avec arthrose médiale diffère de la cinématique du genou sain. La principale observation de la première étude était que le groupe avec arthrose du genou montrait une stratégie de raidissement de la marche en présentant une réduction de mouvement non seulement dans le plan sagittal, mais aussi dans le plan axial. Le groupe avec arthrose de genou a manifesté un mécanisme de "screw-home" altéré, avec une diminution d'excursion dans les plans sagittal et axial : rotation axiale et translation postérieure du tibia. D'autre part, nous avons observé un angle d'adduction augmenté pendant toutes les phases du cycle de marche et une laxité frontale augmentée avec l'augmentation du défaut d'alignement en varus.

Après PTG, les patients avaient de meilleurs paramètres cliniques, spatio-temporels et cinématiques. Ils marchaient plus longtemps, plus vite et avec une meilleure amplitude de mouvement. Malgré les améliorations, la cinématique du genou lors de la marche dans le groupe PTG différait de celle du groupe contrôle. Après une arthroplastie totale du genou, la flexion lors de la marche est plus efficacement récupérée tandis que l'extension est plus difficile à regagner. Le groupe post-OP a montré une augmentation significative de l'angle d'adduction par rapport au groupe contrôle qui peut entraîner des charges dynamiques plus élevées sur le compartiment médial. La rotation interne-externe du tibia dans les genoux après arthroplastie n'a pas été rétablie à sa valeur normale. Au cours de la phase oscillante, le groupe post-OP a montré une translation tibiale postérieure accrue comparativement aux données pré-OP et du groupe contrôle. Une explication possible est que les géométries prothétiques de PTG ne reproduisent pas complètement la morphologie du genou intact.

Il faut noter que la prothèse totale du genou est conçue pour une utilisation dans des conditions normales de marche. La performance à long terme de la prothèse totale de genou dépend de la cinématique de l'articulation du genou. Des déviations anormales de la marche

après PTG peuvent accélérer la dégradation et la détérioration de la prothèse elle-même et ainsi augmenter la probabilité d'une chirurgie de révision.

Dans l'avenir, des études complémentaires devraient être faites pour mieux comprendre la biomécanique des genoux dégénératifs et arthroplastiques. Une étude complexe, qui analyse à la fois la cinématique et la cinétique de différents modèles de prothèses en pré et post-opératoire, avec un délai post-OP plus long, pourrait probablement apporter plus d'informations sur la fonction du genou.

Mots-clés : genou, arthrose, prothese totale de genou, analyse de la marche, cinématique

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PUBLICATIONS

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Gait Knee Kinematic Alterations in Medial Osteoarthritis – 3D assesemen

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Background:

Knee osteoarthritis (OA) is a common cause of functional disability, the medial compartment being the most affected. Patients with osteoarthritis (OA) tend to modify spatial and temporal parameters during walking to reduce the pain and knee joint moments. Although the kinematic changes in the sagittal plane of OA patients have been elucidated, very few studies have analyzed changes in the frontal and horizontal planes. Therefore the aim of this study was to investigate in vivo three dimensional knee kinematics in patients suffering from knee OA during walking.



Fig.1. Knee KG system

Material and methods:

Thirty patients with medial knee OA and a control group with age-matched subjects were prospectively collected for this study. All subjects were examined with KneeKG system, (Fig.1) while walking on a treadmill at a self-selected speed. The KneeKG is composed of passive motion sensors fixed on the validated knee harness, an infrared motion capture system (Polaris Spectra camera, Northern Digital Inc.), and a computer equipped with the Knee3DTM software suite (Emovi, Inc.). In each trial, we calculated the angular displacements of flexion/extension, abduction/adduction, and external/internal tibial rotation.

Results:

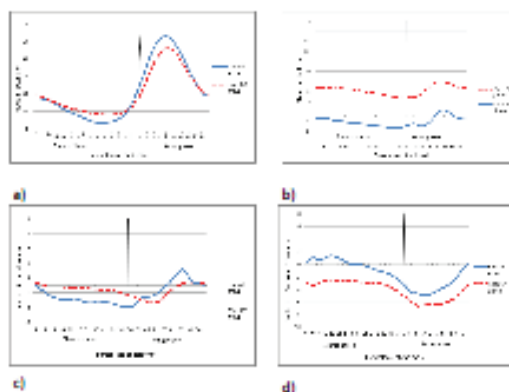


Fig.2. The knee motion during a gait cycle: a) flexion/extension, b) adduction/abduction, c) internal/external rotation, and d) anterior/posterior translation

