Preoperative individualized planning tool to determine 3D tunnel orientation for ACL reconstructions

MSc. Thesis

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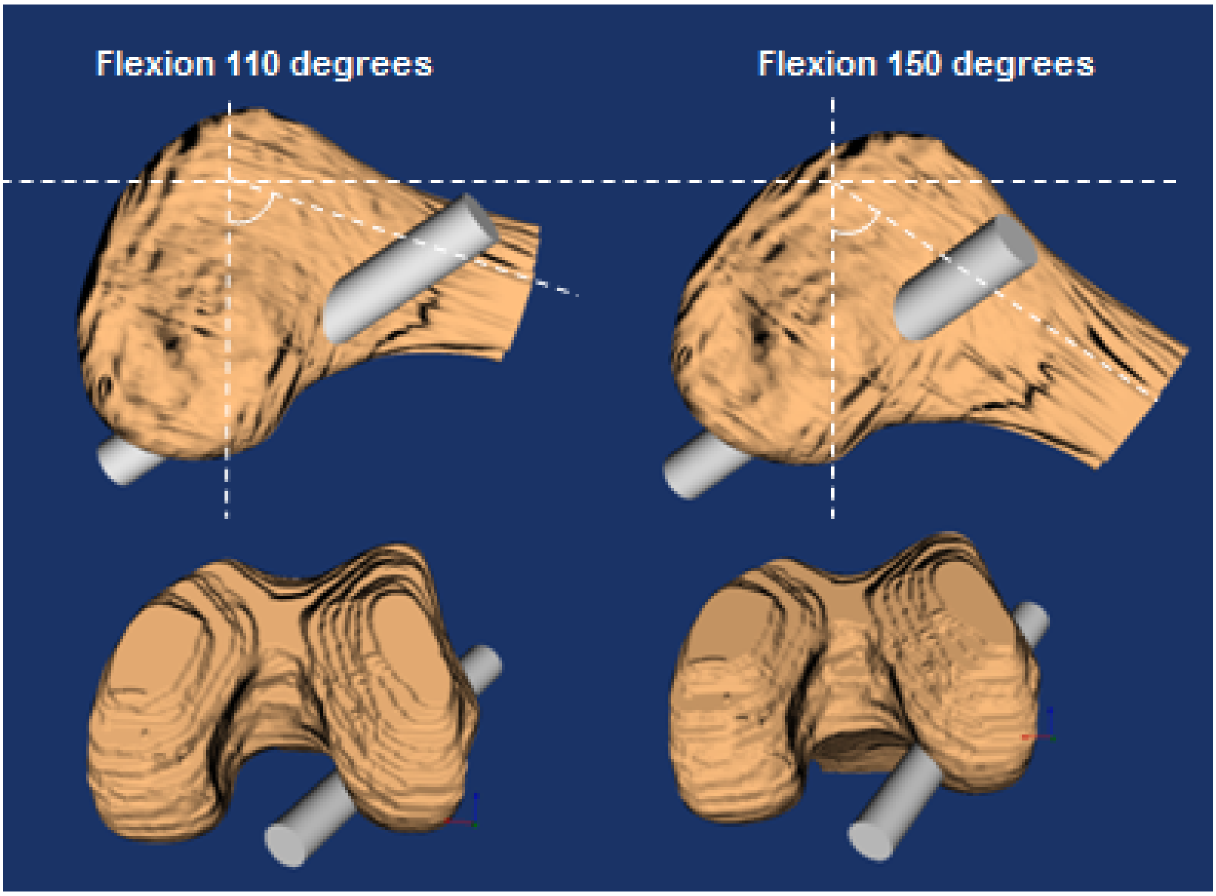

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Abbreviations

ACL	Anterior Cruciate Ligament
AMB	Anteromedial Bundle
DICOM	Digital Imaging and Communications in Medicine
FE	Finite Element
LCL	Lateral Collateral Ligament
MCL	Medial Collateral Ligament
MRI	Magnetic Resonance Imaging
OA	OsteoArthritis
PCL	Posterior Cruciate Ligament
PLB	Posterolateral Bundle
PSM	Patient-specific modeling
TT	Trans Tibial technique
MP	Medial Portal technique
VTK	The Visualization Toolkit

Preface

This is my master thesis for the mastertrack Robotica and Imaging of my study Technical Medicine at the University of Twente (Enschede). I have accomplished my research project at the department of Orthopaedic Surgery of the Onze Lieve Vrouwe Gasthuis (OLVG) in Amsterdam.

A few years ago, shortly before I graduated in Computer Science, I heard about the study Technical Medicine. Directly, I found Technical Medicine an interesting study in which new techniques can be developed and applied for treatment options for patients. My research project illustrate the possibilities arisen from studying Technical Medicine. They are exactly corresponding to the expectations I had at the beginning of my study.

My thesis describes the development of a new preoperative individualized planning tool to determine the native insertion region and orientation of the femur and tibia tunnels for primary ACL reconstructions. This topic fascinates me, because it includes a Technical Medicine problem in which informatics solutions can provide support.
The problem definition of my thesis arose during my 2nd masters internship at the department of Orthopaedic Surgery of the OLVG. In the literature has become clear that non-anatomic ACL reconstruction do not restore normal knee function. Many studies focused on the anatomical positioning of the femoral tunnel at the native insertion site in order to improve tunnel placement. Studies only focused on a 2D orientation and no studies described the 3D orientation for optimal individualized tunnel orientation. This gap in knowledge was the motivation to describe the first steps to develop a 3D individualized planning tool for ACL reconstruction.

Abstract

Background: The Anterior Cruciate Ligament (ACL) is one of the most frequently injured ligaments of the knee. ACL injuries are commonly treated by surgery, which attempts to restore the translational and rotational stability of the knee. The literature report failure rates between 10% and 25% after primary ACL reconstruction of which 70% to 80% is caused by inadequate tunnel placement.

Problem: In recent years, many studies only focused on the anatomical positioning of the femoral tunnel at the native insertion site in order to improve tunnel placement. However, the current surgical arthroscopic ACL reconstruction technique provides no unambiguous protocol for femoral tunnel drilling. Therefore, there is still a considerable freedom in drilling angle, limited by flexion angle of the knee, the femoral notch shape and the location of the endoscopic portals. As a consequence, standards for the tunnel-exit on the lateral wall are not defined, although in the literature is described that a sharp bending angle between the femoral and tibial tunnel by a non-optimal drilling angle results in graft damage due to repetitive bending stress forces at the femoral tunnel aperture.

Hypothesis / Purpose: The clinical outcome after ACL reconstruction will be improved by determining an individualized optimal femoral and tibia tunnel orientation.

Methods of approach: In this study, a 3D computer model is developed to pre-operatively assess the optimal patient-specific tunnel orientation. MRI-scans are used to generate an individual 3D-model of the knee. The optimal patient-specific tunnel orientation is subsequently determined with a focus on a minimal bending angle of the graft at femoral and tibial tunnel aperture over the entire range of motion. The computer model will account for the surgical boundary conditions and patient-specific anatomical characteristics. The 3D model is evaluated by calculating the optimal tunnel orientation using MRI scans of operative patients. These findings were then compared with the actually placed tunnels, because we suggest that tunnel position of failed ACL reconstruct significant differs from the predicted tunnel placement of our developed 3D model.

Results: Optimal patient-specific tunnel orientation were calculated with the developed 3D computer model on MRI-scans of operative patients with known clinical outcome. Tunnel orientation, the effect of femoral tunnel position on surface area and shape of the aperture, graft orientation and inclination, elongation of the graft were compared with knowledge in the literature. Results of calculation were in terms of horizontal and vertical tunnel angles which can be used by drilling the tunnels intraoperatively.

Conclusion: The present study demonstrates a 3D computer model to individualize ACL reconstruction in terms of femoral and tibial tunnel orientation to reproduce the native ACL characteristics as closely as possible. Further work is necessary to determine the relevance of the morphologic orientation of both tunnels at clinical outcome after ACL reconstruction. Some additional studies are needed to actually apply the 3D model in a clinical setting.

Contents

1 Introduction

This first chapter will describe the clinical background of the Anterior Cruciate Ligament. Furthermore, the problem definition, the overall goal of the study and the scope of this thesis will be described.

1.1 Anterior Cruciate Ligament

1.1.1 Anterior Cruciate Ligament injuries

The Anterior Cruciate Ligament (ACL) is one of the most frequently injured ligaments of the knee with an incidence of 30 per 100.000 persons [1]. Seventy percent occur during sports activities with a peak incidence between 16 and 39 years, with higher injury rates in females [2].
Acute ACL injuries can be treated conservatively or by surgery. Treatment options depend on a patient's expectation, age and their level of participation in sports. In young and active patients, a reconstruction of the ACL is often chosen with a physical therapy program preoperatively to strengthen the muscles. Delay in surgery is associated with meniscal or cartilage injuries and it has been suggested that this increases the risk for developing osteoarthritis [3–9]. However, not all studies could demonstrate that ACL reconstruction protects the knee joint against this.

1.1.2 ACL reconstruction

ACL reconstruction is the sixth most commonly performed surgical procedure in orthopedic surgery which attempt to restores the translational and rotational stability of the knee. Each year, 100.000 patients are treated surgically in the United States, compared with 7000 patients in the Netherlands [2, 10].
Several surgical techniques are available for ACL reconstruction, varying in surgical approach, type of graft (such as allo- or autograft and type of tendon; hamstring, quadriceps or patellar tendon), graft fixation method, drilling technique (inside-out or outside-in) and the number of bundles restored (single or double bundle).
The ACL consist of two different bundles, the AnteroMedial Bundle (AMB) and a PosteroLateral Bundle(PLB). The bundles vary in length, width, area, function and the tension of the bundles differ during movement of the knee [11]. Single bundle ACL reconstruction is most frequently performed and restoring mainly the function of the AMB [1]. Double-bundle reconstruction reconstructs both bundles and attempt to restore both the anterior-posterior and rotational translation [12–14]. The double bundle method is technically more complex to perform compared with single bundle reconstruction, accompanied by a longer surgery time. There is considerable discussion regarding the use of single- or double-bundle technique [15]. Studies have found few to no superior effects of a double-bundle reconstruction in terms of patient-perception, knee function and complications [16, 17].
Clinical outcome after ACL reconstruction depends on several factors; such as graft selection, tunnel placement, initial graft tension, graft fixation method, graft-tunnel motion, rate of graft healing and optimal timing of postoperative rehabilitation.
Patient's specific characteristics, such as level of activity, age, gender and expectations also play an important role in clinical outcome [6, 18, 19]. Injuries to secondary ligaments, capsular structures, articular cartilage and meniscus also affect the overall success or failure after ACL reconstruction [19]. The most common complaints seen after primary surgery are persistent instability, recurrent knee pain and a disability to return to preinjury level of (sport)activities [20–22]. One of the determining factor in obtaining a good clinical outcome after ACL reconstruction is secure graft incorporation in bone to prevent re rupture of the graft. Kim et al. [11, 37] described that a sharp bending angle between the femoral and tibial tunnel by a non-optimal drilling angle results in graft damage due to repetitive bending stress forces on the graft at the femoral tunnel aperture. Due to these shearing forces between graft and bone, a delayed graft-tunnel incorporation could occur.

1.2 Problem

The literature report failure rates between 10% and 25% after primary ACL reconstruction of which 70% to 80% is caused by inadequate tunnel placement [6, 18, 23–29]. As a consequence, revision surgery is seen in 10% after primary ACL reconstruction resulting in higher additional social and healthcare costs [18].

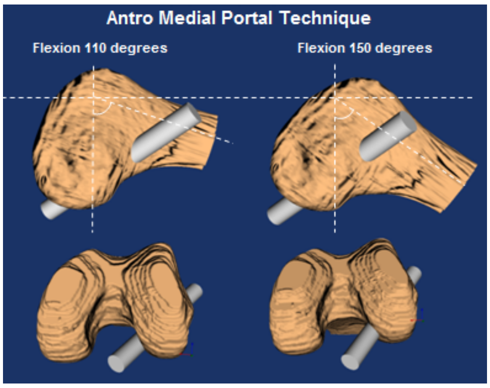


Figure 1: **Femoral tunnel orientation depends on knee flexion angle in which the tunnel is drilled.** A schematic representation of the influence of drilling flexion angle on tunnel-exit position (lateral wall of the femur condyle). Flexion at 110-130 degrees provides a more vertical tunnel in comparison with higher flexion angle which results in a more horizontal and anterior tunnel-exit position.

In recent years, many studies only focused on the anatomical positioning of the femoral tunnel at the native insertion site in order to improve tunnel placement in a 2D orientation. However, no research is done to an appropriate tunnel-exit on the lateral wall. Despite the fact that in the literature is described that a sharp bending angle between the femoral and tibial tunnel by a non-optimal drilling angle results in graft damage due to repetitive bending stress forces at the femoral tunnel aperture. Studies have tried to minimize failure rates by define the optimum tunnel in a 2D orientation.
However, the clinical outcome after ACL reconstruction might be improved by determining the optimal drilling angle for the femoral and tibia tunnel in a 3D orientation, depending on patient-specific characteristics, such as shape and size of the femur condyle [30–35].

1.3 Ultimate goal of the study

The overall goal of the study is to determine pre-operative optimal patient-specific tunnel orientation for both ACL and Posterior Cruciate Ligament (PCL) reconstruction techniques. This will be achieved by creating a 3D computer model that takes into account patient-specific characteristics, boundary conditions, biomechanical aspects of the ligaments (angle, strength, force).
The workflow is outlined in Figure 2 and consists of three different main processes. First, a preprocessing/modeling phase in which a patient-specific 3D computer model of the knee is created from an imaging scan. Patient-specific characteristics, the size of the tunnel aperture and boundary conditions of the operation method will also be included. Based on these data, the most appropriate tunnel orientation will be calculated. Second, the inter-operative phase, in which the calculated patient-specific tunnel orientation in the preprocessing phase will be applied to the operation procedure. In the final phase, we will evaluate the tunnel orientation postoperatively from an imaging scan and compare this to the patients physical function, obtained from physical examination and Patient Reported Outcome Measurement instruments.

1.4 Scope of this thesis

The aim of this thesis is to find a suitable computer platform, specify the boundary conditions, setup the workflow of calculating tunnel orientation in a 3D computer model for single bundle ACL reconstructions. The focus in these thesis is therefore particularly on the preprocessing / planning stage such as described in Figure 2. The research question resulting from the goal of the thesis is:

"Is it possible to determine patient-specific femoral tunnel in 3D orientation preoperatively with a 3D computer model for single bundle ACL reconstruction?".

The hypothesize is that with a 3D computer model an optimal pre-operative tunnel orientation for each individual patient can be determine, resulting in a reduction of ACL reconstruction failure rate.

