Optimization of the Implantation Angle for a Talar Resurfacing Implant: A Finite Element Study

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Optimering av Implanteringsvinkeln för ett Ytåterskapande Talusimplantat: En Studie Utförd med Finita Element Metoden

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Abstract

Osteochondral lesions of the talus (OLTs) are the third most common type of osteochondral lesion and can cause pain and instability of the ankle joint. Episurf Medical AB is a medical technology company that develops individualized implants for patients who are suffering from focal cartilage lesions. Episurf have recently started a project that aims to implement their implantation technique in the treatment of OLTs.

This master thesis was a part of Episurf’s talus project and the main goal of the thesis was to find the optimal implantation angle of the Episurf implant when treating OLTs. The optimal implantation angle was defined as the angle that minimized the maximum equivalent (von Mises) strain acting on the implant shaft during the stance phase of a normal gait cycle. It is desirable to minimize the strain acting on the implant shaft, since a reduction of the strain can improve the longevity of the implant.

To find the optimal implantation angle a finite element model of an ankle joint treated with the Episurf implant was developed. In the model an implant with a diameter of 12 millimeters was placed in the middle part of the medial side of the talar dome. An optimization algorithm was designed to find the implantation angle, which minimized the maximum equivalent strain acting on the implant shaft. The optimal implantation angle was found to be a sagittal angle of 12.5 degrees and a coronal angle of 0 degrees. Both the magnitude and the direction of the force applied to the ankle joint in the simulated stance phase seemed to influence the maximum equivalent strain acting on the implant shaft.

A number of simplifications have been done in the simulation of this project, which might affect the accuracy of the results. Therefore it is recommended that further, more detailed, simulations based on this project are performed in order to improve the result accuracy.
Sammanfattning

Fokala broskskador på talusbenet är den tredje vanligaste typen av fokala broskskador och kan ge upphov till smärta och instabilitet av fotleden. Episurf Medical AB är ett medicintekniskt företag som utvecklar individualpassade implantat för patienter med fokala broskskador. Episurf har nyligen påbörjat ett projekt där deras teknik ska användas i behandlingen av fokala broskskador på talusbenet.


Ett antal förenklingar har gjorts i projektets simuleringar, vilket kan påverka noggrannheten i resultatet. Därför rekommenderas att ytterligare, mer detaljerade simuleringar baserade på det här projektet görs för att förbättra resultatets noggrannhet.
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Terminology

**Isotropic material** – Same material properties in all directions.

**Anisotropic material** – Different material properties in different directions.

**Viscoelastic material** – A material that is both viscous and elastic.

**Dorsiflexion** – When the foot is flexed upwards, towards the shank.

**Plantarflexion** – When the foot is flexed downwards. The opposite to dorsiflexion.

**Inversion (of the foot)** – Turning of the foot medially resulting in the sole of the foot to face the medial sagittal plane.

**Eversion (of the foot)** – Turning of the foot laterally tilting the sole of the foot away from the medial sagittal plane. The opposite of inversion.

**Abduction (of the foot)** – Moving the foot away from the midline of the body.

**Adduction (of the foot)** – Moving the foot towards the midline of the body. The opposite of abduction.

**Anterior** – Front.

**Posterior** – Back. The opposite of anterior.

**Distal** – Located far from an anatomical reference point.

**Proximal** – Located close to an anatomical reference point. The opposite of distal.

**Superior** – Above.

**Inferior** – Below. The opposite of superior.

**Medial** – Middle of the body.

**Lateral** – Left or right of the body. The opposite of medial.

**Transverse** – Horizontal plane through the body.

**Coronal** – Vertical plane going through the body from the right to left.

**Sagittal** – Vertical plane going through the body from the anterior to posterior.
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1. Introduction

The talus, together with the tibia and fibula, forms the ankle joint. The ankle joint can simply be described as a hinge joint and is the main contributor to the dorsiflexion and the plantarflexion of the foot. [1] Around 60 percent of the surface of the talus is covered in cartilage. [2]

Osteochondral lesions of the talus (OLT) refer to damages in the talar cartilage and subchondral bone of the talar dome. OLTs are the third most common type of osteochondral lesion; only lesions at the knee and the elbow are more commonly occurring. [3] The majority of OLT cases have a history of ankle trauma. [4] Some of the symptoms that patients with OLTs can experience are pain and swelling of the ankle. The joint can also be unstable and the patient can experience that the joint is clicking and/or being blocked. [2] Severe OLT cases can, if left untreated, lead to arthrosis. [3]

Today several different methods are used in the treatment of OLTs, involving both non-surgical techniques (e.g. resting and anti-inflammatory drugs) and surgical techniques (e.g. bone-marrow stimulation and grafting). [5] A problem that the most common surgical techniques have in common is the inability to perfectly match the surface contour of the patient’s articular cartilage.

Episurf Medical AB is a medical technology company that develops individualized implants for patients who are suffering from focal cartilage lesions. Episurf have recently started a project that aims to implement their implantation technique in the treatment of OLTs. The developed system will include an individualized cobalt-chrome implant, an osteotomy guide and a drill guide. The osteotomy guide is used to lead the surgeon through a tibial osteotomy, which is needed to access the damaged part of talar surface. The drill guide shows the surgeon where to drill an implantation hole, in which the resurfacing implant is to be placed. The size of the osteotomy and the position of the implant will be decided pre-operatively. The surface of the implant will be custom shaped to fit the patient’s talar topography.

This master thesis is a part of Episurf’s talus project and the main goal of the thesis is to find the optimal implantation angle of the Episurf implant when treating osteochondral lesions of the talus. The optimal implantation angle is the angle that minimizes the strains acting on the implant. To find this angle the talus joint is simulated during the stance phase of a normal gait cycle with the use of the finite element method and an optimization algorithm is designed with the purpose of finding the angle where the strains are minimized. It is desirable to minimize the strain acting on the implant, since a minimized strain can increase the longevity of the implant. It is also of interest to
evaluate how the strain in the implant depends on the implantation angle since the implantation angle determines the required size of the osteotomy needed to access the talar surface. A small osteotomy is desirable, since a smaller osteotomy is a less invasive procedure than a larger.

The implantation angle in both the sagittal and the coronal plane will be evaluated. Only implantation treatment of OLTs located in the most common area of the articular surface of the talus, with a diameter of 12 millimeters will be evaluated in this report.
2. Background

The background section of this report starts with a detailed description of the anatomy and anthropometry of the different components of the ankle joint, followed by some basic solid mechanics concepts and the finite element method. Thereafter the kinematics and kinetics of the ankle joint during gait is described, as well as the structure and biomechanics of bone and cartilage. Some previous finite element simulations of the ankle joint and talus are then illustrated. The section also contains background information about osteochondral lesions of the talus, where topics such as probable causes and diagnosis are discussed. Last in this section OLT treatment methods are described.

2.1. Ankle joint anatomy and anthropometry

The distal end of the fibula and tibia together with the talus bone forms the ankle joint, see Figure 1. The talus is held on its lateral, medial and superior side by the lateral malleolus, medial malleolus and the tibial plafond. The ankle joint is the main contributor to the plantar- and dorsiflexion of the foot. [1]

![Figure 1 Lateral view of some bones of the ankle joint and foot.](image)

2.1.1. Distal fibula and tibia

The lateral malleolus is a part of the distal fibula. At the medial surface of the lateral malleolus there is a triangular shaped articular cartilage surface that articulates with the lateral surface of the talus. [6] The articular cartilage of the distal fibula has a thickness of around 0.95±0.17 millimeters. [7]
The tibial plafond is the inferior surface of the distal tibia and it articulates with the dome shaped upper part of the talus bone, called the talar dome. The medial malleolus is a part of the distal tibia. The medial malleolus lateral surface articulates with the medial surface of the talus. [6] The articular cartilage thickness varies over the distal tibial surface. Over the tibial plafond the cartilage thickness is between 1.20 and 1.30 millimeters, and over the medial malleolus lateral surface the thickness is 0.97±0.16 millimeters. [7]

2.1.2. Talus

Around 60 percent of the talar surface is covered in cartilage. [2] The talus is attached to the adjacent bones only by ligaments and has no muscle attachments. In addition to the distal tibia and fibula, the talus also articulates with the calcaneus and the navicular bone, see Figure 1. The talus consists of three parts: the head (caput), neck (collum) and body (corpus tali), see Figure 2.