Kinematic characteristics	OA (n=30) Mean (SD)	Control (n=30) Mean (SD)	P
Speed	1.25 (0.15)	1.28 (0.12)	<0.05
Relative vertical displacement	14.5 (1.5)	14.8 (1.2)	<0.05
Median knee flexion angle	52.4° (3.1)	53.8° (2.4)	<0.05
Median extension during stance	1.5° (1.0)	2.2° (0.8)	<0.05
Median flexion during swing	48.1° (3.2)	50.4° (2.1)	<0.05
Median knee extension during	1.0° (0.5)	1.2° (0.7)	<0.05
Range of flexion/extension	43.8° (3.1)	52.3° (2.2)	<0.05
Abduction (SD) and Adduction (SD) during stance	1.1° (0.7)	0.8° (0.5)	<0.05
Range motion of abduction	1.6° (1.2)	1.1° (0.8)	<0.05
Internal-rotation (SD) and External-rotation (SD)	0.2° (0.1)	0.1° (0.1)	<0.05
Range motion of internal rotation	0.3° (0.1)	0.2° (0.1)	<0.05
Anterior-translation (SD) and Posterior-translation (SD)	1.2mm (0.5)	0.8mm (0.4)	<0.05
Range of Anterior-posterior Translation	1.2mm (0.5)	0.8mm (0.4)	<0.05

Tab. 1 The kinematic data of the knee OA and control group (* statistically significant)

Conclusion:

Weight-bearing kinematics in medial OA knees differs from normal knee kinematics. Knee OA group showed an altered "screw-home" mechanism by decreased excursion in sagittal and axial tibial rotation and a posterior translation of the tibia. These data will be taken in consideration in a further study that evaluates the kinematic outcomes of the patients after total knee replacement.



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 Ahlback'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
Ahlback'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.

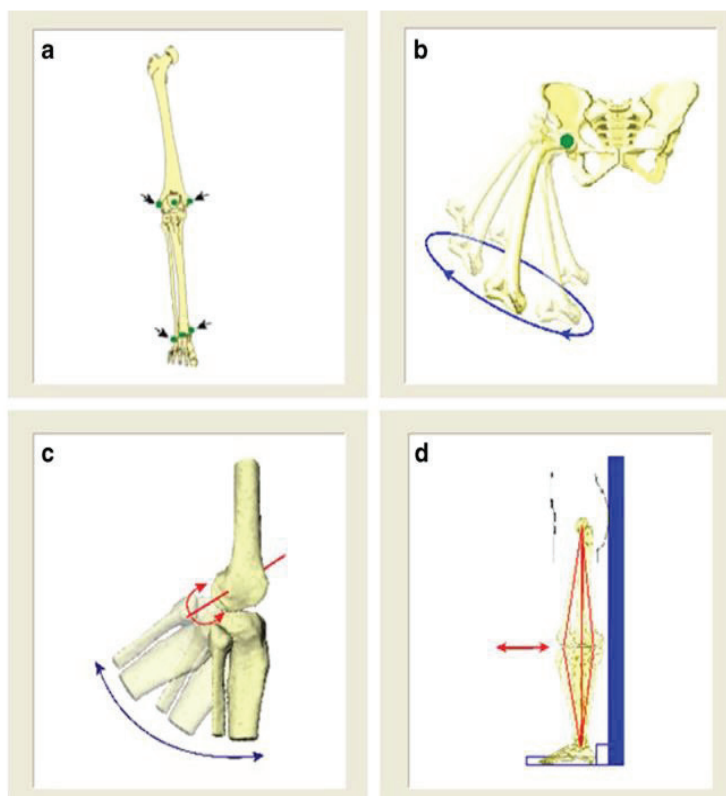
Once system installation and calibration was finalised, the study participant was asked to walk at the preselected speed, and 3D displacements of the reflective markers were recorded by the operator. Once calibration and measurement is performed, the Knee3D suite computes various angle values on anatomical 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 ($7.6^\circ \pm$

