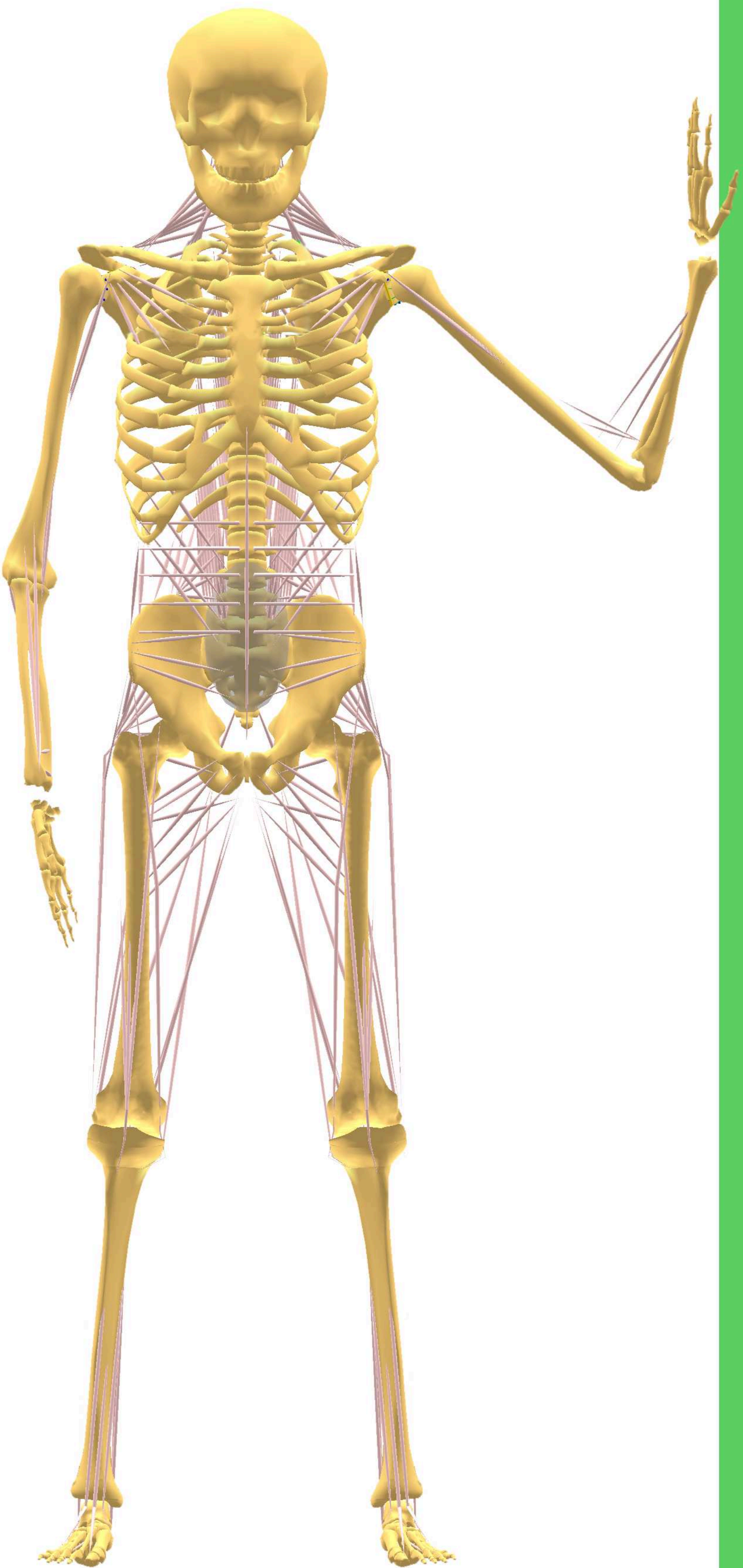
Biomechanical Analysis of Anterior Cruciate Ligament Injury Mechanisms



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Abstract

The anterior cruciate ligament is the most frequently injured knee ligament. The ligament can be injured due to a sudden external impact like a traffic accident, but in most cases the injury is a result of people participating in athletic activities.
This ligament has a significant impact on people's quality of life, because it is a basis for a normal knee function. The patients are mostly young people and the consequences often affect them for the rest of their lives.
Regardless of the great number of injuries, the trauma mechanisms are still unclear. A better understanding of the aetiology might increase the possibilities to prevent the injuries and improve the rehabilitation strategies.

The objective of this project was to determine which trauma mechanisms have the potential to rupture the anterior cruciate ligament by quantifying the strain in the ligament during both voluntary and forced movements.

Although a ligament injury may appear to have been caused by a single inciting event, it may be a complex interaction between internal and external risk factors. The mere presence of these risk factors is not sufficient to produce injury, but they predispose the athlete for the injury to occur in a given situation. The inciting event is the final link in the chain that causes an injury. The project did not attempt to determine the factors that increase the risk of sustaining an injury, but focuses on the inciting event - the injury mechanism.

Anterior cruciate ligament injury mechanisms were studied with four musculoskeletal models made with *The AnyBody Modelling System*™. AnyBody is a general musculoskeletal modelling and optimisation software system based on inverse dynamics. The inverse dynamic analysis determines the unknown forces from the equations of the known motion. Due to the redundancy of the muscle actuator configuration, the muscle recruitment problem is formulated as an optimisation problem.

The musculoskeletal models made it possible to determine the knee shear force during various sports movements and explore the elongation of the anterior cruciate ligament during both natural and forced movements.

In order to investigate which movements that have the potential to rupture the anterior cruciate ligament, it was chosen to quantify the strain in the ligaments and muscles around the knee joint during a forward lunge. The dynamic analysis was applied to a model with the anterior cruciate ligament intact, and to a model without the ligament.

It had been expected that there would be a significant difference between the two models, because studies have shown that anterior cruciate ligament deficient subjects perform a forward lunge differently from healthy subjects. However, the dynamic analysis showed that there was no difference in muscle activity or joint reactions between the two models. The analysis revealed that the knee joint reaction produced an anterior pull in the proximal tibia. In other words, the anterior cruciate ligament was unstrained.

An analysis of a male runner showed that sprint strains the anterior cruciate ligament. However, the knee shear force, which was used to evaluate the ligament strain, was well below the ultimate tensile strength of the ligament. Considering that sprint probably is one of the most intense sagittal plane sports movements, it appears that voluntary contraction is insufficient to injure a healthy cruciate ligament. Even though intense voluntary contraction might be insufficient to injure the anterior cruciate ligament during sprint, the analysis does not rule out the possibility that other sagittal plane movements may put more strain on the ligament. Therefore, a representative selection of various feasible sagittal plane movements were analysed with the sagittal model. The analysis of the sagittal model demonstrated that it is unlikely that sagittal plane mechanisms will rupture the anterior cruciate ligament.

In the lunge model, the runner model and the sagittal model the knee joint was approximated as an ideal hinge. But the relative movements between femur and tibia are far more complex and are related to a complicated interaction between muscles, ligaments and bones. In addition to the knee joint's natural movement, flexion/extension, it can also be forced into hyperextension, valgus or varus positions, increased internal/external rotation and anterior/posterior translation of the tibia.
The advanced knee model made it possible to investigate the elongation of the anterior cruciate ligament for various knee positions and thereby evaluate which movements are most likely to tear the ligament.
The analysis of the model showed that: The ligament is strained the most when the knee joint is flexed $5° - 25°$. Anterior translation of the tibia increases the strain significantly. Valgus and especially varus positions can increase the strain in the ligament significantly and the ligament is therefore likely to tear if the knee joint is forced into either varus or valgus. Rotation of the tibia about its longitudinal axis only produces minor strain and it seems implausible that this mechanism will injure the ligament.

It was found that visual analysis of injury situations does not produce the information necessary to evaluate the strain level in the anterior cruciate ligament at failure.

Preface

This report is the result of a final year project from the Institute of Mechanical Engineering of Aalborg University. The work was carried out from September 2007 till 9th of January 2008.
Anterior cruciate ligament injury mechanisms were analysed with *AnyBody 3.0* which is a software system for computer analysis of the mechanics of the human body. Terms relating to the modelling software and medical terms are explained continuously. Terminology which has not been defined in the report may be found in the nomenclature list in Appendix A
The report is accompanied with a compact disc containing data used or generated during the course of the project and a digital copy of the report and the appendices.

My profound thanks go towards Professor John Rasmussen for his support, inspiration and guidance throughout the project. Thanks to Mark De Zee for inspiring literature and to everyone at *AnyBody Technology* for their support and help with *The AnyBody Modelling System*™.

Contents

CHAPTER 1

Introduction

Every year there is more than 5000 knee ligament injuries in Denmark. The severity of the injuries range from insignificant to disabling. While some require acute surgery, others heal by themselves, however, a large part of the injuries lead to so much discomfort that they require a ligament reconstruction, (SAKS, 2006).
The most frequent injury occurring is a rupture of the anterior cruciate ligament. This ligament has a significant impact on people's quality of life, seeing that the ligament is a basis for a normal knee function, both during everyday activities like walking and intense physical strain. The patients are mostly young people and the consequences often affect them for the rest of their lives. Unfortunately it is only a very few of the acute treatment strategies and the rehabilitation strategies that are based on evidence, they are on the contrary often related to local traditions at the hospitals, (SAKS, 2006).
In 2006 there were 3062 knee ligament operations of which 85% were isolated, first time reconstructions of the anterior cruciate ligament, Lind. (2007). In addition to this, Lind. guesses that there are 1000 - 1500 diagnosed anterior cruciate ligament ruptures that does not get treated with surgery.
Regardless of the great number of injuries the trauma mechanisms are still unclear. A better understanding of the aetiology might increase the possibilities to prevent the injuries and improve the rehabilitation strategies.

1.1 Objective

The objective is to determine which trauma mechanisms have the potential to rupture the anterior cruciate ligament by quantifying the strain in the ligament during both voluntary and forced movements.

1.2 Knee Anatomy

In order to understand how a rupture of the anterior cruciate ligament occurs and how it affects the knee, it is essential to understand the basic anatomy of the knee.
The knee joint, *articulatio genus*, which connects the thigh and the crus, is a modified hinge joint with two degrees of freedom. Its primary movement is flexion-extension and when the knee is bent the crus can rotate medially and laterally[1]. The bones, *Os femoris* (*femur*), *tibia* and *patella*, make a very open and almost plane socket and the stability of the joint is ensured by ligaments, joint capsule and the surrounding muscles.

The joint capsule surrounds *patella* and respectively top and bottom of *tibia* and *femur*. The capsule consists of an outer and an inner membrane separated by a thin layer of fat. The outer membrane, *membrana fibrosa*, consists of strong collagen threads. The inner membrane, *membrana synovialis*, produces synovial fluid which lubricates the joint and gives sustenance to the cartilage cells. The articular cartilage which surrounds the bone ends is shock absorbing and reduces the friction.
Femur and *tibia* are separated by *meniscus medialis* and *meniscus lateralis* which distribute the pressure between the bones and contribute to absorption of shocks. The anatomy of the knee is illustrated on Figure 1.1.

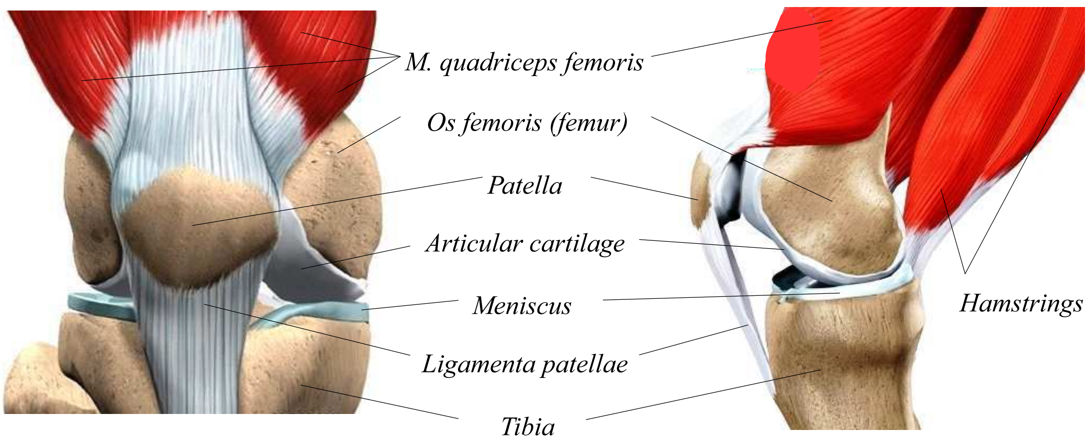


Figure 1.1: Knee anatomy

The flexion-extension motion is a combination of rolling and sliding. When the knee flexes the lateral femoral condyle slides posteriorly, but the medial femoral condyle demonstrates almost no translation which means that the femur rotates laterally.
Four ligaments are crucial to the control and stability of the knee. *Ligamentum collaterale tibiale* and *ligamentum collaterale fibulare*, which runs along the knee one on each side of the joint capsule, prevent varus and valgus motions. *Ligamenta cruciata* are two intraarticular ligaments. *Ligamentum cruciatum anterius* prevents that the *tibia* slides forward with respect to *femur*, while *ligamentum cruciatum posterius* has the opposite effect

[1]Medial and lateral rotation is interchangeable with respectively internal and external rotation.

and prevents that the *tibia* slides backwards. Cadaver studies have shown that when the knee is flexed at 30° the anterior cruciate ligament represents 87% of the total capsular and ligamentous resistance, Butler et al. (1980). The anterior cruciate ligament prevents hyperextension, large rotation of the *tibia* and contributes to the knee's sideways stability, Olsen et al. (2004). The ligaments are illustrated on Figure 1.2.

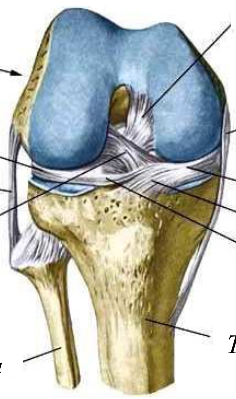


Figure 1.2: Knee anatomy

The surrounding muscles are stabilising the knee as well. Table 1.1 outlines the muscles that flex, extend and rotate the knee. These muscles are described in Appendix B.

Flexion	**Extension**	**Medial rotation**	**Lateral rotation**
M. sartorius	M. rectus femoris	M. sartorius	M. biceps femoris
M. gracilis	M. vastus lateralis	M. gracilis	
M. biceps femoris	M. vastus intermedius	M. semitendinosus	
M. semitendinosus	M. vastus medialis	M. semimembranosus	
M. semimembranosus		M. popliteus	
M. gastrocnemius			
M. plantaris			

Table 1.1: Overview of the muscles that flex, extend and rotate the knee

Femur and *tibia* make a very open and almost plane socket. When the tibial plateau leans forward the natural response from the *condyli femoris* are to slide forward out of the socket. This motion is opposed by *musculus quadriceps femoris*[2], that has the insertion

[2]i.e. *musculus rectus femoris, musculus vastus lateralis, musculus vastus intermedius* and *musculus vastus medialis*

on *tuberositas tibiae* via *ligamenta patellae*, and pulls the proximal *tibia* forward. The hamstrings[3] have the opposite effect and are capable of pulling the *tibia* backwards.

Not only the muscle force but also the gravitational force has an impact on the movement of the knee joint. The gravitational force will extend the knee when the centre of gravity is in front of the knee and must be balanced by a flexor moment from the hamstrings. When the centre of gravity is behind the knee, the gravitational force will flex the knee, which must be balanced by an extensor moment from quadriceps. When the body is accelerated or decelerated compared to the knee, the muscles must respond with extensor or flexor dominance.

Because of the dynamics and the requirement for moment as well as force equilibrium the difference between the force in the quadriceps and the hamstrings might be too large to keep *tibia* and *femur* together. This difference should be absorbed by the cruciate ligaments, *ligamenta cruciata*. See Figure 1.3.

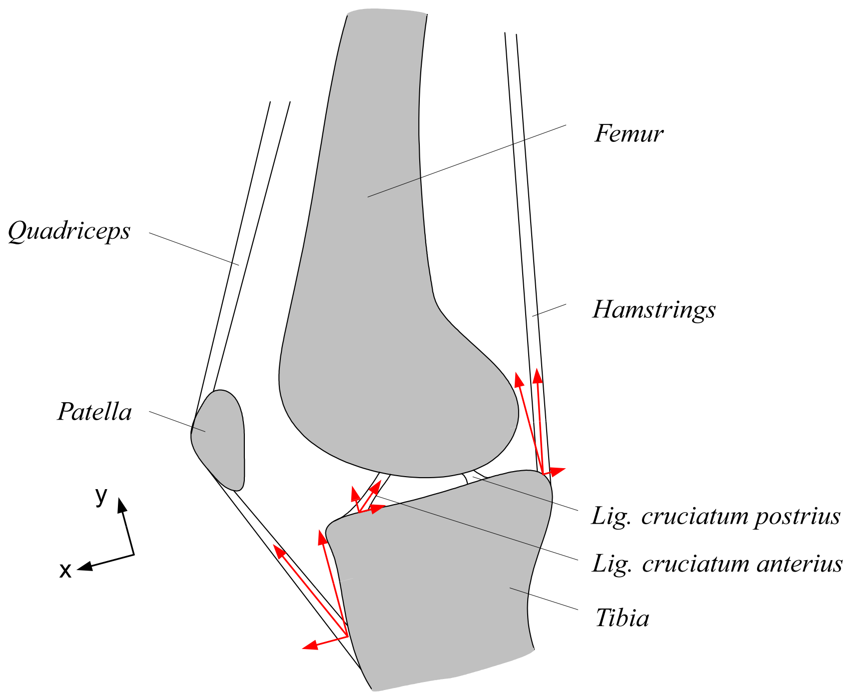


Figure 1.3: The strain on *ligamenta cruciata* arises due to the the difference between the anterior component of the drag from quadricps and the posterior component of the drag from the hamstrings. The figure illustrates the case where the anterior component of the drag from quadricps is larger than the posterior component of the drag from the hamstrings and put strain on *ligamentum cruciatum anterius*

When the anterior component of the drag from quadriceps exceeds the posterior component of the drag from the hamstrings, the force difference is absorbed in the anterior

[3]i.e. *musculus biceps femoris, musculus semitendinosus* and *musculus semimembranosus*

cruciate ligament and in the joint capsule. Co-contraction[4] of the hamstrings prevents that the force exceed the maximum tensile strength of the ligament. Krogsgaard (2002) believes that the cruciate ligaments would be torn during everyday activities if the hamstrings did not co-contract.
The function of the anterior cruciate ligament is not only mechanical. The ligament contributes to the proprioceptive sense that determines the position and the strain on the knee. The neuromuscular function is being disturbed if the ligament is thorn, which reduces the ability to coordinate the muscle activity around the knee. This is termed functional instability, Krogsgaard (2002).
The sensory nerves in the ligament can trigger a muscular reflex in the hamstrings. During knee extension this reflex will inhibit anterior migration of the proximal *tibia*, but the latent time of the reflex is too long to protect the anterior cruciate ligament against sudden strain, and the function of the reflex is probably to coordinate the movement of the knee, Krogsgaard (2002).

1.2.1 Anterior cruciate ligament injuries

Ligamentum cruciatum anterius is probably the most commonly injured ligament of the knee. The ligament can be injured due to a sudden external impact like a traffic accident, but in most cases the injury is a result of people participating in athletic activities.
Numerous types of sports have been associated with injuries on the anterior cruciate ligament, especially those sports where a foot is planted on the ground while the body changes speed or direction rapidly as in team handball, basketball and football. These sports all involve player-to-player contact which can lead to ligament injuries; however, most injuries seem to occur without contact, Olsen et al. (2004), SAKS (2006). Four mechanisms of trauma have been recognised.

Trauma mechanisms:

- Medial rotation of the *tibia* while the knee is slightly flexed.
- Valgus trauma possibly combined with a medial or lateral rotation of the *tibia*.
- Hyperextension trauma.
- Intense deceleration.

The hyperextension trauma seems to be rare; there is, however, a dispute about whether the primary injury mechanism is valgus trauma, medial rotation of the *tibia* or intense deceleration.
As described above contraction of the quadriceps may result in significant anterior shear

[4]Co-contraction is the phenomenon that when a muscle is being activated then the antagonist is also being activated and hereby protects the ligament.

force on the proximal *tibia*. One hypothesis states that the contraction of quadriceps alone may normally be insufficient to tear the ligament, but the addition of a rotation of *tibia* or a valgus movement, which could cause additional strain in the ligament, might result in a rupture of the ligament, Olsen et al. (2004).
A video analysis of female team handball players reviled that the main injury mechanism seemed to be a forceful valgus collapse from a position with the knee close to full extension combined with a slight rotation of the *tibia*. Unfortunately it is not known "whether the consistent valgus collapse observed in the videos was actually the cause of the injury or simply a result of the ACL being torn", Olsen et al. (2004).

Symptoms

The athlete will often experience a forceful crack inside the knee and an intense pain when the ligament tears. Torn blood vessels in the damaged ligament will cause swelling of the knee joint within short time following the injury. The instability caused by the ruptured ligament leads to a feeling of insecurity and giving way of the knee. The pain and swelling from the initial injury will usually be gone after two to four weeks but the knee may still feel unstable.