![Head, neck and body of the talus](image)

Figure 2 Head, neck and body of the talus. Right: superior view. Left: lateral view.

The head of the talus is located in the anterior portion of the talus. The anterior part of the talar head articulates with the navicular bone and the inferior part of the head articulates with the calcaneus bone and the calcaneonavicular ligament.

The neck of the talus connects the body of the talus with the head of the talus. The neck deviates downward relative to the talar body in the sagittal plane in an angle called the inclination angle. The neck also deviates from the long axis of the talar dome in the transverse plane in an angle called the declination angle.

The body of the talus makes up the biggest part of the talus. The superior surface of the talar body faces the distal end of the tibia and its surface curvature interlock with its tibial counterpart. The surface curvature is anteroposteriorly convex and transversely slightly concave. The anterior part of the body surface is wider than the posterior part. The medial edge of the surface is slightly lower then the lateral side and the medial edge are straight, while the lateral is oblique. On the lateral surface of the talar body there is an articular surface facing the lateral malleolus and on the medial surface of the talar body there is a comma shaped articular surface facing the medial malleolus. The inferior
surface of the body of the talus articulates with the calcaneus. [6] The thickness of the articular cartilage varies over the body of the talus. Athanasiou et al. [7] studied the cartilage of the proximal talus and found that the thickness of the superior surface of the talar body varies between 1,01 and 1,45 millimeters, where the posterior cartilage is thicker than the anterior. The thickness of the cartilage on the talar body lateral and medial surface was found to be 1,14±0,23 millimeters and 1,18±0,24 millimeters respectively. Overall, the cartilage was found to be thicker in men than in females.

The physical appearance of the talus bone can vary. By examining 100 tali, Sarrafi-an [6] found an average talar width of 37 millimeters, a minimum of 30 millimeters and a maximum of 45 millimeters, and an average talar length of 48 millimeters, a minimum of 40 millimeters and a maximum of 60 millimeters. Saffarian [6] also found an average inclination angle of 114 degrees, a minimum of 95 degrees and a maximum of 140 degrees, and an average declination angle of 24 degrees, a minimum of 10 degrees and a maximum of 44 degrees.

Isman and Inman [8] made an anthropometric study of the foot and ankle of 46 human cadaver legs. They studied the angle between the midline of the foot and the axis of the talocalcaneal joint and the angle between the horizontal plane and the axis of the talocalcaneal joint and found quite a wide range of angles. The angle between the midline of the foot and the talocalcaneal joint varied between 4 degrees and 47 degrees with an average value of 23±11 degrees, and the angle between the horizontal plane and the talocalcaneal joint varied between 20 degrees and 68 degrees with an average value of 41±9 degrees. Isman and Inman [8] also studied the angle between the midline of the foot and the axis of ankle joint and found that the angle varied 69 degrees and 99 degrees with an average value of 84±7 degrees.

### 2.1.3. Ligaments

The deltoid ligament attaches the medial malleolus to the talus, navicular and the calcaneus bone and stabilizes the ankle medially by resisting the valgus forces to the ankle. [9, 10] The deltoid limits plantarflexion, dorsiflexion, eversion and abduction of the foot.

The tibia and the fibula are joined together by the anterior, posterior and transverse tibiofibular ligaments, as well as the interosseous ligament. [9]

The anterior and posterior talofibular ligament, as well as the calcaneofibular ligament stabilize the lateral side of the foot. [10] The anterior talofibular ligament attaches the lateral malleolus to the talus and it limits plantarflexion and inversion of the foot. The calcaneofibular ligament attaches the lateral malleolus to the calcaneus and limits inversion of the foot. The posterior talofibular ligament attaches the lateral malleolus to the posterior talar surface and it supports the lateral part of the ankle by limiting plantarflexion, dorsiflexion and inversion. [9]
The anterior and the posterior talotibial ligaments are two of the medial ligaments of the ankle. The anterior talotibial ligament attaches the anterior tibia to the talus and it limits plantarflexion and abduction of the foot. The posterior talotibial ligament attaches the posterior tibia to the talus and is limits plantarflexion. [9] Some of the ligaments of the ankle joint can be seen in Figure 3.

![Figure 3 Some ligaments of the ankle joint.](image)

### 2.2. Basic solid mechanics and the finite element method (FEM)

#### 2.2.1. Basic solid mechanics

One of the fundamental concepts in solid mechanics is stress. Stress is defined as force per unit area and has the measurement unit Pascal (Pa). When stress is evenly distributed over an area the force in each point can be found by calculating the mean stress, which is defined as

\[
\sigma = \frac{N}{A}
\]

(1)

Where \( \sigma \) is the mean stress in Pa (N/m\(^2\)), \( N \) is the total force over the area in Newton (N) and \( A \) is the cross-section over which the force is distributed in m\(^2\). [11]

Strain is also a fundamental concept in solid mechanics and is a measurement of the degree of deformation. Nominal strain is defined as the deformation per unit of the original length and has no unit of measurement

\[
\varepsilon = \frac{\delta}{L_o}
\]

(2)

Where \( \varepsilon \) is the nominal strain, \( \delta \) is the deformed length and \( L_o \) is the original length. [11]
Young’s modulus or modulus of elasticity is a material constant that describes the stiffness of the material. In linear elastic materials Hook’s law can be used to describe the relationship between the mean stress, nominal strain and Young’s modulus

$$\sigma = E\varepsilon$$  \hspace{1cm} (3)

Where $\sigma$ is the mean stress in Pa (N/m$^2$), $E$ is the Young’s modulus in Pa (N/m$^2$) and $\varepsilon$ is the nominal strain. [11]

Poisson’s ratio is a material specific constant and is defined as follows

$$\nu = -\frac{\varepsilon_{\text{transverse}}}{\varepsilon_{\text{axial}}}$$  \hspace{1cm} (4)

Where $\nu$ is the Poisson’s ratio, $\varepsilon_{\text{transverse}}$ is the nominal strain in the transverse direction of the applied force/deformation and $\varepsilon_{\text{axial}}$ is the nominal strain in the axial direction of the applied force/deformation. [11]

2.2.2. Finite element method (FEM)

The finite element method (FEM) is frequently used in many engineering science fields, such as fluid dynamics and structural mechanics. FEM is a numerical method that is used to find an approximate solution to problems that are often too complicated to solve analytically. FEM divides the problem into smaller pieces, called elements. Each element has nodes located on the element vertices and edges, which connect the elements to each other. For each element, the solution of differential equations describing the kinematics of the problem is calculated and, by connecting the element simple solutions, an approximate solution for more a complex problem can be found. [12, 13] There are many available computer software that are used for FEM simulations, for instance ANSYS Workbench.

2.3. The ankle joint during gait

2.3.1. Gait

A gait cycle is defined as “the interval from one event on one limb until the same event on the same limb in the following contact” [9, p320]. The gait cycle consists of a stance phase and a swing phase. The stance phase is the part of the gait cycle when the foot is in contact with the ground and the swing phase is the part of the gait cycle when the foot does not have contact with the ground. The stance phase make up around 60 percent of the human gait cycle and the remaining 40 percent is the swing phase. The stance phase is initiated by heel strike, followed by foot flat, heel off and finally toe off.
The angles of the foot during a gait cycle can be seen in Figure 4, where positive values represent dorsiflexion and negative values represent plantarflexion.

![Figure 4 Ankle joint angles during one gait cycle, where positive angles represent dorsiflexion and negative angles represent plantarflexion. Based on data from Winter [14].](image)

Stauffer et al. [15] found that the compressive force acting on the ankle joint varies over the stance phase according to Figure 5 and that it peaks at around 4.5 to 5.5 times the body weight. The normal subjects evaluated by Stauffer et al. [15] had a mean free-walking cadence of 58 strides per minute, giving a mean stance duration time of 0.62 seconds.