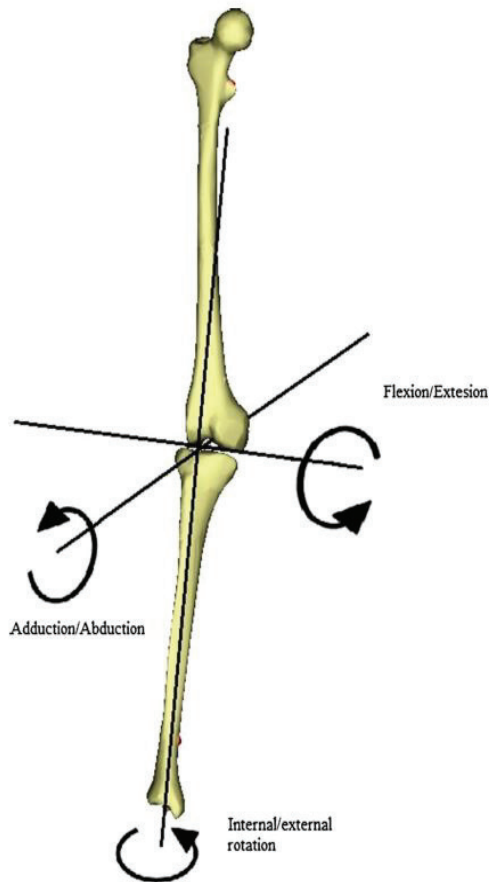


Fig. 3 Knee rotation axes

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 \pm 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

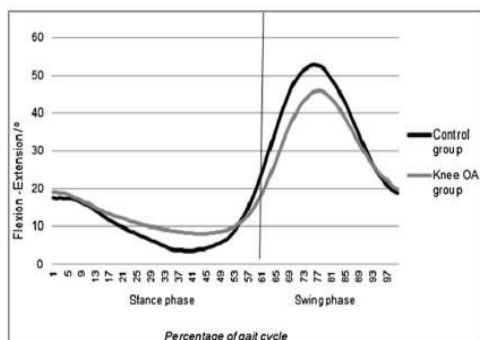


Fig. 4 Knee flexion (+)/extension (-) angle during gait cycle

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

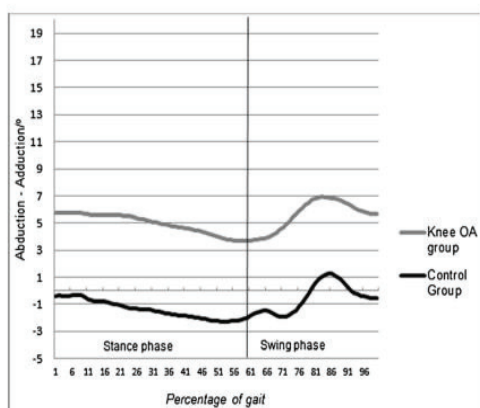


Fig. 5 Knee adduction (+)/abduction (-) angle during gait cycle

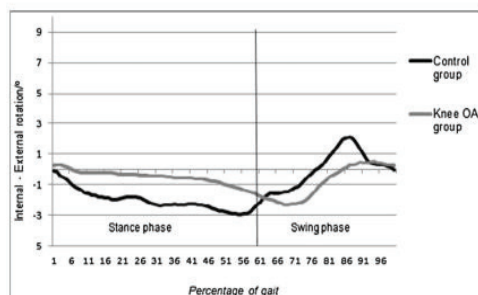


Fig. 6 Knee external (+)/internal (-) rotation angle during gait

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

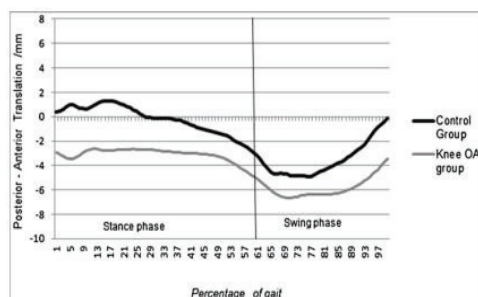


Fig. 7 Tibia anterior (+)/posterior (-) translation (mm) with respect to femur

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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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 (ACLD) 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 ACLD 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 ACLD and control group, respectively). A significant difference in tibial rotation angle was found in ACLD knees compared to control knees

($p < 0.05$). The patients with ACLD 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 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. 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 ACLD 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 (ACLD) knees [2, 7, 15, 29]. In vitro studies using cadavers [28, 34] 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.