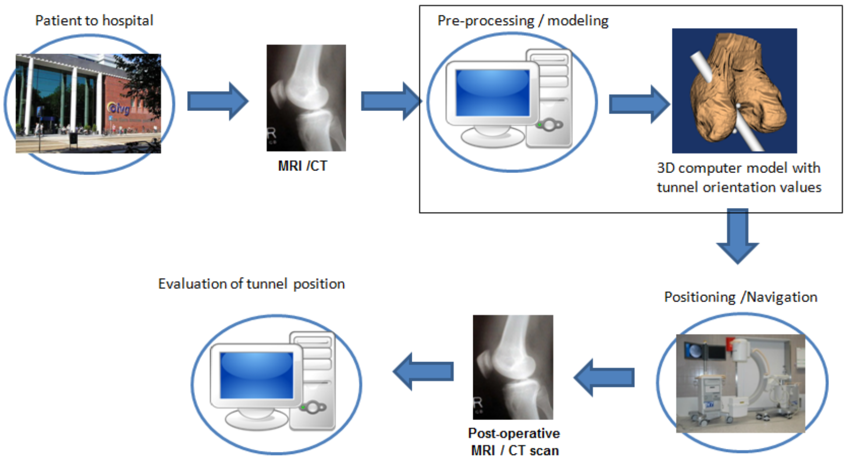


Figure 2: **Ultimate goal of the study.**
A patient with an ACL rupture presents at the outpatient department. A Pre-operative MRI / CT scan is obtained and a 3D computer model is generated from these scan. The most optimal femur and tibia tunnel orientation will be calculated for the patient. Intra-operatively the 3D model is linked to a navigation system to assist by accurate placement of the tunnel. Afterwards a postoperative imaging scan is obtained for the evaluation of the surgical placed tunnels.

From the beginning must be assumed that the design and the software program language of the develop computer program must be suitable to implement in daily clinical practice.

1.5 Clinical relevance

The clinical outcome of an ACL reconstruction could be improved by individualization of tunnel orientation, since optimizing graft-to-bone healing will improve a patients return to preinjury level of activities, satisfaction and both translational and rotational stability of the knee. A reduction of technical failures (70-80%) of the 10-25% failures of primary ACL surgeries could thereby be accomplished.

1.6 Report outline

Background information about ACL injuries and reconstruction, the definition of the problem, the overall goal of the study and the scope of this thesis will be described in Chapter 1. Chapter 2 will provide a description of the method of approach to develop a 3D computer model and calculate optimal tunnel orientation. Schematic overview of processes in the planning phase and input and output parameters are defined. Chapter 3 describes a minor overview of 3D models of the knee in the literature. In chapter 4 is in minor detail the workflow of determination of the optimal tunnel orientation described. Patient characteristics, tunnel length, aperture area and shape, tunnel diameter were take into account to reproduce the native ACL characteristics as closely as possible and compared with knowledge in the literature.
The 3D model is evaluated with MRI-scans of patients with known clinical outcome after ACL reconstruction and whereof surgical placed tunnel can be obtained and compared with model-based calculated tunnel orientation. These (preliminary) results are described in Chapter 5. Finally in Chapter 6, the key findings, limitations and future implications and overall conclusion of these study will be described.

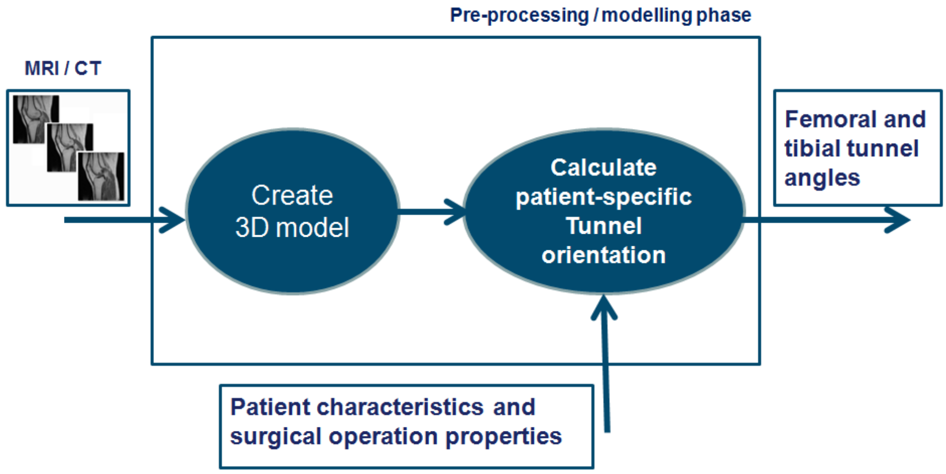


Figure 3: **Preprocessing / planning phase in more detail**
The main focus in these thesis is particularly on the preprocessing / planning phase as outlined in Figure 2. The two most important overall sub-processes are illustrated here. MRI/CT input-data can be used to generate a surface model of both the femur and tibia. Patient-specific tunnel orientation can then be calculated by using the generated surface model and input-date such as patient-specific characteristics and surgical operation boundary conditions. The ouput-data of this process is in terms of horizontal and vertical tunnel angles, that can be used intraoperatively.

2 Methods

In order to develop a 3D model to determine best patient-specific tunnel orientation various steps will be investigated.

First, a literature research will be performed on ACL reconstructions, anatomy of the ACL, tunnel orientations and 3D computer models.

On the basis of the literature results and the standard orthopedic ACL reconstruction protocol an algorithm to calculate patient-specific tunnel orientation needs to be developed.
In Figure 3 is shown a schematic overview of the Preprocessing / planning phase. The input parameter of preprocessing / planning phase as described earlier, is the pre-operative obtained MRI / CT data from which the femur and tibia geometry will be constructed. Generating a correct 3D model of the native anatomy is of high importance, and can be achieved by a quality generated mesh. This meshing of the femur and tibia geometry is of major influence to determine tunnel orientation accurately and provides an acceptable and workable computation time. This sub-process will be described in further detail in chapter 5.
In the second sub-process we will calculate the patient-specific tunnel orientation on the generated mesh. Necessary input parameters are patient-specific properties and boundary conditions related to the surgical method (single bundle reconstruction).
The implementation of this sub-process is largely based on some basic principles of an ACL reconstruction [21].

- The first principle is to define the native insertion sites of the ACL bundles and by placing the tunnels in the true anatomic positions.
- The second principle is to take patient-specific characteristics into account, such as tunnel diameter which is tailored to graft size.
- The third principle is a correct tensioning pattern of the bundle to restore their native tensioning behaviors.

The ouput data of this process is in terms of horizontal and vertical tunnel angles which can be used by drilling the tunnels intraoperatively.

The above described workflow is elaborated in more detail together with the resulting algorithm to calculate patient-specific optimal tunnel orientation should need to be written. In parallel, a suitable computer platform to develop the 3D computer model to calculate, analyze and visualize has to be found.

The last step is to evaluate the developed 3D model with patients MRI data which have undergone an ACL reconstruction. Patient-specific tunnel orientation determined using the 3D model will be compared with tunnels actually drilled during surgery.

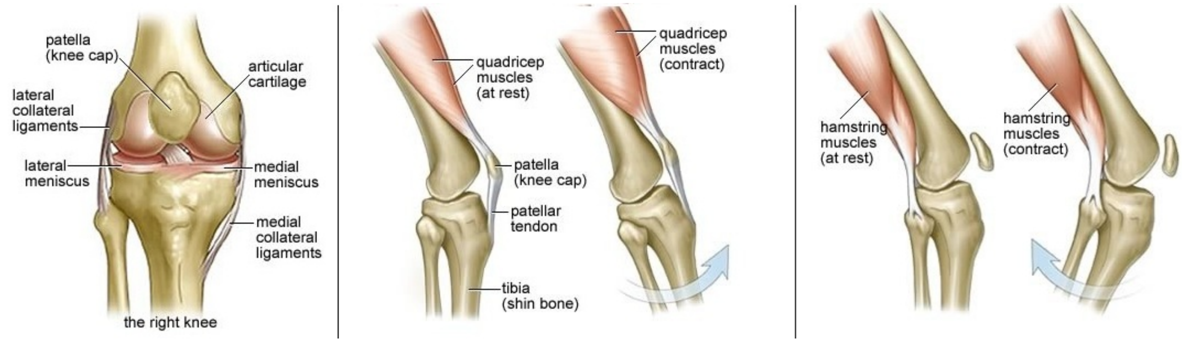


Figure 4: Anatomy of the knee joint. The knee joint consist of three bones; the patella, femur and tibia and a complex arrangement of ligaments, tendons and muscles. Contracting the quadrices or hamstring the knee extend and flex respectively [36].

3 Background Anterior Cruciate Ligament

This chapter will discuss the anatomy of the knee joint, biomechanical function of the ACL bundles and the femoral and tibial tunnel orientation as described in literature. Furthermore, the most commonly used measure-method to determine femoral and tibial insertion site will be described.

3.1 Anatomy of the knee

The knee is one of the largest and most complex joints in the body and consist of three bones; the patella, femur and tibia and a complex arrangement of ligaments, tendons and muscles, see Figure 5 and 6. Excluded from this study is the fibula since its location is lateral of the tibia (and therefore not part of the knee joint).

The surfaces of both femur and tibia are covered with a slippery substance of articular cartilage, numerous bursae, or fluid-filled sacsa and will together with the menisci form a nearly frictionless gliding of the knee joint. The meniscus, a horseshoe-shaped shock absorber on both the medial and lateral side, have a function in load transmission, shock absorption and gives a stabilisation effect between femur and tibia. Several muscles and ligaments control the motion and stabilization of the knee. The quadriceps and hamstring muscles are the main muscles for extending and flexing the knee joint. The quadriceps attaches to the patella and the patellar tendon connects this muscle to the front of the tibia.

The four most important ligaments in the knee are:

- The medial collateral ligament (MCL)
- The lateral collateral ligament (LCL)

and two ligaments in the center of the knee joint:

- The Anterior Cruciate Ligament (ACL), The function of the ACL will be described in detail in the next section.
- The Posterior Cruciate Ligament (PCL)

3.2 Biomechanical function

The ACL consist of two different bundles, an anteromedial (AM) and a posterolateral (PL) bundle. These bundles have been named on to the anatomic positions on the tibia plateau, see also Figure 5. The AMB lies more anterior and proximal in comparison with the PLB which lies relatively more posterior and inferior on the tibia plateau [2]. Due to the anatomical location of the insertion site of the AMB and PLB on the femoral condyle and tibia plateau, both bundles cross when the knee is flexed and become parallel when the knee is extended, see Figure 5.

Most of the present knowledge of the biomechanical function of the ACL and its bundles is derived from in vitro cadaveric studies [18, 37–39] and have different stabilization effect on the knee. The AMB

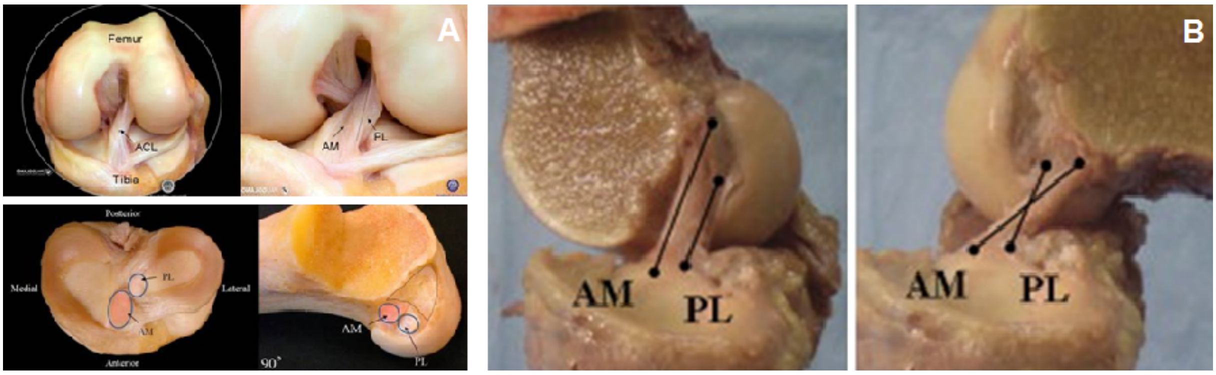


Figure 5: **Anatomy of the Anterior cruciate ligament**
A) Location of the anterior cruciate ligament in the intercondylar notch and insertion sites on the femur and tibia plateau[10 , 11].
B) The anterior cruciate ligament consisting of two different bundles, an anteromedial (AM) and a posterolateral (PL) bundle. Both bundles are parallel to each other, when knee is in extension, but when bending the knee both two bundles cross each other.

mainly functions to prevent anterior-posterior translation, whereas the PLB mainly functions to prevent rotational shearing forces.
Three characteristics of the ACL bundles; function of the bundle, in-situ force and length, are described below, subdivided into the AM and PL bundle.

- Function of the bundle
  Several studies investigated the role of both bundles during knee movements. Zantop et al [37] investigated the role of both bundles in Anterior Tibial Translation (ATT) by resection of these bundles. Resection of the AMB resulted in an increased ATT at 60 and 90 degrees of flexion, indicating its role in ATT at deep flexion. Resection of the PLB increased the ATT at minimal flexion up to 30 of flexion.

- In-situ force
  Gabriel et al. [40] and Sakane et al. [41], have investigated the in-situ forces of both bundles in response to 134 N at different knee flexions. The AMB had a lower in situ force than the PLB at full extension. In-situ forces of the AMB were significantly greater at 60 and 90 degrees of flexion in comparison with full extension and 15 degrees of flexion. The in-situ forces of the PLB performed in the opposite direction. A sharp bending angle of the graft at flexion angle with an elongated graft in combination with a large in-situ force will have to be avoided.

- Length
  The variation in length during flexion and extension are shown in Figure 6 for both individual bundles. The AMB and PLB are elongated in the fully extended position. The AMB is also elongated in more than 90 degrees of flexion, however is less elongated between 20 - 60 degrees of knee flexion. The PLB relaxes during knee flexion.

In conclusion, above described characteristics needs to be taken into account for setting up an algorithm to calculate optimal femur and tibia tunnels.