![Figure 5 Compressive force acting on the ankle joint during the stance phase of the gait cycle, in percent of body weight. From Stauffer et al. [15].](image)

### 2.3.2. Axes of rotation

Many researchers have investigated whether the ankle joint can be described as a hinge joint with a fix axis or not. Barnett and Napier [16] suggest that the ankle joint axis of rotation can be found by studying the curvature of the upper articular surface of the talus. In their study they measured the upper articular surface of 152 tali and found a difference in the radii of the curvature of the medial and lateral profiles of the surfaces. They concluded that the ankle joint axis of rotation is changing during motion and that during dorsiflexion the axis is inclining downwards and laterally, while during plantarflexion the axis is inclining downward and medially. Arndt et al. [17] could also find indications that the ankle axis of rotation changed during the gait cycle by doing kinematic evaluation of data collected from intracortical pins inserted in the tibia and talus of three living subjects. In contrast Singh et al. [18] found that the ankle joint have
a single fix axis of rotation. After evaluating the motions of six cadaver joints they found that the axis of rotation was located just below the distal tips of the malleoli.

In their literature review of the ankle joint biomechanics and ankle joint modeling, Dettwyler et al. [19] found that more detailed kinematic investigations of the ankle joint axis of rotation showed that the ankle joint cannot simply be described as a hinge joint, while less detailed studies found that the ankle joint acts as a hinge joint. They concluded that when it comes to simple modeling of the ankle joint, the joint can be described as a hinge joint with a fix axis, and when it comes to more detailed modeling the joint should be described as a joint with a varying axis of rotation.

2.4. Structure and biomechanics of bone and cartilage

Bone is composed of a dense outer layer of cortical bone and an interior of more porous cancellous bone. Around 80 percent of the total weight of human bone is made up of cortical bone, and the rest of cancellous bone. [9]

2.4.1. Cortical bone

Cortical bone is made up of parallel cylindrical structures called osteons or Haversian systems. In long bones the osteons run parallel to the long axis of the bone. In the center of each osteon there is a central canal called the Haversion canal. [9, 20] The Haversion canal contains blood vessels, lymphatic fluid and nerves and is surrounded by several layers of concentric lamellae. [20] The lamellae in turn consist of collagen fibers. The collagen fibers in a single lamella run parallel to each other, while the collagen fibers of neighboring lamellae run in different directions. [9]. The bone’s collagen matrix is filled with inorganic calcium compounds, mainly hydroxyapatite. [21]

Cortical bone has anisotropic mechanical properties due to the orientation of the osteons and their lamellae. The cortical bone strength and stiffness in the longitudinal direction is of greater magnitude than in the transverse direction. Stronger in compression than in tension, the cortical bone is suited for typical loading conditions, where it is usually loaded along its axis in compression. [22]

Both cortical bone and cancellous bone have viscoelastic material properties, making the bone stronger and stiffer when subjected to higher strain rates. However, to large extent, the bone can be considered as a linear elastic material. [22]

2.4.2. Cancellous bone

Cancellous bone, just as cortical bone, is made out of lamellas, but unlike cortical bone cancellous bone does not contain any osteons. Instead the lamellae form irregular patterns of thin beams, called trabecule. The collagen in lamellae of cancellous bone
runs along the axis of the trabecule, which give the cancellous bone tensile and compressive resistance. [20] Between the trabecule of the cancellous bone there is space, which makes cancellous bone less dense than cortical bone. [9] The space between the trabecule is filled with red bone marrow in blood cell forming bones and with yellow bone marrow in other bones. [20]

The pattern of the trabecule changes over time and adapts to the direction of the stress the bone is subjected to. Since bone adapts its geometry to the stresses it is subjected to the introduction of an implant in the bone, which can cause a reduction of the normal stresses acting on the bone, can lead to a loss of bone density, a phenomenon called stress shielding.

The mechanical properties of cancellous bone depend strongly on the trabecule relative density and architecture, which in turn differs between anatomical sites, ages etc. Cancellous bone behaves as linear elastic materials at small strains. At larger strains the trabecular matrix starts to collapse, resulting in an end to the linear elastic behavior and a plateau in the cancellous bone compressive stress-strain curve. At even larger strains the trabecule come in contact with each other, eliminating the space between the trabecule, putting an end to the plateau and making the stress rise steeply in the compressive stress-strain curve. [22]

When comparing cancellous and cortical bone mechanical properties one find that cortical bone has more strength than cancellous bone and that cancellous bone can undergo more deformation before failure than cortical bone. [9]

2.4.3. Articular cartilage

Articular cartilage covers articulating surfaces of freely moving joints and provides the joint with an almost frictionless joint surface. Articular cartilage consists of cells called chondrocytes and a collagen and proteoglycan matrix, and contains between 60 to 80 percent of water. Articular cartilage is avascular and gets its nutrition from the joint synovial fluid (joint fluid). [9]

Articular cartilage can be divided into four zones: the superficial, middle, deep and calcified zone. The superficial zone is located near the articulating part of the cartilage and make up around 10 to 20 percent of the cartilage thickness. In the superficial zone the collagen fibers run parallel to the articular surface in dense fibrils. In the zone below the superficial zone, called the middle zone, the fibrils are less organized. The middle zone makes up around 40 to 60 percent of the cartilage thickness. The deep zone make up the next 20 to 30 percent of the cartilage thickness and the zone's collagen fibrils are organized perpendicular to the subchondral bone. The fourth zone, the calcified zone, binds the articular cartilage to the subchondral bone. [22]

Articular cartilage is just as cortical and cancellous bone anisotropic. When loaded, cartilage also acts as a viscoelastic material. Its stiffness increases and it deforms more
slowly when it is subjected to a rapid load and deforms immediately when loaded with a
low or moderate force. The articular cartilage thickness has an influence over how
forces are distributed over an articular surface. The forces over an articular surface in
turn decide the stress distribution in the cartilage. [9]

2.5. Other FEM simulations of the ankle joint and talus

Finite element simulations of the talus and the ankle joint have been used over the years
with the purpose to evaluate the kinetics, kinematics and mechanical performance.

Dul and Johnson [23] modeled the kinematics of the lower human leg that included two
joints: the ankle joint and the subtalar joint. The model consisted of the shank, talus and
foot and was joined together with the ankle joint and the subtalar joint. The two joints
was modeled a hinge joints, i.e. they had one rotational degree of freedom and no
translation. The locations of the ankle joint axis of rotation were approximated as the
line between the two malleoli of the foot.

Many of the FEM simulations of the ankle joint are simulations of total ankle
replacements (for more details regarding total ankle replacements see 2.7.2. Surgical
treatment methods). Falsig et al. [24] did three simulations of total ankle replacements
with different tibial total ankle replacement components and evaluated the stresses in
the distal tibia. Falsig et al. [24] modeled cortical bone as a homogeneous, isotropic and
linear elastic material and cancellous bone either as a homogeneous, isotropic and linear
elastic material or a heterogeneous material. In their simulation they loaded the tibial
component joint surface with a single force of approximately three times the body
weight. Reggiani et al. [25] made simulations of a total ankle replacement that included
the ligaments of the ankle and evaluated the overall kinematics, contact pressures and
ligament forces in unloaded conditions and during a simulated stance phase of gait.
Bouguecha et al. [26] developed a strain-adaptive FE models to evaluate the bone loss
due to stress shielding caused by the total ankle replacement implants. Bouguecha et al.
[26] modeled bone as a homogenous and elastic material. The joint was subjected to a
static force approximately of 5.2 times the body weight.

Anderson et al. [27] have developed a finite element model that aimed to predict the
contact stresses in the ankle joint during the stance phase of normal gait. The geometry
of their model was based on patient-specific CT-data. The model included the distal part
of the tibia and the talus. In the model the bone was modeled as rigid and the articular
cartilage was modeled with isotropic linear elastic material properties (\( E = 12 \) MPa and
\( \nu = 0.42 \)) and a uniform thickness of 1.5 millimeters. Their model did not include any
ankle ligaments. In the model the rotation of the ankle joint was not simplified to a
hinge joint, instead the rotation of the joint was determined by the tibio-talar
articulation, by letting the talus rotate in multiple planes, only restricted by springs, while
the tibia was brought through dorsi- and plantarflexion according to a normal gait cycle.
Anderson et al. [28] was able to validate the above model by comparing the contact results of the finite element simulation with the results from cadaver experiments.

The same modeling approach as Anderson et al. [27] was used in a study by Anderson et al. [29] to evaluate the implantation accuracy of the HemiCAP focal resurfacing implant. In the study Anderson et al. [29] evaluated the effect that the implantation height had on the articular contact area and contact stress.