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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 pre-morbid 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

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	<i>p</i> value
Flexion–extension- <i>SP</i> (mean)	13.2° ± 2.1°	7.3° ± 2.7°	<i>p</i> < 0.05
Range of flexion–extension	45.9° ± 7.3°	54.2° ± 4.5°	<i>p</i> < 0.001
Internal–external rotation- <i>MSP</i> (mean)	−1.4° ± 0.2°	0.2° ± 0.3°	<i>p</i> < 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	<i>p</i> < 0.05

SP Stance phase, *MSP* Mid-stance phase

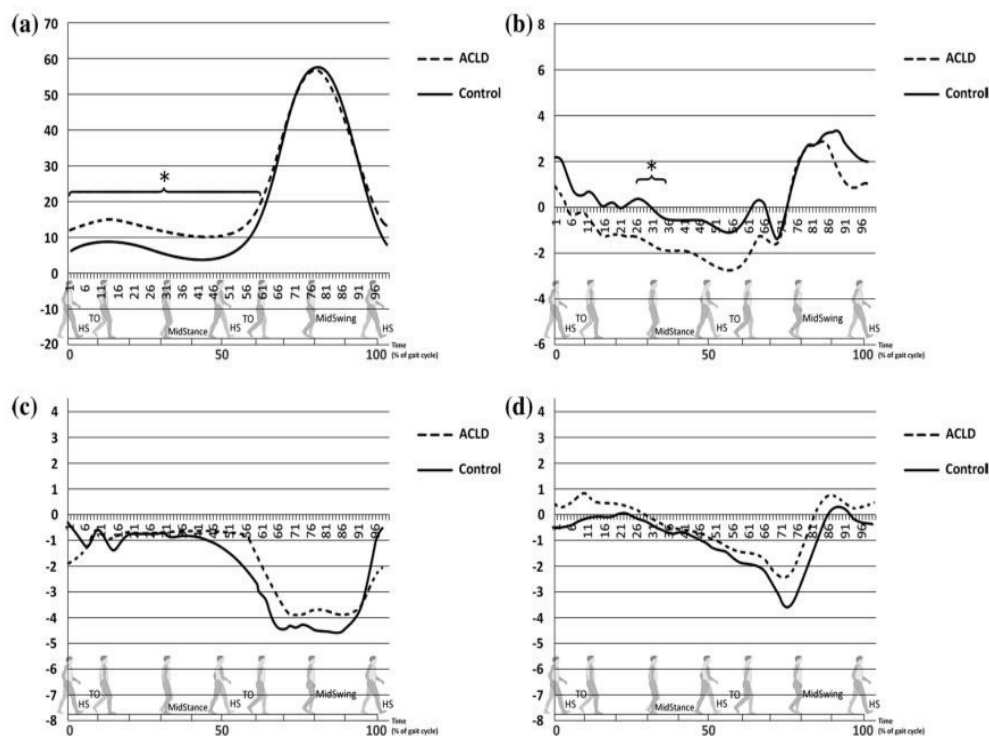


Fig. 1 Tibiofemoral kinematics of ACLD knees and healthy control knees. **a** Sagittal plane. **b** Axial plane. **c** Anteroposterior tibial translation. **d** Coronal plane; (*) Indicates significant difference between ACLD knees and control knees

and 34 % of the gait cycle ($-1.4^{\circ} \pm 0.2^{\circ}$ 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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Résumé de la thèse :

Le but de cette thèse était d'étudier, in vivo, la cinématique en 3D du genou lors de la marche sur des patients souffrant d'arthrose du genou et de quantifier l'apport de l'arthroplastie totale du genou (PTG) sur la restauration d'une cinématique normale.

Trente patients et un groupe de contrôle composé de 12 participants du même âge ont été inclus dans la première étude. Sur ces 30 patients, nous avons obtenu des évaluations de suivi après l'arthroplastie totale du genou sur 20 patients, avec un délai moyen de 11 mois. L'analyse cinématique tridimensionnelle du genou a été réalisée en utilisant le système KneeKGTM.

Cette analyse de la marche a révélé que la cinématique de genou avec arthrose médiale diffère de la cinématique du genou sain. Le groupe avec arthrose du genou montrait une stratégie de raidissement de la marche en présentant une réduction de mouvement non seulement dans le plan sagittal, mais aussi dans le plan axial. Après PTG, les patients avaient de meilleurs paramètres cliniques, spatio-temporels et cinématiques. Malgré les améliorations, la cinématique du genou lors de la marche dans le groupe PTG différaient de celle du groupe contrôle.

Mots-clés : genou, arthrose, prothèse totale de genou, analyse de la marche, cinématique

DISCIPLINE :

Biomécanique

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