3.3 Femoral and Tibial Tunnel orientation

Tunnel orientation
Tibial and femoral tunnels can be classified as either anatomic or non-anatomic. Anatomic ACL reconstruction refers to the tunnels that were placed in the center of the native femoral and tibial insertion sites. Several studies report that anatomic tibial and femoral tunnel positions after ACL reconstructions is associated with better clinical outcome [42].
Besides correct positioning of the tunnel in the native insertion sites, orientation of the tunnel is of major interest. Correct orientation of both femoral and tibial tunnel is relevant in clinical outcome preventing

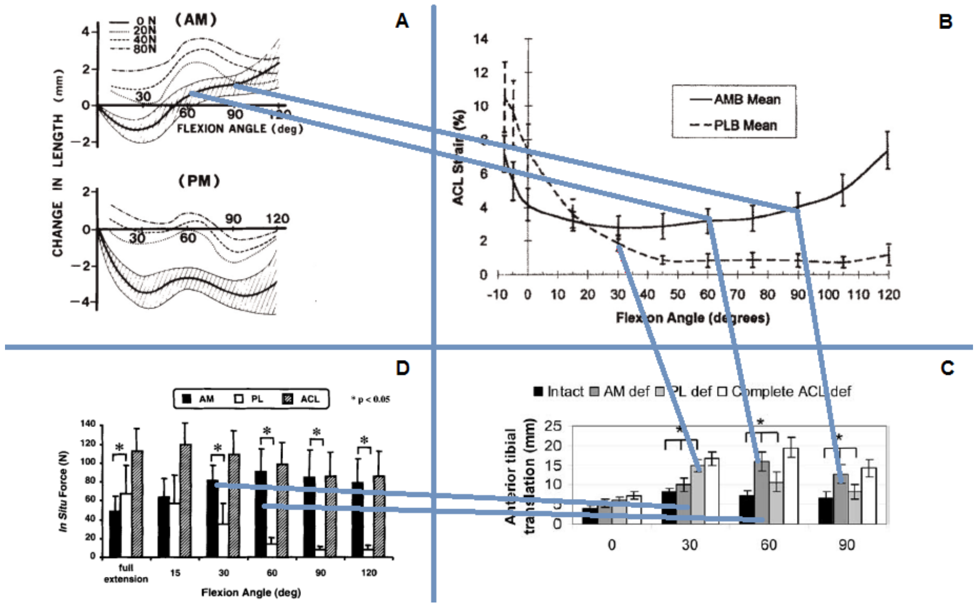


Figure 6: **Biomechanical function of the ACL.**
PL bundle: The strain in the PL bundle decrease in extension up to 30 degrees of flexion and then remains roughly constant (B). The in-situ force of the PLB at these flexion is slightly higher (D).
AM bundle: The AM bundle has a reasonably constant pattern, with slightly increase from 90 degrees of flexion (B). The in-situ force of the AMB at 60 up to 90 degrees of flexion is higher and also the length of the graft increase (A,D). The anterior tibial translation is increase at a deficient AMB between 60 degrees of flexion (C).

laxity of anterior tibial translation and rotational instability [43, 44]. Biomechanical studies has supported a more horizontal femoral tunnel position where controlled both anterior translation and internal tibial rotation. Risk of a horizontal orientation tunnel is too short tunnel length which result in reduced length of tendon graft and impair early graft incorporation [44, 45]. Anterior tibial translation can be well controlled with a vertical femoral tunnel orientation, but tunnel aperture become a more oval shaped and longer tunnel length [15, 43, 44, 46]. A longer distance between graft fixation points result in graft motion within the tunnel and may cause tunnel widening [47].
Bone tunnel widening after ACl reconstruction has been reported [20, 47, 48]. Tunnel widening usually occurs in the first 6 months after surgical operation and becomes stabilized within a year. Mechanical and biological multifactorial factors playing a role for the widening. The risk of the occurrence of this phenomenon is related with the type of graft. The use of hamstring grafts as opposed with patella tendon bone grafts has higher risk [20, 48]. The type of fixation and fixation device, the angle of the tunnels, motion of the graft within the tunnel, stress shielding of the graft and aggressive rehabilitation have all been associated to widening of the tunnel [20, 48]. Femoral and tibial tunnel widening are significant correlated with each other and tunnel widening are greater with more anterior, more proximal, and more vertical femoral tunnels. Tunnel widening might a problem by revision because a tunnel can overlap with the revision tunnels [15, 26, 31, 42, 47, 49–51].

Surgical techniques
Two commonly used techniques to drill both tunnels are the TransTibial (TT) and Medial Portal technique (MP). Positioning femoral tunnel with TT technique is restricted by the orientation of the tibial tunnel, because the femoral tunnel orientation cannot be made independently [52]. Independent drilling of the femoral tunnel is improved with the introduction of the MP technique [15]. A reported risk of the MP technique is short tunnel length as well as to avoid damage to the nerves and the cartilage [44, 45, 53]. Although there is an going discussion about the optimal technique [15, 51, 54].
Independent drilling of the femoral tunnel implies that the surgeon must drill the femoral tunnel under a certain flexion angle of the knee. A patient-specific guideline in which flexion angle the surgeon must drill the femoral tunnel is missing. Through this, Farrow et al. [16] and Basdekis et al. [31] evaluated femoral tunnel orientation with respect to knee flexion angle during single bundle reconstruction through

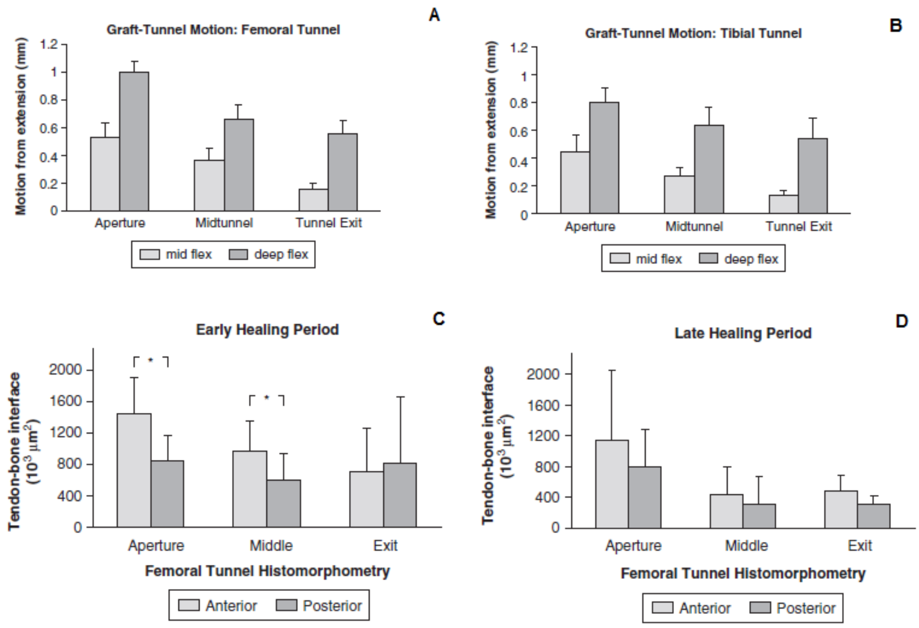


Figure 7: **Motion of a graft in femoral and tibial tunnel impair graft incorporation [55].** Graft-tunnel motion varies within a tunnel and is greater in the femoral tunnel than in the tibial tunnel (A, B). Tunnel motion is greater at tunnel apertures which influencing graft healing. More rapid tissue ingrowth at the tunnel exit is shown (A,C,D).

a MP technique. These authors recommended drilling at 110 degrees of knee flexion, because drilling at 90 degrees of knee flexion could result in too short tunnel length and posterior wall blow out [16]. At 130 degrees of knee flexion, the authors found the increased acuity of the femoral tunnel with respect to the lateral wall of the intercondylar notch. On the other hand, hyperflexion is not the optimal knee flexion angle when drilling the femoral tunnel through the AM portal [16, 50].
Studies above described a potential general guideline that applies to all patient. Because bone geometry differs for each individual, a patient-specific 3D tunnel orientation must be determined. Reported risk as described above, such as short tunnel length and posterior wall blow out, must be taken into account.

3.4 Healing of the graft

A successful clinical outcome of ACL reconstruction requires sufficient healing of the graft in both femur and tibia tunnels. Literature describes that graft-tunnel motion occurs within both femoral and tibial tunnel and may impair early graft incorporation. An aspect after ACL reconstruction is the optimal time to start mobilization, physical therapy, range of motion exercises and daily activities. A delay protect the healing graft from excessive loads and improve bone-graft incorporation. Improvement in graft healing to bone is critical to allow earlier rehabilitation and an earlier return to daily activities.
Rodeo et al. [55] has investigated the amount of graft-tunnel motion in femur and tibia between full extension and full flexion. Graft-tunnel motion was greater in the femoral tunnel than in the tibial tunnel and varies within the tunnel. Motion was greater at tunnel apertures in both tunnels, see Figure 7. The study concluded that the maximum graft tunnel motion in the femoral tunnel is 11.8% of the length of the femoral tunnel and 8.0% in the tibial tunnel.
As previously described, motion of the graft influences the healing of the graft into the bone. Although, the widths between graft and tunnel decreased over time throughout all 3 zones of the tunnel, with more rapid tissue ingrowth at the tunnel exit. The study concluded also, both in the early and late healing period, that the width between the tendon graft and bone tunnel was greater in the anterior half of the tendon-bone than in the posterior half at both the aperture and the tunnel exit. In addition, the anterior interface was significantly wider in the femoral tunnel aperture than in the tunnel exit, but there were no significant differences between the tunnel aperture and the tunnel exit in the posterior half of the

interface.
In conclusion, tunnel orientation influences graft incorporation. For example, a more vertical tunnel results in longer tunnel length which induces more graft motion and thus slow graft incorporation. With this fact must be taken into account in the development of a 3D computer model. Especially, at tunnel aperture and the anterior side of the femoral tunnel.

3.5 Method for localization insertion area of both femur condyle and tibia plateau

Several techniques are available to define native insertion area for both femur condyle and tibia plateau. The ruler is easily and commonly used in arthoscopic surgery, see Figure 8A. In addition, Bernard et al. [32] has described a radiographic quadrant method to define the insertion area postoperatively or intraoperatively.
Several studies have also investigated fluoroscopic and computer-assisted surgery, however no unequivocal conclusion that this increase in either accuracy or precision of the tunnel placement. The time-consuming problems, exposure to radiation and cost related to the use of these devices are still major obstacles to their widespread use in clinical practice [25, 29].

Preoperative methods of determining femur and tibia insertion site, which can be used in the 3D model is described below.

Method for localization femoral insertion area
Bernard et al. [32] has described a radiographic quadrant method to determine the correct femoral tunnel position on the femur condyle. The method is based on a correlation of the lateral condyle shape on the sagittal plane and the original femoral insertion place. The visualization orientation must into a true lateral position so that both condyles were superimposed. The anatomical insertion areas of both bundles can be determine with a coordinate system based of anatomic landmarks of the subchondral bone. This method is independent of knee form and size and the film-focus distance of the imaging technique.

Parameters are defined as follow, see also Figure 8B:

- Distance **t**, measured along the Blumensaat's line and was limited by the intersections between this line and the ventral and dorsal borders of the femoral condyle.
- Distance **h**, this defines the height of the intercondylar space. This is the distance between the Blumensaat's line and the border parallel of this line.
- Distance **a**, this defines the distance between the center of the of the ACL insertion and the Blumensaats line along **h**. These value is describes as a percentage of height **h**.
- Distance **b**, this defines the distance between the center of the of the ACL insertion and the border most posterior of the condyle. These value is describes as a percentage of length **t**.

A comment is that several studies reported different percentage values of distance **a** and **b**, see Table 1.

Table 1: Overview of different positions of the femoral insertion area reported in the literature, based on the quadrant method of Bernard et al.

Study	Length t (mm)			Length h (mm)		
Colombet et al. [30]	26.4	32.3	29.35	25.3	47.6	36.45
Zantop et al. [33]	18.5	29.3	23.9	22.3	53.6	37.95
Tsukada et al. [34]	25.9	34.8	30.35	17.8	42.1	29.95
Yamamoto et al. [35]	25	29	27	16	42	29
Bernard et al. [32]			24.8			34.35
Mean value of the studies (mm)			27.08			32.37

Method for localization tibial insertion area
The central points of the tibial footprints of each bundle were measured from the anterior edge and the

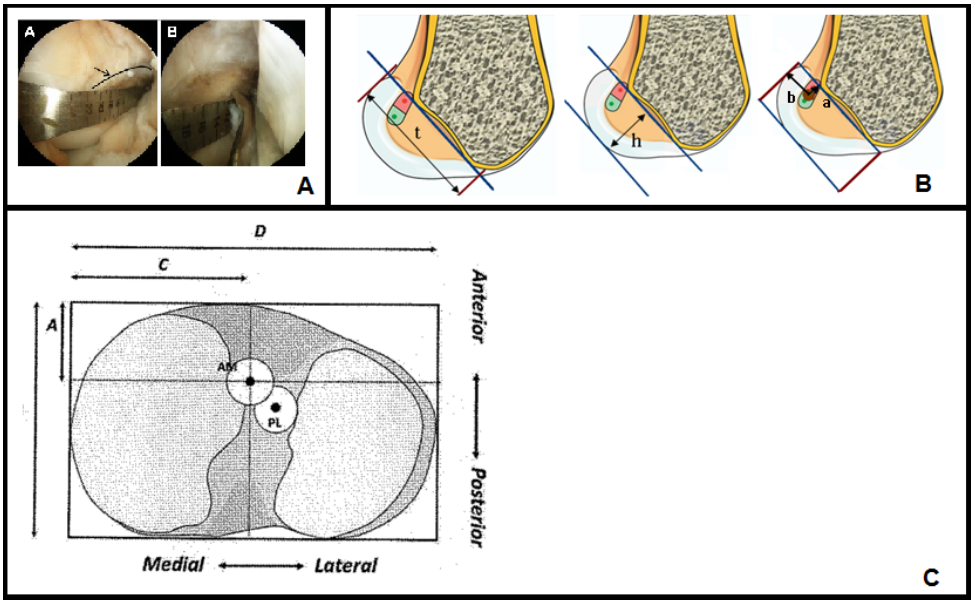


Figure 8: **Method for localization insertion area**
A) Arthoscopic measurement [56].
1. The length of the intercondylar notch measured along the Blumensaat's line.
2. The bottom of the ruler aligns with the border of the condyle. The height is corresponding with the width of the ruler.
B) Bernard et al. [32] method to determine the anatomical position of the ACL. The method is based on a correlation of the lateral condyle shape on the sagittal plane and the original femoral insertion place.
C) Tibial measurement of central point of the AM and the PL bundle. The horizontal line (c) was defined by the line between the most posterior margin of the lateral tibial plateau and the medial tibial plateau. The vertical line was defined by the line perpendicular to the horizontal line (a).

medial edge of the tibial plateau and expressed as a percentage of the horizontal and vertical distance on the tibial plateau, see Figure 8C. The horizontal line was defined by the line between the most posterior margin of the lateral tibial plateau and the medial tibial plateau. The central point of the AM bundle was located at 37.6% and the PL bundle at 50%. The vertical line was defined by the line perpendicular to the horizontal line and the AM bundle was located at 46.5% and 51.2% of the PL bundle.