Parr et al. [30] did a series of simulations where they compared a model that included the cancellous network of the talus with solid non-porous models, by evaluating the mechanical performance of the talus bone. They found that the model that included the cancellous network of the talus was significantly stiffer than the solid non-porous model, suggesting that including the cancellous network in simulations is of importance, since it affects the mechanical performance of the bone.

2.6. Osteochondral lesions of the talus (OLTs)

Osteochondral lesions of the talus (OLTs) refer to damages in the talar cartilage and subchondral bone of the talar dome.

2.6.1. Demographic and causes

The occurrence of OLTs is most frequent in the ages 20-30 years and OLTs are slightly more common in men than women. [31] OLTs are the third most common type of osteochondral lesion; only lesions at the knee and the elbow are more commonly occurring. [3, 32]

There has been an indication that around 6 percent of severe ankle sprains lead to OLTs [33], but this number is probably an underestimation since many OLTs goes undetected. [34] Initially Berndt and Harty [35] suggested that ankle sprains are the single cause of OLTs after being able to reconstruct four types of OLTs in cadavers, but later it has been found that not all OLTs are associated with ankle trauma. Tol et al. [4] conducted a systematic reviewed of data from 1966 to 1998 where 582 OLT patients were included and found that a history of ankle trauma was reported in 76 percent of the patients, where lateral OLTs was more frequently associated with ankle trauma (94 percent) than medial OLTs (62 percent). Therefore other OLT causes have been suggested including repetitive microtrauma, abnormal vascularization, hormonal factors, genetic predisposition and metabolic disorders. [36-38]
2.6.2. Diagnosis

OLTs caused by a traumatic ankle injury are most often not diagnosed acutely, but are usually diagnosed some time after the injury. The lesion may not be visible on plain films or be disregarded in the case of an acute traumatic event. [39]

Some symptoms that can be present in the case of an OLT, are deep aching ankle pain, joint instability, a decreased range of motion and the experience of clicking and/or locking of the joint. Weight bearing intensifies these symptoms. [2, 39] The presence of a loose osteochondral fragment in the joint or a severe OLT can also be indicated by other mechanical symptoms such as grinding or catching. A displaced osteochondral fragment might lead to arthrosis if left untreated. [3, 40] Magnetic Resonance Imaging (MRI) is the non-invasive image modality of choice to diagnose OLTs. [40, 41]

2.6.3. Locations and sizes

The occurrence of OLTs over the talar dome area has been investigated by Elias et al. [32]. With the use of a grid scheme the authors have divided the surface of the talar dome into nine different areas. By examining 428 MR imaged tali that had been reported to have OLTs they could conclude that around 63 percent of the lesions where located in the medial part of the talar dome and that around 33 percent of the lesions where located on the lateral part. The most common OLT location is the mid part of the talar domes medial part, where around 53 percent of OLTs are located. Lesions located on the medial part of the talar dome also have larger surface areas than lesions located on the lateral side of the talar dome. Lesions at the most common location of the talar dome (mid part of the medial side) have an average transverse size of 7.6 millimeters and a anterior-posterior size of 10.3 millimeters. Lesions located on the medial side of the talar dome tend to be cup-shaped and deep whereas lesions located on the lateral side tend to be wafer-shaped and shallow. [42]

2.6.4. Injury classification

The first OLT classification system was formulated by Berndt and Harty [35] in 1959 and was based on four types of injuries that they were able to reproduce in cadaveric ankles by simulating traumatic ankle injuries. Their classification system is based on plain radiography and includes four classification stages:

- Stage I: Subchondral compression
- Stage II: Partial detachment of osteochondral fragment
- Stage III: Completely detached non-displaced fragment
- Stage IV: Detached and displaced fragment
2.7. Available OLT treatment methods

Today several different methods are used in treatment of OLTs, involving both non-surgical techniques and surgical techniques. What treatment method is used depends on the lesion type and size.

2.7.1. Non-surgical treatment methods

Non-surgical treatment is the first option to be considered when it comes to treatment of OLTs. [43] Non-surgical techniques are used for treatment of Berndt and Harty type I and II lesions and small type III lesions. The treatment involves immobilization of the ankle joint and non-weight bearing for around six weeks. In some cases the treatment also include treating the patient with anti-inflammatory drugs. The six weeks of non-weight bearing is followed by physical therapy and increasing weight bearing. [5] In a systematic review of treatment strategies of OLTs Tol et al. [4] found a success rate of 45 percent for non-surgical treatments of OLTs. If non-surgical treatment of Berndt and Harty type I or II lesions fail or for lesions of type III or IV, surgical treatments methods might be necessary. [44]

2.7.2. Surgical treatment methods

In OLT cases when a separated fragment of the talar dome is present, like in Berndt and Harty stage II to IV, a technique called excision can be used. In this treatment method the separated fragment of the talar dome is excised, while the damages area is left untreated. Excision can be combined with curettage. In the excision and curettage method first an excision is done and the necrotic tissue is curettaged from the subchondral bone. [4, 5, 43] This technique is most often used to treat Berndt and Harty stage III or IV lesions and occasionally stage II lesions. [43] Excision and curettage can also be combined with bone marrow stimulation. In this method the excision and curettage is followed by drilling or microfracturing, which create small holes in the subchondral bone. The holes are supposed to increase vascularization to stimulate formation of new cartilage. [4, 5] A drawback with bone marrow stimulation is that the new cartilage formed is fibrocartilage, which do not have the same mechanical properties as hyaline cartilage, which articular cartilage usually is made up of. [45] In the treatment of Berndt and Harty type I or II lesions with a more or less intact articular cartilage, transmalleolar or retrograde drilling technique can be used. The purpose with both techniques is to stimulate growth of new bone. [5] When it comes to surgical treatment methods of OLTs there are indications that the methods excision, excision and curettage, excision, curettage and bone marrow stimulation and subchondral drilling give inadequate results for OLTs with a size larger than 1,5 cm², and are therefore recommended for smaller lesions.

For talar lesions with a size larger than 1,5 cm² the surgical treatments osteochondral transplantation, mosaicplasty and autologous chondrocyte implantation are
recommended. [46] Osteochondral transplantation and mosaicplasty are OLT treatment techniques that reconstruct bone with one or more cylindrical osteochondral grafts, taken from the less weightbearing parts of the knee. With this method the mechanical, biochemical and structural properties are hoped to be reproduced. These methods are most often used on large OLTs, or where previous treatments have failed. [43] A drawback with the osteochondral transplantation and mosaicplasty is the difficulty of matching the surface contour of the grafts with the talar articular surface. Since grafts are taken from the knee, both the surface contour and cartilage thickness differ from the talus. [45] In the autologous chondrocyte implantation method the patient’s own chondrocytes, retrieved from a biopsy of healthy cartilage, is used to treat the OLT. The damaged area of the talus is excised and a periosteal flap (piece of the outer layer of the bone) from the patient’s tibia is sutured in place to cover the area where the lesion was before. Thereafter the retrieved chondrocytes are injected beneath the flap. [43] A technique similar to autologous chondrocyte implantation is matrix-based chondrocyte implantation. In this technique the chondrocytes are embedded in a collagen membrane bilayer, instead of being injected beneath a periosteal flap. The membrane is then used to cover the damaged area of the talus. An advantage with this technique over the autologous chondrocyte implantation is that it does not require an osteotomy. [5]

Refractory lesions larger than 3 cm² can be treated with autologous chondrocyte, allografts, arthrodesis (ankle fusion) or total ankle replacement, where the two latter options are recommended for patients older than 50 years [46, 47] In the allograft treatment method the damaged part of the talus is replaced with cadaveric talar grafts. An advantage with this method is that it can replicate the ankle joint anatomy. Some disadvantages are limited availability of grafts, high costs and the risk of immune rejection. [48] In the arthrodesis treatment method the patient’s ankle joint is fused. The fusing makes the patient’s gait pattern abnormal and slows down the patient’s gait velocity. The fused ankle can also make it difficult for the patient to run or walk on inclines. [49] When treating patients with the total ankle replacement method the patient ankle joint is replaced with a joint prosthesis. One advantage that total ankle replacement has over arthrodesis is that it some joint movement is maintained. The treatment method also has disadvantages like risk of infection, delayed wound healing and occasional fracture. [50]

Another treatment alternative for OLTs are artificial implants that replaces the damaged part of the talar surface. One such implant is used in the Talus HemiCAP implant system developed by Arthrosurface. The implant consists of two parts: an articular component that is supposed to replace the damaged articular surface and a fixation component in the form of a screw that is supposed to fixate the articular component to the bone. The fixation component is supposed to be implanted in an angle perpendicular to the articular surface of the damaged area, which is found during the surgical procedure. A number of standardized premade options of articular components with different topographies are available for the surgeon to choose from during the surgical procedure in order to find the component that best matches the articular
surface topography. In this treatment a malleolar osteotomy is performed in order to access the articular surface of the talus. [51]

2.8. Episurf’s OLT treatment method

A problem that many of the above describe treatment methods (e.g. osteochondral transplantation, mosaicplasty and HemiCAP’s talus implant, see 2.7 Available treatment methods) have in common is the difficulty or inability to produce a congruent articular surface of the talus. With a treatment method from Episurf Medical AB a congruent articular surface of the talus can be reproduced.