Diagnosis

The injury is identified on the basis of dialog between the patient and a doctor, about the trauma and the anamnesis, followed by a physical examination to evaluate the looseness of the joint. It is often difficult to identify the right diagnosis and the extent of the injury, Ottosen (2007). A significant twist trauma of the knee joint that leads to a rupture of the anterior cruciate ligament will often induce fracture of other ligaments or menisci. A valgus trauma will often rupture both *ligamentum collaterale tibiale* and *ligamentum cruciatum anterius*. It has been observed that about half of the patients who get a reconstruction of *ligamentum cruciatum anterius* will also have a lesion on *meniscus medialis* or *meniscus lateralis*, Jakobsen.

Treatment

The ruptured anterior cruciate ligament cannot heal and cannot be sutured. Rehabilitation will improve the compensatory functional stability, but it will not stabilise the knee mechanically. Most patients will be able to obtain sufficient control of the knee to perform everyday activities but they will still experience pain when the knee is in extreme positions or when the knee has to support excessive load. It is necessary to reconstruct the ligament if the patient wants to perform intensive sports activities or has a job that requires a mechanically stable knee. Even with a reconstruction of the ligament the patient might still experience pain and still have difficulties performing sport on a serious level.

Non-surgical treatment

The initial treatment focuses on decreasing pain and swelling of the knee. The patient will normally receive physiotherapy. The duration and extent of the rehabilitation programs depends on the hospital and seems to vary a lot.

Surgery

Most patients will attend physiotherapy before the surgery in order to reduce the swelling, regain muscle strength and secure full range of motion. This practice reduces the risk of scarring inside the joint and can accelerate the recovery after the surgery.
The torn ligament is replaced with a piece of tendon. The graft is usually cut out from the patient's patellar tendon or the hamstring tendons. The surgery doesn't require the surgeon to open the joint, but can be accomplished with arthroscopy. The patient will be involved in a rehabilitation program after the reconstruction.

Physiotherapy

The objective of the physiotherapeutic treatment is to ensure that the patient has a functional, stable knee, which does not give way during everyday activities. The patient should regain full range of movement and muscle strength. With a reconstructed ligament the patient should ideally be able to return to the same level of activities as before the accident, but that is strongly dependent on the extent of the injury and the patient's motivation.
The treatment includes movement, balance, strength and function training, which comprises training of proprioception. There has been developed a lot of different treatment programs but it is very difficult to compare their effect. The rehabilitation is to a significant extent controlled by the local traditions at the hospitals rather than by evidence, (SAKS, 2006). The physiotherapeutic treatment is described in more detail in Appendix C.

Result

Everyone who has a knee injury has an increased risk to develop osteoarthritis in the injured knee compared with the healthy knee. Reconstruction of the anterior cruciate ligament will not prevent osteoarthritis, but it reduces the risk of damaging articular cartilage and meniscus in the ligament-deficient knee, and in case of meniscus lesion it will increase the healing potential of a fixed meniscus. There is an increased risk of developing osteoarthritis in the ligament deficient knee in case of meniscus resection, (SAKS, 2006).

The patients that do not go through surgery will recover faster, but the knee is more instable and 50% will experience a bad result or recurrence. A larger percentage of the patients, who get a reconstruction of the ligament, will have good results. Approximately 80% of elite sportsmen may be expected to be able to return to their sports activities after 12 months, but only 50% will retrieve the same level as before, (SAKS, 2006).

Ottosen states that a more extensive rehabilitation program might give better results for both patients with and without ligament reconstruction.

Copers and non-copers

The anterior cruciate ligament deficient patients can be divided in two groups: The copers and the non-copers. Only half of the patients - the copers - are able to return to their everyday and sports activities, while the other half of the patients - the non-copers - are unable to take part in the same activities as they did before the injury.

Patients with a ruptured anterior cruciate ligament have different ways of compensating for the missing ligament. Alkjær et al. (2002) investigated copers', non-copers' and healthy subjects' different movement pattern, when performing a forward lunge[5]. The non-copers performed the movement significantly more slowly and loaded the knee joint less than the copers and the control subjects. The copers moved more slowly during the knee flexion phase of the forward lunge, but just as fast as the control subjects during the knee extension.

The non-copers had a reduced peak angular velocity of the knee joint during both the extension and flexion phase of the movement. Alkjær et al. (2002) interpreted the slow movement as an attempt to reduce the quadriceps force needed to accelerate and decelerate the body mass, seeing that a reduced quadriceps force might decrease the anterior translation of *tibia*.

The exact mechanisms responsible for the dynamic knee joint stability are still unclear, but the different movement patterns might indicate that the anterior cruciate ligament plays an important role in performing a forward lunge.

In order to investigate which movements that have the potential to rupture the anterior cruciate ligament, it was chosen to quantify the strain in the ligaments and muscles around the knee joint during a forward lunge.

[5]Performing a forward lunge means to stand in an upright position, then take one step forward and flex both knees and subsequently extend the knee in front and push oneself back into the upright starting position.

CHAPTER 2

The Lunge Model

This chapter treats a model of the human body performing a forward lunge. The strain in the ligaments and muscles around the knee joint was quantified during a forward lunge in order to gain a better understanding of which movements that have the potential to rupture the anterior cruciate ligament. The software system that was used to create the model is described and it is explained why and how the model was made the way it was.

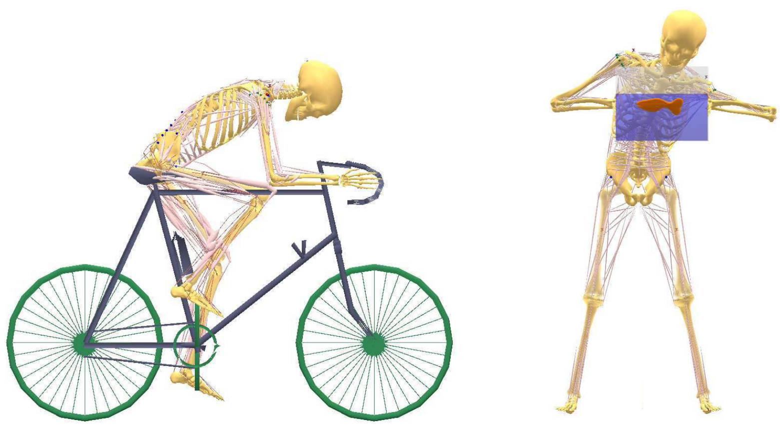
2.1 The AnyBody Modelling System™

AnyBody 3.0 is a software system for computer analysis of the mechanics of the human body. The system can model the musculoskeletal system and the environment that interacts with the human body; compute forces in individual muscles, elastic energy in tendons, joint reactions, antagonist muscle action etc. A body model consists of rigid segments - the bones, joints between the segments and tendon-muscles units. Figure 2.1 shows an example of a human model in two different environments.
The models are constructed with *AnyScript*, which is an object-oriented programming language developed for the *AnyBody* system. *AnyScript* also contains facilities for definition of movements, constraints and external forces.

Kinematics

The kinematic analysis determines the position of every segment in the body at all times. An unconstrained segment has six degrees of freedom, but the segment can be constrained by adding joints or a driver, which is a motion predefined by the user, i.e. a function of time determining the position of a joint or some other kinematic measure[1] at any given time. This means that a system with n segments has 6n degrees of freedom and it is necessary with 6n equations (joints or drivers) that constrain the system in order to solve the

[1]*AnyBody Technology* invented the concept of kinematic measures as a way of describing dimensions in a kinematic model. This enables the user to study the development of selected dimensions or control the kinematic measure by adding a driver.



(a) Human model riding a bike (b) Standing human model

Figure 2.1: Models developed with the AnyBody Modeling System

kinematic analysis. The kinematic analysis also determines velocities and accelerations.

Dynamics

In a mechanical system the forces can in principle be determined by setting up equilibrium equations and solving them, but a biomechanical model is often more complex because the problem often is statically indeterminate and because the muscles are only able to pull.
In a musculoskeletal system there will be infinitely many different muscle force combinations that can balance the external loads and it is not known how the human body distributes the forces between the redundant muscles. This means that there are not enough equilibrium equations to solve the problem.

In AnyBody this is solved by adding an optimality criterion that determines the muscle recruitment, Rasmussen et al. (2001), Damsgaard et al. (2001). The optimality criterion builds on the assumption that the human body attempts to recruit the muscles in such a manner that fatigue is postponed as long as possible. This is obtained by minimising the maximum muscle activity, where muscle activity is defined as muscle force divided by muscle strength.

Model Repository

It is an exhaustive job to develop a model of the human body from scratch. Fortunately it will often be possible to reuse other people's models, which can be found in the AnyBody Model Repository. The repository is divided into an Application Repository and a Body Repository. The Body Repository contains a full body and a selection of subsets of the body, like an arm, a single leg, the upper or lower extremities etc. None of the models have specified supports, movement or external forces. The Application Repository contains body models that are connected to some sort of environment and have supports, movements and external forces. Figure 2.1 shows two different models from the application repository. The model used to asses the differences between a healthy knee and an anterior cruciate ligament deficient knee is a modification of an existing model. The model will be described in the following.

2.2 Lunge model

In Section 1.2.1 it was described that there is a significant difference between how anterior cruciate ligament deficient subjects perform a forward lunge. Studies have shown that the copers, who were able to return to the same activity level as before the injury, moved more slowly during the knee flexion phase of the forward lunge, but just as fast as the healthy control subjects during the knee extension, while the non-copers performed the movement significantly more slowly and loaded the knee joint less than the copers and the control subjects, (Alkjær et al., 2002). The different movement patterns might indicate that the anterior cruciate ligament plays an important role in performing a forward lunge and on basis of theses studies it was decided to investigate a model of the lower extremities of the human body performing a forward lunge, Figure 2.2. The upper extremities were omitted in order to reduce the computational effort. The AnyBody model will be described in the following.

(a) Take one step forward from an upright position

(b) Flex knee

(c) Maximum knee flexion. Start to extend the knee and push back

Figure 2.2: Model of the lower extremities performing a forward lunge.

2.2.1 Segments

The model contained seven segments: Pelvis, thighs, shanks and feet. The pelvis consists of three bones but it was modelled as one ridged segment. The thigh bone, *femur*, was modelled as one segment. *Tibia* and *fibula* are the two bones in the crus. The body weight is transferred through *tibia* from *femur* to the foot, and *tibia* is significantly stronger than *fibula* which is not connected to *femur*. The shank was modelled as a single *tibia* segment and *fibula* was omitted. The foot consists of more that twenty bones and was likewise modelled as a single segment.

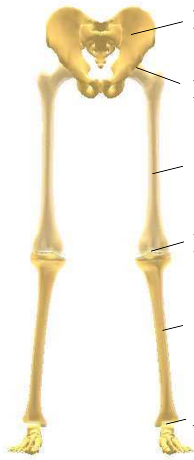


Figure 2.3: Segments and joints in the model

2.2.2 Joints

The model had a hip, a knee and an ankle joint in each leg. The hip joint was a spherical joint which inhibited translation but allowed rotation, i.e. three degrees of freedom. The ankle was a universal joint which allowed rotation about two perpendicular axes, i.e two degrees of freedom, plantar/dorsal flexion and inversion/eversion.
A real knee has two degrees of freedom. Its primary movement is flexion-extension and when the knee is bent it can rotate medially and laterally. It adds further complexity to the movement that the flexion-extension is a combination of roll and a slide inside the joint capsule. However, the joint can be approximated as an ideal hinge. I the model the knee was a revolute joint that only allowed rotation about one axis, i.e. one degree of freedom.
A mechanical joint provides the same number of kinematic constraints as reactions, because it is the reaction forces that enforce the kinematic constrains. This is not necessarily the case in a biomechanical joint. The movement of the knee can for many applications be approximated by a hinge joint, but the load-carrying mechanisms in the joint are far from a mechanical hinge. The reaction forces of the knee are related to a complicated interaction between muscles, ligaments and bones. The AnyBody Modelling System allows definition of a joint that only provides the kinematic constrains but not the reaction forces. This ability was utilised for simulating a knee without an anterior cruciate ligament (Section 2.2.6).

2.2.3 Muscles

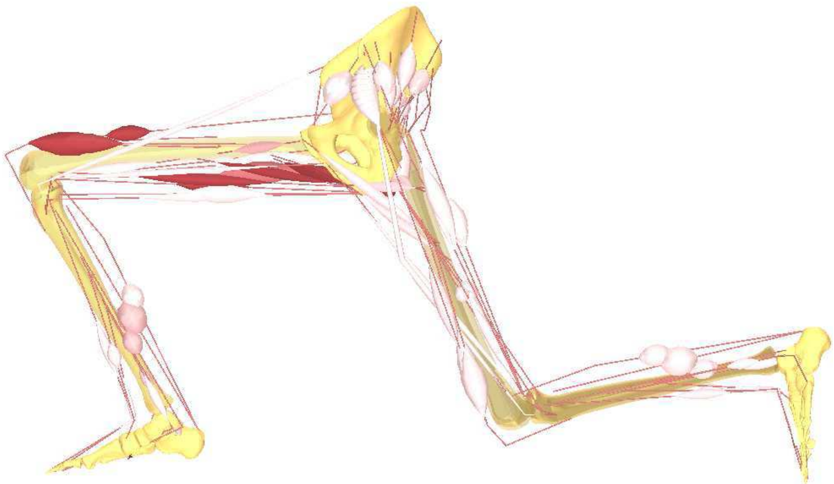


Figure 2.4: The Lunge Model

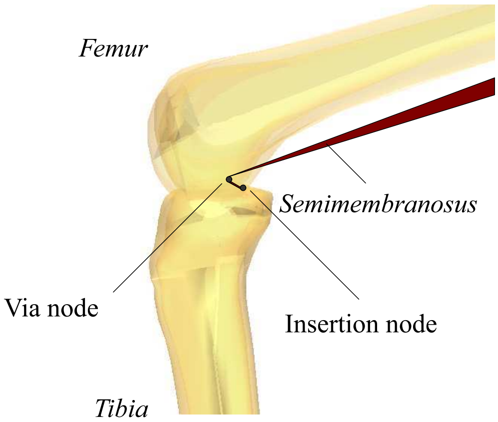
Each leg in the AnyBody model contained 27 different muscles, Figure 2.4. Some of the muscles had multiple insertions or origins and were therefore subdivided into more muscle-tendon units. *Gluteus maximus* was for example split up into gluteus maximus 1, gluteus maximus 2 and gluteus maximus 3. As a result each leg consisted of 35 muscle-tendon units. Note that several muscles were omitted. The lower extremities of the human body contain 38 different muscles. The muscles that were omitted in the model includes: *adductor brevis, inferior gemellus, obturator externus, obturator internus, pectieus, peroneus longus, peroneus tertius, plantaris, popliteus, quadratis femoris, superior gemellus.*

Excluded muscles

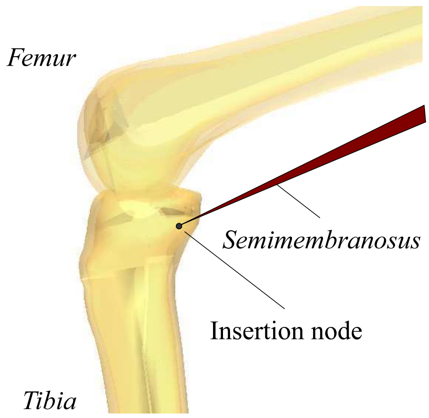
Popliteus' primary action is medial rotation of the knee. This muscle wouldn't have much effect in the model seeing that the knee was a hinge joint and therefore unable to rotate medially. *Plantaris* plantar flexes the ankle joint and flexes the knee. This muscle could be omitted because it has the same action as *gastrocnemius* which is significantly bigger. *Peroneus tertius* dorsi flexes and everts the foot, while *peroneus longus* plantar flexes and everts foot. The muscles included in the model were sufficient to simulate plantar and dorsal flexion of the ankle joint, but they couldn't support the sideways movements of the ankle joint. Therefore the model included an "Ankle Brace" that transferred the moment from the foot to the shank segment.
The actions of the other excluded muscles were all related to the hip joint. The selection of which muscles that were included and which that were omitted won't be discussed further because the selection process wasn't related to this project.

Relocation of muscles

An examination of the muscles in the model showed that some of the insertions nodes and via nodes[2] were misplaced and therefore had to be relocated. The most obvious example was the placement of the insertion point and via point of the *semimembranosus*, which is shown on Figure 2.5(a). The via point was in front of the insertion point which meant that the reaction force in the insertion point would pull the *tibia* forward, when it should be pulling backwards.



(a) Original insertion



(b) New insertion

Figure 2.5: Insertion of *semimbranosus*

The new insertion was found from a new data set (Horsman et al., 2007), which contained information about the location of joints and muscles insertions/origins and was considered to be more precise, than the data set used to create the original leg model. To use information from the new data set, two problems had to be solved. The data sets were obtained from two different cadaver studies, and because the cadavers did not have the same size it was necessary to determine a scaling factor in order to use data from the new set in the lunge model. The second issue that had to be solved was that there was no information regarding how the reference coordinate systems in the two data sets were located with respect to each other.

Scaling

The scaling factor between the two data sets was found by comparing the distances between the joints and muscle insertions/origins in one set with the same distances in the other data set. Unfortunately there was no consistency in the results as the scaling factor was ranging from 0,959 to 1,41. The location of the joints were considered to be the most accurate in both sets, so it was determined to use the distance from knee joint to ankle joint to find a scaling factor.

[2] A muscle's via node, is a point where the considered muscle is in contact with a segment and reaction forces are generated.

$$\text{Scale factor} = \frac{d}{d_{newset}} \tag{2.1}$$

$$\Downarrow \tag{2.2}$$

$$\text{Scale factor} = \frac{0,4600}{0,4098} = 1,222 \tag{2.3}$$

d is the distance from knee joint to ankle joint in the existing model

d_{newset} is the distance from knee joint to ankle joint in the new data set

Coordinate transformation

The positions of the muscle insertions/origins in the new data set (Horsman et al., 2007) were referring to a coordinate system located in the hip joint, while the existing joints and muscle insertions/origins on *tibia* were referring to a local *tibia* coordinate system. In order to use data from the new data set it was necessary to transform the position vectors from one coordinate system to another.

$$\vec{\mathbf{r}}_{tibia} = \vec{\mathbf{r}}_{tibhip} + \underline{\mathbf{A}} \cdot \vec{\mathbf{s}}_{hip} \tag{2.4}$$

$\vec{\mathbf{r}}_{tibia}$ is the position vector in the local *tibia* coordinate system

$\vec{\mathbf{r}}_{tibhip}$ is the positon vector from the local *tibia* coordinate system to the hip system

$\underline{\mathbf{A}}$ is the transformation matrix

$\vec{\mathbf{s}}_{hip}$ is the position vector in the hip coordinate system

The transformation matrix and the position vector, from the local *tibia* coordinate system to the hip coordinate system, gives twelve unknowns and needs twelve equations to be solved. Nine equations were obtained by selecting three points that is described in each coordinate system and apply the relation in equation (2.4). The remaining three equations were obtained by requiring that the three direction vectors that comprise the transformation matrix had to be unit vectors. The system of equation was solved in a MATLAB® program with a Newton-Raphson solver.

The inconsistent scaling made it difficult to solve the problem. Most points would make the system singular and impossible to solve, and the points that would allow a solution would result in a transformation matrix where the direction vectors where not strictly perpendicular to each other as they should be. Fortunately the matrix and position vector were good enough to determine a better location for *semimembranosus'* insertion node. The new data set had no via point for *semimembranosus* and it was therefore removed from the lunge model. In addition to this *biceps femoris caput breve* and *biceps femoris caput longum* had their vianodes removed and new insertions were defined.

The MatLab program, which can be found on the enclosed CD, contains detailed information about the system of equations and how it was solved and examined.

Muscle model

The AnyBody modelling system has three different muscle models readily available. The models differ in complexity and how accurately they represent real muscles. The Any-MuscleModel3E was chosen because it is the most accurate. The muscle model was based on the classical work by A.V. Hill and consists of three elements: A contractile element representing the active properties of the muscle fibres, a parallel-elastic element representing the passive stiffness of the muscle and a serial-elastic element representing the elasticity of the tendon, see Figure 2.6. As a result this model takes elasticity of the tendon and passive elasticity of the muscle into account.

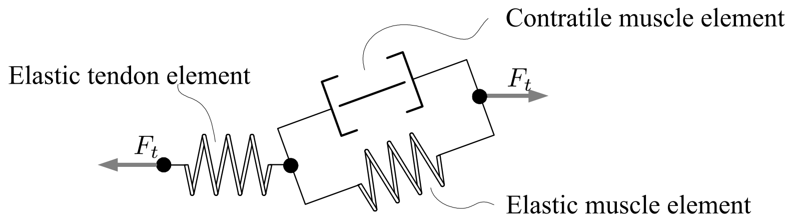


Figure 2.6: The chosen muscle model: AnyMuscleModel3E. F_t is the force in the tendon.

2.2.4 Movement

The movement of the model can be imposed by drivers, which each determine a kinematic measure. A kinematic measure is a way of describing dimensions in a kinematic model; this could be the position of a point, a joint angle or a distance between two selected points. In order to drive a complicated model accurately, the user must know the motion in detail.

