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BIOMECHANICAL ANALYSIS OF SITTING DOWN AND RISING UP FROM A CHAIR WITH RESPECT TO THE CRUCIATE LIGAMENT FORCES.

Claudia M. Schaap (idnr. 343279)
Practical training report nr. 96.004

At the Oxford Orthopaedic Engineering Centre
Nuffield Orthopaedic Centre
Department of Engineering Science, University of Oxford

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Abstract

As knee injuries and degenerative diseases are very common, it is interesting to investigate rehabilitation exercises.

The sitting down and rising up from a chair motion was evaluated using kinematic, force plate, and electromyographic (EMG) data. The kinematic data was collected with a stereophotogrammetric system. It consists of infra-red video cameras that track the position of infra-red reflective markers attached to certain bony landmarks of the limb segments of three young and healthy test subjects. A Vicon computer system recorded the three-dimensional trajectories. A floor-mounted dynamometer force plate measured the ground reaction force. Muscle activity was monitored using surface electromyography, this information was used for qualitative validation.

A two-dimensional mathematical model (Lu [12]) applying inverse dynamics was used to calculate the internal forces in the ankle, knee, and hip joint occurring during sitting down and rising up from a chair. The resultant forces and moments were then distributed between the structures of the joints.

Various ways to perform half squats were also recorded, but not analysed.

It was found that, cocontraction at the hip, knee, and ankle occurred during sitting down and rising up. The external dorsiflexing moment round the ankle was of the order of 20 to 50 Nm, the external flexing moment at the knee was between 20 to 30 Nm, for the hip the external flexing moment was 90 to 140 Nm. The model calculated several possible solutions, the most likely ones (the force in the quadriceps is bigger than the force in the hamstrings) did not include ligament forces. The less likely ones (the force in the quadriceps is smaller than the force in the hamstrings) included PCL forces varying from subject to subject between 0.8 and 3.6 times body weight. The impression gathered from these results is that cocontraction in the knee might reduce or even obviate the need for ligament forces.

Cycling is another exercise used for rehabilitation. To be able to perform a similar experiment to calculate the internal knee forces, a pedal force dynamometer is needed. A literature search was performed, and it was found that all the six load components have to be measured. Otherwise, errors in axial and varus/valgus moments at the knee will occur. As the ACL resists internal rotation and anterior draw, it will be affected by out of plane loads.

The six-component pedal dynamometer developed by Hull and Davis [9] is recommended. It consists of four subsystems: the dynamometer containing straingauges, The pedal position measurement subsystem with single turn continuous rotation potentiometers, the signal conditioning electronics using carrier amplification, and a mini computer to digitise, store and analyse the data.
1. Introduction

The human knee is a very complex joint. Every day it is subject to high loads. Knee injuries are among the most common injuries occurring today. Another well known phenomenon is the occurrence of degenerative diseases, such as osteoarthrosis. Patients suffering from these injuries and diseases are often successfully treated with rehabilitation exercises. Current rehabilitation practice has developed largely as a result of trial and error. It is very interesting to understand why some of the protocols are successful and why others fail. With this knowledge, new and improved protocols can be developed. To come to this understanding, the internal knee joint forces occurring during rehabilitation exercises should be determined. Knowing these, the new protocols can be based on the loading requirements and capabilities of the knee structures.

At the moment there is a PhD-project that aims to devise a method for the calculation of internal joint forces during movement (D.Toutoungi [22]). These forces are carried by the ligaments and muscle tendons and are transmitted between the articular surfaces. This method will be used to investigate a selection of the most commonly used rehabilitation exercises.

The project can be summarised as follows: A mathematical two-dimensional model of the lower limb has been developed (T.-W. Lu [12]). The relative positions of the model segments are determined using a stereophotogrammetric system. This system consists of two or more infra-red video cameras, which track the position of infra-red reflective markers attached to the limb segments of a test subject. A Vicon computer system records the three-dimensional trajectories of the markers. The software then calculates the velocities and accelerations of the segments using the mathematical model. A floor-mounted dynamometer force plate measures the external force on the lower limb. With inverse dynamics, the forces and moments at the knee joint can be determined. The next step is to distribute the resultant force and moment between the structures out of which the knee joint consists. As there are many sets of muscles and ligaments spanning the knee joint, this is not easy to do. The system is indeterminate. To reduce the number of unknowns the muscles with similar function are grouped. The lines of action and moment arms of the muscles in terms of the position of the knee joint are determined from the model. To find a limited number of solutions, electromyograms, also recorded during the exercises, will be used to verify the solutions qualitatively.