Episurf medical AB is a medical technology company that develops individualized articular resurfacing implants and surgical instruments to be used in the treatment of patients suffering from focal cartilage lesions. The Episurf resurfacing implant is a “plug”-like implant that is supposed to replace the damaged area of the articular cartilage in a joint. Based on the patient’s MRI or CT images, the surface of the implant is customized to fit the patient’s cartilage topography and its position is decided preoperatively. Before the implant can be put in position a hole for the implant must be drilled in the area that will be treated. A preoperatively designed drill guide is used to guide the surgeon in the drilling of the implant hole, to ensure that the implant will be placed in the preoperatively planned position. So far Episurf have been able to implement their technique in the treatment of focal cartilage damages of the knee joint. When treating focal cartilage damages in the knee the Episurf implant is placed to be recessed 0.5 millimeter below the surrounding cartilage to prevent damages to the opposing articular surface, which a more proud implant placement might give rise to. [52]

The Episurf resurfacing implant is made out of the material ASTM-F1537, which is a chrome-cobalt-molybdenum alloy often used in surgical implants. The outer surface of the implant that is in contact with bone and articular cartilage is coated with the two materials titanium (ASTM-F1580) and hydroxyapatite (ASTM-F1185), where the latter make up the outer layer. Together the titanium and hydroxyapatite have a thickness of 30±10 micrometer. Both titanium and hydroxyapatite are osseointegratable materials, meaning that they have the ability to bind to bone. Coating titanium with hydroxyapatite has been shown to improve bone ingrowth. [53]

Episurf have recently started a project that aims to implement their resurfacing technique in the treatment of OLTs. In this treatment method, the Episurf implant will be placed recessed to the surrounding cartilage, just as in the treatment of focal cartilage damages in the knee. In addition to the resurfacing implant and the drill guide the developed treatment system also include an osteotomy guide, since an osteotomy of the tibia is necessary to access the damaged part of the talar dome surface. The osteotomy guide is, just as the implant and the drill guide, designed preoperatively to fit the patient’s anatomy. The required size of the osteotomy depends on the angle in which
the implant will be inserted. An implantation angle perpendicular to the surface of the damaged area of the talus will require a larger osteotomy than an implantation angle slanted in the anterior and/or the medial direction. A small size of the osteotomy is desirable since a smaller osteotomy is a less invasive procedure than a larger. On the other hand, a slanted implantation angle might increase the strain acting on the implant shaft, which in turn can lead to an increased risk of implantation failure and a reduced longevity of the implant. Therefore it is of great interest to evaluate how the implantation angle affect the strain acting on the implant shaft, and find the implantation angle that minimizes the strain.
3. Materials and Methods

The main goal of this project was to find the optimal implantation angle of a focal resurfacing implant in the treatment of focal cartilage damages of the talus. The optimal implantation angle was defined as the angle that minimizes the strain acting on the implant shaft during normal gait. In this project the finite element method has been used to approximate the strain acting on the implant shaft. By constructing a finite element model of an ankle joint treated with the implant while it was being brought through the stance phase of a simulated normal gait cycle, the strain acting on the implant shaft could be evaluated. The optimal implantation angle could then be found with the help of an optimization algorithm.

3.1. Simulation software

In this project the finite element simulation software ANSYS Workbench 14.5 was used for the ankle joint simulation. In ANSYS Workbench’s computer-aided design (CAD) software DesignModeler the user can create and/or import geometric models. ANSYS Workbench’s simulation software Mechanical is used to conduct simulations of the geometric models. In Mechanical the geometric model is divided into finite elements, i.e. a mesh is generated for the geometric model, and loads and supports are defined. Mechanical is also used to set up the solution and view the simulation results. ANSYS Workbench’s optimization software DesignXplorer can be used to find an optimal model design. [13]

3.2. Geometric model

3.2.1. Ankle joint

As a first step in the simulation design assumptions about the ankle joint geometry had to be made. The simulated ankle joint was simplified to only consist of the talus and the distal tibia, and did not include the distal fibula. The simulations did not either include any ligaments or other structures, other than the tibia and talus bone and their articular cartilage. These simplifications were made based on previous FEM simulations described in section 2.5. Other FEM simulations of the ankle joint and talus. The majority of these simulations [23, 24, 26, 27] did not include the fibula or any ligaments.

The geometry of the ankle joint model used in the simulations was provided by Episurf. The ankle joint geometry was based on CT data of the ankle joint of a 49-year-old man.
without any articular damages. The provided geometry consisted of the talus bone and the distal part of the tibia with no articular cartilage included. The CT data was treated and made into solids using SolidWorks, ready to be imported to the simulation software.

For geometry modifications ANSYS Workbench’s computer-aided design (CAD) software DesignModeler was used. The solid bodies provided by Episurf were imported into the DesignModeler. Using the DesignModeler, solid bodies representing the articular cartilage of the tibia and talus was generated. Both the tibial and the talar articular cartilage was assumed to have a constant thickness of 1.5 millimeters. The choice to have a constant cartilage thickness was based on a previous finite element simulation of the ankle joint by Anderson et al. [27]. The thickness approximation was based on the imaging data of the joint. Now the geometric model consisted of four parts; the bone of the tibia and talus and the articular cartilage of the tibia and talus. The four different parts of the joint were all modeled as solid bodies with isotropic linear elastic material properties. In the model the bone bodies of the tibia and the talus was assumed to only consist of one type of bone, cortical bone, instead of the more realistic scenario where the inner part of the bone consists of cancellous bone and the outer part consists of cortical bone. This simplification reduces the total number of elements used in the model, since additional bodies would require more contact elements. This reduction was done to reduce the simulation computational time, which is highly dependent on the number of elements. The simplification to model the bone as solid bodies was based on previous finite element simulations. [24, 26]

As a starting point for the gait simulation the ankle joint needed to be positioned in a neutral angle, i.e. have an angle of 90 degrees between the foot and shank. Using the DesignModeler the tibia and talus was positioned in way that was approximated to correspond to a neutral foot position.

In order to reduce the total number of elements in the simulation, and so save simulation computational time, the distal end of the tibia, the lower part of the talus and the head and neck of the talus was cut away in the geometric model of the ankle joint. This created a flat top surface of the tibia and a flat bottom surface of the talus, both perpendicular to the tibial longitudinal axis.

3.2.2. Implant

The DesignModeler was also used to create the implant. The implant had a diameter of 12 millimeters. The length of the shaft was 16.5 millimeters. The length of the whole implant varied depending on the implantation angle. The upper surface of the implant was generated to create a congruent articular surface of the talus. The implant was placed to be in flush with the surrounding cartilage.

The geometry of the implant was somewhat simplified. Some of the curvatures of the implant, for instance the curvature between the implant shaft and the upper part of the
implant, were simplified to increase the quality of the mesh later generated in Mechanical. The real implant also has a rounded edge between the upper articular surface and the side of the upper part of the implant that is facing the surrounding cartilage. This rounded edge was not included in the implant model; instead the edge was left sharp. The geometric model of the implant can be seen in Figure 6.