The movement pattern of the forward lunge was measured with optical marker trajectories on a healthy female subject and recorded by a motion capture system. The subject was instructed to stand in an upright position in front of a force plate and perform the forward lunge by talking one step forward, placing the foot on the force plate, flexing the knee to approximately $90°$ and subsequently extending the knee and pushing herself back into the upright standing position. The MOCAP[3] data was recorded by Alkjær et al. and was utilised to drive the model by Andersen.

The movement sequence was divided into 41 time steps. In the first time step, when the right foot touched the ground, i.e. the force plate, the right knee was flexed $39,5°$. At the 23rd time step the knee had a maximum flexion of $116°$ and started to extend. Figure 2.7 illustrates the knee flexion as a function of time, measured at each time step.

Each segment in the model had a reference coordinate system. For both tibia and femur it applied that the x-axis was directed anteriorly and the y-axis was parallel with the segment's longitudinal axis, proximally directed. The knee joint angle was measured as

[3]Motion capture

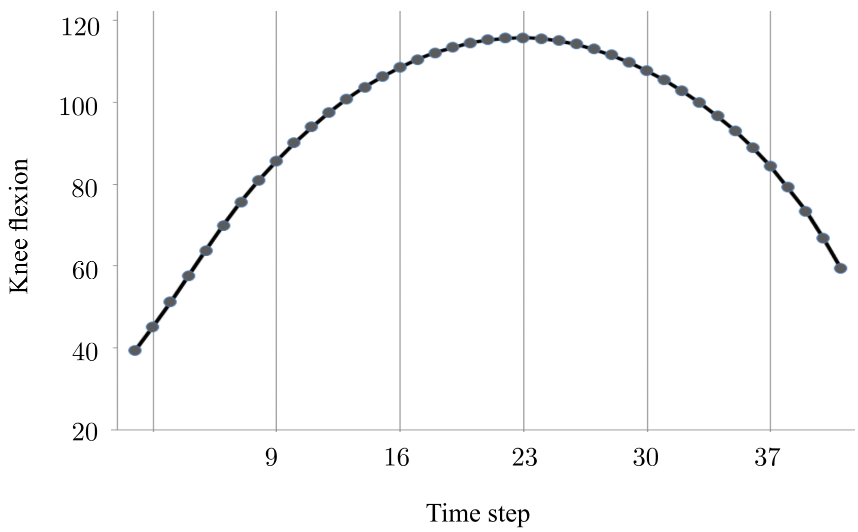


Figure 2.7: Flexion of the right knee during a forward lunge. Five selected time steps is emphasised.

the angle between the y-axes (clockwise from femur's to tibia's y-axis). Positive angles means that the knee is flexed, $0°$ means that the knee is fully extended, while negative angles indicate hyper extension.

2.2.5 External Forces

The ground reaction forces were recorded simultaneously with the measurement of the movement, by instructing the test subject to place the foot on a force plate.

2.2.6 Removing the cruciate ligament

It is possible to define a joint that only provides the kinematic constrains but not the reaction forces. Figure 2.8 shows the *tibia* and its local reference coordinate system. When the two joint surfaces in a real knee are sliding with respect to each other the movement follows the tibial plateau, which approximately is parallel to the xz-plane of the *tibia's* local coordinate system. It is this forward and backward sliding motion that the cruciate ligaments restrict.

An anterior cruciate deficient knee was imitated by aligning the joint coordinate system with the *tibia's* coordinate system, and removing the knee reaction force in the x-direction. The forces that were transmitted by the cruciate ligaments in the healthy knee must consequently be transmitted trough the muscles. However, the model should still be able to transmit forces through the posterior cruciate ligament. The posterior ligament was included with a unilateral constraint inside the knee joint which acts in the positive x-direction of the *tibia's* local coordinate system, i.e. the ligament could only pull the tibia

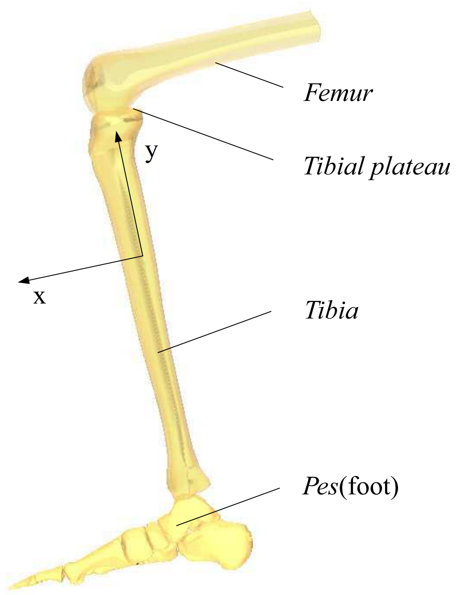


Figure 2.8: Tibia

forward.

2.3 Analysis of the lunge model

The lunge model was analysed with and without the anterior cruciate ligament. The dynamic analysis calculated the forces in the tendon-muscles units and the reactions in the joints. Every single force on the tibia from the muscles and joints was transformed into the *tibia's* local coordinate system and compared. The diagram on Figure 2.9 treats a model with a healthy knee and provides an overview of which muscles and joint reactions that pulled the tibia in an anterior or posterior direction. The moments around the ankle joint is illustrated in Figure 2.10. Similar calculations were made for a lunge model with a ligament deficient knee. The analysis revealed that there was no difference in muscle activity or joint reactions between the two models.
Additional results from the dynamic analysis is in Appendix E.

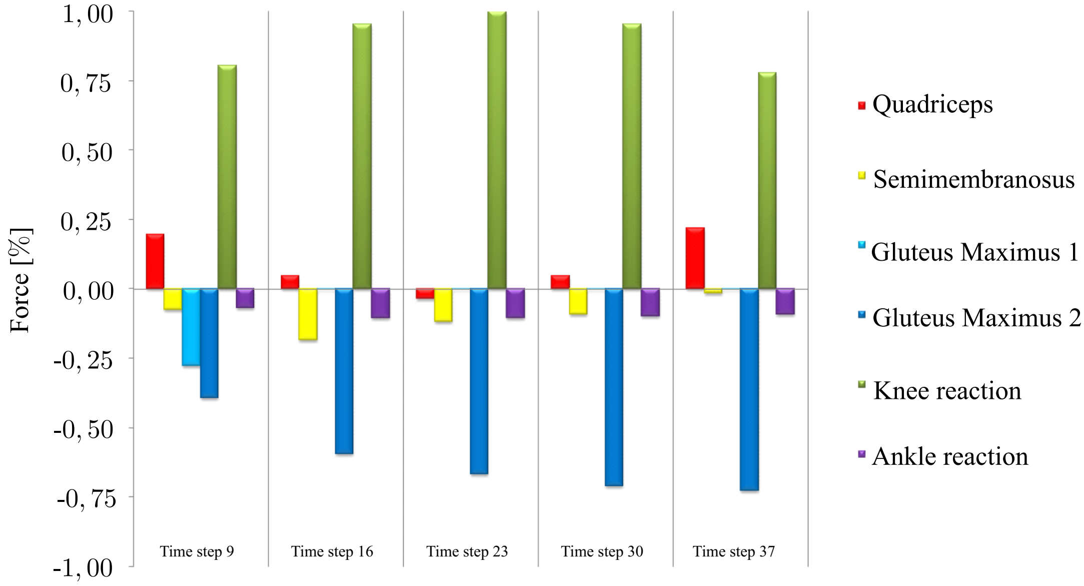


Figure 2.9: Reaction forces on the tibia in five selected time steps. The knee had reached maximum flexion in the 23rd time step. A positive force value indicates an anterior drag on the tibia, while a negative value indicates a posterior drag. Note that the force is given as a fraction of the total anterior (or posterior) drag. For instance in time step 16, *gluteus maximus 2* was responsible for approximately 60% of the posterior drag in tibia which corresponds to -0,6 on the graph. Muscle-tendon units which where responsible for less than 10% of the total anterior (or posterior) drag in all time steps were omitted from the graph. They were, however, included in the calculation of the total drag. The five selected time steps is shown in Figure 2.7. The results were similar for healthy and anterior cruciate ligament deficient knees.

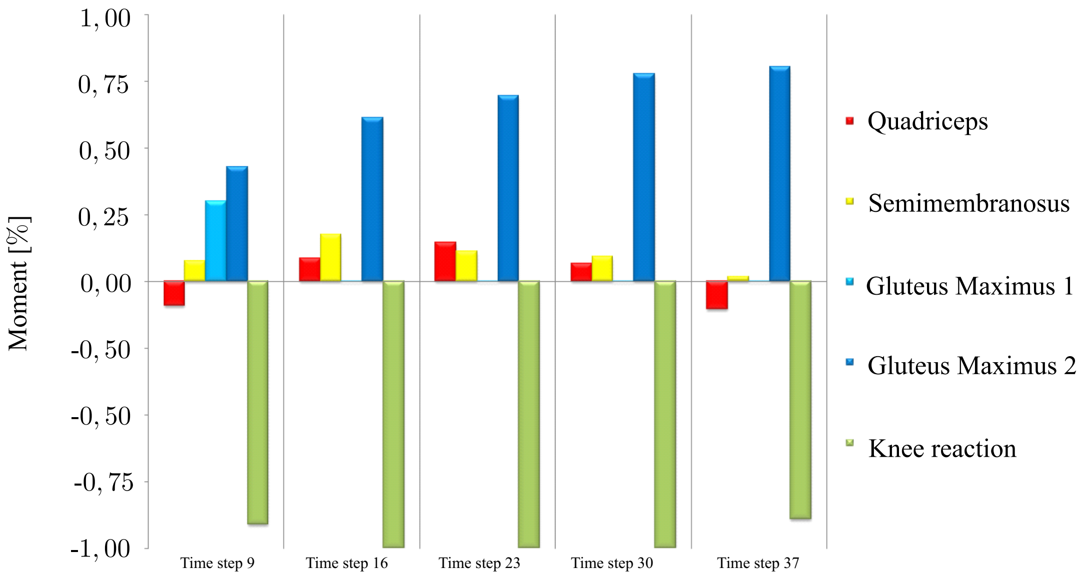


Figure 2.10: Moment around the ankle joint from the reactions forces on tibia in five selected time steps. The knee had reached maximum flexion in the 23rd time step. A negative moment indicates that the proximal *tibia* is pulled in an anterior direction, while a positive value indicates a posterior drag in the proximal *tibia*. Note that the moment is given as a fraction of the total positive (or negative) moment. For instance in time step 37, *quadriceps* was responsible for approximately 10% of the negative moment (which pulled the proximal *tibia* in an anterior direction) this corresponds to -0,1 on the graph. Muscle-tendon units which where responsible for less than 10% of the total positive (or negative) moment in all time steps were omitted from the graph. They were, however, included in the calculation of the total moment. The results were similar for healthy and anterior cruciate ligament deficient knees.

2.4 Discussion of the lunge model

It had been expected that there would be a significant difference between the two models, seeing that studies have shown that the anterior cruciate ligament deficient subjects perform a forward lunge differently from healthy subjects.
The similar results can be explained from Figures 2.9 and 2.10. They show that the reaction in the knee creates an anterior pull in the proximal tibia, i.e. there is no strain in the anterior cruciate ligament.
This is in agreement with Beynnon and Flemming (1998), who measured strain behaviour of a normal anterior cruciate ligament by arthroscopic implantation of a differential variable reluctance transducer in the ligament while subjects were under local anaesthesia.
The strain in the anterior cruciate ligament was measured at four different knee ankles: $15°, 30°, 60°$ and $90°$. In comparison to a condition with the muscles completely relaxed, isomeric contraction of quadriceps produced a significant increase in strain in the ligament at a knee flexion of $15°$ and $30°$, and no changes at $60°$ and $90°$. Isometric contraction of

the hamstrings didn't produce a significant change at any of the four selected knee ankles. Simultaneous contraction of quadriceps and hamstrings created a significant increase in the ligament strain at 15°, at no change at 30°, 60° and 90° of flexion.
Beynnon and Flemming also measured the strain in the anterior cruciate ligament at active flexion/extension of the knee, in the interval $10° - 90°$. The ligament strain values were larger during the flexion phase than during knee extension. See Figure 2.11.

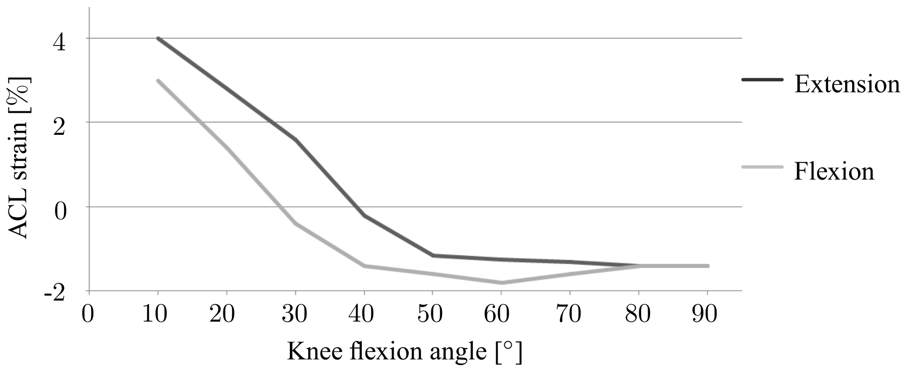
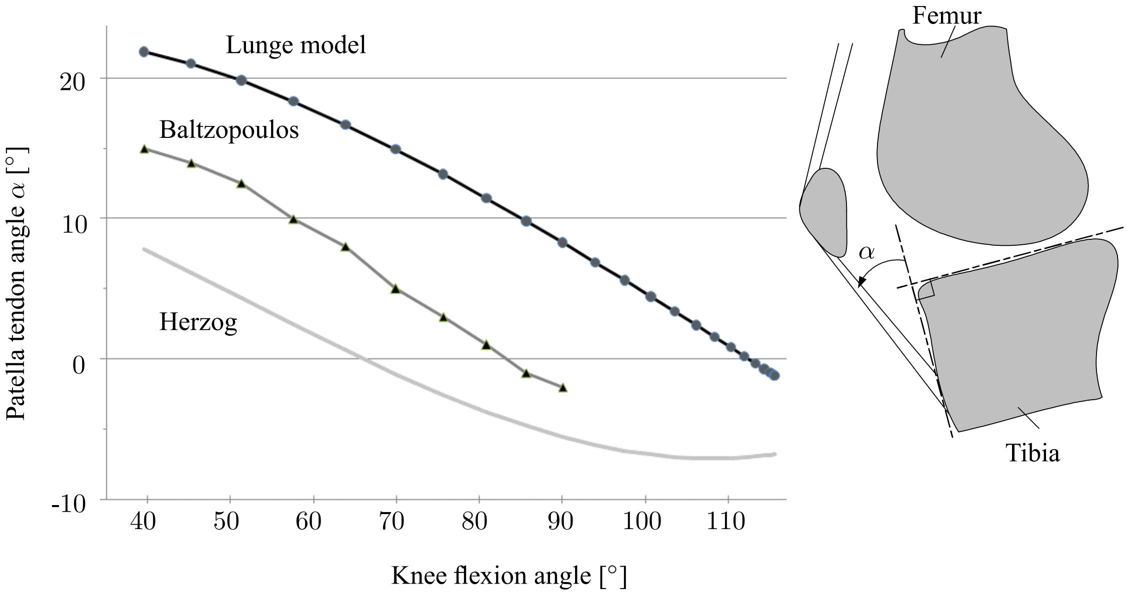


Figure 2.11: Active flexion-extension of the knee.
Replication of graph (Beynnon and Flemming, 1998, page 522, Fig. 4)

Extension of the knee by contraction of the quadriceps did not increase the strain in the ligament when the knee angle was between approximately 27° and 90°. Knee flexion did not increase the strain in the ligament when the knee angle was between approximately 38° and 90°. Unfortunately the paper from Beynnon and Flemming (1998) does not describe the effects of increased weight. One might expect that increased load will not only increase the strain in the anterior cruciate ligament, but also influence the angle interval where the ligament is unstrained.

As described in Section 1.2 the quadriceps is normally considered to produce an anterior drag, however in Figure 2.9 it can be seen that the force on the tibia from the quadriceps had a posterior direction in the 23rd time step. This was never the less in agreement with Baltzopoulos (1995), who measured the patella tendon angle during knee extension using videoflouroscopy in vivo: "for knee flexion angles between 0 and 90 degrees, knee extensor activity translates the tibia in an anterior direction relative to femur. For knee flexion angles above 90 degrees, however, a posterior translation of the tibia by the knee extensors is indicated."
Herzog and Read (1993) conducted a cadaver study to identify the lines of action, in the sagittal plane, of the major force-carrying structures crossing the human knee. The lines of action were determined as a function of the knee joint angle and were expressed using polynomial regression equations. The patella tendon angle varied a lot between the five cadavers. The knee joint angle where the patella tendon is perpendicular to the tibial plateau, i.e. the angle of transition between anterior-posterior drag from the quadriceps, was in the interval $60° - 90°$.
The polynomial regression equation for the patella tendon angle as a function of knee

joint angle can be seen in Figure 2.12(a) together with the patella tendon angle measured by Baltzopoulos (1995) and the angle in the lunge model. The patella tendon angle was given as the smallest angle between the tendon and a line perpendicular to the tibial plateau, Figure 2.12(b).



(a) Patella tendon angle as a function of knee flexion (b) Patella tendon angle α

Figure 2.12: Patella tendon angles from the lunge model compared with results from Baltzopoulos (1995) and Herzog and Read (1993). A positive angle indicates an anterior drag in tibia, while a negative angle indicates a posterior drag. The patella tendon angles, determined with videofluoroscopy, was read from a graph (Baltzopoulos, 1995, Figur 7, page 90), while the polynomial regression equation was given in (Herzog and Read, 1993, Table 1, page 220). In the lunge model the patella tendon angle was given as the smallest angle between the tendon and a line perpendicular to the tibial plateau Figure 2.12(b). The two studies used two different references when measuring the tendon angle, so the angles had to be translated in order to be able to compare them, see Appendix D

Baltzopoulos (1995) was considered to be more accurate seeing that theses results were obtained from in vivo studies while the results from Herzog and Read (1993) were obtain from cadaver studies. No matter which experiment was more accurate it was clear that the orientation of patella tendon in the lunge model is more anterior than a real knee. Therefore it was deduced that the anterior drag on the tibia from the quadriceps is not larger than the lunge model predicted.
Seeing that it is unknown how the body distributes the forces between the redundant muscles, one might argue that the muscle recruitment could be wrong. Nevertheless the force in the in quadriceps must be sufficiently accurate because quadriceps is the only knee extensor and the only muscle that has the potential to pull the proximal tibia forward.

It seemed unlikely that the drag from the quadriceps will increase the strain in the anterior cruciate ligament during a forward lunge and it was chosen to investigate the ligament strain at smaller knee flexion with a running model, which is described in Chapter 3.

CHAPTER 3

The Runner Model

Sagittal plane mechanisms have been proposed to be able to tear the anterior cruciate ligament (DeMorat et al., 2004). Such postulates can be based on the fact that sports movements like running and side-cutting[1] give rise to large quadriceps forces at relatively small knee flexion angles; a combination known to induce anterior force on the proximal tibia.
One of the most intense and physical demanding sagittal plane movements is probably sprint; because it requires very high muscle activity to gain maximum forward propulsion. This chapter describes an attempt to investigate to what extend, the intense muscle activity during sprint, put strain on the anterior cruciate ligament. The study was based on an AnyBody model of the lower extremities; driven by motion capture data from a male sprinter. The joint reactions and the forces in the muscles around the knee were quantified and it was discussed whether this kind of sagittal plane movement has the potential to tear the ligament.

3.1 Model structure

3.1.1 Segments, joints and muscles

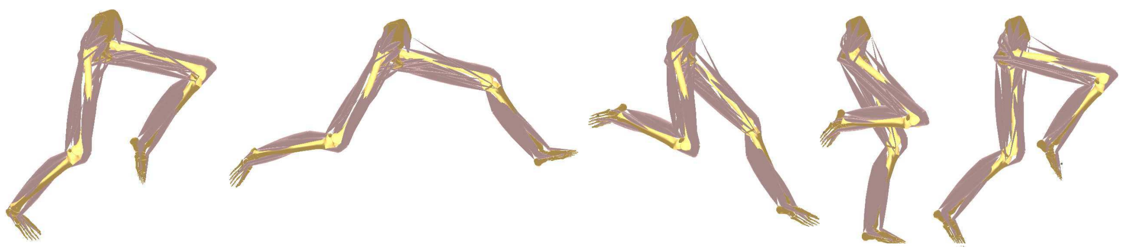
Just as the lunge model, the runner was based on a model of the lower extremities of the human body. The segments and joints were equivalent to those of the lunge model, but the muscles were slightly different. The insertions of *semimembranosus*, *biceps femoris caput breve* and *biceps femoris caput longum* were modified and their via nodes were removed like in the lunge model, but it was necessary to choose a different muscle model. The AnyBody Modelling System has three different muscle models readily available. At first it was chosen to apply the muscle model AnyMuscleModel3E, which was used in the lunge model. The strength of this model is that it takes the muscle's immediate length and the contraction velocity into account, but the data defining the properties of muscle model

[1]Side-cutting is a feint which is performed in sports like team handball, basketball and American football. The purpose of the side-cutting manoeuvre is to pass an opponent by faking the direction opposite to the intended movement. Normally, a right-handed shooter will approach the opponent head on, brake the forward movement with the left foot and step to the right side.

originated from an average male and not a top athlete like a sprinter. When the model was driven to sprint like a top athlete, the muscle activity showed that the muscles were unable to perform this fast movement. Rather than manufacturing properties for a top sprinter, it was chosen to apply the more simple muscle model, AnyMuscleModel, which is independent of muscle length and contraction velocity. The properties of AnyMuscleModel are based on the muscles' isometric strength[2].