4 Computer models in literature

This chapter describes a literature search to other studies which developed a computational model of the human knee joint to study the ACL reconstruction.

The first publications about Computer-Assisted surgery started during the 1990s. For instance several studies had investigated fluoroscopic-based navigation systems and computer-assisted surgery in ACL reconstruction [29]. In some reports is assumed that computer assistance would permit more accurate and more precise femoral and tibial tunnel placement compared with conventional ACL reconstructions. Others showed no significant improvement in clinical outcome with the use of computer-assisted surgery in ACL reconstruction [29]. In conclusion, most of studies on tunnel placement suggest that, for tibial placement, an experienced surgeon can achieve comparable results with or without navigation, whereas only a few authors have shown an improvement in tibial drill hole placement with navigation [29]. In contrast, for femoral placement, most of articles show improved positioning in the navigated ACL compared with the conventional technique.

4.1 Ligaments

Various computational models of the ACL have been developed to evaluate ligament behavior in health, disease and injury knees. In particular, the finite Element(FE) method has been shown to be an effective approach characterizing stress distributions in the ACL in response to loading and tibio femoral movements.
Direct measurement of muscle, ligament, and articular-contact forces at the knee in vivo is impracticable. Many studies have calculate ligament strains [1,2], ligament forces [3], contact pressures [4,5] and contact forces [6]. However, the loading applied to the cadavers in these experiments is different from the forces applied by the muscles during activity. Mathematical models have been developed to calculate the forces in the soft tissues in and around the knee, but these models do not account fully for the interactions between the muscles, ligaments, and bones.
Direct measurement of the stress or strain distribution within the ACL is difficult and various techniques have been used. There are a number of limitations with the existing experimental methods. Experimental sensors tend to alter the natural geometry of the ligament and thus cause errors in the measurements. Also, in the majority of cases, the measurements are taken at discrete locations, rather than continuously over the entire surface of the ligament.
AN alternative to experimental studies is the use of computer methods. In particular, the use of finite element analysis is popular as it allows the threedimensional structure to be analysed and the stress or strain distribution to be visualised anywhere within the model. Comparatively few full three-dimensional (3D) finite element (FE) models of the ACL have been developed. The major limitation of finite element-based studies is the lack of adequate validation. In the majority of finite element studies of the ACL to date only the stress distribution patterns have been reported for simulated passive knee flexion or drawer tests. The predicted stress and strain distributions are difficult to validate, because of the problems in experimentally measuring these parameters, as discussed earlier.
The ACL is a complex structure, therefore most studies have modulate the ACL as a single tensile element or multiple bundles [39, 57]. Most studies, investigate the length and/or force changes during knee range of motion, orientation and inclination angle. Various methods have been used to represent ligaments in computational models including finite element techniques and elastic springs. Modeling the ligaments as elastic springs is the most computationally efficient method and several studies have looked at how these elastic springs should be defined. Some have defined the springs as completely linear while others used non-linear springs to represent ligaments. Studies have shown that ligaments have a non-linear toe region which occurs because of the initial crimping of the ligament fibers and this toe region ends when all of the fibers have become taut. At that point the ligaments behave as a linear spring with a specific stiffness parameter [58]
Hensler et all. [59] has investigated the effects of the drill-bit diameter, the transverse drill angle and the knee flexion angle on the aperture size and orientation of tunnels in medial portal drilling approach. Young Kim et all. [11] and Zaffagnini **??** has both investigated the change in length of ACL at different knee flexion angle on healthy persons. Virtual measurements were done using a 3D finite element method for double bundle reconstruction.

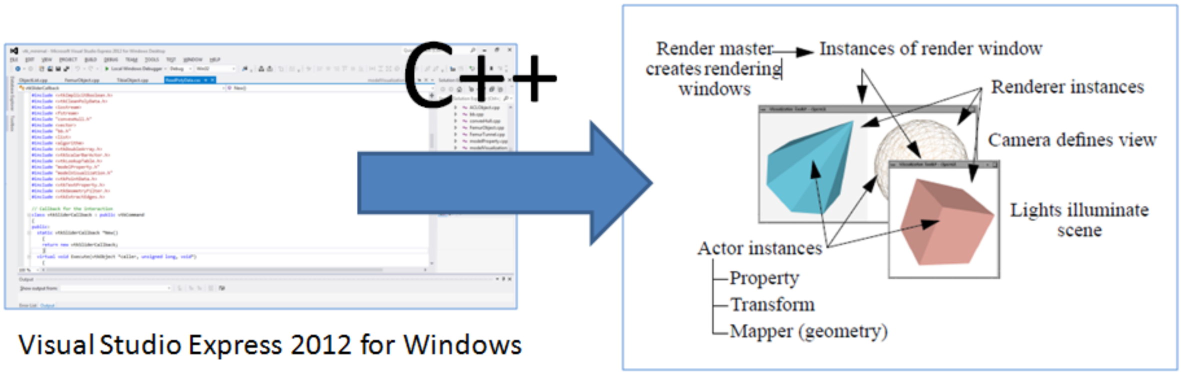


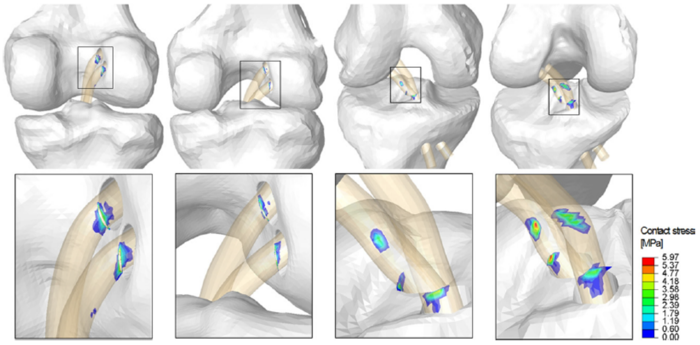
Figure 9: **A schematic overview of the graphics model of VTK [60]**
A variety of open-source software-modules / Libraries were used for this purpose, such as ITK-SNAP and The Visualization Toolkit which include image processing tools and supports a wide variety of visualization algorithms. The C++ programming language in combination with the free Visual Studio Express 2012 for Windows Desktop environment was used to linked each library.

4.2 Computer platform

Patient-specific modeling (PSM) has not yet become a standard of care in clinical practice, the costs and the time-consuming problems, because many manual and high technical steps in a workflow from data acquisition to a model, are still the major obstacles.
The use of MATLAB is an example of this, the environment is usefull for analyze data and develop algorithms, but not user-friendly environment to use in clinical care. Moreover, some large algorithm in MATLAB taking hours and days to run. One of the underlying ideas is that the work in these thesis can be used in a clinical setting. Ideally, all essential steps must be implemented in an all-in-one system. This is a motivation to search to a more efficient language such as C++. The disadvantage is some experiences in specific computer language is required.
The C++ programming language in combination with the free Visual Studio Express 2012 for Windows Desktop environment to build and developed algorithms is used in this study. In addition, The Visualization Toolkit (VTK) an open-source, freely available software toolkit for interactive, image processing and supports a wide variety of visualization algorithms is implemented. In Figure 9 is shown the essentials of this 3D graphics toolkit to apply algorithms, define the camera view position and amount of light, the rendering windows and objects, such as lights, cameras and actors. A major concern when developing a 3D graphics system is the amount of memory consumed by creating objects and compute time to calculate algorithms and visualize 3D objects.
ITK-SNAP (Library of Medicine - Insight Segmentation and Registration Toolkit) is an open source software application used to segment structures in 3D medical images with tools for fast post-processing of segmentation results. ITK-SNAP provides a number of supporting file formats to import MRI/CT scans, such as DICOM-files. This a stand-alone application currently, however the ITK segmentation C++ library's may be optionally integrated at a later stage in the developed software application.

4.3 Contact pressure

Kim et al. [11] investigate the contact stress of the AM and PL bundle after ACL double bundle reconstruction at different knee flexion angles by use of a 3-dimensional finite element model. see Figure 10.
The contact stresses between the AM and PL bundle and femur condyle and tibia were investigated. The maximum contact pressure of the graft in the normal and anterior femoral tunnels was observed at the anterior portion when the knee was in full extension and in the posterior portion with deep knee flexion. The knee flexion angle at which the contact pressure shifted from the anterior to the posterior portion of the femoral tunnel was highly correlated with the knee flexion angle at which alignment of the femoral and tibial tunnels occurred in each instance.



Stress distribution within AM and PL bundles at different knee angles in a representative case. The contour shows the level of stress on the grafts. At 0° and 45° of flexion (left 2 figures, posterior view), most of the contact force was generated at the sharp edge of the anterior margin of the femoral tunnel. At 90° and 135° of flexion (right 2 figures, anterior view), contact force caused by both grafts being deformed and rerouted by the lateral intercondylar tubercle was noted, which may be an important factor in maintaining tension within both grafts despite a decrease in direct length at deep flexion.

Figure 10: **A 3-dimensional finite element model of contact stress described in the literature [11].**

The authors suggest that knowledge of tension changes and contact stress is an useful basis for an improved anatomic ACL reconstruction. Contact forces is of major interest by determination of optimal tunnel orientation and will need to be taken into account in the development of 3D model.

4.4 Conclusion

Such as described in this chapter, other research has investigated biomechanics in normal ACL or tunnel orientation postoperative or in healthy persons. To my knowledge, no other study has determine patient-specific femoral and tibial tunnel orientation preoperatively.

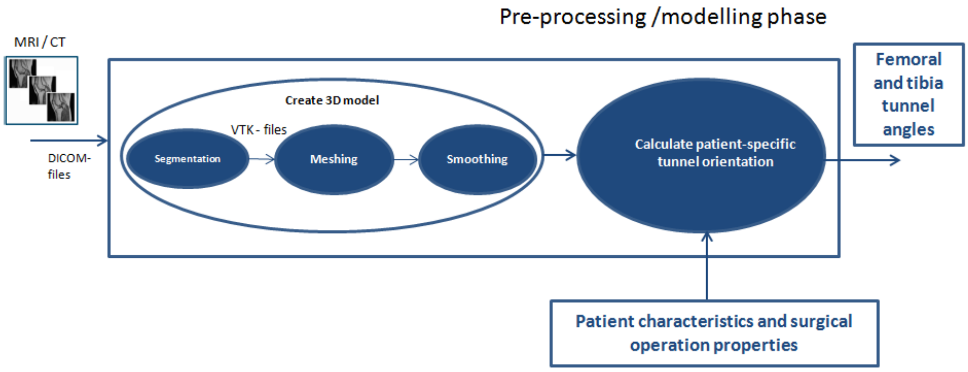


Figure 11: **A schematic overview of sub-processes in the pre-processing / modelling phase**
The pre-processing / modelling phase can be divided into two sub-processes; *Creating a surface model* and *Calculating a patient-specific tunnel orientation*. The femur and tibia are (semi)-manually segmented with ITK-SNAP software from MRI/CT scans stored in DICOM files. ITK-SNAP software export the segmented femur and tibia separately in different VTK-files. These files will be imported in the software application and meshed and smooth before patient-specific tunnel orientation can be calculated.
Patient characteristics and surgical operation properties have to be included to calculate individualized tunnel orientation.

5 Computer Modelling

This chapter describes the patient-specific characteristics and boundary conditions which must be taken into account to set up the workflow of calculating patient-specific tunnel orientation. In addition, the workflow and corresponding subprocesses will be described in more detail.

5.1 Workflow to define tunnel orientation

A number of essential steps are needed to calculate patient-specific tunnel orientation from MRI/CT scans. In this thesis we focus on the process "Pre-processing /modelling phase" which is described in earlier chapters and shown in Figure 3. This process can be divided into two sub-processes which is shown in Figure 11:

1. Creating a surface model
2. Calculating a patient-specific tunnel orientation.

Both of these processes will be explained in more detail in the sections below.

1) Creating a surface mesh
In Figure 11, the first important steps to create the surface model are illustrated. Both magnetic resonance imaging (MRI) and computed tomography (CT) can be used to acquire bone geometry. MRI can provide more detailed images of soft tissue structure and CT provides accurate boundaries of bone. The ability to calculate tunnel orientation through the 3D model precisely, will mainly depend on the quality of the scan methods and accuracy of the segmentation method.
The first step is to segment the femur and tibia from MRI/CT scans, which are stored in DICOM (Digital Imaging and Communications in Medicine) files. As this study is particularly focused on the Pre-processing /modelling phase, we have chosen to segment the scans (semi)-manually with ITK-SNAP software. By using ITK-SNAP software, data can be easily formatted and write to commonly use data file formats. However, in order to transfer data between VTK software, it is easier to use their own file format. This file format(.vtk) is a consistent data representation scheme for a variety of dataset types.
Extracting surface mesh models from MRI scans is a common and important step for visualization and intervention planning effects. Manual or automatic segmented data are often contaminated with noise, which can result in reconstructed surfaces with low geometric quality. Surface mesh smoothing is required to remove this noise. One of the most common methods of smoothing a surface is Laplacian smoothing

in which the position of a point is modified, depending on the local connectivity and position of the neighboring points in triangular meshes. In addition, the Delaunay triangulation is one of the most popular and most used methods related to the generation of meshes. This method is based on the fact that angles in triangles should not be too small or too large, and the triangles should not be much smaller than necessary, nor larger than desired.
Several parameters have an influence on how to obtain a qualitatively correct mesh. After meshing and smoothing the data from MRI/CT, the geometry can be used to calculate tunnel orientation.

2) Calculating a patient-specific tunnel orientation
The next process can be subdivided into minor sub-processes, what is outlined in Figure 12. These processes are designed on the basis of the overall steps during a surgical ACL procedure, see Figure 12 Arthroscopic procedure.