The Episurf resurfacing implant is made out of a chrome-cobalt-molybdenum alloy and have a 30±10 micrometers thick titanium and hydroxyapatite coating. Due to the limited thickness of the titanium and hydroxyapatite coating the implant was modeled as a solid body entirely made out of the chrome-cobalt-molybdenum alloy.

![Figure 6](image)

**Figure 6** Left: Geometrical model of the implant from above. Right: Geometrical model of the implant from below.

### 3.3. Simulation

#### 3.3.1. Material properties

ANSYS Workbench’s simulation software Mechanical was used for the simulations. The bone bodies and the cartilage bodies of the ankle joint model were modeled as solid bodies with isotropic linear elastic properties, based on previous simulations of the ankle joint. [24, 27] The bone bodies were assumed to only consist of cortical bone. The resurfacing implant was also modeled as a solid body with isotropic linear elastic properties. The material properties used in the simulations are presented in Table 1.

The friction coefficient between two cartilage surfaces is 0.01 [54] and around 0.02 for cartilage articulating with cobalt-chromium alloys [55]. Since the magnitude of the friction coefficients is small the contact between the two cartilage surfaces and between the cartilage and the implant was assumed to be frictionless in the simulation.
<table>
<thead>
<tr>
<th>Material</th>
<th>Young’s modulus (MPa)</th>
<th>Poisson’s ratio</th>
<th>Source</th>
</tr>
</thead>
<tbody>
<tr>
<td>Cortical bone</td>
<td>20 700</td>
<td>0.3</td>
<td>Rho et al. [56]</td>
</tr>
<tr>
<td>Cartilage</td>
<td>12</td>
<td>0.42</td>
<td>Hayes and Mockros [57]</td>
</tr>
<tr>
<td>ASTM-F1537 implant</td>
<td>210 000</td>
<td>0.3</td>
<td>MakeltFrom.com [58]</td>
</tr>
</tbody>
</table>

Table 1 Mechanical properties of the materials used in the simulations.

3.3.2. Contacts

Using Mechanical, frictionless and bonded contacts were created between the different bodies. In bonded contacts the faces of the bodies are not allowed to slide or be separated from each other. In frictionless contacts the faces of the bodies are allowed to separate from each other and free sliding is allowed between the bodies. [13] Bonded contacts were created between the talus bone and the talar articular cartilage and between the tibial bone and the tibial articular cartilage. Bonded contacts were also used between the implant and the talar bone. The implant and the tibial articular cartilage were connected with a frictionless contact. Frictionless contacts were also used to connect the talar articular cartilage to the tibias articular cartilage, as well as the implants upper surface to the tibial articular cartilage.

3.3.3. Mesh

Mechanical was also used to generate the mesh, i.e. divide the geometry model into finite elements. The bone and cartilage bodies of the talus and the tibia were meshed with tetrahedral elements with the element type SOLID187 and had an element size of 3 millimeters. Each SOLID187 element has 10 nodes and uniform reduced integration is used, where hourglass control is incorporated. The solid body of the implant was also meshed with tetrahedral elements with the element type SOLID187 and had an element size of 3 millimeters. The contact elements between the bodies were meshed with the element types CONTA174 and TARGE170. The number of solid elements was 31297 and the number of contact elements was 14237. Some elements where distorted, however the average skewness (a measurement of element quality) was around 0.35, which is considered very good.

3.3.4. Axis of rotation and forces

In Mechanical the stance phase of gait was simulated. The simulation consisted of 10 load steps, where the first load step had the purpose of initiating the first contact between the talus and the tibia, while the nine following load steps represented the
stance phase of gait. Each load step was 0.07 seconds, giving a total length of the stance phase of 0.63 seconds, which is in the same range as found by Stauffer et al. [15].

The bottom flat surface of the talus was fixed during the simulation while the tibia upper flat surface was subjected to a time varying axial force with similar pattern to Stauffer et al. [15], with a maximum force of 1000 Newton. The reason why the axial force had a maximum of 1000 Newton and not a more realistic value of 3400 Newton (4.5-5.5 times body weight according to Stauffer et al. [15]), was simulation problems. The ankle joint was modeled as a hinge joint and was therefore only allowed to rotate around a defined axis of rotation. During the simulation the tibia was rotated around the ankle joint axis of rotation according to Winter [14]. The time varying axial force and ankle joint rotation is presented in Figure 7. The ankle joint axis of rotation was assumed to be fixed. The axis of rotation was based on the curvature of the talus articular surface, and was found using a method similar to the method used by Barnett and Napier [16]. A circle was drawn on a plane on the talus lateral and medial side, respectively. The upper edge of each circle coincided with the articular curvature of the talus at a neutral position. The ankle joint axis of rotation was defined as the line connecting the center points of the two circles.

![Graph of time-varying force and ankle angle](image)

**Figure 7** The upper graph shows the applied axial time varying force on the tibia and the lower graph shows the rotation of the tibia around the defined ankle joint axis of rotation, where positive values represent dorsiflexion and negative values represent plantarflexion. The first load step (from time 0 to 0.07 seconds) is to initiate contact between the talus and the tibia and is not part of the stance phase of the gait cycle.
3.4. Optimization

ANSYS Workbench’s optimization software DesignXplorer has a system called Response Surface Optimization. Response Surface Optimization has three steps: Design of Experiments, Response Surface and Optimization. In Design of Experiments a number of design setups with varying input parameters are generated and simulations with the different design setups are performed. The input parameters are the parameters that are allowed to vary in a range decided by the user in order to optimize the design. For each set of input parameters output parameters are generated. [13] In this project the input variables were the coronal and the sagittal implantation angle and the output variable was the maximum equivalent (von Mises) strain acting on the implant shaft. Both the coronal and sagittal implantation angle was assigned a lower bound of 0 degrees, which corresponded to the normal direction of the cartilage surface where the implant was placed (see Figure 8). The coronal and sagittal implantation angle was also assigned an upper bound of 25 degrees in the medial and anterior direction respectively. The range of the sagittal and coronal implantation angles was chosen based on the surgical accessibility of the ankle joint and that the head of the implant needed to be submerged in at least 2 millimeters of bone, as a margin of security for implantation. The Design of Experiments step is the most time-consuming optimization step, since this is the step in which all the simulations are performed. In this project the Design of Experiments step consisted of nine simulations, where each simulation took around 6 hours to complete, adding up to 54 hours in total.

In the Response Surface step response surfaces are generated based on the input and output parameters from the previous step. The response surfaces are generated with the help of mathematical functions that approximately predict the output parameters for all possible input parameters in the assigned range. Each output parameter generates one response surface. The response surfaces can provide a graphical representation of how the output parameters depend on the input variables. [13] In this project one response surface, predicting the maximum equivalent strain acting on the implant shaft for all possible input parameter combinations, was generated.

In the Optimization step the goal or restraints of the output parameters are defined. When the goal or restraints are defined the program use an algorithm to search through the response surface to find the set of input parameters that best fulfill the goals or restraints of the output parameters. [13] In this project the goal of the optimization was to minimize the maximum equivalent strain acting on the shaft of the implant.
3.5. Indication of validity

Some form of indication of model validity is needed to be able to suggest that the results from the simulations can be trusted. Indications of validity can be found by comparing simulation results with results from cadaver studies. Unfortunately the author could not find any cadaver studies with the exact same setup as described above. However, cadaver studies evaluating contact stresses of intact ankle joints were found. [59, 60] Therefore a simulation of an untreated ankle joint, without the implant, was generated. In this simulation the ankle joint was brought through the same time varying dorsiflexion and plantarflexion angles as described above, but with an axial force of 600 Newton. During the first two load steps the axial force went from 0 to 600 Newton, thereafter the force was held constant at 600 Newton. The tibial axial force was chosen to resemble the cadaver studies. Contact stress results from the simulation were to be compared with contact stress results from the two cadaver studies, to give indications of model validity.
4. Results

In the optimization a response surface, showing approximations of the peak maximum equivalent (von Mises) strain acting on the implant shaft for various implantation angles, was generated, see Figure 9. This response surface was searched through to find the optimal implantation angle.

![Response Surface](image)

**Figure 9** The response surface showing the peak maximum equivalent strain acting on the implant shaft for various angles.