3.1.2 Movement and external forces

Generally, running is a sequence of strides and each stride can be divided into three phases: support, drive, and recovery. The support phase starts when the foot touches the ground. During this phase, the supporting foot is slightly ahead of the point that lies directly below the body's centre of gravity. The knee joint is at its largest extension prior to the support phase and begins to flex just before contact is made with the ground. The knee continues to flex, while the supporting hip extends, and as the centre of gravity passes above the supporting foot, the drive phase begins. During the drive phase the knee joint and the hip of the supporting leg extends, so that the toes maintain contact with the ground as the leg trails behind the body. The foot pushes backward and down. The recovery phase begins when the driving foot loses contact with the ground. The hip flexes and drives the thigh forward. When the hip reaches maximal flexion, the lower leg rapidly unfolds and the knee joint is almost fully extended. The movement continues back into the support phase when the foot touches the ground.
The sprint movement was measured with optical marker trajectories on a male runner and recorded with a motion capture system. The movement sequence lasted less than one third of a second and was divided into 180 time steps. The sequence is illustrated and described in Figure 3.1.



(a) Drive (b) Recovery phase (c) Support and drive

Figure 3.1: The sprinter's movement sequence. The sprinter's left leg was in the recovery phase when the movement sequence started, while the right leg was in the drive phase (a). Subsequently, the runner leaped through the air and both legs were in recovery (b). The left leg entered the support phase as the foot touched a force plate and continued into drive (c). The sequence ended as the runner once again leaped through the air and both legs were in recovery. Note that the movement sequence was not a complete cycle.

[2]The static force that a muscle can exert at its optimum length

The ground reaction forces were recorded simultaneously with the measurement of the movement, but only when the left foot touched the ground, and the analysis of the model was therefore reduced to the left leg. Note that the remaining part of the chapter only treats the left leg, i.e. the denomination 'leg' is referring to the left leg unless otherwise stated.

The knee flexion as a function of time step is shown in Figure 3.2.

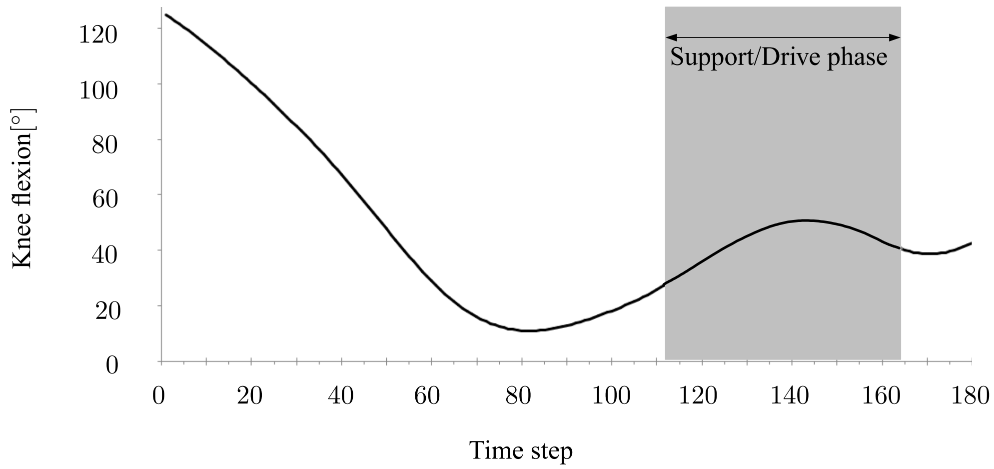


Figure 3.2: Knee flexion as a function of the time. The grey square marks the duration of the support and drive phase from the 112th - 164th time step, when the left foot was in contact with the forceplate.

3.2 Analysis of the runner model

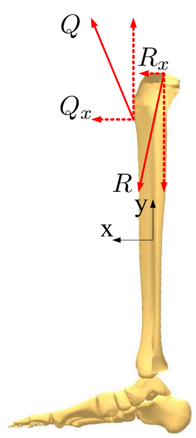


Figure 3.3: Tibia's reference coordinate system. Q is the drag from quadriceps. R is the knee joint reaction.

The dynamic analysis calculated the forces in the tendon-muscles units and the reactions in the joints. Every single force on the tibia from the muscles and joints was transformed into the tibia's local reference coordinate system, Figure 3.3. It was expected that running would give rise to large quadriceps force and strain the anterior cruciate ligament. Figure 3.4 shows the variation of the extensor moment in the knee joint (M_z), the anterior drag from the quadriceps on the tibia (Q_x) and the anterior knee joint reaction (R_x).

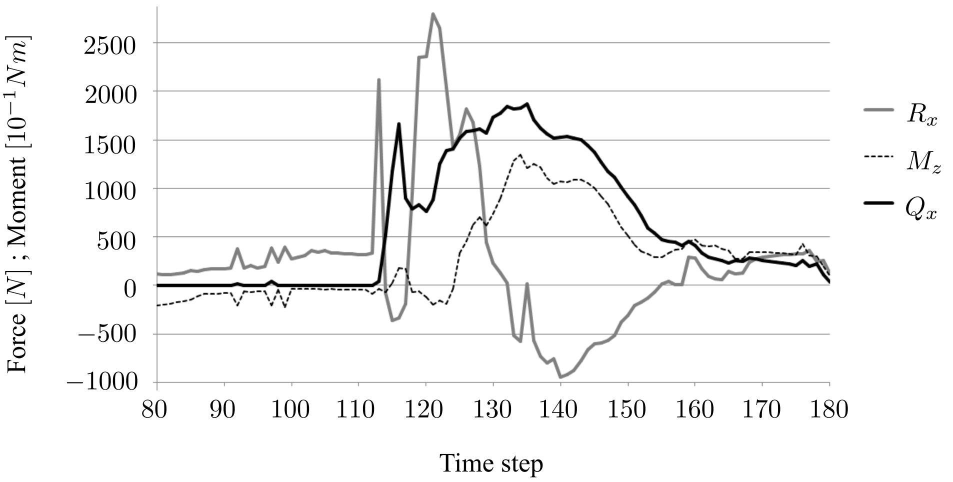


Figure 3.4: Knee moment M_z, quadriceps drag Q_x and joint reaction R_x as a function of time. A positive and negative moment indicates extensor and flexor moment respectively. However, the sign of the moment does not necessarily indicate the direction of movement, and accordingly an extensor-dominated moment, observed during flexion, indicates an eccentric contraction of quadriceps. Note that the moment is scaled on the graph. Q_x is the component of the drag from the quadriceps that is parallel with the tibia's local x-axis. When Q_x is positive the quadriceps pulls the tibia forward. The joint reaction, R_x, is the component of the total reaction force that is parallel with the tibia's local x-axis. Positive values indicate that the posterior cruciate ligament is strained, while negative values indicate that the anterior cruciate ligament might be strained. The graph only shows the time steps 80 -180 because the anterior cruciate ligament was unstrained in the first half of the movement sequence.

The quadriceps force increased significantly when the foot touched the ground (112th-164th time step) and the quadriceps' pull in the tibia was directed anteriorly. Negative knee joint reactions (R_x), which indicated that the anterior cruciate ligament might be strained, occurred twice during the movement sequence. In the 114th-117th time step, which belonged to the support phase, the minimum reaction force was -$359N$ in step 115. In the 133rd-154th time step, which belonged to the drive phase, the minimum reaction force was -$943N$ in step 140. The intervals where the ligament might be strained are shown in Figure 3.5.

The diagram on Figure 3.6 provides an overview of which muscles and joint reactions pull the tibia in an anterior or posterior direction.

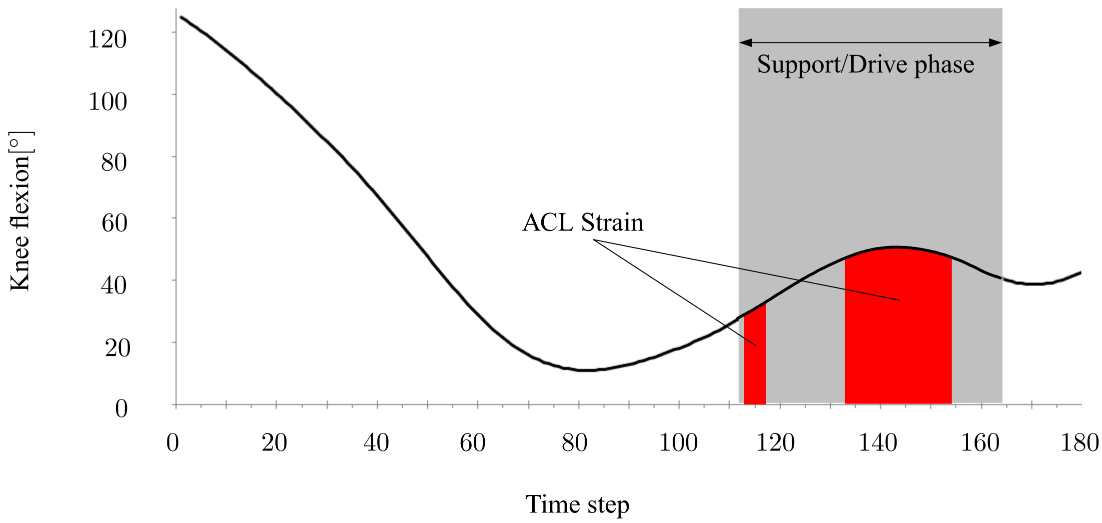


Figure 3.5: Knee flexion as a function of time step. The time steps, where the anterior cruciate ligament might be strained, is emphasised.

3.3 Discussion

In order to evaluate to what extent the anterior cruciate ligament was strained during the sprint, one must consider that it is not the anterior cruciate ligament alone that restricts the relative movements between femur and tibia. According to Butler et al. (1980), the anterior cruciate ligament on average provides 86 % [3] of the total capsular and ligamentous resistance. It should also be taken into account that the way the ligament is constrained in a real knee is quite different from the test conditions when the strength of the ligament is determined. In a real knee the ligament is not aligned with the direction of the knee shear force, R_x, but is wrapped around the surfaces of the bones. However, the strength of the ligament is usually determined by stretching the ligament along its longitudinal axis in a tensile testing machine until failure. Studies have shown that the tensile strength of the ligament is within the range $1725 - 2195N$ [4]. The strength depends on the subject's age, gender and activity level and it can therefore be anticipated that the sprinter's ligament strength is relatively high. The test result is also effected by measurement techniques, specimen orientation and loading rate.

The anterior cruciate ligament strain was evaluated on basis of the size and direction of the knee joint reaction force calculated in the dynamic analysis. The component of the reaction force parallel to the tibia's y-axis was compressive; hence it did not strain the ligament. As mentioned previously, the component parallel to the tibia's x-axis, R_x, was negative twice during the movement sequence, corresponding to a strained ligament.
The force in the anterior cruciate ligament would be slightly bigger than $|R_x|$ due to the

[3] 85% at $90°$ knee flexion and 87% at $30°$ knee flexion.

[4] Woo et al. (1991): $2160N$; Noyes and Grood (1976): $1730N$; S.Karmani and Ember (2003): $1725 - 2195N$; Chandrashekara et al. (2006): $1818N$ (male) , $1266N$ (female), $1526N$ (male and female)

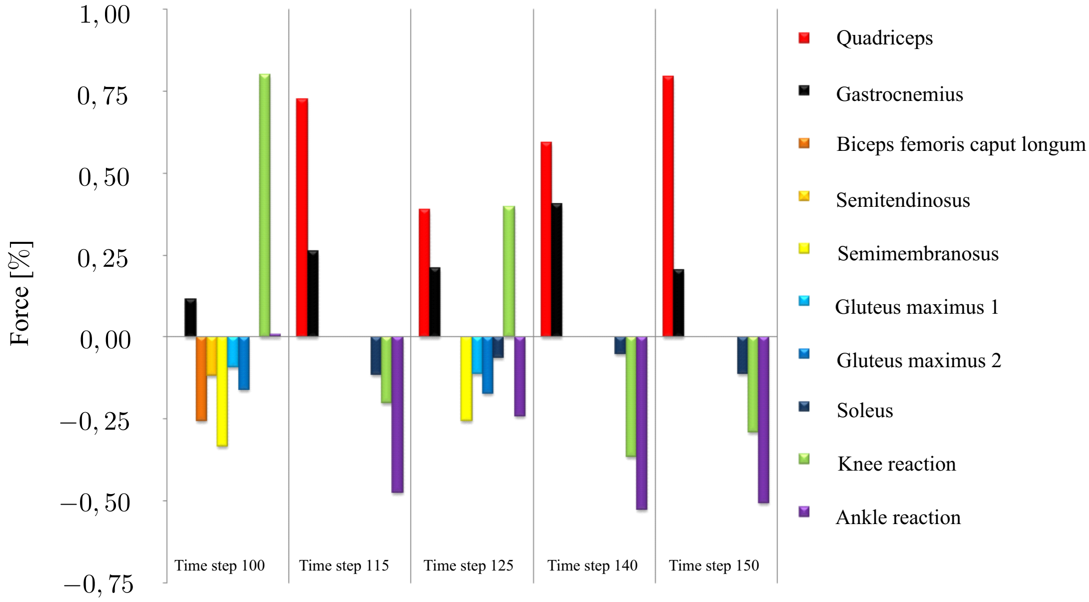


Figure 3.6: Reaction forces on tibia in five selected time steps.
In the 100th time step the leg is in the recovery phase. In the 115th time step the foot touches the ground and is in front of the body's centre of gravity, i.e. the leg is in the support phase. The 125th time step is a transition between support and drive. The 140th and 150th time step is in the drive phase.
A positive force value indicates an anterior drag in tibia, while a negative value indicates a posterior drag. Note that the force is given as a fraction of the total anterior (or posterior) drag. For instance in time step 100, *semimembranosus* was responsible for approximately 30% of the posterior drag in tibia, which corresponds to -0, 3 on the graph. Muscle-tendon units, which where responsible for less than 10% of the total anterior (or posterior) drag in all time steps, were omitted from the graph. They were, however, included in the calculation of the total drag.

ligament's orientation, provided that the ligament alone had been responsible for resistance of anterior translation of tibia. However, because of the joint capsule's and the other knee ligament's ability to resist anterior translation of the tibia, it was estimated that the anterior cruciate ligament would not tear as long as the magnitude of the posterior directed joint reaction R_x was less than the ultimate tensile strength of the ligament determined in a tensile testing machine.

$$R_x > 0 \quad \Rightarrow \text{No strain}$$
$$R_x < 0 \ , \ |R_x| < \text{Ultimate tensile strength} \Rightarrow \text{No tear}$$

Since the minimum value of R_x was -$943N$ compared with tensile strength of $1725 - 2195N$, it seemed unlikely that running would tear the anterior cruciate.

McLean et al. (2004) stated that sagittal plane loading mechanisms during sports activities are accompanied by a large ground reaction force and suggested that the posteriorly directed ground reaction, which is transferred to the tibia, helps to protect the ligament. The connection between ground reaction and knee loading will be treated in Chapter 4.

Considering that sprint probably is one of the most intense sagittal plane sports movements, it appears that voluntary contraction is insufficient to injure a healthy cruciate ligament. A rupture of the ligament would require that the athlete was predisposed for the injury or was subject to exterior factors, which could force the joint into an injuring position.
Factors such as gender, age, history of previous injuries, physical fitness, anatomy and body composition may influence the risk of sustaining injuries, and predisposing the athlete to injury. There is a significant gender disparity in non-contact injuries. Female athletes are reported to suffer anterior cruciate ligament injuries 2-7 times more often than male athletes in sports such as team handball, football and basketball (McLean et al., 2005), (Hewett et al., 2006b), (Hewett et al., 2006a), (Bahr and Krosshaug, 2005). Advancing age also increase the risk of ligament injuries, because ageing leads to a significant reduction in the ligament's strength and stiffness properties(Noyes and Grood, 1976). Sports equipment and environment may also affect the injury risk and make the athlete even more susceptible. Olsen et al. (2004) provided a video analysis of injury mechanisms for anterior cruciate ligament injuries in team handball and found that the injury risk is higher on synthetic floors (generally having a higher friction) than on wooden floors.

Even though intense voluntary contraction might be insufficient to injure the anterior cruciate ligament during running, the analysis does not rule out the possibility that other sagittal plane movements may put more strain on the ligament. Therefore, it was chosen to investigate various feasible sagittal plane movements in Chapter 4.

CHAPTER 4

Sagittal Model

This chapter treats a model of a single human leg. The model was used to study various sagittal plane movements in order to determine whether these mechanisms have the potential to rupture the anterior cruciate ligament. The sagittal plane movements were imitated by systematically varying the ground reaction force and the flexion angles of the hip joint, ankle joint and knee joint. The magnitude of R_x, that is the component of the total knee joint reaction parallel with the tibia's local x-axis, was used to determine whether the ligament would rupture or not. It was stipulated that the anterior cruciate ligament would tear if the magnitude of the posterior directed joint reaction R_x exceeded the ultimate tensile strength of the ligament.

4.1 Creating the model

The underlying basis for the sagittal model was a one-legged model from the AnyBody Model Repository. The model's four segments, pelvis, thigh (femur), shank (tibia) and foot, were defined and joined similar to the segments and joints in the runner and the lunge model. The analysed positions and external forces will be described in the following. Note that the sagittal model was a right leg and all illustrations in this chapter refer to the right leg.

(a) (b) (c)

Figure 4.1: The sagittal model in three random postures.

4.1.1 Movement

Performing a thorough analysis of every conceivable sagittal plane movement was considered infeasible in terms of time consumption. As an alternative it was chosen to study various static leg positions. The flexion angles of the hip joint, ankle joint and knee joint were varied systematically to analyse a representative section of conceivable leg positions during sagittal plane sports movements. In other words, instead of analysing different movement sequences, it was chosen to analyse different sequences of various static leg positions. The hip joint flexion angle and the knee joint flexion angle were varied from $0° - 80°$; the ankle joint dorsal/plantar flexion angle was varied from -30° $-$ 30° (Figure 4.2).

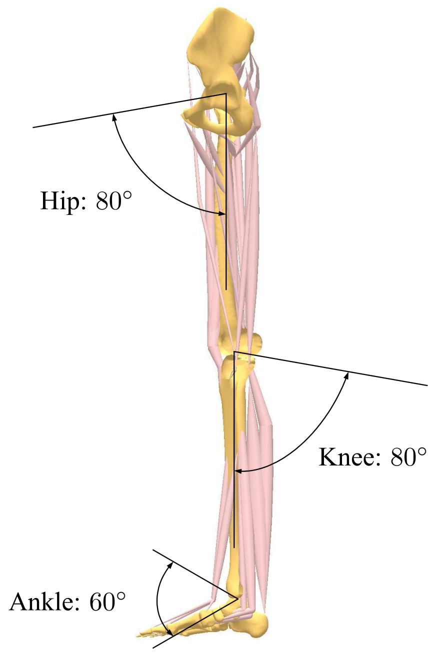
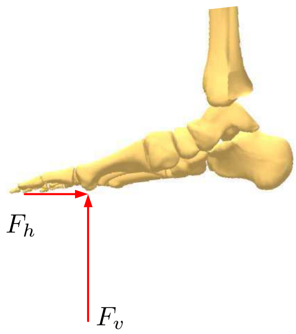


Figure 4.2: The three joints' range of motion. The flexion angle of the hip joint was varied systematically from $0° - 80°$. The flexion angle of the knee joint was also varied within the range $0° - 80°$, but the hip flexion angle was the controlling parameter. In some sequences the hip and the knee flexion was synchronous, in other sequences the knee flexion was up 30° smaller or larger than the hip flexion.

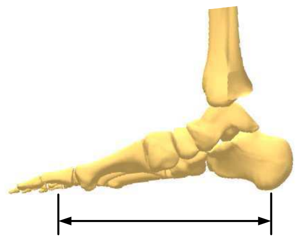
4.1.2 External forces

The only external force was the ground reaction on the foot. The ground reaction force was divided into a vertical and horizontal component (Figure 4.3(a)). During level walking the vertical component is upwards of 100% of the body weight and the horizontal component is $10 - 20\%$ of the body weight. Running increases the vertical component to $200 - 300\%$ of the body weight, while the horizontal component only is about 25%.

However, intense deceleration can increase the horizontal component to approximately 150% of the body weight.
The point of action might be on the ball of the foot, on the heel or anywhere in between, depending on the specific movement. Therefore, the point of action was tentatively positioned from the heel to the ball of the foot for all leg postures, see Figure 4.3(b).