The first step of surgery is to arthroscopically inspect damage of different tissue in the knee, such as the meniscus, cartilage and ligaments. Treatment of these injuries and notchplasty can be performed at the same procedure. Regardless of which measurement method is used by the surgeon (see Chapter 1), the native insertion sites of both femur condyle and tibia plateau can be determined after preparation. These insertion sites should also be determined in our 3D model using the Bernard el al. [32] method as described in Chapter 1.
The second step is to determine the length and thickness of the tunnel. Our 3D model in this thesis is based on the use of autograft hamstring tendons replacement of the graft for reconstruction. The semitendinosus and gracilis tendons were harvested from the knee. Depending on thickness of the material, a few parallel strands were constructed to increase the stiffness and strength of the graft. Tunnel length and the drill bit size is dependent on length and thickness of the graft. Tunnel length is of clinical relevance because minimal amount of graft is required for successful incorporation and prevention movement within the tunnel, see chapter 1.
Tunnel orientation not only influences the length of the tunnel, but also the area and shape of the intra-articular aperture. The next step is to calculate aperture shape and area on both femur condyle and tibia plateau to reproduce the native characteristics as closely as possible. Prevention of bone avulsion and wall blow out must be taken into account.
In arthroscopic surgery, tunnel orientation depends mainly on which knee flexion angle the tunnel will be drilled. In the model, tunnel orientation will pre-operatively be calculated with minimal intra-articular graft bending angles over the entire range of motion which take into account patient-specific characteristics and boundary conditions.
The entire workflow will be described in more detail in the next sections. However, the most important (surgical) conditions will now be discussed:

- The 3D computer model will be based on a single bundle technique with the use of hamstring tendons and an EndoButton fixation.
- The femoral tunnel has a minimal length of 30mm (with Endobutton fixation), because the shortest loop length of a EndoButton is 15mm. [61]
- The ACL must have a minimal length of 15 between 20mm in femoral tunnel, a tunnel less result in imperfect graft healing within the tunnel [31, 45].
- The ACL graft is tensioning at 30 degrees of flexion of the knee.
- Distance from wall and articular surface to prevent wall blow out must be minimal 1.5mm
- The drill bit diameter varies [43].
- The area of the aperture at the medial wall and aperture of the tibia the diameter must be approximately equal the native insertions size described in the literature [62].

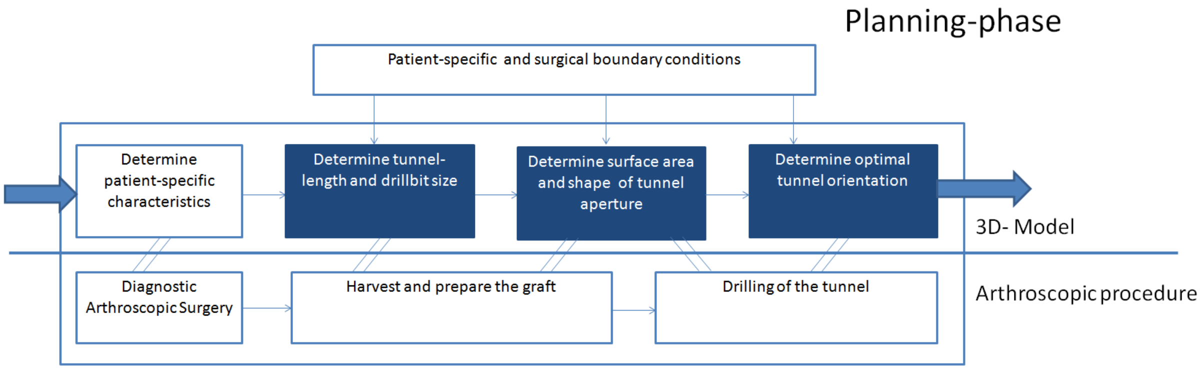


Figure 12: **Steps to calculate a patient-specific tunnel orientation are designed on the basis of the overall steps during a surgical ACL procedure**
At the upper side is shown several steps in our 3D model, in accordance with the steps during a standard orthopedic ACL procedure shown at the bottom.
Native insertion sites of both femur condyle and tibia plateau must be determined in both arthoscopic surgery and in the 3D model. Tunnel diameter and aperture shape is depending on thickness and length of the harvested semitendinosus and gracilis tendons. Currently in arthoscopic surgery is tunnel orientation dependent of flexion angle of knee. In the 3D model, the tunnel orientation will be calculated preoperatively.

5.2 Patient-specific properties applied to an 3D model

An important aspect of the study is individualization of both femur and tibia tunnel orientation since the geometric differences between individuals. Morphological characteristics between individuals were investigated by several studies [30, 32–35, 63]. Terzidis et al [63] have investigated differences in size of the femur and tibia in cadaver knees, containing five anatomical measured characteristics: the femur bicondylar width, intercondylar width, the femur intercondylar depth, medial and the lateral condylar depth. Differences between individuals were found and most characteristics were significantly greater in men than in women. There was no significant differences between the two sides of the body. These anatomical measured characteristics are of clinical relevance, because for example smaller intercondylar notches have been associated with smaller ACL width and more frequent ACL ruptures. As femur and tibia geometric do vary between individuals, patient-specific tunnel placement are of interest,

Determination of patient-specific anatomical landmarks in our 3D model
Two important aspects are of interest to determine the patient-specific anatomical landmarks in our 3D model: determining of the insertion site at the medial wall and determining the tunnel-exit on the lateral wall. Both will be explained in more detail below.

- **Determining of the insertion site at the medial wall**
  In the operation theatre most used arthoscopic measurement method to define the anatomic insertion site is measurement with a arthoscopic ruler. However, the Bernard et al. [32] method will be used to define these locations in the 3D model, such as described in Chapter 1. Three anatomical landmarks must have to be determined by clicking these points in the software application: the most anterior and the most posterior point at the border of the femoral condyle, which define the virtual Blumensaat's line. Third point is perpendicular to the Blumensaat's and the femur condyle border. The selection of these anatomical points in the 3D model is illustrated in Figure 13A.

- **Determining the tunnel-exit on the lateral wall**
  In computer graphics are objects represented as triangulated polyhedra. A corner of the triangle is named as a vertex. Every triangle has three vertices and is not only associated with 3D positions but also with other graphical properties to render the object correctly, such as color of the vertex and available light projection on the object.
  The anatomic landmarks on the medial wall on the femur condyle are now well defined. However, there is still a substantial freedom in tunnel-exit position on the lateral wall of lateral femur condyle.

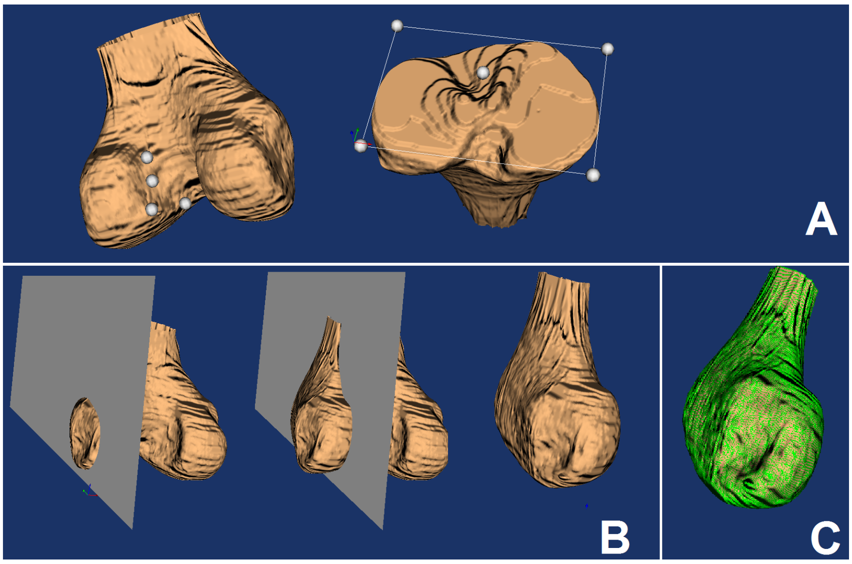


Figure 13: **Patient-specific properties applied to 3D model.**
a) Femoral and tibial insertion area has to be defined by the methods described in section 3.3. Three anatomical markers must be defined by clicking markers on both the femur condyle and tibia plateau.
b) The surface area of the lateral wall has to be selected and clipped by a be plane in the software application.
c) The selected surface area will be used to determine the optimum position in a later phase. Each vertices, illustrated as green dots, represented a potential location for optimal tunnel placement.

Initially, each position on the surface lateral wall corresponds with a potential point for optimal patient-specific tunnel orientation. These selected surface area consist thus of a large number of vertices, and each vertex point representing a potential location for optimal tunnel placement, see Figure 13C.

5.3 Mathematical description of a computer graphic objects

As described in the section above, computer graphic objects, such as a femur or tibia, are constructed from a large number of vertex points.
The femur object can therefore be described as follow, assuming X is a dataset of n vertex points:

$$X = \{\mathbf{x}_1, \mathbf{x}_2, ..., \mathbf{x}_n\} \tag{1}$$

Every vertex point $\mathbf{x}_i$, (wherein $i = 1, .., n$), includes a 3D coordinate. The lateral wall of the lateral femur condyle is of major interest in determination of the tunnel orientation. In the software application the lateral wall of the lateral femur condyle is manually selectable by moving a plane across the x-axis, see Figure 13B. Let P_{plane} be a point on the x-axis which the position of the plane describes. The lateral wall of the femur can now be described as a subset of all vertex points of X, defined as $X_{lateral} \subseteq X$ for which all $\mathbf{x}_i$ the x-position $\leq P_{plane}$.

5.4 Process of determining the most optimal tunnel orientation

In Figure 14 the process of determining the most optimal tunnel orientation is shown. After segmentation, meshing and smoothing a suitable 3D model can be used to find the the most optimal tunnel orientation. In the beginning, a subset $\mathbf{X}_{lateral}$ is selected with n potential vertex points. Then, the length of the tunnel by all potential vertex points in $\mathbf{X}_{lateral}$ will be verified by predefined conditions. Some vertex points are excluded, so that a smaller subset of $\mathbf{X}_{lateral}$ remains. The process is repeated again and again until all conditions are verified and there is a consensus of tunnel orientation found.
In the following chapters, all conditions which are going to be verified will be discussed in detail and these are:

- The effect of femoral tunnel length

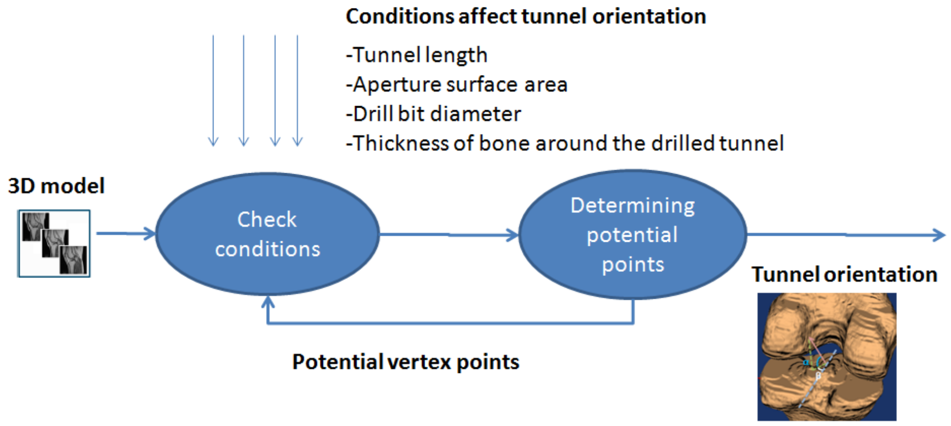


Figure 14: **Process of determining the most optimal tunnel orientation**
After segmentation, meshing and smoothing a suitable 3D model can be used to find the the most optimal tunnel orientation. Several predefined conditions must be verified. This process of verifying is repeated again and again until all conditions are verified and there is and a consensus of tunnel orientation found.

- The effect of femoral tunnel position on surface area and shape of the aperture
- The effect of varying drill bit diameters
- Amount of surrounded bone over the entire length of the tunnel to prevent wall blow out and avulsion of bone.
- Optimal tunnel orientation which controls both anterior translation and internal tibial rotation stability

5.5 Drilling of the femoral tunnel

Determination of the native insertions site at both femur and tibia were described in above sections. The following step is to define the tunnel in the 3D model. The tunnel must be orientated in the way that he is identical to the defined anatomic positions. In addition, the diameter of the tunnel depends of the harvested thickness of the semitendinosus and gracilis from the knee. In Figure 15B,C are shown a cylinder (similar to a tunnel) at the well defined position on the medial wall and an arbitrary point on the lateral wall.

Tunnel orientation in the literature
Due to the drilling of the virtual tunnel, The optimal tunnel position will be adjusted accordingly.
In the literature is femoral tunnel orientation described on postoperative radiographs or MRI/CT. Disadvantages are the lack on an unambiguous measurement protocol and measurement of tunnel angles in a 2D plane. Therefor, it is difficult to compare the findings of 3D tunnel orientation in our 3D model with values known in the literature. Results of femoral tunnel angles (jaw) in the literature, in a 2D plane orientation, were found in the average range of 33.9 and 46.1 degrees and tunnel inclination angles (pitch) in the average range of 46.4 and 54.5 degrees[42, 61].

Mathematical description of the virtual 3D tunnel orientation
The tunnel orientation will be described using Euler angles. Since a tunnel is rotational symmetric, the roll of this object is not relevant, and two angles, the pitch (α) and the jaw (β), suffice to define the orientation. The reference coordinate system is defined with respect to the femur. The femur is located at the origin of the coordinate system. The z-axis is aligned with the longitudinal direction of the femur. The x-axis is the latitudinal direction of the femur. The tunnel has an initial reference orientation, $\alpha = 0$ and $\beta = 0$, aligned with the y-axis of the reference coordinate system. Thus, an arbitrary point, $\mathbf{p}_{init}$, of the axis of the tunnel in its reference orientation is given by

$$^{ref}\mathbf{p}_{init} = \lambda \mathbf{e}_y \tag{2}$$

Figure 15: **Virtual drilling of the femoral tunnel**
a) The anatomical markers are defined in the 3D model and an arbitrary point on the lateral wall is illustrated. The cylinder must be orientated so that the virtual tunnel is identical to the the certain defined anatomic positions. A virtual tunnel will be drilled through these points.
b) Tunnel orientation in terms of vertical α and horizontal tunnel β angles which can be used by drilling the tunnels intraoperatively.

where $\mathbf{e}_y = [0\ 1\ 0]^T$ is the unit vector in the y-direction and $\lambda \in \mathbb{R}$ is any real number. To define the tunnel direction, we use:

$$\begin{aligned} \mathbf{p} &= \mathbf{R}_z(\beta)\mathbf{R}_x(\alpha)^{ref}\mathbf{p}_{init} + \mathbf{b}_{begin} \\ &= \lambda\mathbf{R}_z(\beta)\mathbf{R}_x(\alpha)\mathbf{e}_y + \mathbf{b}_{begin} \end{aligned} \tag{3}$$

$\mathbf{R}_x(\alpha)$ and $\mathbf{R}_z(\beta)$ are rotation matrices that rotate around the x-axis and z-axis, respectively with the pitch α and jaw β. The vector $\mathbf{p}_{begin}$ is a support which makes sure thats the tunnel will enter the femur at the correct anatomic location. It is understood, that if $\lambda = 0$, the $\mathbf{p} = \mathbf{p}_{begin}$ is the start point of the hole defined by the standard protocol described in an earlier chapter. The direction of the tunnel is defined between $\mathbf{p}_{begin}$ and $\mathbf{p}$.