The optimal implantation angle, i.e. the angle that minimized the maximum equivalent (von Mises) strain acting on the implant shaft over time was a sagittal angle of 12.5 degrees and a coronal angle of 0 degrees. The optimized maximum equivalent strain over the course of the stance phase of one gait cycle is shown in Figure 10 (full line). The strain results from one of the placements that generates the largest maximum equivalent stress on the implant shaft over the stance phase of one gait cycle (double line) as well as from an implantation angle perpendicular to the cartilage surface, i.e. sagittal and coronal angle of 0 degrees, (dotted) is shown in Figure 10.

During the first load step in the simulation (time 0-0.07 seconds) contact is initiated between the tibia and talus. The stance phase of the gait cycle is from load step 2 to 10
(time 0.07-0.7 seconds). Maximum plantarflexion occurs at time 0.21 seconds and maximum dorsiflexion and maximum axial load occur at time 0.56 seconds.

The implantation angle that causes one of the largest maximum equivalent strains on the implant shaft is around 40 percent larger than the strain generated by the optimized implantation angle. An implantation angle perpendicular to the cartilage surface (sagittal and coronal angle of 0 degrees) generate a maximum equivalent strain that is around 20 percent larger than the strain generated by the optimized implantation angle.

![Graph showing maximum equivalent (von Mises) strain acting on the implant shaft through the stance phase of one gait cycle.](image)

**Figure 10** Maximum equivalent (von Mises) strain acting on the implant shaft through the stance phase of one gait cycle. Full line is for the optimized implantation angle, dotted line is for an implantation angle perpendicular to the cartilage surface and the double line is for sagittal angle 12.5 degrees and coronal angle 25 degrees.

Peak stresses from load step 3 to 10 in the simulation of the ankle joint without the implant, i.e. the simulation used to indicate model validity, have been evaluated. The two first load steps where excluded since the constant axial force of 600 Newton was not reached until the third time step. In the simulation of the ankle joint without the implant the contact stress had an average peak value of 2.26 MPa. During maximum plantarflexion (time 0.21 seconds) the contact stress has its lowest value of 1.64 MPa, see Figure 11. In the dorsiflexion part of the motion (time 0.35-0.7 seconds) the contact stress have very limited variation (2.41±0.05 MPa), see Figure 11. Over all the contact stress have no major variation over time (2.06±0.40 MPa).

![Graph showing peak contact stress in the simulated ankle joint without the implant.](image)

**Figure 11** Peak contact stress in the simulated ankle joint without the implant.
5. Discussion

This section will start by discussing the results of the optimization simulation and the validity of the model. This is followed by a discussion of factors that affect the solution accuracy. Some future studies will be recommended. The section ends with a brief discussion of ethical and environmental aspects of the project.

5.1. The simulation results

The results showed that the optimal implantation angle is a sagittal angle of 12.5 degrees and a coronal angle of 0 degrees. This angle minimizes the peak maximum equivalent strain acting on the implant shaft over time (see Figure 10).

The maximum equivalent strain response for some different implantation angles shown in Figure 10 seem to indicate that the strain acting on the implant shaft depend on the axial force acting on the tibia (see Figure 7). The general shape of the graphs shown in Figure 10 seem to depend on the magnitude of the axial force acting on the tibia, however there are clear differences between the curves, indicating that the direction of the axial force relative the direction of the implant shaft also affect the maximum equivalent strain acting on the implant shaft. In the beginning of the stance phase of the gait cycle (from around time 0.14 to 0.42 seconds) the optimized maximum equivalent strain have a larger magnitude than the strain generated by a sagittal and coronal implantation angle of 0 degrees. During this time the tibia is first rotated to -5 degrees of plantarflexion and then rotated to around 3 degrees of dorsiflexion (see Figure 7). At these angles the implant shaft direction generated by sagittal and coronal implantation angles of 0 degrees is closer to the axial force direction than the shaft direction generated by the optimized implantation angle. This is probably the reason why, at this time, the optimized maximum equivalent strain is larger than the strain generated by a sagittal and coronal implantation angle of 0 degrees. One could expect that the minimal maximum equivalent strain acting on the implant shaft is obtained when the external force acting on the implant is directed in the shaft direction, since other force directions might make the implant work as a lever and thus cause increased strain on the implant shaft. Therefore the shaft direction generated by a sagittal and coronal implantation angle of 0 degrees would generate a smaller strain than a shaft direction that is less close to the axial force direction, like the direction generated by the optimized implantation angle, during this time. However as the dorsiflexion proceeds (from around time 0.42 seconds) the optimized shaft direction is closer to the axial force direction than the shaft direction generated by a sagittal and coronal implantation angle of 0 degrees, making the optimized strain smaller than the strain generated by a sagittal and coronal implantation angle of 0 degrees. Large values of the coronal implantation angle will cause the shaft to
have a direction that diverges from the axial force direction through all of the stance phase, which will probably result in a relatively large strain throughout the motion, as seen in the case of an sagittal implantation angle of 12.5 degrees and a coronal angle of 25 degrees, see Figure 10. The response surface in Figure 9 also indicates that small coronal implantation angles together with sagittal angles close to the optimized sagittal angle will generate low peak values of the maximum equivalent strain acting on the implant shaft.

The optimized implantation angle is able to minimize the maximum equivalent strain acting on the implant shaft, but it does not allow for as small osteotomy size as larger sagittal and coronal implantation angles would. However, the optimized implantation angle still allows for a smaller osteotomy than, for instance, an implantation angle perpendicular to the cartilage surface. If the resulting size of the osteotomy required for the optimized implantation angle is acceptable, then the optimized implantation angle might be recommended. If the resulting osteotomy size is not accepted other implantation angles might have to be considered.

5.2. Model validation

As described under the topic 3.5. Indication of validity, a simulation of an intact joint was done to generate results that could be compared with two cadaver studies in order indicate model validity. The cadaver studies were made by McKinley et al. [59] and Hunt et al. [60] and both evaluated contact stresses in the ankle joint. In the study made by McKinley et al. [59] the contact stresses at ten intact cadaveric ankle joints was measured while the tibia was subjected to a constant axial force of 600 Newton and the ankle joint angle varied to simulate the stance phase of gait. In the study made by Hunt et al. [60] the contact stresses at eight intact cadaveric ankle joints where measured at a neutral angle and at 15 degrees of dorsiflexion while the tibia was subjected to a constant axial force of 686 Newton. The mean peak stress response for the joints evaluated by McKinley et al. [59] had a small variation over time, with a mean peak stress of 2.7 MPa. Hunt et al. [60] found a peak stress of 2.63±0.23 MPa at a neutral angle and 2.85±0.65 MPa at 15 degrees of plantarflexion of the intact specimens. The average peak stress for the validation simulation was 2.26 MPa, which is roughly the same magnitude as found in the cadaver studies. The curve of the peak stress values over time for the validation simulation also showed quite small variations (2.06±0.40 MPa), especially in the dorsiflexion part (time 0.35-0.7 seconds), see Figure 11. The fact that the average peak stress is roughly the same magnitude as the cadaver studies and that the peak stress has limited variation over time indicate that the geometry of the modeled ankle joint and the approximation of the joint axis of rotation are valid.
5.3. Solution accuracy

The simulations in this project are built on approximations and simplifications, all of which potentially can limit the solution accuracy. The FEM is used to find an approximate solution to problems that are often too complicated to solve analytically. One factor that affects the accuracy of the FEM solutions are the number of finite elements in which the problem is divided into, i.e. the size of the mesh. Increasing the number of elements would produce a more accurate solution, but would also increase the computational time of the simulation. In this project the number of elements was kept low in order to reduce the simulation computational time, which could have limited the accuracy of the solution. The implantation angle optimization is also based on approximations, as described in 3.4. Optimization, which further can limit the solution accuracy.

Other approximations and simplifications that can limit the solution accuracy are simplifications of material properties and geometry. In this project simplifications have been made of several material and geometrical properties.