(a) Vertical and horizontal components of the ground reaction force. (Global coordinates)



(b) The point of action of the ground reaction force was moved from the heel to the ball of the foot

Figure 4.3: Ground reaction force

By changing the components of the ground reaction forces it was attempted to answer the following questions:

- What is the knee joint reaction R_x, when the ground reaction lies within typical values?
- Does increased ground reaction force have the potential to tear the anterior cruciate ligament?
- Does the horizontal component of the ground reaction help to protect the anterior cruciate ligament? It might appear straightforward that posteriorly directed forces counteract quadriceps' drag in the proximal tibia and hence protects the ligament, but posteriorly directed forces induce larger quadriceps force due to increased knee moment.

4.2 Analysis

It was observed that some combinations of leg position and ground reaction force would require muscle activation beyond 100%, i.e. the muscles were not strong enough to hold the static position. Excessive muscle activation occurred for instance when the point of action was under the ball of the foot, the vertical component was $200 - 300\%$ of the body weight and the horizontal component was $0 - 10\%$. This ground reaction induced a large moment at the ankle joint which had to be balanced by the plantar flexors[1]. In some leg postures the muscle activity of gastrocnemiums exceeded 400% and gastrocnemius' via-node on the tibia would experience an anterior directed push from the muscle that would never take place in the human body. As a result all combinations of leg positions and ground reaction forces, which led to muscle activity beyond 100% were omitted in the calculations of the knee joint reaction force R_x. Figure 4.4 gives an example of the consequence of only including the situations with a plausible muscle activity.

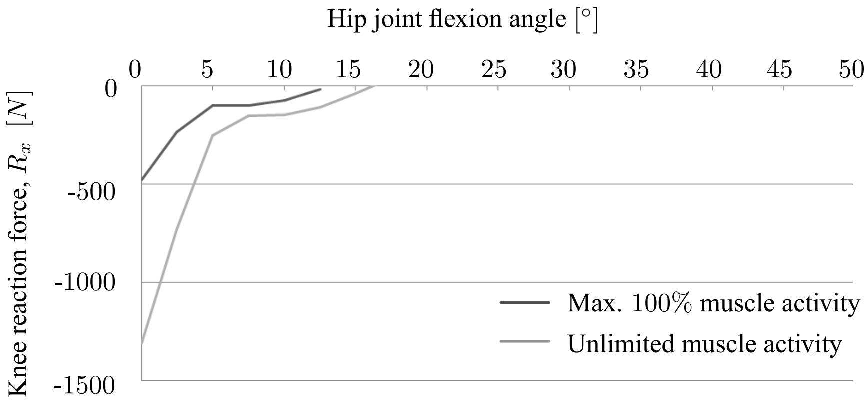


Figure 4.4: The knee joint reaction, R_x, as a function of the flexion angle of the hip joint. In this sequence the knee and hip joint flexion was synchronous and the ankle joint was fixed in a neutral position, i.e. the longitudinal axis of the tibia was vertical like in Figure 4.1(a). The ground reaction force was vertical, $F_v = 1473N$ and $F_h = 0N$. Every leg position was combined with points of action running from the heel to the ball of the foot. The combination which gave the highest reaction in the knee joint was used to produce the curves. The black curve comprises the combinations where the muscle activity did not exceed 100%, while the light grey curve comprises all combinations regardless of muscle activity.

McLean et al. (2004) stated that sagittal plane loading mechanisms during sports activities are accompanied by a large ground reaction force and suggested that the posteriorly directed ground reaction, which is transferred to the tibia, helps to protect the ligament. This postulate was probed by applying a ground reaction to the foot in the sagittal model,

[1] *musculus tibialis posterior, musculus flexsor hallucis, musculus flexsor digitorium longus, musculus gastrocnemius, musculus soleus, musculus plantaris, musculus peroneus longus* and *musculus peroneus brevis*

retaining magnitude of the vertical component and tentatively changing the magnitude of the horizontal component.
Figure 4.5 shows an example where the vertical component was $700N$ and the horizontal component was respectively $0, 50, 100, 150$ and $200N$.

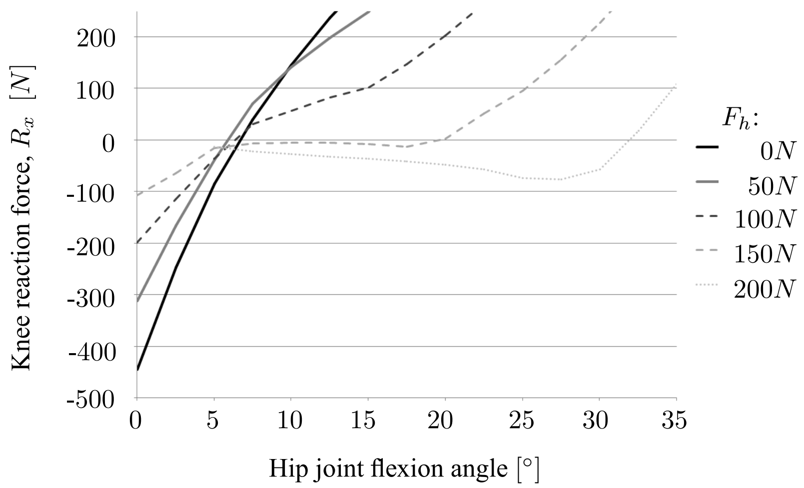


Figure 4.5: The knee joint reaction, R_x, was affected by direction of the ground reaction force, F. In this example the hip and knee joint flexion angle was synchronous and the angle joint was fixed in a neutral position. The point of action was fixed between the heel and the ball of the foot. The vertical component was $F_v = 700N$ and the horizontal component, F_h was $0, 50, 100, 150$ and $200N$ respectively, corresponding to the five different curves. For flexion angels smaller than $5°$, it was an advantage with an increased horizontal component. For knee flexion angles above $5°$, it was a disadvantage when the horizontal component was increased from $150N$ till $200N$

Similar simulations were made with different leg positions, increased ground reaction force and various points of action. The shape of the curves looked different from the curves in Figure 4.5, but the interpretation was the same: In some leg positions it is an advantage if the horizontal component is increased, in other positions it is a disadvantage.

4.2.1 Knee reaction force at typical ground reactions

All leg postures were subjected to 12 different ground reaction forces. The combinations of horizontal and vertical components are outlined in Table 4.1. In addition to this, each ground reaction force had 18 different points of application, i.e. each leg posture was subject to 216 different external forces.

The knee reaction force, R_x, was computed for each combination of leg posture and external force. Due to the large number of possible combinations it was chosen to display the results as compactly as possible, by showing results from several different simulations

Horizontal component F_h		Vertical component F_v	
Percentage of body weight [%]	Force [N]	Percentage of body weight [%]	Force [N]
0	0	200	1473
20	147	200	1473
50	368	200	1473
75	552	200	1473
100	737	200	1473
150	1105	200	1473
0	0	300	2210
20	147	300	2210
50	368	300	2210
75	552	300	2210
100	737	300	2210
150	1105	300	2210

Table 4.1: Horizontal and vertical components of the applied ground reaction forces. The sagittal model was based on a single leg and the pelvis. The size of the model was given by the AnyBody Modelling System's standard scaling which means that it fitted a 75 kg human body. The force corresponding to 100% body weight was calculated as $75kg \cdot 9,82\frac{m}{s^2} \cdot 1,00 = 737N$.

in one diagram. Figures 4.6 and 4.7 are two diagrams displaying representative results. The remaining diagrams are in a spreadsheet on the enclosed CD.

One might argue that it was too restrictive only to consider the situations where the muscle activity did not exceed 100%. But even if the allowable muscle activity was set to 150%, the knee reaction force R_x was above -$1000N$.

4.2.2 Knee reaction force at increased ground reactions

Increased ground reaction did not induce increased knee reaction forces, because the muscles did not have the required strength to balance the external forces.

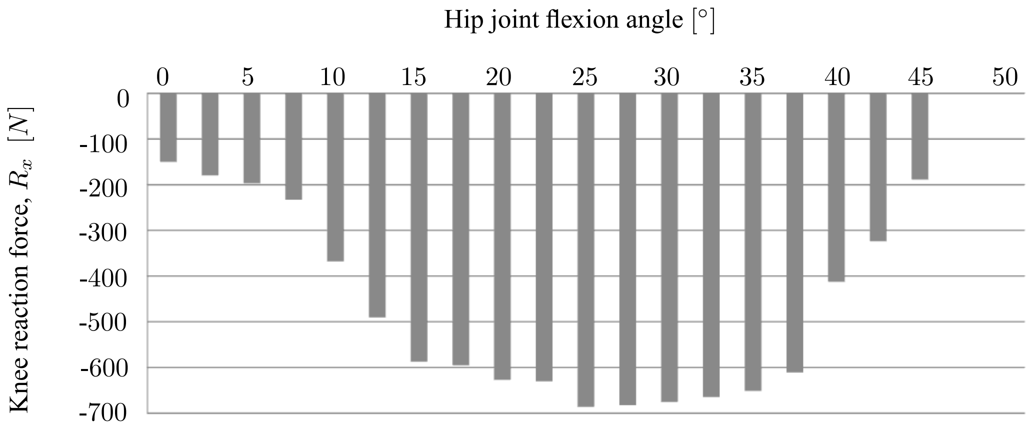


Figure 4.6: Knee joint shear reaction as a function of hip joint flexion.
This diagram includes all leg posture where the position of knee joint flexion angle was given as the hip flexion plus 15°, i.e. the knee was in front of the foot like in Figure 4.1(c). The vertical component was 200% of the body weight and the horizontal component was $0, 20, 50, 100$ and 150% of the body weight.

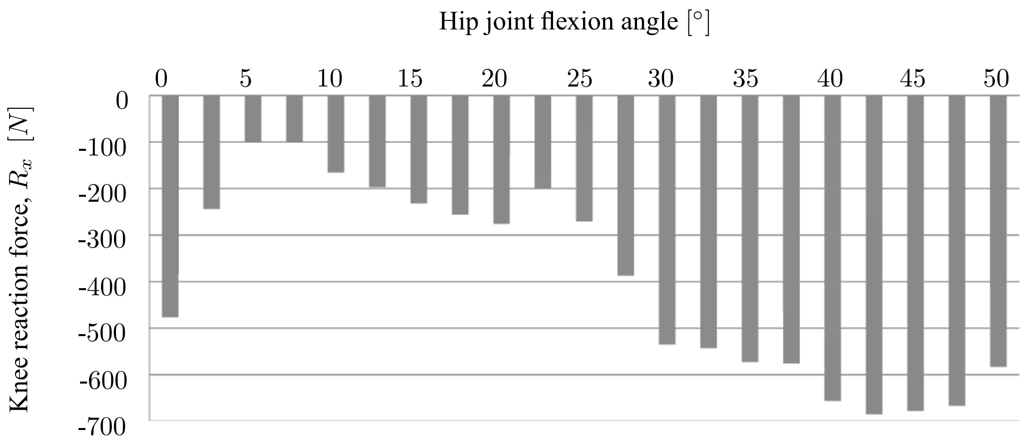


Figure 4.7: Knee joint shear reaction as a function of hip joint flexion.
This diagram includes all leg posture where the position of knee joint flexion angle was equal to the hip flexion. The vertical component was 200% of the body weight and the horizontal component was $0, 20, 50, 100$ and 150% of the body weight.

4.3 Discussion

The ligament strain was evaluated on basis of the size and direction of the knee joint reaction force calculated in the dynamic analysis. The component of the reaction force parallel to the tibia's y-axis was compressive and did not strain the cruciate ligaments. The sign of the component parallel to the tibia's x-axis, R_x, decided whether the posterior or the anterior cruciate ligaments were strained. The posterior ligament was strained when R_x was positive, while the anterior ligament was strained when R_x was negative.

The force in the anterior cruciate ligament would be slightly bigger than $|R_x|$ due to the ligament's orientation, provided that the ligament alone had been responsible for resistance of anterior translation of tibia. However, because of the joint capsule's and the other knee ligament's ability to resist anterior translation of tibia, it was estimated that the anterior cruciate ligament would not tear as long as the absolute value of the joint reaction, $|R_x|$, was less than the ultimate tensile strength of the ligament.
When the leg model was subject to typical ground reaction forces, the knee joint reaction, R_x, was above -$800N$, which was considered to be insufficient to tear the ligament.

An additional aspect which was not taken into account when estimating the ligament strain, was articular contact forces. The muscles around the knee joint compress the articular surfaces when they contract and because the tibial plateau is slightly concave this might give rise to contact forces which protect the cruciate ligament. Studies have shown that compressive loads on the knee joint decrease anterior knee laxity, but it is not known to what extend the articular contact forces have the ability to protect the knee ligaments during sports movement (Hewett et al., 2006b) (Fleming et al., 2003).

Regardless of the size of the articular contact forces, the analysis of the sagittal model demonstrated that it is unlikely that sagittal plane mechanisms will rupture the anterior cruciate ligament, which is in agreement with McLean et al. (2004) and Simonsen et al. (2000).
McLean et al. (2004) used subject specific forward dynamic musculoskeletal models, to study whether sagittal plane knee loading during sidestep cutting could injure the anterior cruciate ligament. The potential for sagittal plane loading as an injury mechanism was quantified as the number of simulations where the peak anterior drawer force exceeded $2000N$. McLean et al. found that the ligament was not even strained during normal sidestep cutting movements, as opposed to Simonsen et al. (2000) who estimated that the ligament load was $520N$, yet insufficient to rupture the ligament.
The forward dynamic analysis enabled McLean et al. (2004) to imitate variation in neuromuscular control, with random perturbations in initial segment kinematics and muscle activation patterns. The perturbations significantly increased the anterior drawer force, the valgus moment and the internal rotation moment, but the anterior drawer forces were well below $2000N$, hence the sagittal plane loading did not have the potential to rupture the ligament. However, McLean et al. found that the coronal plane[2] loads could injure the ligament; especially knee valgus loading was sensitive to neuromuscular perturbations. This subject was treated in Chapter 5.

[2] A coronal plane (or frontal plane) is any vertical plane that divides the body into ventral and dorsal sections.

CHAPTER 5

Advanced knee model

I the previous models the knee joint was approximated as an ideal hinge with a revolute joint that only allowed rotation about one axis. However, the relative movement between femur and tibia is far more complex. When the knee flexes the joint surfaces both roll and slide with respect to each other and the tibia will naturally undergo a slight internal rotation. In addition to the knee's natural movement it can also be forced into hyperextension, valgus or varus positions, increased internal/external rotation and anterior/posterior translation of tibia.
The relative movement between femur and tibia, natural or forced, is related to a complicated interaction between muscles, ligaments and bones. The purpose of the advanced knee model was to improve the imitation of the movement of a real knee. This was done by adding knee ligaments, imposing a contact condition between the segments and changing the way the model was driven.
The advanced knee model made it possible to investigate the elongation of the anterior cruciate ligament for various knee positions and hereby evaluate which movements are most likely to tear the ligament.

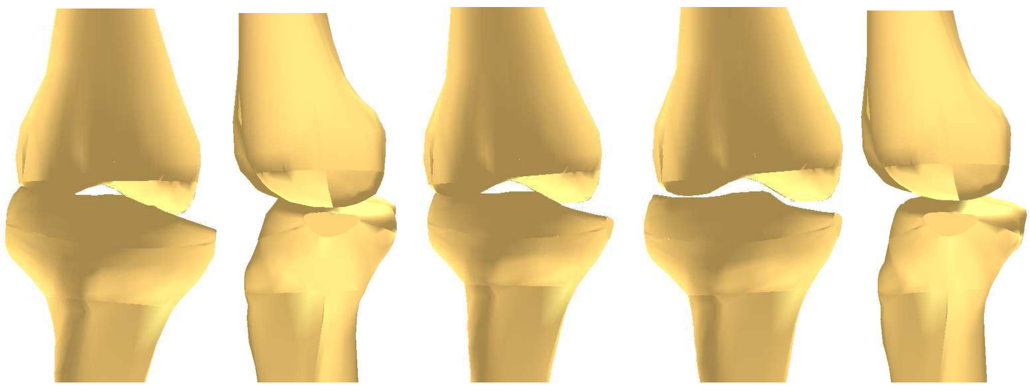
5.1 Creating the model

The underlying basis for the model was a one legged model from the AnyBody Model Repository and it had to undergo thorough modifications which will be described in the following. Some of theses changes entailed that it was only possible to analyse the model kinematically and it was impossible to determine the forces. Note that the model was a right leg and all illustrations in this chapter refer to the right knee.

5.1.1 Segments

The model contains four segments: pelvis, thigh (femur), shank (tibia) and foot. Seeing that the purpose of the model was to quantify the strain in the anterior cruciate ligament and not the muscles, these could be excluded from the model. As follows it was sufficient with just *femur* and *tibia* in the model to investigate the strain in the knee ligaments;

however, the remaining segments were retained in order to utilize the structure in the model from the repository.
A visual inspection of femur and tibia in the model repository revealed that the size of the two bones did not match and the way they were assembled did not look quite right, Figures 5.1(a) and 5.1(b). The discrepancy in the dimensions probably occurred because the data defining the two bones were not obtained from the same cadaver and it was improved by scaling the size of tibia with a factor $0,9$, see Figure 5.1(c). The joint was improved further by changing the location of the knee joint node[1] on the femur, see Figures 5.1(d) and 5.1(e).
The fact that the bones originate from two different cadavers, made it difficult to obtain perfect congruency. This would inevitably induce some inaccuracies which had to be taken into account when the results were assessed.



(a) (b) (c) (d) (e)

Figure 5.1: Modification of the knee joint.
(a)Front view of the original knee joint configuration
(b)Medial view of the original knee joint configuration
(c)The size of tibia is reduced, but the joining of femur and tibia is wrong
(d)Front view of the new joint configuration
(e)Medial view of the new joint configuration

5.1.2 Contact condition

The interaction between the joint surfaces was taken into account by introducing a contact condition between the tibial plateau and the femoral condyles, *condyli femoris*. The condition ensured that there was a minimum distance of $0,01m$ between the surfaces and that there always would be at least one contact point, i.e. a point on each surface where the intermediate distance is exactly $0,01m$. The surface of the tibial plateau was identified by

[1]All the segments have joint nodes. These nodes are used to assemble the segments because they define the locations and reference coordinate systems of the joints.

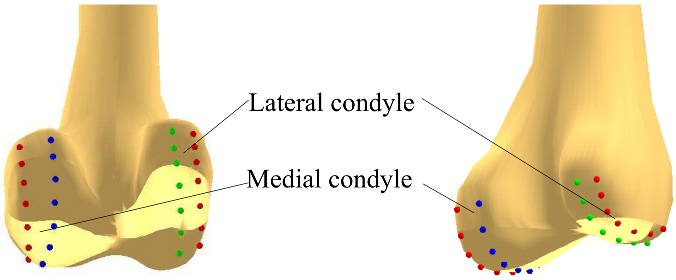
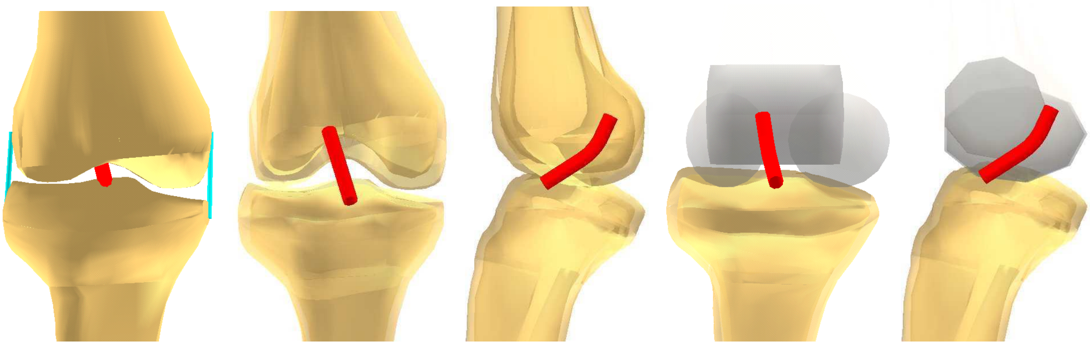


Figure 5.2: Illustration of the contact condition. The points on the femoral condyles were used to identify the surface.

two points, while the surface of the femoral condyles was identified by 28 points which were distributed over the surface, see Figure 5.2.

5.1.3 Ligaments

The knee joint was equipped with the anterior cruciate ligament and the two collateral ligaments[2], Figure 5.3(a).



(a) (b) (c) (d) (e)

Figure 5.3: Knee joint ligaments.
(a)Front view of knee joint with the anterior cruciate ligament and the two collateral ligaments
(b)Front view of the anterior cruciate ligament in the knee joint, the segments are transparent
(c)Medial view of the anterior cruciate ligament in the knee joint, the segments are transparent
(d)The three surfaces the anterior cruciate ligament had to pass
(e)The anterior cruciate ligament wraping over the surfaces

[2] *Ligamentum collaterale tibiale* and *ligamentum collaterale fibulare* - also denoted respectively medial and lateral collateral ligament.