If we define l as the tunnel length, than the tunnel exit point($\mathbf{p}_{exit}$) on the lateral wall of the femur is:

$$\mathbf{p}_{exit} = l\mathbf{R}_z(\beta)\mathbf{R}_x(\alpha)\mathbf{e}_y + \mathbf{p}_{begin} \tag{4}$$

5.6 Determination of femoral tunnel length

Femoral tunnel orientation influences the length of the femoral tunnel. Tunnel length is of clinical relevance, because minimal amount of graft is required for successful incorporation and healing of the graft. Longer length between graft fixation points resulted in significantly greater tunnel motion which induces tunnel widening [43, 44, 55, 64]. Graft tunnel motion was greater in the femoral tunnel than in the tibial [59]. In addition, Scheffler et al. [2] and Ishibashi et al. [65] reported lower stiffness and a decreased stability by longer lengths between fixation points. The minimum length of graft within the tunnel has not been reported in humans, but an animal study showed that relatively short graft-tunnel lengths of up to $15mm$ did not adversely affect graft healing. Ilahi et al. [45] reported no differences in biomechanical strength of tunnels with 15 mm in length in a goat model. Evidence is lacking as to whether a graft-tunnel length of less than $15mm$ in the femoral tunnel can be safely used for ACL reconstruction in humans. Boundary conditions in our 3D model anchored that femoral tunnel length must minimal $30mm$ with EndoButton fixation.

Tunnel length in the literature

Several cadaveric studies reported variable mean lengths from 21.3 to 43.3 mm [45, 50, 52, 53]. Several studies compared tunnel length between TransTibial and Anteromedial Portal Techniques. Shorter femoral tunnel lengths were observed when drilling through a medial arthroscopic portal approach. Short tunnel length is associated with horizontal placed femoral tunnels and as early described, has this influence on knee stability.

Mathematical description of tunnel length

Tunnel length (l) can be written as:

$$l = \mid \mathbf{p}_{exit} - \mathbf{p}_{begin} \mid \tag{5}$$

As earlier described, the lateral wall is defined as $X_{lateral}$ with all potential tunnel-exit points included. The length of the femoral tunnel of all vertex points in dataset $x_{lateral}$ can be calculated using Formula 5. Results of calculating these tunnel length can be described in L. Assuming L is a dataset of k tunnel lengths.

$$L = \{\mathbf{l}_1, \mathbf{l}_2, ..., \mathbf{l}_k\} \tag{6}$$

As described above, short and long tunnel lengths have clinical impact. Therefore, every tunnel length in $L_i < 30mm$ (wherein $i = 1, ..., k$) will be excluded. In Figure 16, tunnel length of all potential points in $X_{lateral}$ and a subset of tunnel length in L is shown in different colors.

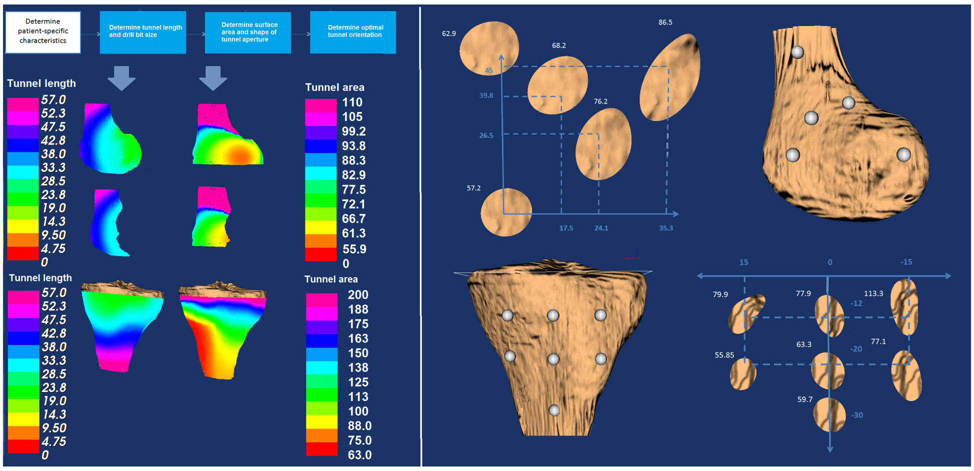


Figure 16: **Effect of femoral tunnel length and surface area and shape at different lateral wall positions.**
A) At all potential points at the lateral wall (wall blow out is not taken into account) the length of the femoral tunnel and area of tunnel aperture at the medial wall of the lateral femur condyle are calculated. Tunnel length must be minimal 30mm with EndoButton fixation, as can be seen in the figure at the bottom. Only the potential positions were showed in which tunnel length is longer than 30mm.
B) Four different positions on the lateral wall were simulated varying in horizontal and vertical tunnel orientation. Horizontal tunnel produces more circle-shaped aperture and more vertical tunnel orientation produces more oval-shaped tunnel orientations. Five different positions on the tibia wall were simulated varying in different tunnel orientation.

5.7 The Effect of Femoral Tunnel Position on surface area and shape of the aperture

Changing tunnel orientation on the lateral femur wall and/or drill bit size of the tunnel will lead to alteration of surface area and shape of the aperture of both femur and tibia drill holes [59]. To calculate the most optimal tunnel orientation, tunnel aperture dimensions that resembles the native attachment site in terms of size and orientation need to be taken into account. In addition, destructing of important anatomic structures, such as the meniscal roots, the tibial tubercles and the plateau cartilage must be prevented. The lateral and medial roots of the meniscus are intimately associated with the posterior and lateral border of the ACL, thus increasing the size of the bone tunnel aperture increases the risk of damaging these surrounding structures.

Surface area of the aperture
In Figure 16 is described in different colors the surface area of the aperture at the medial wall. A perfect circle (a smooth surface cannot be expected using patient-specific data) would give an surface area of 50.3mm^2 using a drill bit diameter of 8mm^2. Illustrated in this Figure, a horizontal tunnel orientation produces an area between 60mm^2 and 80mm^2. A more vertical tunnel orientation will produce greater areas, between 80mm^2 and 120mm^2 or higher. In comparison with the attachment site areas of the AM and PL bundles on the medial wall of the lateral femoral condyle described in the literature, wherein a range from 65mm^2 to 90mm^2 is found. [2, 59].

Shape of the aperture
The insertion sites of both bundles on the femoral condyle are located on the medial wall of the lateral femoral condyle in an oval with an area ranging from 65mm^2 to 150mm^2 and an average of 136.0mm^2. The attachment area of the AMB is larger than the PLB at a ratio of 3:2, see Figure 5 [2, 30]. The average femoral insertion site is 8.9mm wide and 16.3mm long [59].
The tibial insertion sides are anatomically located on the anterior aspect of the tibial plateau in the Area Intercondylaris Anterior (AIA). Insertions are in an oval with an average area of 144mm^2 (range between 67mm^2 and 259mm^2). The average width of the ACL is 10mm and the average length is 14mm. The AMB has a larger area (67mm^2) than the PLB (52mm^2) [66].
In Figure 16, the shape of the aperture at the medial wall of the lateral condyle is showed by different

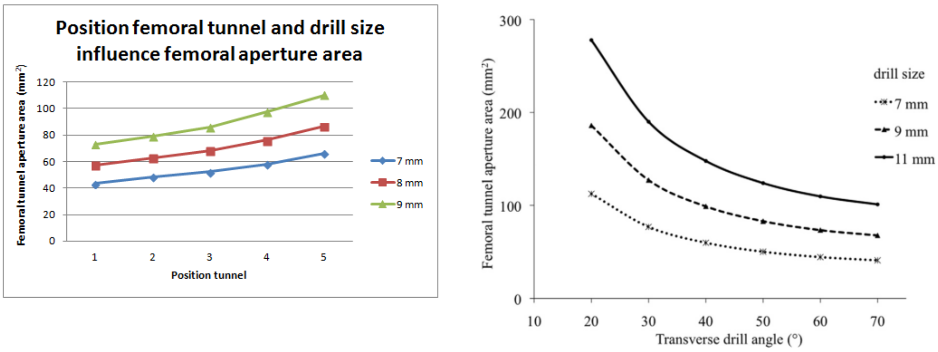


Figure 17: **Effect of drill bit size on femoral aperture area.**
Left: The effects of varying drill bit diameters (7,8,9mm) on aperture area at several positions on lateral wall.
Right: The same effect by varying the drilling flexion angle is described in the literature [59]. In both is illustrated, that a more vertical orientation (lower drill flexion angle, but higher tunnel orientation angle) and drill bit size is related with a increase in femoral aperture surface area.

positions at the lateral wall using a simulated drill bit diameter of $8mm^2$. As shown, a more vertical tunnel orientation gives a more oval-shape (with a greater surface area). On the other hand a more horizontal orientation gives a more circle-shape, corresponding to the shape of the drill bit.

Effect of varying drill bit diameters
In Figure 17A the effect of varying drill bit diameters (7,8,9mm) on aperture area at several positions on lateral wall in the 3D model is shown. A decrease of 29.2% in tunnel aperture area while change the tunnel orientation from vertical to horizontal orientation with same drill bit size is illustrated. In addition, a increase of drill bit size results in a increased tunnel aperture area. The same effect is described in the literature [59], see Figure 17B.

Mathematical description of tunnel area
Computer graphics objects (such as femur and tibia) are represented as triangulated polyhedra. Tunnel aperture area on the medial wall of the lateral condyle can be derived from the formula of Heron:

$$O = \sqrt{2(a^2b^2 + b^2c^2 + a^2c^2) - (a^4 + b^4 + c^4)} \tag{7}$$

with a b, c corresponding to the side of a triangle.
As earlier described, the lateral wall is defined as $X_{lateral}$ with all potential tunnel-exit points included. Different tunnel-exit points in dataset $X_{lateral}$ will lead to alteration of surface area and shape of the aperture, accomplished by changed tunnel orientation. Assuming A is a dataset of k vertex points which describes the tunnel aperture surface area.

$$A = \{\mathbf{a}_1, \mathbf{a}_2, ..., \mathbf{a}_k\} \tag{8}$$

Femoral tunnel aperture area for different tunnel orientations can be calculated using Formula 7. Results of calculating these tunnel aperture area can be described in A_{res}. As shown in Figure 16B, horizontal tunnel produces more circle-shaped aperture, more vertical tunnel orientation produces a more oval-shaped tunnel aperture. In the literature an aperture area ranging from $65mm^2$ to $150mm^2$ is described. Therefore, every aperture surface area wherein $150mm^2 \leq A_{res} \geq 65mm^2$ will be excluded. In Figure 16A, tunnel aperture area of all potential points in $X_{lateral}$ and a subset of tunnel aperture area in A_{res} are shown in different colors.

In conclusion, the surface area and shape of tunnel aperture at the medial wall on lateral condyle are dependent on the position of the lateral wall and the drill bit diameter. A non-optimal shape of the tunnel aperture can partly result in repetitive graft bending stress on the tunnel aperture caused by the contact area force of the femoral bone during movement of the knee joint. To avoid complications such

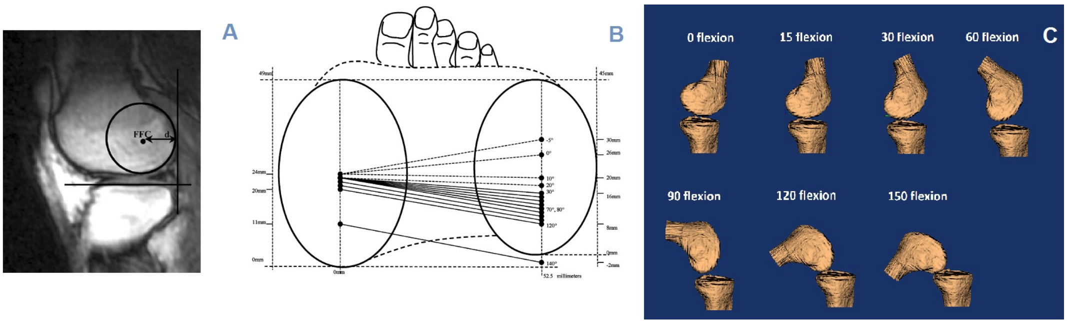


Figure 18: **The pattern of tibio-femoral movement during knee flexion.**.
A representation of femoral condyle movement relative to the tibia during knee flexion based on the literature [68].
A) Circular surfaces at the posterior femoral condyle. The centres of these circles are called the 'flexion facet centre'.
B) Position of the femoral condyle relative to tibia in weight-bearing knee at different flexion.
C) The tibio-femoral movement applied in the 3D model is based on the findings in [68]. A representation of femoral condyle movement relative to the tibia during knee flexion in the 3D model is illustrated.

as posterior tunnel blowout or tunnel-footprint mismatches, the effects of drill size on tunnel aperture and location must be taken into account.

5.8 Tibio-femoral movement over the range of knee movement

Several studies have investigated the kinematics of the tibio-femoral joint in cadaveric and non- and weight-bearing living knees. Most of the studies use MRI or CT to describe the femoral and tibial articular surfaces and the movement of the femoral condyle during the whole range of knee movement [67, 68].
Johal et al. [68] have concluded that the pattern of tibio-femoral movement during knee flexion is independent of gender and knee-side, however the movement pattern differs in non- and weight bearing knees. The motion patterns of the human knee joint are related to a combination of the bony geometry, ligamentous constraints, muscle activity and external forces during specific knee joint functions.
Weigth-bearing investigations in this studies means that tibio-femoral movement in scans were obtained in a standing posture, wherefore subject's weight, divided over both knees, can be taken into account. This provides an accurate approximation of tibio-femoral movement during normal knee joint movement. Anyhow, some caution would be appropriate in extrapolating these results to activities such as walking or running.
Iwaki et al. [67] and Pinskerova et al. have been described a reliable reference point for the position of the femoral condyles in reference to the tibia at the entire range of motion. Circular surfaces at the posterior femoral condyle, visible in the sagittal plane, have been described by these studies, see Figure 18. The centres of these circles are called the 'flexion facet centre'.
Based on these studies, the tibio-femoral movement is implemented in the 3D model. Relative femur condyle positions related to tibia plateau and flexion angle were calculated from findings described in the study [68]. The relative positions were stored in the application, so that it can be applied for each individual geometry. The femoral condyle position can be calculated using these relative values, because the size of the tibia plateau is defined at an earlier phase, see Chapter 4.2, see Figure 13.