Rehabilitation exercises often used are squatting and sitting down on and rising from a chair. During this project, some squatting experiments and sitting and rising experiments were done in order to compare the forces in the knee ligaments. An analysis of the sitting and rising part was attempted.

Another often used rehabilitation exercise is cycling. In order to use the same protocol to calculate the forces in the ligaments while a subject is cycling, data are necessary about the foot-pedal reaction force. The dynamometer force plate can not provide these. Therefore a different kind of dynamometer is needed, one that is built in a pedal. I looked into the feasibility of making an instrumented pedal, but did not pursue this work for lack of time.
In this report the experiments are described and the results for the sitting and rising are discussed. It also contains a description of the instrumented pedals.

The report is built up the following way: Chapter two describes the anatomy of the lower limb. Chapter three is about Gait analysis. The optometric method will be explained. Chapter four introduces the two-dimensional mathematical model of the lower limb used for the analysis. In chapter five, the squatting and sitting and rising experiments are described and the outcome is discussed. In chapter six, various instrumented pedals, that have been found in the literature, are discussed. Chapter seven gives a description of the selected pedal.
2. The anatomy of the lower limb

This chapter contains a short description of all the important structures that together form the lower limb. For more detailed information the reader is referred to [13].

2.1 The bones

The so called long bones are present in the limbs. The metaphysis is a cylinder of compact bone surrounding a medullary cavity which is filled with some spongy bone and a large amount of yellow fatty marrow. The epiphysis is formed of spongy bone with a thin outer shell of compact bone. These bones are formed to provide maximum support and consist of a minimum amount of material. Both genetic and local factors influence the shape and size of a bone. Adjacent muscles or organs mould the bone to some extent.

In figure 2.1.1 the skeletal system of the lower limb is shown.

Figure 2.1.1: The skeletal system of the lower limb.

Through the pelvis the weight of the trunk is transmitted to the lower limb. The proximal end of the femur (the thigh bone) consists of a head, a neck and greater and lesser trochanters. The head is quasi spherical and articulates with the acetabulum of the pelvis to form the hip joint. This is a synovial joint of the ball and socket variety.
The trochanters are bony projections and they serve as attachment points for some of the major thigh and buttock muscles. They increment the lever-arms of the muscles about the hip. The distal end of the femur is expanded and forms two large masses, the **medial** and **lateral femoral condyles**. They articulate with the **tibial condyles** and the **patella**. The femoral condyles are separated posteriorly by the **intercondylar notch**. The condyles are anterosuperiorly united and there they articulate with the patella. On the outer nonarticulate surface of each condyle is a small elevation, the **epicondyle**. These give attachments to the **medial** and **lateral ligaments** of the knee joint.

The patella is a **sesamoid** bone, triangular in shape. A sesamoid bone means a bone that forms in a tendon. The patella is attached to the **patellar tendon** and the **quadriceps tendon**. The anterior surface is roughened and subcutaneous. The posterior surface is smooth and has a smaller medial and a larger lateral facet for articulation with the femoral condyles.

The lower leg consists of two roughly parallel bones, the **tibia** on the medial side and the **fibula** on the lateral side. The upper end of the tibia is expanded to form two masses, the **medial** and the **lateral tibial condyles**. A central **anteroposterior ridge** across the upper surface, the intercondylar area, separates the articular surfaces and gives attachment to the **cartilaginous menisci** and to the **anterior and posterior cruciate ligaments**. The prominent **tibial tuberosity** is situated anteriorly below the superior surface and receives the attachment of the patellar ligament. The posteroinferior surface of the lateral condyle bears an oval facet for the head of the fibula.