The insertion and origin of the anterior cruciate ligament was estimated from anatomical drawings and a physical anatomic skeleton model. This procedure could introduce some inaccuracies motivating an investigation of the sensitivity of the ligament strain against the position of the insertion and origin.
In a real knee the ligament is wrapped over the surface of the femur and it will slide on the bone surface when the knee moves. This effect was taken into account by defining three surfaces which the ligament had to pass. If the surfaces block the way between the origin and the insertion the ligament will find the shortest path around the surface, see Figure 5.3. The shortest path around the surfaces was found with an optimality criterion. The criterion determined the position numerically while the velocity was derived from the position. The software could not determine the acceleration and was therefore unable to set up the equilibrium equation required to perform a dynamic analysis.
The collateral ligaments are an integrated part of the joint capsule and therefore it is difficult to identify specific insertions and origins on the bones. The insertions and origins of the collateral ligaments were loosely estimated from drawings in various anatomy books and wrapping around bones was not taken into account.

5.1.4 Driving the model

When the knee flexes the joint surfaces both roll and slide with respect to each other, but the lateral and the medial condyle undergo different translations causing the femur to experience a slight external rotation. The tibio-femoral movement during knee flexion has been measured and described by Iwaki et al. (2000) and Johal et al. (2005)[3].
Johal et al. (2005) found that when the knee flexes from -$5°$ to $120°$, the lateral femoral condyle translates $22mm$ in posterior direction, while the medial condyle demonstrates minimal translation and the femur rotates externally. The external rotation of the femur occurred both for loaded and unloaded conditions, but the rotation was larger and occurred earlier when the knee was bearing weight. Johal et al. (2005) assess that gender has no effect and that there is no difference between the right and the left knee.
The translation of each condyle was given by the distance between the "ipsilateral posterior tibial cortex" and the centre of "the femoral posterior circle", see Figure 5.4. The locations of the references points were based entirely on the descriptions and multiple resonance images in Iwaki et al. (2000) and Johal et al. (2005), seeing that the term "ipsilateral posterior tibial cortex" could not be found elsewhere.

Figure 5.5 shows the translation of the reference points in the condyles. By adding similar reference points to the model these data could be used to drive and constrain two degrees of freedom - namely anterior/posterior translation along the tibial plateau and internal/external rotation.

[3]The measurements from Johal et al. (2005) were used to drive the model, but the location of their references points where referring to the work of Iwaki et al. (2000)

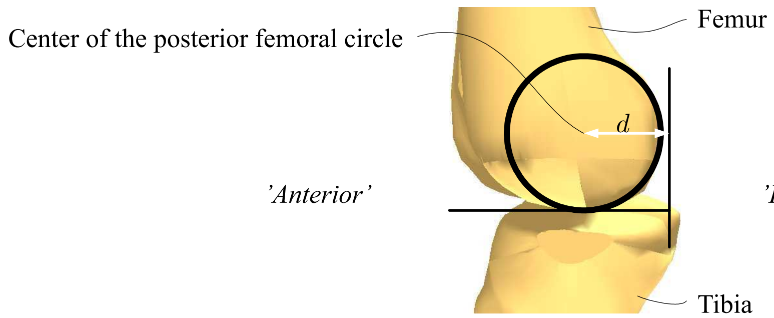


Figure 5.4: Distance between the "'ipsilateral posterior tibial cortex"' and the centre of the posterior femoral circle, denoted d. (Iwaki et al., 2000, page 1190, Fig. 1.)

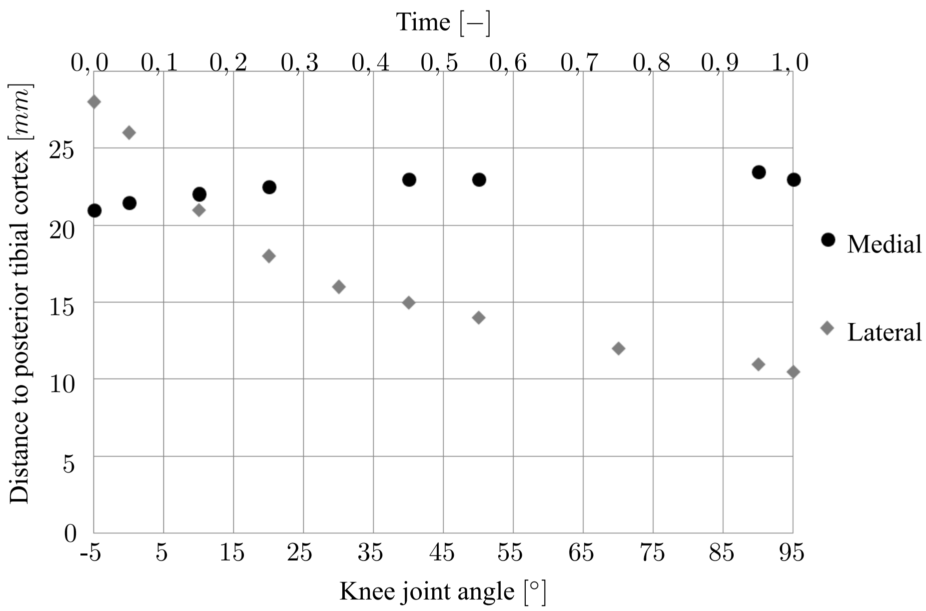
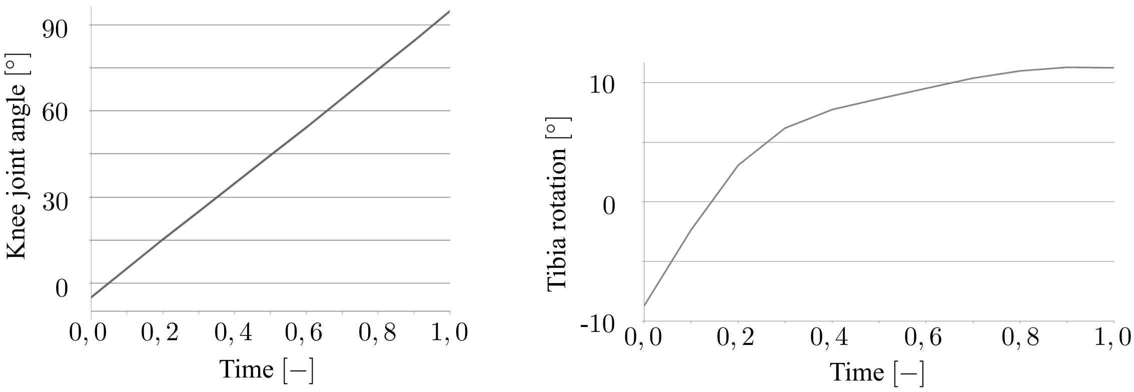


Figure 5.5: Translation of the medial and lateral condyle as a function of knee joint angle or time.
The position of the femoral condyles was read from a graph (Johal et al., 2005, page 271, Fig. 2.).

The translation of the reference points were given as a function of knee flexion, but the model had to be driven as a function of time (with an interpolation driver) and it was therefore necessary to determine the translation at selected time steps so they matched with the driver which controlled the knee's extension/flexion.
The contact condition described above restricts translation perpendicular to the tibial plateau, while translation in the knee's transversal direction was constrained. The remaining degrees of freedom were controlled by driving the knee's abduction/adduction.
Figure 5.6 shows the extension/flexion angle and tibia rotation as a function of time when the model is driven on basis of the measurments in Johal et al. (2005).



(a) Knee joint angle as a function of time. Negative values indicate hyperextension.

(b) Tibia rotation as a function of time. Negative values denote external rotation, positive values denote internal rotation.

Figure 5.6: Position of the knee joint as a function of time

5.2 Analysis of the advanced knee model

The measurements from Johal et al. (2005) were used as a basis for analysing the elongation of the anterior cruciate ligament. It was chosen to investigate the knee joint flexion range from -5° to 95° and divide the movement sequence into 11 time steps, i.e. there was 10° between each step. The length of the ligament (Figure 5.7) is subject specific and it was therefore considered to be more valuable to describe the strain.

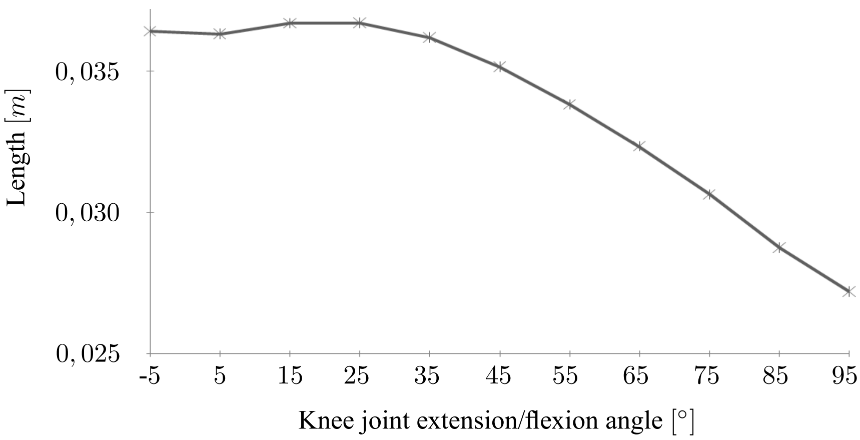


Figure 5.7: Length of the anterior cruciate ligament as a function of knee joint angle. 0° is full extension, i.e. -5° is hyper extension

Unfortunately it was not known for which knee joint configuration the ligament could be expected to be unstrained. Figure 5.8 shows the strain as a function of the knee joint angle, calculated with two different knee joint angles as reference for the unstrained knee joint configuration.

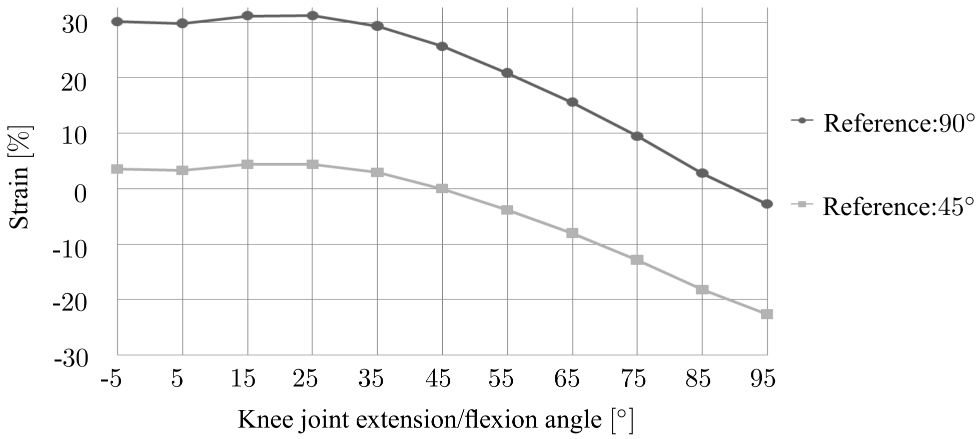
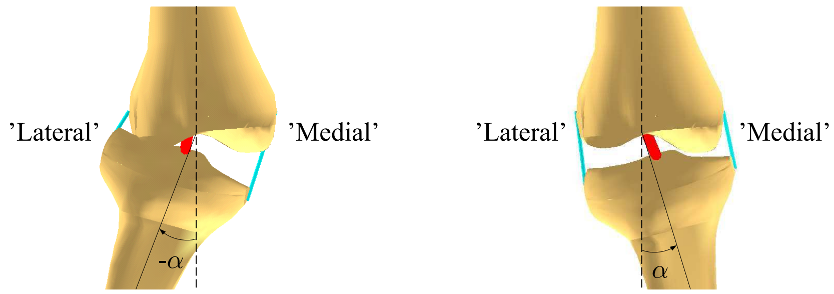


Figure 5.8: Strain in the anterior cruciate ligament as a function of knee joint flexion angle. The darker curve represents the strain when the unstrained length is set at 90°, while the lighter curve gives the strain when the unstrained length is set at 45°.

It was chosen to use the ligament length at 45° knee flexion as reference to calculate the strain and compare different knee positions. This reference was chosen because the cruciate ligament was expected to endure strain up to approximately 10%[4].
If the reference ligament length was set at a knee angle larger than 55° the ligament would experience unnatural large strain near full extension, while there would be almost no strain near full extension if the reference length was set at a knee angle smaller than 35°.

5.2.1 Varus/Valgus

The terms varus and valgus refer to the direction that a joint's distal segment points.



(a) Valgus position (b) Varus position

Figure 5.9: Knee joint in valgus and varus position.

[4]Several papers treat the maximum strain the anterior cruciate ligament can endure before failure sets in, however, there are significant variations in the failure strain values. The failure strain was given as approximately 10%, 15%, 30%, and 60% in theses papers: S.Karmani and Ember (2003), Chandrashekara et al. (2006), Noyes and Grood (1976).

The knee is in a valgus position, *genu valgum*, when the tibia is rotated laterally in relation to the femur, resulting in a knock-kneed appearance, Figure 5.9(a). As follows the knee is in varus position, *genu varum*, when the tibia is rotated medially in relation to femur, resulting in a bowlegged appearance, Figure 5.9(b).
The strain during knee flexion for varus, valgus and neutral position was compared, Figure 5.10.

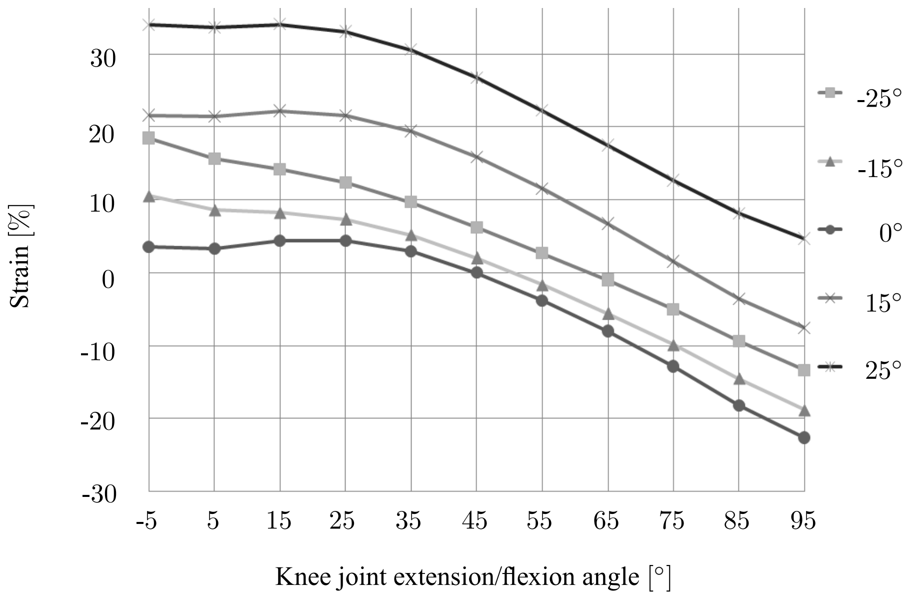


Figure 5.10: Strain in the anterior cruciate ligament as a function of knee joint flexion angle. The different curves represent different varus or valgus positions: -25°, -15° is valgus angles, 15°, 25° is varus angles, while 0° is neutral position.

It was observed that the strain increased significantly for both valgus and varus positions, however, the varus position increased the strain the most. There was a notable difference between the shapes of the five curves in Figure 5.10, this is even more evident in Figure 5.11 where the strain is a function of varus/valgus angle.

Varus positions produced higher strain levels than the corresponding valgus positions. For particular knee joint flexion angles the strain actually decreased for valgus angle from 0° to -5°. For knee joint angles above 40deg this tendency was reduced and the curves attained a shape similar to the curve for 0° knee flexion.
Except from the case with hyperextension, -5°, the valgus angle had to exceed -5° to -10° before the strain exceeded the level at neutral valrus/valgus position.

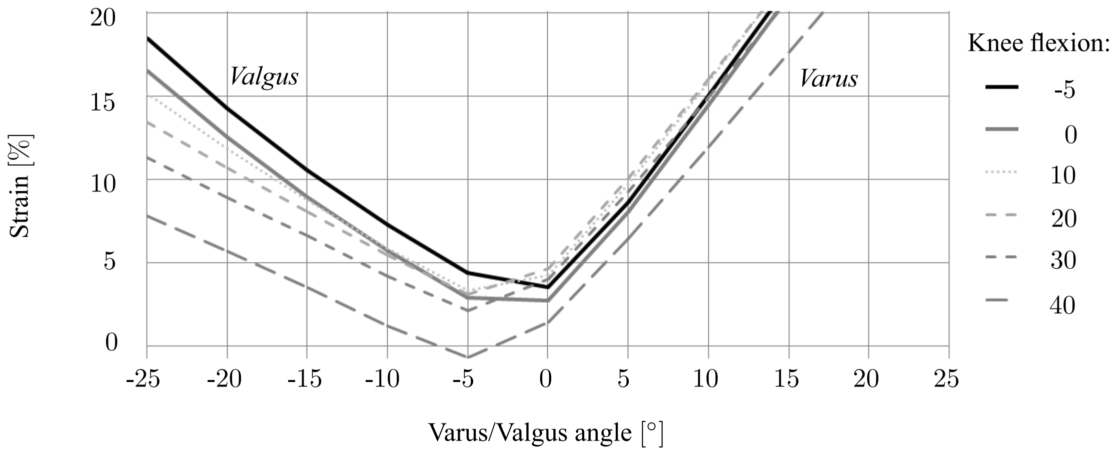


Figure 5.11: Strain in the anterior cruciate ligament as a function of varus/valgus angle. Negative and positive angles indicate respectively valgus and varus positions. The different curves represent selected knee joint flexion angles: -5, 0, 10, 20, 30, 40

5.2.2 Rotation

The rotation of tibia about its longitudinal axis was controlled by the translation of the reference points in the condyles, hence the translation of the condyles was modified in order to increase the rotation. One approach could be to set the rotation to a constant value, e.g. 10deg internal rotation, during the entire knee flexion range from -5° to 95°. However, it was chosen to investigate the effect of tibia's rotation by introducing a constant rotation increment, e.g. natural rotation during knee flexion plus an increment of 10° internal rotation. The result of increased internal and external rotation is shown in Figure 5.12.

Increased external rotation reduced the strain in the anterior cruciate ligament while increased internal rotation increased the strain. The changes in strain were relatively small in view of the severe rotation of the tibia.

5.2.3 Translation

The anterior translation of tibia was increased by reducing the distance between the reference points in the femoral condyles and the reference points on tibia, 'ipsilateral posterior tibial cortex'. Even a slight anterior translation of tibia increases the strain in the anterior cruciate ligament significantly, Figure 5.13.

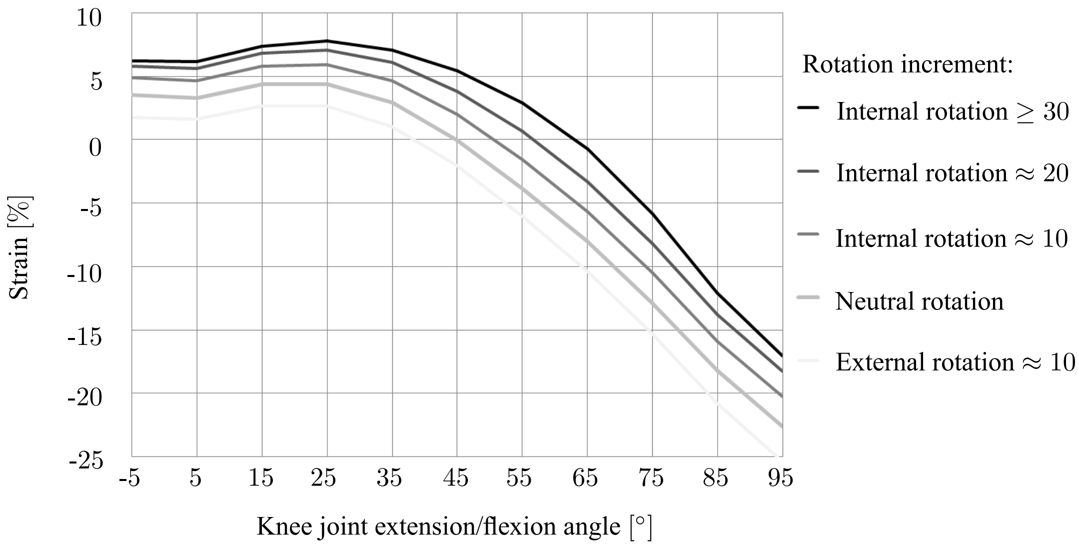


Figure 5.12: Strain in the anterior cruciate ligament as a function of knee joint flexion angle. The different curves represent different increments of the tibial rotation: No increment, increased external rotation of approximately 10°, increased internal rotation of approximately 10°, 20° and 30°.