Mathematical description of Tibio-femoral movement
The femur has his own coordinate system, wherein the z-axis is aligned with the longitudinal direction of the femur and the x-axis in the latitudinal direction. The origin of this coordinate system is defined in the 'Flexion facet centre' of the femur condyles in the latitudinal direction (along the x-axis). Anatomically, this is located in the intercondylar notch of femur.
The tibio-femoral movement during various arcs of knee flexion will be described in a rotation angle θ and translation distance $\mathbf{c} = \{x_c, y_c, z_c\}$ along the y-axis of the femur in reference to the tibia. In addition, $\mathbf{c} \in \mathbb{R}$ and will never be larger than the size of the tibial plateau. $\mathbf{R}_x(\theta)$ is the rotation matrix along the x-axis and $\theta \in \{0, ..., 150\}$, assuming that the flexion angle of the femur is between 0 and 150 degrees.

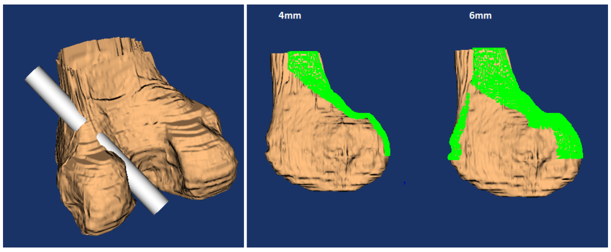


Figure 19: **Thickness of bone around the drilled tunnel**
A drilled tunnel must be surrounded by bone thickness over the entire length of the tunnel.
Left: There is not enough thickness around the tunnel to prevent avulsion of bone.
Right: In green dots is illustrated which potential position will be excluded, because the minimum amount of bone surrounding the entire length of the tunnel. Increasing drill bit size (4 and 6mm) affects the amount of bone surrounding the tunnel.

Let $\mathbf{P}_{femur}$ the actual position of the femur object. The orientation and movement of the femur object in reference with the overall coordinate system can be described as:

$$\mathbf{P}_{femur} = \mathbf{R}_x(\theta)\mathbf{P}_{foriginal} + \mathbf{c} + \mathbf{T} \tag{9}$$

With $\mathbf{T}$ is the translation matrix from the origin of the coordinate system of the femur object to the overall coordinate system and $\mathbf{P}_{foriginal}$ is a vertex point in $\mathbf{X}$. This will have to be applied on all $\mathbf{X}_i$ with $i = 1, .., n$, so that the entire femur object will rotate and translate in reference to the tibia.

5.9 Thickness of bone around the drilled tunnel

A drilled tunnel must be surrounded by bone over the entire length of the tunnel to prevent wall blow out and avulsion of bone. A tunnel orientation such as shown in Figure 19 is thus not permitted. For each potential point on the lateral wall there must be checked if there is enough bone thickness around the entire length of the virtual drilled tunnel. The influence of the drill bit diameter and thickness value around the tunnel on exclude potential tunnel position at the lateral wall is showed in Figure 19.
Mathematically, the limitation of surround bone is described as follow. Let the femoral tunnel described by a line with two points $\mathbf{P}_{begin} = \{x_{begin}, y_{begin}, z_{begin}\}$ and $\mathbf{P}_{exit} = \{x_{exit}, y_{exit}, z_{exit}\}$. A vector along the tunnel is given by

$$v = \begin{vmatrix} x_{begin} + (x_{exit} - x_{begin})t_x \\ y_{begin} + (y_{exit} - y_{begin})t_y \\ z_{begin} + (x_{exit} - z_{begin})t_z \end{vmatrix}$$

wherein a point $\mathbf{X}_0 = \{x_0, y_0, z_0\}$ is perpendicular to the vector along the tunnel and $\mathbf{t}$ the reference point on these vector.

The squared distance can now be describes as:

$$d^2 = [(x_{begin}-x_0)+(x_{exit}-x_{begin})\mathbf{t}]^2+[(y_{exit}-y_0)+(y_{exit}-y_{begin})\mathbf{t}]^2[(z_{begin}-z_0)+(z_{exit}-z_{begin})\mathbf{t}]^2 \tag{10}$$

When minimize the distance

$$\mathbf{t} = \frac{(\mathbf{P}_{begin} - x_0) \cdot (\mathbf{P}_{exit} - \mathbf{P}_{begin})}{\mid \mathbf{P}_{exit} - \mathbf{P}_{begin} \mid^2} \tag{11}$$

The minimum distance can then be found by

$$d_{min} = \frac{\mid (\mathbf{P}_{begin} - x_0) \mid^2 (\mathbf{P}_{exit} - \mathbf{P}_{begin})^2 - [(\mathbf{P}_{begin} - x_0) \cdot (\mathbf{P}_{exit} - \mathbf{P}_{begin})]^2}{\mid \mathbf{P}_{exit} - \mathbf{P}_{begin} \mid^2} \tag{12}$$

Let $\mathbf{X}_0$ be all vertex points ($\mathbf{X}$) in the femur object (wherein $i = 0, ..., n$) and the minimum thickness around the tunnel be $Min_{thickness}$. All points fulfill the condition of formula 13 are included in subset H.

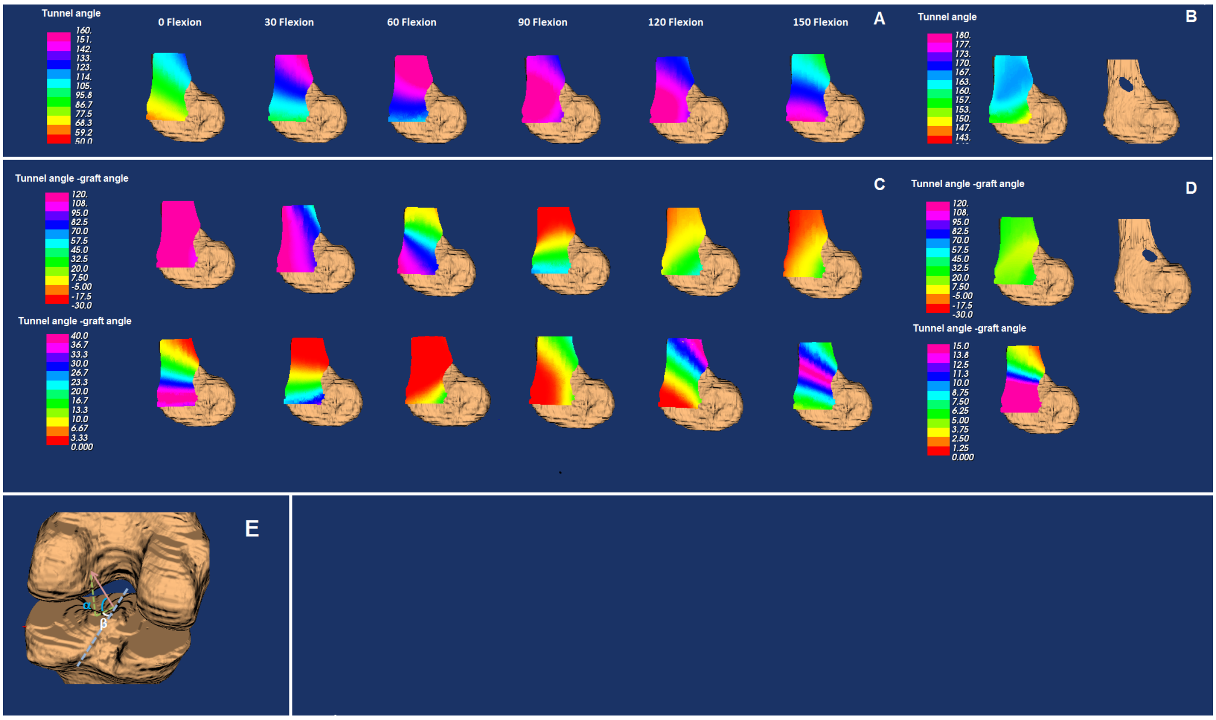


Figure 20: **The bending angle of the point on the lateral wall, femoral aperture and tunnel aperture on tibia plateau in different flexion angles**
A) The angle expressed in a single angle.
C) The angle described in terms of a horizontal and vertical angle. The optimal angle between these points would be an angle of 180 degrees, although this is dependent on the graft orientation at flexion angle.
B,D) The average angle calculated over the entire range of motion.
E) Orientation of the graft and both tunnels can be expressed in a rotational (α) and elevation (β) angles. Here the graft orientation was illustrated only.

$$\frac{|\ (\mathbf{P}_{begin} - x_0)\ |^2\ (\mathbf{P}_{exit} - \mathbf{P}_{begin})^2 - [(\mathbf{P}_{begin} - x_0) \cdot (\mathbf{P}_{exit} - \mathbf{P}_{begin})]^2}{|\ \mathbf{P}_{exit} - \mathbf{P}_{begin}\ |^2} \geq Min_{thickness} \tag{13}$$

In addition to the thickness around the tunnel, it is also important to take into account of posterior wall blow out at the medial wall of the femoral condyle. In the literature is described that risk of wall blow out occurs at a certain drilling knee flexion angle. Basdekis et al [31] suggest at least 4 mm in front of the posterior wall. In the 3D model must thus be taken into account that tunnel aperture, which is influenced by the drilling angle, will not approach the femoral condyle border.

5.10 Calculate Femoral and Tibial Tunnel orientation

In the literature is described that ACL graft and tunnel orientation are proposed as important factors of clinical outcome. Scanlan et al. [69] suggest that walking mechanics are correlated with ACL graft orientation. The peak external knee flexion moment is influenced by the coronal ACL graft angle during walking. If the ACL graft is placed in a more vertical, and thus less anatomical, orientation, a greater reduction in the peak external knee flexion moment is observed. Reduced knee flexion moment during walking can be considered as a marker of an ACL-deficient knee [69].
As previously described, the native ACL consists of two bundles which both have different stabilizing effects. The AMB takes most of the load during anterior tibial translation at high flexion angles, whereas the PLB takes the rotatory force at low flexion angles [70]. This study only focuses on a single bundle reconstruction. The single bundle reconstruction implies that both functions of the AM and PL bundles must be contained in one single bundle. Using a single bundle results mostly in replication of only the AM bundle function of the native ACL.
In the literature is found that anterior tibial translation can be well controlled with a vertical tunnel ori-

entation. However, tunnel aperture becomes more oval shaped and a longer tunnel distance between graft fixation points is observed. These aspects result in graft motion within the tunnel. In contrary, a more horizontal femoral tunnel position did better control both anterior translation and internal tibial rotation [47]. Risk of a horizontal orientation tunnel will become too short, resulting in reduced length of tendon graft. This will impair early graft incorporation. Therefore, the goal is to find an optimal tunnel orientation which includes all above described aspects in earlier chapters, and which takes also into account the fact that the tunnel orientation controls both anterior translation and internal tibial rotation stability.

Tunnel orientation described in different angles

The angle between the point on the lateral wall $\mathbf{p}_{exit}$, the point at femoral aperture $\mathbf{p}_{begin}$ and tunnel aperture on tibia plateau varies with flexion angle. If these angles are expressed in a single angle, the optimal angle between these points would be an angle of 180 degrees, because repetitive bending stress forces on the graft at the femoral tunnel aperture is then minimal.

Through movement of the joint, the 180 degree can not be achieved in all flexion angles at the same location on the lateral wall on the lateral femur condyle. The bending angles over the entire knee flexion in steps of 30 degrees are shown in Figure 20A. The average angle, which is calculated from angles over the entire flexion angle is illustrated in 20B.

Tunnel orientation can also be described in terms of elevation (α) and rotational (β) angles (see Figure 20E), taking into account that the anterior tibial translation can be well controlled with a vertical tunnel orientation and a horizontal femoral tunnel orientation which better controlled internal tibial rotation [47].

In Figure 20B, these bending angles are separately calculated (α and β). In the Figure is shown that these angles are different in both vertical and horizontal calculations at the same position on the lateral wall and at different flexion angles. With cognizance that the PLB has mainly a rotatory stability function and the in-situ forces for these bundle is greatest at 0 to 15 degrees of flexion. The AMB gives an more anteriorposterior translation stability by a horizontal orientation, whereof the in-situ forces increase at 30 to 60 degrees of flexion. Assuming an increased in-situ force results in a more contact force between bone and a graft at a particular bending angle.

The potential optimal angle can now be calculated by minimizing horizontal and vertical angles at associated flexion range and the average overlapping positions.

6 Preliminary results

In this chapter will be described how the above-developed 3D model has been evaluated by calculating differences between surgically placed and model-based calculated tunnel orientation in patients after ACL reconstruction.

Postoperative MRI data from patient after single bundle reconstruction were retrospectively obtained from the hospital. Prerequisite was that the surgical drilled tunnels in femur and tibia were still visible and tunnel orientation can be measured.
The patient-specific geometry of the knee were manually segmented from MRI with ITK-SNAP software and stored separately twice. The first segmentation contains the tibia, femur and both tunnels pre-operatively drilled in the operating theater. In the other, both tunnels were not included and removes manually with ITK-SNAP software from the geometry. With the reason that optimal tunnel orientation can be determined with our 3D model as described in this study.
The basic concept of the algorithm is to determine the optimal patient-specific tunnel orientation with a minimal bending angle of the graft at femoral and tibial tunnel aperture over the entire range of motion. This algorithm will be applied of each included patient and only on the saved geometry models of which the tunnels are removed.
Femoral and Tibial tunnel length will be measured in both geometry of each patient, additionally the graft length in full extension will be measured. Femoral and tibia tunnels and graft orientation will be described in terms of rotational (α) and elevation (β) angles, see Figure 20.
The hypothesis is that the model-based calculated tunnel orientation in failed ACL reconstruction differs from the surgically placed tunnel.

Table 2: Overview of femur and tibia tunnel angle in our patients.

Patient	Status	Femur angle 1 α	Femur angle 2 β	Femur length (mm)	ACL angle 1 α	ACL angle 2 β	ACL length (mm)	Tibia angle 1 α	Tibia angle 2 β	Tibia length (mm)
1	no compl.	17	19	26.6	79	51	39.6	37	56	42.0
2	no compl.	23.7	23.7	34,3	66.8	55.0	26.8	13,2	54.9	38.7
3	no compl.	24.4	21.1	41.60	67.9	48.2	38.1	32.3	49.9	40.4
4	no compl.	13.5	16.9	27.3	81.7	59.7	35.1	54.3	69.6	31.3
5	compl.	45.0	33.3	33.6	34.5	64.5	73.7	73.3	54.0	29.0
Mean value		21,7	21,3	34,2	71,2	51,4	34,8	27,5	53,6	40,4
SD		3,1	1,6	5,0	5,2	2,4	5,3	9,5	2,5	1,1

6.1 Results

5 patients were included, 1 patient with failed ACL recontruction and 4 patients (as far as known) with good clinical outcomes. Currently, there were only a small number of MRI data in the hospital available and useable to be evaluate in the 3D model. The MRI data sequence was not entirely appropriate and must need to be adjusted to specific requirements for this 3D model. The ability to calculate tunnel orientation precisely through the 3D model is particularly depend on the quality of the scan. Because of a long follow-up time of period (six months or more) of ACL reconstruction and with respect of the time remaining of the thesis was at this moment not possible to making new scans. A post-operative MRI scan is not a standard procedure after ACL reconstruction, in order to investigate the study a review board must be notified and give permission .