The **foot** is an arched platform consisting of a number of separate bones bound together by ligaments and muscles. The arched nature of the foot supports the body’s weight, acts as a lever which is sufficiently rigid to propel the body forwards and yet is resilient enough to absorb sudden shocks. The talus forms the connecting link between the bones of the leg and those of the foot, it has no muscular attachments. The **calcaneus** is an irregular bone, its posterior part forms the prominence of the **heel**. The upper surface has posterior, middle and anterior articular facets for the talus. Several ligaments, short muscles of the sole, the **flexor hallucis longus tendon** and the **Achilles’ tendon** are attached to this bone. The **navicular** lies on the medial side of the foot between the talus and the **cuneiform** bones. The **tendon tibialis** and a ligament are attached to this bone. The **cuboid** lies on the lateral side of the foot. Posteriorly it articulates with the calcaneus, anteriorly with the bases of the 4th and 5th **metatarsal bones**, and medially with the lateral cuneiform and sometimes the navicular. The **long plantar ligament** is attached to it. Distal from the metatarsals is sesamoid bone present. Cuneiforms are three wedge-shaped bones. Posteriorly they articulate with the navicular and anteriorly with the bases of the medial three metatarsal bones. Parts of the tendons of **tibialis anterior**, **tibialis posterior** and **peroneus longus** are attached to the medial cuneiform. The metatarsal bones and **phalanges** are the bones that form the toes.
2.2 The joints

A joint is a union between two or more bones. According to [13] there are four types of joints: bony, fibrous, cartilaginous or synovial. In the bony type, the elements are joined by bony union, this is the case with the three elements of the pelvis. In a fibrous joint, the bony surfaces are united by fibrous tissue, as is the case in the inferior tibiofibular joint. There is little or no movement. The cartilaginous joints are divided into two groups: the primary and the secondary. In primary cartilaginous joint the bony surfaces are united by hyaline cartilage. In the secondary cartilaginous joint the bony surfaces are covered with hyaline cartilage and united by a fibrocartilaginous disc. Again there is little or no movement. The bony articular surfaces in a synovial joint are covered with hyaline cartilage. A fibrous capsule is attached near to the articular margins of the bones. There is a cavity in the joint which makes the synovial joints freely movable. The surfaces of the interior of the joint, except those covered by cartilage, are lined by a delicate vascular synovial membrane which secretes a watery synovial fluid into the joint cavity.

The knee joint is a synovial joint between the distal end of the femur, the patella and the proximal end of the tibia. In the joint are two menisci present. These are crescent shaped pieces of fibrocartilage with thickened outer margins. Each lies on a tibial condyle. The joint is capable of flexion, extension and a little rotation.

The ankle joint is also a synovial joint. It connects the inferior surface of the tibia and the facing surfaces of the medial and lateral malleoli (surfaces on the distal end of the tibia) and, inferiorly, the upper, medial and lateral surfaces of the talus. The ankle joint may be plantarflexed and dorsiflexed.

As is mentioned before is the hip joint also of the synovial variety. It is capable of flexion, extension, abduction, adduction, circumduction, and medial and lateral rotation.

2.3 The ligaments

Ligaments are cord- or band-like bundles of fibrous connective tissue that link two bones together at a joint. They maintain its stability by guiding and restricting the movements of the bones relative to each other. They maintain contact between the articulating surfaces and provide reinforcement to the capsule. When stretched they transmit a tensile force between the bones.

The ligament at the hip is the capsule. It consists of the following thickenings: The iliofemoral ligament, the pubofemoral ligament and the ischiofemoral ligament.

The capsule in the knee joint is attached to the femur. Posteriorly it is attached above the intercondylar notch. It is also attached to the patellar margins and to the periphery of the upper end of the tibia. The patellar ligament passes from the distal pole of the patella to the tibial tuberosity. The oblique popliteal ligament is an extension of the semimembranosus tendon and passes upwards and laterally from the medial tibial condyle across the posterior surface of the capsule. There are two major pairs of ligaments: the medial and lateral collateral ligaments and the anterior and posterior cruciate ligaments. The first pair strengthen the joint capsule on its medial and lateral aspects and are responsible for the transverse stability of the knee. The latter pair ensures the antero-posterior stability of the joint and allow hinge-like
movements to occur while keeping the articular surfaces together. Their geometry determines the shape of the femoral condyles in all three planes.

The medial collateral ligament (MCL), also known as the tibial collateral ligament, runs obliquely inferiorly and anteriorly from the postero-superior aspect of the outer surface of the medial femoral condyle to the medial aspect of the upper end of the tibia. Its anterior fibres are distinct, but its posterior fibres blend with those of the capsule at the medial border of the medial meniscus. The attachment area on the femur is nearly circular and the attachment area on the tibia resembles a long, narrow band.