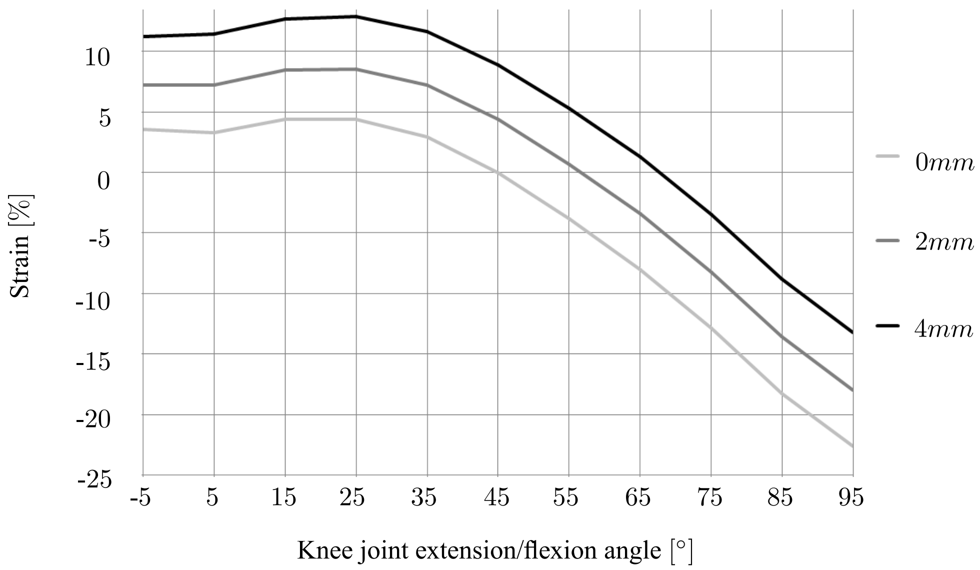


Figure 5.13: Strain in the anterior cruciate ligament as a function of knee joint flexion angle. The different curves represent increments of anterior tibial translation, respectively 0mm, 2mm and 4mm.

5.2.4 Comparison with video analysis

Olsen et al. (2004) studied anterior cruciate ligament injuries in female athletes by analysing videotapes of injury situations from European handball. The team analysed the playing situations that lead to the injury and the injury mechanism with particular attention to the knee position, specifically they estimated knee flexion, internal/external rotation and valgus/varus position. The knee model was placed in positions similar to the knee positions Olsen et al. (2004) observed at the time of the injury, to investigate whether there was a pattern in the strain levels, Table 5.1.

Knee Position			**ACL strain**
Flexion	Valgus	Rotation	
[°]	[°]	[°]	[%]
5	-5	-5	2,4
5	-10	10	7,0
10	-15	5	8,9
10	-20	-10	10,7
10	-10	-10	4,1
15	-20	10	11,5
15	-15	5	8,2
15	-15	10	8,8
15	-15	10	8,8
15	-15	-10	6,3
15	-10	-5	7,0
20	-15	15	8,9
20	-15	-10	5,5
20	-15	-10	5,5
20	-10	-10	2,6
20	-10	-10	2,6
20	-10	-10	2,6
25	-15	-15	4,3
25	-10	-10	1,6

Table 5.1: Estimated knee positions when the anterior cruciate ligament ruptured (Olsen et al., 2004, page 1006, table 4) and the calculated strains.

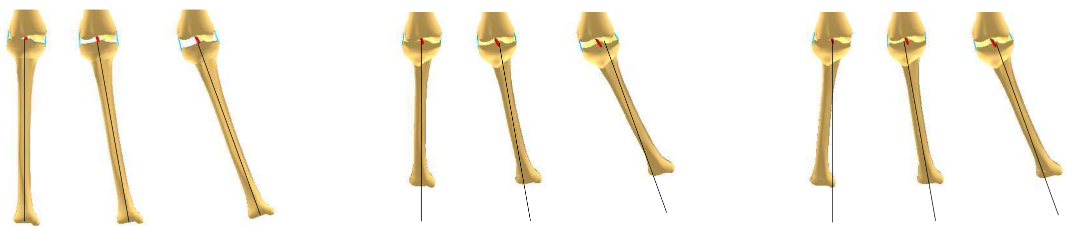
There was no consistency in the computed strains. The strain at the time of the injury varied from 1,6 - 11,5%.

5.3 Discussion

In order to determine the strain in the anterior cruciate ligament it is crucial to know for which knee joint configuration the ligament is unstrained. Examining the length of the ligament as a function the knee joint flexion angle lead to the assumption, that the ligament is unstrained when the knee is flexed between $35°$ and $55°$. The reference length for the unstrained ligament was set at a $45°$ knee joint angle. If the length should have been determined at $35°$ or $55°$ this would introduce a deviation up to respectively 3% and 4% strain. However, the curve representing the strain as a function of knee joint flexion angle would on the whole have the same shape regardless of the chosen unstrained ligament length.
The highest strain value was computed when the knee was flexed $15°$, i.e. the peak strain had to be found at a knee flexion angle of $5° - 25°$.
The shape of the curve representing the strain as a function of knee joint flexion angle is in agreement with Li et al. (1999) who investigated the in-situ forces in the anterior cruciate ligament. Li et al. (1999) measured the in-situ forces in the ligament in response to isolated quadriceps load as well as combined quadriceps and hamstring loads when the knee was flexed respectively $0°$, $15°$, $30°$, $60°$, $90°$ and $120°$ and found that the peak strain occurred when the knee was flexed $15°$ regardless of muscle loading. Generally the forces Li et al. measured in the ligament varied in a way similar to manner the strain varied in the knee model, which supports that the strain variation in the model during knee extension/flexion is probable.

The strain as a function of valgus/varus angle (Figure 5.11) appeared less probable. Valgus and varus positions were expected to increase the strain so it seemed peculiar that the minimum strain was to be found in the interval -$5°$ to $0°$ valgus angle.
The sensitivity of the position of the origin and the insertion of the ligament was investigated by tentatively changing the positions. It was observed that the exact position of origin/insertion had a significant impact on the strain level; however, the shape of the curves remained unchanged, i.e. the minima were in the interval -$5°$ to $0°$ valgus angle.
As described in Section 5.1.1 the position of the knee joint node on femur had been changed because the original assembly of the joint was incorrect. It appeared that the location of the minimum strain was very sensitive to the way the joint was assembled. It was possible to move the minima of the curves to a neutral valgus/varus angle by a slight modification of the orientation of the knee joint node. Unfortunately it was unclear whether this assembly of femur and tibia was more or less accurate. The accuracy of model might be improved if it was rebuilt with data obtained from just one cadaver study.

Another parameter that influenced the result was the interpretation of the anatomic joint positions. Abduction and adduction are only clearly defined when the knee is fully extended. There are several ways to interpret abduction/adduction when the knee is flexed.
The abduction/adduction of the tibia with respect to femur was controlled with the kinematic measure "AnyKinRotational" and the output type "RotAxesAngles". In other words, the orientation of tibia was given by Cardan angles that measure a three dimensional rotation by three angles of planar rotation about the reference system's axes, in this case the axes of the knee joint node on femur. The sequence of these rotations was very important because any rotation subsequent to the first rotation was a rotation about a local axis developed by the previous rotations. It was chosen to use the default sequence of rotation: first rotation around the z-axis, second rotation around the y-axis, third rotation around the x-axis; i.e. the abduction/adduction was last in the sequence and thus affected by the two previous rotations.
Alternatively the kinematic measure could have had the output type "RotVector", which is a Cartesian rotation vector that measures the three dimensional orientation of tibia as a single rotation about the x-axis of the knee joint node on femur.
Figure 5.14 illustrates how the joint position is affected by the chosen definition of abduction/adduction. There is no difference between the two definitions when the knee is extended (Figure 5.14(a)), but the difference between the two definitions is clear when the knee is flexed (Figure 5.14(b) and 5.14(c)).



(a) Knee joint flexed 0° and tibia adducted respectively 0°, 10° and 20°

(b) Knee joint flexed 40° and tibia adducted respectively 0°, 10° and 20° (Cardan angles)

(c) Knee joint flexed 40° and tibia adducted respectively 0°, 10° and 20° (Cartesian rotation vector)

Figure 5.14: The knee joint configuration is dependent on how the anatomic joint positions are defined. There is no difference between the two definitions when the knee is flexed 0°, but there is a clear difference when the knee is flexed and rotated. The black lines are inclined 0°, 10° and 20° and indicate the angulation of tibia when the knee is flexed 0°.

If the abduction/adduction had been controlled by a Cartesian rotation vector the strain as a function of varus/valgus position had been slightly different, Figure 5.15. In particular, the minimum strain had been found at a neutral valgus/varus position for most knee flexion angles.

It remained unclear whether the minimum strain is at a neutral varus/valgus angle or is somewhere in the interval -5° to 0°. Regardless of the joining of femur and tibia and the chosen definition of abduction/adduction, the results indicated that valgus and especially varus positions can increase the strain significant. This is in agreement with

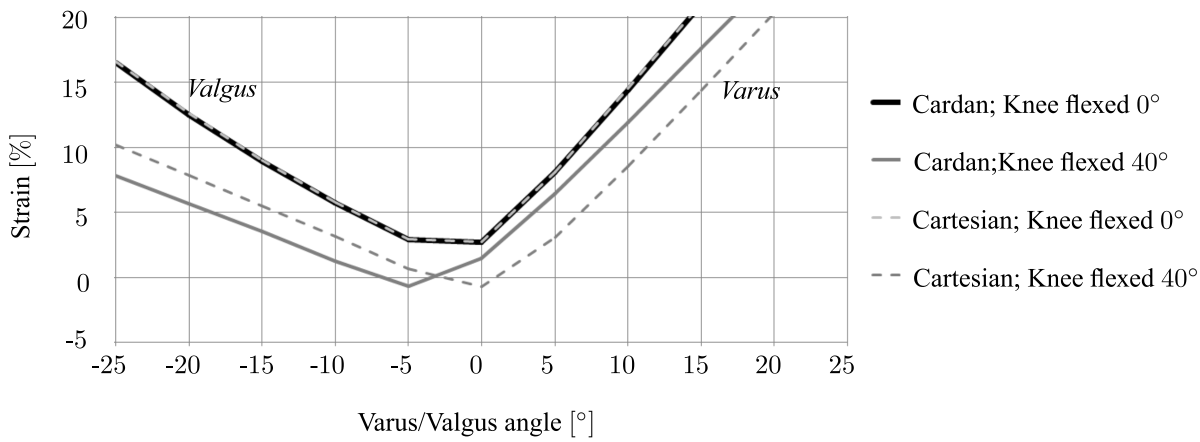


Figure 5.15: Strain in the anterior cruciate ligament as a function of varus/valgus angle. The dotted lines shows the strain when abduction/adduction was defined by a Cartesian rotation vector, while the unbroken lines shows the strain when abduction/adduction was defined by Cardan angles. There is no difference in strain when the knee is flexed 0°.

S.Karmani and Ember (2003), who stated that varus forces strained the the ligament more than valgus forces.

Rotation of tibia was also expected to increase the strain in the anterior cruciate ligament. However, external rotation of tibia slightly reduced the strain. This could be explained from the position of the ligament's origin and insertion and that the strain in the ligament was computed as the shortest distance between origin and insertion. The origin was on the lateral condyle, while the insertion was on the medial side of tibia, which meant that the insertion moved towards the origin when tibia rotated externally. It should be noticed, that the strain caused by the twist of the ligament was not taken into account, but it was considered to be limited.
Internal rotation of tibia increased the strain, but the strain increments were very limited for even severe rotations. Therefore it seems unlikely that rotation alone will cause a rupture of the anterior cruciate ligament.
When the effect of tibial rotation was examined, it was assumed that the centre of rotation was in between the femoral condyles. In reality the centre of rotation might be located in the medial or lateral condyle, which corresponds to a rotation plus a translation. The analysis of the knee model showed that even a small anterior translation of tibia increases the strain level significantly, which entail that an internal rotation of tibia around the medial condyle (or a external rotation around the lateral condyle) can rise the strain level substantial.

The strains computed at the injuring knee positions which Olsen et al. (2004) estimated from a video analysis were compared, but there was no consistent pattern in the strains at the time of the injuries. One might argue that the inconsistent strain level was a result of the accuracy of the estimated knee positions. Even small diversions in knee flexion or valgus angle would change the strain level significantly. Another explanation could be that the analysis did not produce any information about anterior translation of tibia with respect to femur, or information about the centre of rotation in connection with internal/external rotation of tibia. When the strains caused by the injuring knee positions were computed it was assumed that the centre of rotation was between the femoral condyles, but it might as well have been somewhere in the medial or lateral condyle, which would change the strain level significantly. The accuracy of the estimated knee positions might be an issue, but it is important to note that the video analysis did not produce the information that was essential to evaluate the strain level in the anterior cruciate ligament at failure.

CHAPTER 6

Discussion of Injury Mechanisms

The anterior cruciate ligament injury mechanisms were analysed with four different leg models. This discussion was written to summarise the previous discussions in the sections: 2.4, 3.3, 4.3 and 5.3.

6.0.1 Voluntary movements

The quadriceps is important, because it is the only muscle that has the potential to pull the proximal tibia forward.
Considering that it is unknown how the human body distributes the forces between the redundant muscles, the accuracy of the muscle recruitment may be discussed. However, the recruitment of the quadriceps was considered to be sufficiently accurate since the quadriceps is the only knee extensor.
The orientation of the patella tendon (that connects the quadriceps with the tibia) was compared with two empirical studies (Baltzopoulos (1995) and Herzog and Read (1993)), and it was found that the orientation of the patella tendon in the AnyBody models was more anterior than in a real knee. Therefore, it was deduced that the anterior drag on the tibia from the quadriceps is not larger than the models predicted.

Beynnon and Flemming (1998) measured strain behaviour of a normal anterior cruciate ligament by arthroscopic implantation of a differential variable reluctance transducer in the ligament, while subjects were under local anaesthesia. They measured the strain in the ligament during active flexion/extension of the knee and found that the ligament is unstrained for knee flexion angels larger than $38°$. Unfortunately the paper from Beynnon and Flemming (1998) did not describe the effects of increased weight. One might expect that increased load will not only increase the strain in the anterior cruciate ligament, but also influence the angle interval where the ligament is unstrained. The analysis of the lunge model was found to be in agreement with Beynnon and Flemming (1998) because the ligament was unstrained during the entire movement sequence[1]. However, the fact that the ligament was unstrained during the lunge movement was not necessarily exclusively a result of the knee flexion range. The relatively low muscle activity in the hamstrings and in the quadriceps[2] may also have been a decisive factor. This postulate is

[1]Note that the knee joint flexion angle was $39° - 116°$
[2]For knee flexion angels below $75°$

supported by the analysis of a male sprinter, which showed that the ligament was strained for knee flexion angels not only below but also above $38°$, namely up to $50°$.

The sprint gave rise to large quadriceps force and strained the anterior cruciate ligament. However, the knee joint reaction, which was used to evaluate the ligament strain, was well below the ultimate tensile strength of the ligament. Considering that sprint probably is one of the most intense sagittal plane sports movements, it appears that voluntary contraction is insufficient to injure a healthy cruciate ligament. Even though intense voluntary contraction might be insufficient to injure the anterior cruciate ligament during sprint, the analysis does not rule out the possibility that other sagittal plane movements may put more strain on the ligament. Therefore, a representative selection of various feasible sagittal plane movements were analysed with the sagittal model.
The analysis of the sagittal model demonstrated that it is unlikely that sagittal plane mechanisms will rupture the anterior cruciate ligament.

6.0.2 Forced movements

In the lunge model, the runner model and the sagittal model the knee joint was approximated as an ideal hinge. But the relative movements between femur and tibia are far more complex and are related to a complicated interaction between muscles, ligaments and bones. In addition to the knee joint's natural movement, flexion/extension, it can also be forced into hyperextension, valgus or varus positions, increased internal/external rotation and anterior/posterior translation of the tibia. The advanced knee model made it possible to investigate the elongation of the anterior cruciate ligament for various knee positions and thereby evaluate which movements are most likely to tear the ligament.

The flexion/extension movement of the knee was driven with data from an empirical study (Johal et al., 2005) and the resulting elongation of the ligament seemed to agree with Li et al. (1999), who investigated the in-situ forces in the anterior cruciate ligament. Generally the forces Li et al. (1999) measured in the ligament varied in a manner similar to how the strain varied in the knee model, which supports that the strain variation in the model during knee extension/flexion was probable.

It was more difficult to evaluate the strain as a function of varus/valgus position, or as a function of increased internal/external rotation of the tibia. These movements are unnatural and they were not outlined by empirical studies, as opposed to flexion/extension which has been described in several papers. The anatomic joint movements are not even well-defined, except from the case when the knee is fully extended and was therefore defined by the author.
The data used to define the femur and the tibia came from two different cadaver studies and it was therefore difficult to assess whether the segments were scaled and joined correct. The ligament strain proved to be very sensitive to the way the segments were joined and the uncertain knee joint configuration was therefore problematic. The accuracy of model might be improved if it was rebuild with data which was obtain from just one cadaver study.

Regardless of the definition of the anatomic joint movements and the knee joint configuration the analysis of the model showed that:

- The ligament is strained the most when the knee is flexed $5° - 25°$
- Valgus and especially varus positions can increase the strain considerably.
- Rotation of tibia about its longitudinal axis produce minor strain.
- Anterior translation of the tibia increases the strain significantly.

When the effect of rotation of the tibia was examined, it was assumed that the centre of rotation was in between the femoral condyles. In reality the centre of rotation might be located in the medial or lateral condyle, which corresponds to a rotation plus a translation. The analysis of the knee model showed that even a small anterior translation of tibia would increase the strain level significantly, which entail that a rotation of tibia can raise the strain level substantially if the centre of rotation is different from the assumed centre of rotation.

The knee model was applied to anterior cruciate ligament injury situations from European handball (Olsen et al., 2004) in order to investigate whether there was a pattern in the strain levels at the time of the injuries.
There was no consistency in the computed strains. The inconsistent strain level could be a result of the accuracy of the knee joint positions that were estimated in the video analysis. Even small diversions in knee flexion or valgus angle would change the strain level significantly. Another explanation could be that the analysis did not produce any information about anterior translation of the tibia or about the centre of rotation in connection with internal/external rotation. In other words, the video analysis did not produce the information that was essential to evaluate the strain level in the anterior cruciate ligament at failure.

CHAPTER 7

Conclusion

Anterior cruciate ligament injury mechanisms were studied with four musculoskeletal models. The models made it possible to determine the knee shear force during various sports movements and explore the elongation of the anterior cruciate ligament during both natural and forced movements.

The injury mechanisms are much debated and there is a dispute about whether the primary injury mechanism is valgus trauma, medial rotation of the tibia or intense deceleration.
The inverse dynamic analysis revealed that the knee shear force during sagittal plane sports movements is well below the ultimate tensile strength of the ligament. In other words, the analysis eliminate that the sagittal plane mechanism during intense deceleration will tear a healthy anterior cruciate ligament.
Studying the elongation of the ligament showed that it is strained the most when the knee joint is flexed $5° - 25°$.
Valgus and especially varus positions can increase the strain in the ligament significantly and the ligament is therefore likely to tear if the knee joint is forced into either varus or valgus.
Rotation of the tibia about its longitudinal axis only produces minor strain and it seems implausible that this mechanism will injure the ligament.

APPENDIX A

Nomenclature

ACL
Anterior cruciate ligament

ACLD
Anterior cruciate ligament deficient

Aetiology
The study of why things occur. In medicine in particular, the term refers to the causes of diseases or pathologies.

Anterior
The front, as opposed to the posterior.

Coronal plane
Coronal plane (or frontal plane) is any vertical plane that divides the body into ventral and dorsal sections.

Distal
Further from the beginning, as opposed to proximal.

Dorsal
The back, as opposed to ventral.

Dorsiflexion
Ankel flexion towards the back of the foot

In vivo
In vivo refers to experimentation done in or on the living tissue of a whole, living organism as opposed to a partial or dead one.

In vitro
In vitro refers to the technique of performing a given experiment in a controlled environment outside a living organism.

In situ
In situ means to examine the phenomenon exactly in place where it occurs. This usually means something intermediate between in vivo and in vitro.

Joint node
All the segments have joint nodes. These nodes are used to assemble the segments because they define the locations and reference coordinate systems of the joints.

Contraction
Contraction: When a muscle fibre generates tension. While under tension, the muscle may lengthen, shorten or remain the same.
Dynamic contraction: The muscle changes length as it contracts.
Concentric contraction: The muscle shortens as it contracts.
Eccentric contraction: The muscle lengthens as it contracts.
Isometric contraction: The muscle remains the same length despite building tension.
Isokinetic contraction: The muscle contracts with constant velocity.

Forward Lunge
Performing a forward lunge means to stand in an upright position, then take one step forward and flex both knees and subsequently extend the knee in front and push oneself back into the upright starting position.

Kinematic measure
AnyBody Technology invented the concept of kinematic measures as a way of describing dimensions in a kinematic model. This enables the user to study the development of selected dimensions or control the kinematic measure by adding a driver.

Knee joint movements
Flexion-extension: Movements in the sagittal plane around a transversal axis.
Adduction-abduction: Movements in the frontal plane around a sagittal axis.
Rotation: rotation around a longitudinal axis.

Ligaments
Ligaments connect articulating bones and keep joints assembled. Unlike muscles they don't contain an active contractile element which makes them passive structures that only provide forces when they are stretched by the relative movement of the bones they connect.

Lateral
Toward the left or right side of the body, as opposed to medial.

Medial
In the middle or inside, as opposed to lateral.