Results are described in Tabel 2. The average graft length measured in 5 patients is 34,8mm in the range of 26.8 - 73.7mm. In the literature an ACL length of 30.1mm at full extension is described [57]. The average femoral tunnel length is 34.2mm in the range of 26.6 - 41.60mm. In the literature were described of an average lengths in the range of 21.3 to 35.6mm [45, 50, 52, 53].

The rotational (α) and elevation (β) angles were measured separately and average femoral orientation angles are 21.7 and 21.3 degrees respectively. The tibia rotational and elevation were 27.5 and 53.6degrees and graft angles were 71.2 and 51.4degrees.

As suggested in our hypothesis, tunnel orientation in failed ACL reconstruction differs from the surgically placed tunnel. As can be shown in Table 2, there are differences in the values of the angles of the patients with and without complication. At this stage of the project is it not yet possible, to make firm conclusions about the described differences. More research is needed and the algorithm must be adjusted accordingly.

7 Discussion

7.1 Key findings

The present study demonstrates a workflow applied in a 3D computer model to optimize ACL reconstruction in terms of femoral and tibial tunnel orientation to reproduce the native ACL characteristics as closely as possible. To our knowledge, there are no other studies which have applied a computer model for optimal tunnel orientation to clinical practice.
The major principle on which our model is based, is minimization of the bending angle at both the femoral and tibial tunnel aperture over the whole range of motion of the knee. Herein, the geometry of both femur and tibia, the effects of the graft diameter, tunnel aperture shape and size, the possibility of wall blow out and thickness of bone around the tunnel to prevent any avulsions were taken into account. Patient-specific modeling (PSM) has not yet become a standard of care in clinical practice. Major obstacles are high costs, time demanding manual and high technical steps in a workflow from data acquisition into a 3D model. A variety of software products are necessary. In general, a pre-operative planning system will only be used in clinical practice if there is a balance between efficiency, time, costs and if it will influence the clinical outcome. In this study it was attempted to develop an easy to use software application.

7.2 Comparison with literature

Currently, the most two accurate methods for determining femoral and tibial tunnel position after ACL reconstruction are CT and/or MRI scans. Bone in CT scans can be well obtained, however an ACL graft is obviously better perceptible on a MRI. In addition, patients are exposed to increased radiation with CT scans. Therefore, the workflow is developed with data from a CT-scan and the 3D model is evaluated with MRI scans of patients with known clinical outcome after ACL reconstruction.
In this study the femoral insertion sites position in the intercondylar notch was assessed by the method proposed by Bernard et al [32]. In the literature was described different percentage values with the use of this method [30, 32–35]. However, in this study was used the percentage values described in Colombet et al. [30]. These values can be adjusted accordingly, although it does affect the location of the tunnel on the femoral condyle and tunnel orientation consequently.
Several studies described post-operative tunnel positions and compare the results with known characteristics in the literature or related it on the influence of clinical outcome. No other study has determine patient-specific optimal tunnel orientation pre-operatively, although correct femoral and tibial tunnel positioning is one of the major factors in the overall success of ACL reconstructions.
Previous described characteristics, such as aperture shape, graft length, tunnel angle and length, which are of major importance to define optimal tunnel orientation, are compared with in literature described knowledge. The native insertion site of both bundle at femoral condyle is in an oval with an area ranging from $65mm^2$ to $150mm^2$. Average area found in patient was $72.1mm$. The tibial insertion is also an oval with an area found between $67mm^2$ and $259mm^2$ and average area found in patient was $82.4mm$. These both parameters fall thus in the native area range.
In the literature were native ACL elevation angles in coronal and sagital plane reported in the range of 66 and 74 degrees and in de range of 43 and 59 degrees, respectively. Hosseini et al. [28] and Fuijimoto et al. have compared the elevation angle of the graft in failed ACL reconstructions with native ACL orientation. Elevation angles in coronal plane were found in the range of 75.9 and 79.5 degrees and in the sagittal plane of 69.6 degrees. In general, higher elevation angles were found in failed ACL reconstructions.
Illingworth et al. [42] has investigate the relation between anatomic placed tunnel and tunnel and graft inclination angle. The study concluded that a decreasing in femoral tunnel angle and increasing in inclination angle is correlated with a non-anatomic ACL reconstruction. A tunnel angle more than 32.7 degrees and inclination angle less than 55 degrees fell within an anatomic range. These data were described in a two-dimension, however tunnel orientation in this study were in terms of horizontal and vertical tunnel angles. This is an entire different approach of describing the angles, but provides a more accurate description of graft and tunnel orientation. In this study, tunnel orientation were obtained from MRI data wherein the tunnel is surgical placed and whereof tunnel were calculated on model-based. These tunnel angles are therefore not comparable with angles described in the literature. Tunnel orientation found in horizontal plane of average in 71.2 degrees and vertical in 51.4 degrees in range 66.8-81.7 degrees and 48.2- 59.7 degrees respectively.

Several cadaveric studies reported variable femoral mean tunnel lengths from 21.3 to 43.3mm. One of the described boundary conditions was that the length of the tunnel is at least 30mm. Tunnel length obtained in the patient were in average of 34.2mm and thus fall in these range.

7.3 Limitations

There are several important limitations to this study. The C++ programming language in combination with the free Visual Studio Express 2012 for Windows Desktop environment to build and developed algorithms is used in this study. In addition, The Visualization Toolkit (VTK) and ITK-SNAP, both open-source, freely available software toolkits for segmentation, interactive, image processing and supports a wide variety of visualization algorithms was implemented. The use of these product, in instead of for example MATLAB, was the consideration of using an user-friendly environment in clinical care and all essential steps could be implemented in an all-in-one application. Moreover, an important consideration was reducing compute time to calculate complex algorithm in C++. However, in retrospect, there is a lot of time spent in setting up the computer platform and elaboration of certain features, such as object-oriented programming of knee objects, set up the communication between The Visualization Toolkit (VTK) and ITK-SNAP and reduce time to compute algorithms.
Tibia and femur will be manual segmented from MRI or CT scan and mesh to a 3D model. In this study is particularly focused on the pre-processing /modelling phase and therefore there is currently still chosen to segment the scans with time-consuming (semi)-manually algorithms. In order to achieve an all-in-one application, the segmentation process must also be implemented in the application.
The ability to calculate precisely tunnel orientation through the 3D model, will particularly depend on the quality of the scan methods and accuracy of the segmentation and smoothing algorithm. Currently, there were only a small number of MRI data in the hospital available and useable to evaluate in the 3D model. Beside, the MRI data sequence were not entirely appropriate and must need to be adjusted to specific requirements. Due to the follow-up of six months in order to describe the clinical outcome and with respect of the time remaining of the thesis, it was not possible to include new patients with the appropriate MRI sequence. Because of the influence of small geometric differences on tunnel orientation, the segmentation and smoothing algorithm must have to improve in future research in order to make use of a qualitative geometry.
Notchplasty is frequently performed during ACL reconstruction, however corrections is not taken into account in this model at the time. The effect of notchplasty on tunnel placement and knee biomechanics with ACL reconstruction is investigated by Keklikci et al [71]. Small geometric differences in order of 2mm has influence on the anteriorposterior laxity and in situ force graft force compared with pre- and post-notchplasty knees. Fu et al. [41] suggested that an aggressive lateral notchplasty might displace the femoral insertion of the ACL laterally, leading to abnormal knee kinematics. The femoral and tibia geometry segmented from pre-operative MRI's, which is used to determine optimal tunnel orientation, has thus changed with the bone removal. In addition, changed location of the tunnel entrance has influence on the femoral tunnel length, graft-bending angle and tunnel aperture geometry and consequently also the optimal tunnel orientation.
A drilled tunnel must surrounded by bone over the entire length of the tunnel to prevent wall blow out and avulsion of bone. In the literature were no true values found for thickness around the bone to prevent avulsion of bone. This is of importance in the determination of optimal tunnel orientation, because position at the lateral wall depends on drill bit size in combination with thickness around the bone. Also, wall blow out is of major interest and Basdekis et al [31] suggest at least 4mm in front of the posterior wall. Further investigation must done to these parameters.
In this study is focused on optimal tunnel orientation related to minimize acute tunnel angle over entire range of motion. In addition, of major interest is the contact and friction caused by the graft impingement against sharp bone edges of the tunnels at different knee flexion angles. In the literature is also described that this had a direct effect on the tension in grafts [11]. The contact force is not implemented in the 3D model and must be taken into account in order to calculate optimal tunnel orientation.
The optimal angle is admittedly calculates over entire range of motion. However by time-consuming calculation restrictions, were results not continuously calculated, but were obtained of static positions of knee flexion angle (0, 15, 30, 60, 90, 120, 150 degrees). To more accurately calculate the optimal position, the calculations will be done over a continuous range of knee motion.
Injuries to secondary ligaments, capsular structures, articular cartilage and meniscus also affect the overall

success or failure after ACL reconstruction [19]. Surrounding structures were not implemented in the 3D model.
Several investigations in the literature regarding computer-assisted ACL reconstruction contains a variety of conflicting outcome data. Meuffels et al, a recently published study, did not show significant increase in either accuracy or precision of the tunnel placement in computer-assisted surgery compared with conventional surgery. With this study we want to determine the ideal tunnel orientation individually, however it does not mean that orthopedic surgeons can drill these accurate and precise.
In current surgical approach, there is still a considerable freedom in drilling angle of femoral tunnel orientation influenced by femoral notch shape, the associated placed endoscopic portals and the flexion angle of the knee. The 3D model calculate an optimal angle for each individual patient without these limitations. Basdekis et al. [31] evaluated femoral tunnel orientation with respect to knee flexion angle during single bundle reconstruction. Drilling the femoral tunnel at 110 degrees of knee flexion were recommended, while drilling at 90 degrees of knee flexion there is a risk of posterior wall blow-out [31]. In the literature is high flexion angle recommended to drill the femoral tunnel. Although, calculated angle are provided in terms of horizontal and vertical femur and tibia tunnel angles and can not mapped with to the flexion angle, such as described in literature.

7.4 Future implications

Further work will be necessary to determine the influence of the model-based tunnel orientation on clinical relevance and its impact on clinical outcome after ACL reconstruction. A number of steps are needed to actually apply the 3D model in a clinical setting.
The initial tension applied to the graft is one of the factors that affects joint contact forces and knee laxity which may be responsible for long-term complications such as OA. There is no consensus regarding the optimal tension to be applied on the graft during ACL reconstruction [18]. Some have recommended a low initial graft tension, however this will induce postoperative anterior laxity and thus not provide adequate joint stability. Others have advocated high tension, however, over-tensioned grafts may not only lead to a limitation of the range of the motion, but is susceptible of early graft failure [18]. In order to determine the ideal femoral and tibial tunnel orientation, in situ forces on the graft are needed to take into account, since it affects the amount of shearing forces between graft and bone over the whole range of motion. These properties should be incorporated into the 3D model to do actual pronunciation about the ideal tunnel placement. In addition, our suggestion is that the tension applied to the graft should not be generalizable for each patient, but needs intraoperative determined for each individual patient on the basis of the load-elongation curve.
The algorithm as described is applicable for single bundle reconstruction with autograft and Endobutton fixation. Customization of several properties and anatomical characteristics in current algorithm can thereafter be used for example in PCL reconstruction or double bundle reconstruction. PCL injuries occur in up to 44% of all acute knee injuries and clinical failure rate after reconstruction described of approximately 11.6%. In accordance with the ACL reconstruction, the failure rate would be possible reduce with individualized tunnel placement. Since the entire workflow is already developed, we can use the model with minor changes for this condition as well.
In this study are the above-developed 3D model already applied to a few patients after a single bundle reconstruction. Optimal tunnel orientation could be calculated with the algorithm and compared with the tunnel orientation as actually surgical placed in the patient. Our hypothesis is that tunnel position of failed ACL reconstruct significant differs from the predicted tunnel placement of our developed 3D model. A retrospective cohort study must therefore set up with more number of patients to actually be able to make significant pronouncements. After a follow-up of minimal six months to be able to describe the clinical outcome and compare these with tunnel orientation.
The previous described study will compare tunnel orientation in patients with failed and good clinical outcome. The hypothesis is that tunnel position of failed ACL reconstruct significant differs from the predicted tunnel placement of our developed 3D model. However, it does not mean that calculated tunnel orientation from our model is clinically and anatomically correct. Validation of our 3D model in an anatomical cadaver study is therefore necessary.
In this study is focused on the pre-process phase, such as described earlier in Figure 2. It implies that we are able to calculate an optimal tunnel orientation of each individual patient from an imaging scan. Now these data is available, in terms of horizontal and vertical tunnel angles, it will be necessary to investigate whether the current surgical instruments are suitable for drilling these angles. In all probability new

drilling methods of surgical instruments must be developed.
After evaluation of the computer model as mentioned above, the next step is to validate the model and determine its clinical relevance in a patient trial. In this trial the clinical outcome of the conventional surgical ACL reconstruction method will be compared with the clinical outcome after computer determined tunnel orientation. Our hypothesis is that individualized ACL reconstruction based on a 3D model is associated with a reduction of ACL reconstruction failures. Wherein clinical outcome can be measured with patients physical function, obtained from physical examination and Patient Reported Outcome Measurement (PROMs) instruments.

7.5 Conclusion

Most of the previous described studies have focused on replicating the native footprints of the ACL bundle. Results of these studies were of major influence on the surgical approach for ACL surgery. However, these studies have mainly focused on a two-dimensional orientation. Our suggestion is to use a more reliable approach in 3D orientation to improve ACL reconstruction. The present study is a first step to determine femoral and tibia tunnel orientation pre-operatively with a 3D computer model to individualize ACL reconstruction. We focused on a minimal bending angle between femoral and tibial tunnel aperture over the entire range of knee motion. Patient characteristics, aperture area, tunnel length, graft inclination and graft diameter to reproduce the native ACL characteristics as closely as possible were take into accounts.
The 3D model has been evaluated by calculating differences between surgically placed and model-based calculated tunnel orientation in patients after ACL reconstruction. Result were compared with knowledge in the literature.
Further work is necessary to determine the clinical relevance of the morphologic orientation of both tunnels after ACL reconstruction. Several additional studies will be needed prior to application of the 3D model into clinical practice.

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