5.3.1. Geometrical simplifications

One of the most significant geometric simplification of the ankle joint model was to only include the talus and the distal tibia in the simulations, and exclude the fibula and the ankle joint ligaments. A problem that might arise from excluding ligaments in simulations of the ankle joint is to simulation of forces acting in the ankle joint. Fortunately Stauffer et al. [15] was able calculate the axial force acting on the ankle joint during the stance phase of normal gait (see section 2.3.1 Gait). The force pattern calculated by Stauffer et al. [15] was used in the optimization simulation in this project, however the maximum axial force was 1000 Newton, instead of a more realistic value of 3400 Newton (around 5 times a body weight of 70 kilos), which correspond better to the results of Stauffer et al. [15] (4.5-5.5 times body weight). The reason why an axial force of 1000 Newton was chosen over the more realistic value of 3400 Newton was simulation problems. Element distortion occurred when the larger force was applied in the simulation. The contact stiffness was reduced in an attempt to solve this, however this resulted in large penetrations, which could not be accepted. These problems did not occur for the reduced axial force of 1000 Newton, therefore this axial force was chosen for the optimization simulation. The reduced force probably did not affect the result of the optimization simulation, since the task was to find the implantation angle that minimized the maximum equivalent strain acting on the implants shaft, rather than the particular strain value itself.

Another significant geometric simplification that could limit the accuracy of the solution was the cutting of the tibia and talus in order to reduce the total number of elements in the simulation, see section 3.2.1. Ankle joint. The solution accuracy can for instance be limited by the fact that the bottom flat surface of the talus was fixed during the
simulation. This is not a realistic scenario, since a real talus is held in place by ligaments and articulating bones and not by a fixed surface inside the bone.

In this project the ankle joint has simply been modeled as a hinge joint with a fix axis. Even though the movement of the ankle joint is very similar to that of a hinge joint, the exact kinematics of the ankle joint cannot simply be described as a hinge joint. According to Dettwyler et al. [19] simulations where the ankle joint is modeled as a hinge joint will have limited accuracy. Even though the results from the validation simulations in this project could be validated when comparing the contact results with cadaver studies, simulations with more detailed ankle joint kinematics probably would have produced more accurate results.

Another geometric simplification that was done in this project was to model the bone bodies of both the talus and tibia as solids made out of only one type of bone, instead of a more realistic scenario where the bone have an outer layer of cortical bone and an inside of cancellous bone. This simplification was done (as described in the section 3.2.1. Ankle joint) in order to save simulation computational time. However, there is another type of modeling that generates more accurate results. Parr et al. [30] compared simulations of the talus bone that included the geometry of the cancellous network with solid non-porous models of the talus and found that models including the cancellous geometry were stiffer than the solid non-porous models. Therefore it is desirable to include the cancellous geometry when modeling bone. However, models including the geometry of the cancellous network require more finite elements, thus require more simulation computational time, than solid non-porous models. In this project a too long computational time could not be allowed, further motivating the choice of modeling bone as solids.

The geometry of the cartilage of both the talus and tibia has also been simplified in the simulations. The cartilage of the tibia and talus have been modeled with a constant thickness of 1.5 millimeters. However, as described in section 2.1. Ankle joint anatomy and anthropometry, the cartilage thickness varies over both the tibial and the talar surface. Therefore the choice of modeling the cartilage with constant thickness might limit the accuracy of the solution.

The geometry of the implant was also simplified in the simulations, which also could limit the solution accuracy. The implant was also modeled to be in flush with the surrounding cartilage. However, according to Episurf’s treatment method of OLTs (see 2.8. Episurf’s OLT treatment method) the implant should be recessed to the surrounding cartilage, similar as in their treatment method of the knee. Due to convergence difficulties this recession could not be achieved in the simulations, therefore the implant was placed in flush with the surrounding cartilage. This might limit the accuracy of the solution.
5.3.2. Simplifications of material properties

The bone and cartilage in the simulations of this project was modeled with isotropic linear elastic properties. For bone this is a valid assumption since bone to large extent can be considered as a linear elastic material (see section 2.4.1 Cortical bone), but cartilage, however, has anisotropic and viscoelastic material properties. The assumption of isotropic linear elastic properties in the model can therefore limit the accuracy of the simulation results.

The bone bodies in the simulation were chosen to be modeled as solids. The result from Parr et al. [30] suggest that a more realistic model, i.e. a model including the cancellous network, is stiffer than a less realistic model, i.e. a solid non-porous model. This was taken into account when choosing the material properties of the bone in the project simulations, and the stiffer material properties of cortical bone (Young’s modulus 20700 MPa [56]) was chosen over the less stiffer material properties of cancellous bone (Young’s modulus 14800 MPa [56]).

Another simplification in the project simulations was the modeling of the contact between the two cartilage surfaces and the between the cartilage and the implant. These contacts were modeled as frictionless, even though the have small friction coefficients (see section 3.3.1. Material properties). To model the contacts as frictional instead of frictionless might produce a more accurate result, but since the frictional coefficients are so small it is thought to have limited effect on the results.

5.4. Future studies

A future, more detailed study in which the above described approximations are improved is suggested. By refining factors like mesh size, material properties and geometry approximations more accurate solutions can be found. A more realistic bone model, which include the geometry of the cancellous network as suggested by Parr et al. [30], could be used to describe the mechanical behavior of bone more accurate. Another factor that can contribute to a more realistic model is to model the articular cartilage with varying thickness, according to patient-specific data. The movement of the ankle joint through gait could also be described in more detail, and not simply be described as a hinge joint with a fix axis, to improve the accuracy of the simulation. More realistic forces acting on the ankle joint through simulated gait should also be used to enable the generation of more realistic simulation results.

In this project only implant placements at the central part of the medial talar surface was evaluated. This placement was chosen since it is the most common OLT position. However, OLTs can occur at any part of the talar surface. [32] Therefore it is of interest to do similar implantation angle optimizations as has been done in this project at other talar locations. Only one implant size was evaluated in this project. A future study could evaluate if the optimal implantation angle varies between different implant sizes. In this
project only one ankle joint geometry was evaluated. The geometry of, for instance, the talus bone can vary between subjects (see section 2.1.2. Talus), which might affect the simulation results from implantation angle optimizations. Therefore it is of interest to do similar implantation angle optimizations as has been done in this project for different ankle joint geometries.

This project only evaluated the strains acting on the implant shaft for the stance phase of one gait cycle. A future project might consist of developing a strain-adaptive ankle joint model to evaluate how the implant will work over time and how it will influence bone remodeling.

5.5. Ethical and environmental aspects

In this project computer simulations were used to find the optimal implantation angle. To perform computer simulations instead of physical experiments can be positive in an environmental point of view, since it might require fewer resources than experiments.

When it comes to implantation treatments it is of great importance to do everything possible to increase the longevity of the implant. This is ethical since it reduces the risk of patient suffering that might come with implant failure. It is also positive in an economical point of view, since additional surgical treatment that might come with implantation failure are costly. By finding the implantation angle that minimizes the strain acting on the implant shaft this project can contribute to an increased longevity of the resurfacing implant.
6. Conclusion

In this project a finite element model of an ankle joint have been developed and the stance phase of a normal gait cycle has been simulated. This model was used to find the implantation angle of a talar resurfacing implant, which minimized the maximum equivalent (von Mises) strain acting on the implant shaft. This angle was found to be 12.5 degrees in the sagittal plan and 0 degrees in the coronal plane.

The magnitude of the applied force on the modeled ankle joint seems to have great influence over the resulting strain on the implant shaft. Another factor that seem to influence the strain acting on the implant shaft is the direction of the applied force relative to the direction of the implant shaft, where more similar directions seem to reduce the strain acting on the implant shaft.

A number of simplifications have been done in the finite element model of this project, which might affect the accuracy of the results. Therefore it is recommended that further, more detailed, simulations based on the current simulation are done in order to improve the result accuracy. A more detailed simulation can be used to decide the implantation angle in the preoperative planning of treatments of OLTs with Episurf’s focal cartilage resurfacing implant, in order to reduce the strain acting on the implant shaft and thus improve the longevity of the implant. A more detailed simulation can also be used to determine the size of the tibial osteotomy required to access the talar surface in the Episurf talar treatment method.
7. References


Appendix

Appendix A – Optimized strain response on implant shaft

Time 0.07 seconds

Time 0.14 seconds

Time 0.21 seconds

Time 0.28 seconds
Appendix B – Contact stress response validation simulation: Talus subchondral perspective

Time 0.14 seconds

Time 0.21 seconds

Time 0.28 seconds

Time 0.35 seconds

Time 0.42 seconds

Time 0.49 seconds