Plantarflexion
Ankel flexion towards the sole of the foot

Posterior
The back or behind, as opposed to the anterior.

Proximal
Toward the beginning, as opposed to distal.

Sagittal
A vertical plane passing through the standing body from front to back. The mid-sagittal, or median plane, splits the body into left and right halves.

Side-cutting
Side-cutting is a feint which is often performed in team handball, basketball and American football. The purpose of the side-cutting manoeuvre is to pass an opponent by faking the direction opposite to the intended movement. Normally, a right-handed shooter will approach the opponent head on, brake the forward movement with the left foot and step to the right side. During the braking action, the m. quadriceps femoris contracts eccentrically causing an anteriorly directed shear force on the tibia.

Transverse
A horizontal plane passing through the standing body parallel to the ground.

Varus/Valgus
The terms varus and valgus always refer to the direction that the distal segment of the joint points. A valgus deformity is a term for the outward angulation of the distal segment of a bone or joint. A knee is in a valgus position (genu valgum) when tibia is turned outward in relation to femur, resulting in a knock-kneed appearance. A varus deformity is a term for the inward angulation of the distal segment of a bone or joint. A knee is in varus position (genu varum) when tibia is turned inward in relation to femur, resulting in a bowlegged deformity.

Ventral
Pertaining to the abdomen, as opposed to dorsal.

Via node
A muscle's via node, is a point where the considered muscle is in contact with a segment and reaction forces are generated.

APPENDIX B

Muscles

This appendix describes action, origin and insertion of the muscles that is related to the movement of the knee or just attached to *tibia*.

B.1 Muscles that flex, extend and rotate the knee

Musculus rectus femoris
Action: Extends the knee and flexes the thigh.
Origin: Spina iliaca anterior inferior, limbus acetabuli
Insertion: Lig. patellae to tuberositas

Musculus vastus lateralis
Action: Extends the knee
Origin: Proximal part of linea intertrochanterica, trochanter major, labium laterale linea aspera
Insertion: Lig. patellae

Musculus vastus intermedius
Action: Extends the knee
Origin: Anterior and lateral surface of corpus ossis femoris
Insertion: Lig. patellae

Musculus vastus medialis
Action: Extends the knee
Origin: Distal part of linea intertrochanterica, labrum mediale linea aspera
Insertion: Lig. patellae

Musculus sartorius
Action: Flexes and rotates the hip joint laterally. Flexes the knee and rotates the tibia medially
Origin: Spina iliaca anterior superior

Insertion: Pes anserinus

Musculus gracilis
Action: Adducts the thigh, flexes the knee and helps to medially rotate the tibia
Origin: Ramus inferior ossis pubis
Insertion: Pes anserinus

Musculus biceps femoris
Action: Flexes the knee and rotates the tibia laterally. Extends the hip joint, adducts and laterally rotates the thigh
Origin: Caput longum: Tuber ischiadicum, Caput breve: Labium laterale linea aspera and septum intermusc. laterale
Insertion: Caput fibulae

Musculus semitendinosus
Action: Extends, adducts and medially rotates the thigh. Flexes the knee, and rotates the tibia medially, especially when the knee is flexed
Origin: Tuber ischiadicum
Insertion: Pes anserinus, distal for musculus gracilis and musculus sartorius

Musculus semimembranosus
Action: Extends and adducts the thigh. Flexes the knee, and rotates the tibia medially, especially when the knee is flexed
Origin: Tuber ischiadicum
Insertion: Posterior surface of the medial tibial condyle

Musculus popliteus
Action: Rotates knee medially
Origin: epicondylus lateralis femoris
Insertion: Facies posterior tibiae over linea musculus solei

Musculus gastrocnemius
Action: Plantar flexion of ankle. Flexes the knee.
Origin: Caput mediale: condylus medialis ossis femoris, facies poplitea. Caput laterale: epicondylus lateralis ossis femoris, facies poplitea
Insertion: Tuber calcanei via tendo calcaneus

Musculus plantaris
Action: Plantar flexion of ankle. Flexes the knee.
Origin: Proximal for condylus lateralis ossis femoris

Insertion: Tuber calcanei

Musculus tensor fascia lata
Action: Stabilize the hip and knee joints by putting tension on the iliotibial band of fascia
Origin: Spina iliaca anterior superior, outer lip of anterior iliac crest and fascia lata
Insertion: Iliotibial band (tractus iliotibialis)

Musculus gluteus maximus
Action: Major extensor of hip joint
Origin: Posterior to linea glutealis posterior, posterior superior crista iliaca, posterior inferior aspect of os sacrum and os coccygis, and ligamentum sacrotuberous.
Insertion: Tractus iliotibialis and turberositas glutealis

Flexion	**Extension**	**Medial rotation**	**Lateral rotation**
M. sartorius	M. rectus femoris	M. sartorius	M. biceps femoris
M. gracilis	M. vastus lateralis	M. gracilis	
M. biceps femoris	M. vastus intermedius	M. semitendinosus	
M. semitendinosus	M. vastus medialis	M. semimembranosus	
M. semimembranosus		M. popliteus	
M. gastrocnemius			
M. plantaris			

Table B.1: Overview of the muscles that flex, extend and rotate the knee

B.2 Other muscles attached to the *tibia*

Musculus tibialis anterior
Action: Dorsiflexes foot at ankle and inverts foor
Origin: Lateral condyle of tibia, lateral surface of tibial shaft, interosseous membrane, and fascia cruris
Insertion: Medial and plantar surfaces of 1st cuneiform and on base of first metatarsal

Musculus extensor digitorium longus
Action: Dorsiflexes foot at ankle and extends toes 2-5.
Origin: Lateral condyle of fibula, medial fibular shaft surface, interosseous membrane and fascia cruris

Insertion: Splits into 4 tendon slips which each insert on dorsum of middle and distal phalanges

Musculus extensor hallicus longus
Action: Dorsiflexes and inverts foot at ankle. Extends toe no. 1
Origin: Anterior surface of the fibula and the interosseous membrane
Insertion: Base and dorsal center of distal phalanx of toe no. 1

Musculus peroneus tertius
Action: Dorsiflexes and everts foot at ankle
Origin: Medial fibular shaft surface
Insertion: Dorsal surface of the base of the fifth metatarsal

Musculus soleus
Action: Plantar flexes the foot
Origin: Posterior aspect of fibular head, upper 1/4 - 1/3 of posterior surface of fibula, middle 1/3 of medial border of tibial shaft, and from posterior surface of a tendinous arch spanning the two sites of bone origin
Insertion: Achilles tendon, inserting on the middle 1/3 of the posterior calcaneal surface

Musculus flexor hallucis longus
Action: Plantar flexes foot and flexes toe no. 1
Origin: Inferior 2/3 of posterior surface of fibula, lower part of interosseous membrane
Insertion: Plantar surface of base of distal phalanx of toe no.1

Musculus flexor digitorium longus
Action: Plantar flexes foot and flexes toes no. 2-5
Origin: Posterior surface of tibia distal to popliteal line
Insertion: Splits into four slips which insert on plantar surface of bases of 2nd - 5th distal phalanges

Musculus tibialis posterior
Action: Plantar flexes, inverts and addducts foot
Origin: Posterior aspect of interosseous membrane and posterior surface of fibula.
Insertion: Splits into two slips which inserts on the tuberosity of the navicular bone; deeper slip divides again into slips inserting on plantar sufraces of metatarsals 2 - 4 and second cuneiform

Musculus proneus longus
Action: Plantar flexes and everts foot at ankle
Origin: Head of fibula, lateral surface of fibular and fascia curis

Insertion: Plantar posterolateral aspect of medial cuneiform and lateral side of 1st metatarsal base

Musculus proneus brevis
Action: Plantar flexes and everts foot at ankle
Origin: Inferior 2/3 of lateral fibular surface; also anterior and posterior intermuscular septa of leg
Insertion: Lateral surface of styloid process of 5th metatarsal base

Physiotherapeutic treatment

This appendix describes the rehabilitation strategies at a specific hospital in Denmark, Ålborg Sygehus. The information is based on a meeting with physiotherapist Anette Ottosen, who works at the physiotherapeutic ward at Ålborg Sygehus. The writer has also participated in one physiotherapist Gitte Brandtofts rehabilitation sessions at Ålborg Sygehus for patients with knee injuries.

C.1 Rehabilitation

The purpose of the rehabilitation program is, that the knee regains full range of movement, muscle strength and does not give way during everyday activities. The treatment consists of movement, balance, strength and function training, which comprises training of proprioception.

Training principles

- No pain
- No swelling
- Everyday movements: The exercises must imitate the movements the patients have to perform in everyday life. It does not help to strengthen to muscles with unnatural movements that the patient normally would not do.
- Stability and neuromuscular control before strength: The muscles should initially be strengthened with low weight and a lot of repetitions. If the load is too high the stronger muscles will take over and the weaker muscles will not improve. There is particularly focus on strengthening the lower part of vastus medialis - vastus madialis obliqus - since this muscle is believed to be important for the stability of the knee.
- Stretching: The muscles are stretch because overactive muscles inhibit other muscles. There should be an "optimal" muscle balance.

- Uniformness: The condition in the damaged knee should be as identical to the healthy knee as possible. The knees should for example have equal mobility. A hyper extended knee is generally bad; but if the healthy knee can be hyper extended so should the damaged knee. This is done because of the idea that if the knees are alike they will be loaded equally. If the patient had a bad habit of hyper extending the knees and the damaged knee no longer can be hyper extended, the patient will hyper extend just the healthy knee, which means that this knee will be exposed to excess loading.
- Alignment: The legs should have a good alignment, i.e. no valgus or varus posture. If the patient's feet are overpronating, tibia naturally fall into a valgus position. I that case alignment of the legs can be improved by improving the patient's pronation. Weak buttock muscles will allow femur to rotate medially which likewise lead to a valgus movement and should be counteracted by strengthening the buttocks.
- No compensation.

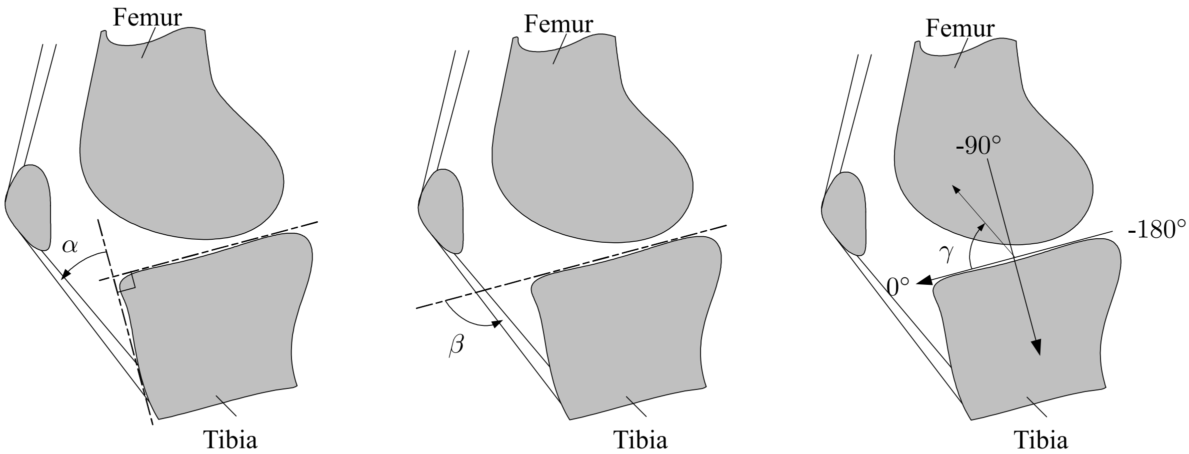
The training target the body as a whole and not just the knee. The rehabilitation does not focus on how to stabilise an anterior cruciate ligament deficient knee, but instead on which exercises generally benefit the knee and the rest of the body. It was observed that patients with different knee injuries was assigned to the same rehabilitation programs and was instructed to perform the same exercises.
Ottosen confirmed that everyone has an opinion on, what the best way the rehabilitate a knee is, and there has been developed a lot of different treatment programs. Unfortunately it is very difficult to compare the effect of the different treatment strategies, which is probably the reason why the rehabilitation programs seem to be controlled by the local traditions at the hospital and not evidence.

The lack of evidence can be exemplified by training of proprioception. Proprioception is the body's ability to perceive muscle force, position and movement of the body parts. The proprioceptive sense is believed to be composed of information from sensory neurons located in the inner ear and in receptors in muscles, tendons, skin, and joint-supporting ligaments.
Earlier it was believed that the cruciate ligaments only had a mechanical function. However, there has been found mechanoreceptors in the ligaments, which indicate that the ligaments contribute to the proprioceptive sense. In addition to this there has been prove a reflex in the ligaments which suggests that the receptors is related to the neuromuscular function of the knee.
At the physiotherapeutic ward at Ålborg Sygehus, it is believed that training of proprioception is important to regain control of the knee and that it is possible to obtain new receptors in a reconstructed ligament, but there is no evidence for these ideas. Clinical examinations have not been able to prove reduced proprioception after an anterior cruciate ligament rupture, while some experimental studies indicate that there is no difference in proprioception and some indicate that it has been reduced. It is therefore unclear whether the mechanoreceptors are necessary for the proprioceptive perception of the knee, (SAKS,

2006). Rehabilitation of knee injuries should comprise training of proprioception, because it has been proven to have a positive effect on strength and a prophylactic effect on the risk of knee injuries, (SAKS, 2006), but this doesn't seem to be the reason why the knee patients at Ålborg Sygehus are instructed to improve proprioception.

Patella tendon angle

The patella tendon angles as function of knee flexion was computede in the dynamic analysis of the lunge model and compared with results from (Baltzopoulos, 1995, Figur 7, page 90) and (Herzog and Read, 1993, Table 1, page 220). In the lunge model the patella tendon angle was given as the smallest angle between the tendon and a line perpendicular to the tibial plateau (Figure D.1(a)), but the two studies used two different references when measuring the tendon angle, why the angles had to be translated in order to be able to compare them.



(a) Lunge model (b) Baltzopoulos (c) Herzog and Read

Figure D.1: Three different definitions of the patella tendon angle

In the lunge model the tendon produce an anterior drag in the tibia when $\alpha > 0°$ and a posterior drag when $\alpha < 0°$, i.e. the angel of transition between anterior-posterior drag from the quadricaeps was $\alpha = 0°$.

Baltzopoulos (1995) provided the orientation of the patella tendon as the angel between the tendon and a line parallel with the tibial plateau, Figure D.1(b). The tendon produce an anterior drag in the tibia when $\beta > 90°$ and a posterior drag when $\beta < 90°$, i.e. the angel of transition between anterior-posterior drag from the quadricaeps was $\beta = 90°$.
In order to compare the results from Baltzopoulos (1995)with the lunge model, the patella tendon angle was computed as $\alpha = \beta - 90$

Herzog and Read (1993) used the tibial reference system shown in Figure D.1(c) and approximated the patella tendon's line of action with the polynomial regression equation (D.0), where θ is the knee flexion.

$$\gamma = -0,744 \cdot 10^2 - 0,575 \cdot 10^{-1} \cdot \theta - 0,475 \cdot 10^{-2} \cdot \theta^2 + 0,309 \cdot 10^{-4} \cdot \theta^3 \qquad \text{(D.0)}$$

The tendon produce an anterior drag in the tibia when 0° $>\gamma>$-90° and a posterior drag when -90° $>\gamma>$-180°, i.e. the angel of transition between anterior-posterior drag from the quadricaeps was $\gamma=$-90°.
In order to compare the results from Herzog and Read (1993)with the lunge model the patella tendon angle was computed as $\alpha = \gamma + 90$

Figure D.2 shows the patealla tendon angles α, β, γ together with the translated angels.

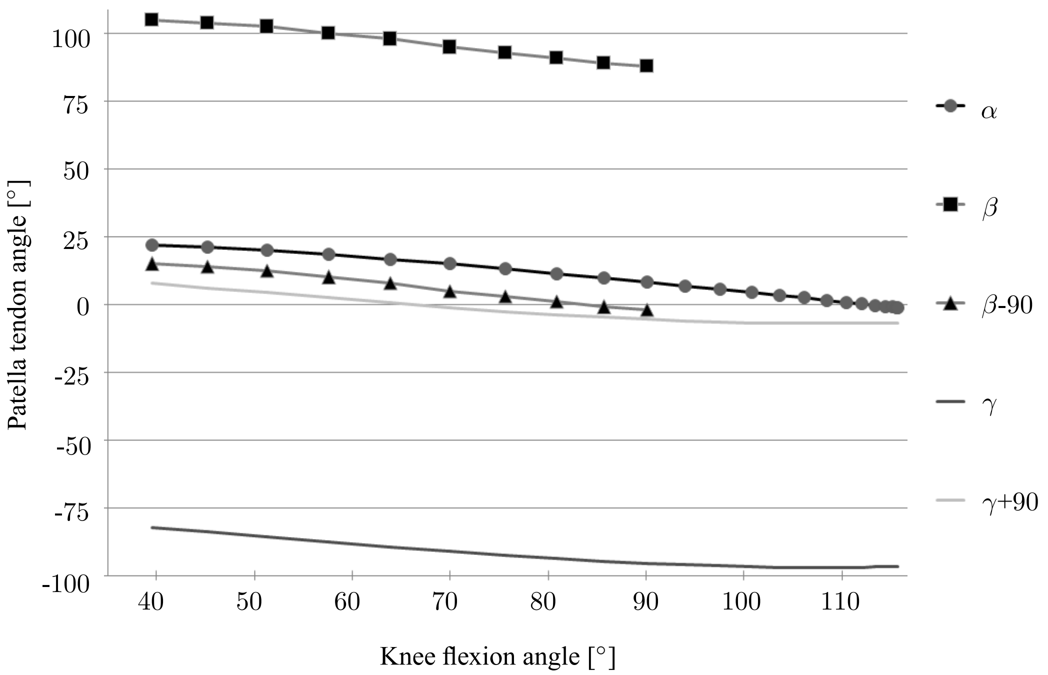


Figure D.2: Patella tendon angles from the lunge model compared with results from Baltzopoulos (1995) and Herzog and Read (1993). α, β, γ refer to Figure D.1

Quadriceps force

This appendix contains additional results from the analysis of the lunge model.

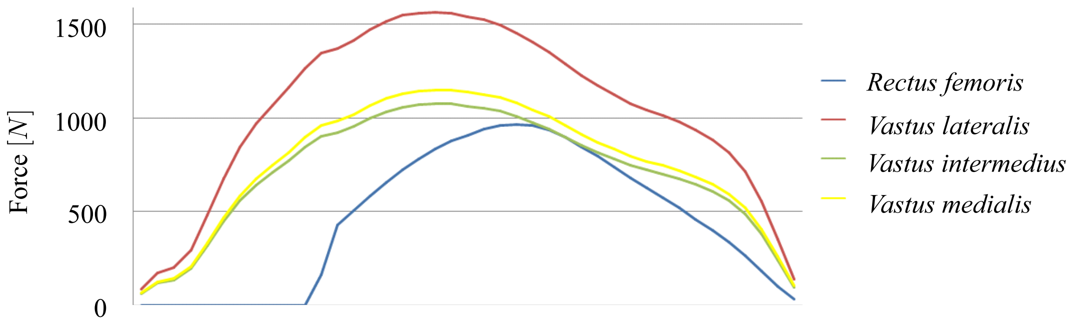


Figure E.1: Forces in muscle-tendon units in the quadriceps during the lunge movement

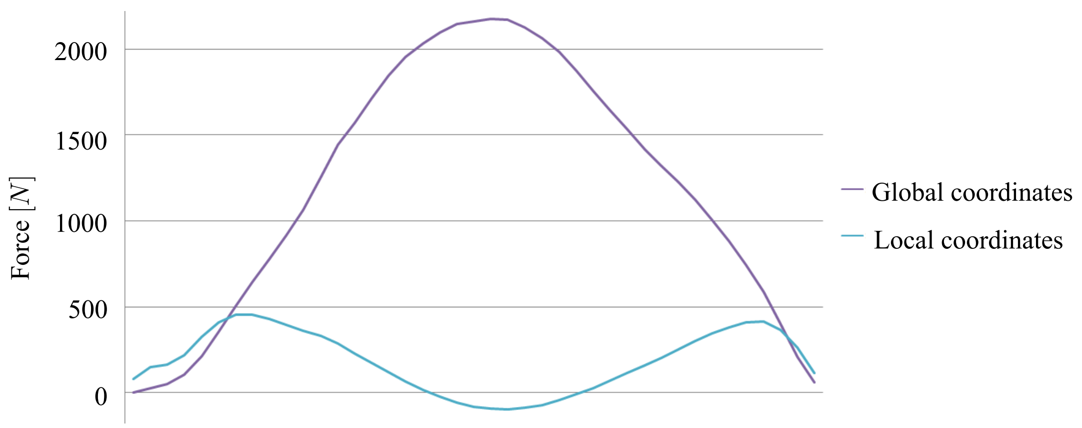


Figure E.2: The x-component of the quadricps drag in the tibia, Q_x. Q_x depends on the chosen reference coordinate system. One curve gives Q_x in global coordnates, while the other curve refer to the tibia's local coordinate system.

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