The lateral collateral ligament (LCL), also known as the fibular collateral ligament, runs obliquely inferiorly and posteriorly from the postero-superior of the lateral femoral condyle to the head of the fibula, anterior of the apex, and deep to the insertion of biceps femoris. It is separate from the capsule for its entire length. Both the attachment areas are roughly circular in shape.

The anterior cruciate ligament (ACL) is attached to the anterior part of the medial tibial condyle, between the insertions of the anterior horns of the medial and lateral menisci. It runs obliquely superiorly and laterally and attaches to the internal aspect of the lateral femoral condyle. It is generally described as consisting of three bands, the antero medial band, the posterolateral band, and the intermediate band, and as a whole the ligament is twisted on itself. The femoral attachment area resembles a segment of an ellipse, tilted slightly from the vertical. The tibial attachment area is nearly elliptical with the major axis of the ellipse running antero-posteriorly.

The posterior cruciate ligament (PCL) lies behind the ACL, within the intercondylar notch. It is attached to the posterior part of the tibial plateau, well posterior to the insertions of the posterior horns of the medial and lateral menisci. It runs obliquely medially, anteriorly and superiorly to be inserted partly into the intercondylar notch and partly onto the edge of the lateral surface of the medial femoral condyle along the line of the articular cartilage. It is often said to consist either of two parts, an antero-medial band and a postero-lateral band, sometimes even of four bands including the anterior band of Humphrey and the menisco-femoral ligament of Wrisberg. Other authors claim that it has no divisions at all [22].

In the ankle are the following ligaments: The capsule, it is attached to the articular margins except anteriorly where it extends to the neck of the talus. The capsular thickenings are the medial ligament, the lateral ligament, the anterior and posterior ligaments and the inferior transverse tibiofibular ligament.

2.4 The muscles

Muscle is a contractile tissue that causes the body movement. The skeleton muscles are composed of unbranched fibres of sarcoplasm limited by a membrane, the sarcolemma, and contains many nuclei. Each fibre has a motor end plate and contains many contractile units, the myofibrils.

The muscles are attached at each end, usually to bone, either directly or through tendons and aponeuroses. They cross one or more joints. Muscle fibres are arranged either parallel to the direction of the action or obliquely to it. Parallel arrangement gives greater movement but less power than an oblique arrangement. The more proximal attachment of a muscle is often called its origin, and the more distal attachment its insertion. When a movement occurs at a joint, the muscles concerned in
producing it are known as its prime movers, those opposing it, as antagonists. Muscles contracting to steady the joint across which movement is occurring, and other joints involved are known as synergists.

Accurate knowledge of the positions of the muscle origins and insertions is important for the determination of the muscle force line of action and moment arm with respect to the centre of rotation of a joint.

The most important muscle groups in the lower limb are mentioned in the following part.

Flexion of the hip is caused by the iliopsoas, assisted by tensor fasciae latae, rectus femoris, sartorius and pectineus. Extension is caused by the gluteus maximus, assisted by gravity, and the hamstrings. Abduction is caused by the gluteus medius and minimus. Adduction is caused by the adductors of the thigh, gracilis and gravity. Medial rotation happens due to the anterior fibres of gluteus medius and minimus assisted by the iliopsoas. Lateral rotation is due to the short muscles at the back of the joint.

Flexion of the knee is caused by the hamstrings aided by the gastrocnemius. Extension is due to the quadriceps and iliotibial tract muscles. Rotation is produced by the hamstrings when the knee is flexed at a right angle and by the popliteus.

Plantarflexion of the ankle is produced by gastrocnemius and soleus. Dorsiflexion is caused by the tibialis anterior.
3. GAIT ANALYSIS

To determine the relative positions of the model segments of the lower limb a stereophotographic system is used. This system consists of two or more infra-red video cameras, which track the position of infra-red reflective markers attached to the limb segments of a test subject. A computer system (Vicon, Oxford Metrics) records the three dimensional trajectories of the markers. The software then calculates the velocities and accelerations of the segments using a mathematical model of the lower limb. A dynamometer measures the external force on the lower limb. With inverse dynamics the forces and moments at the knee joint can be determined.

This system has been used to examine the forces during walking. Therefore it is referred to as gait analysis system. This system however can also be used for other experiments such as squatting and cycling. The motion of almost any major part of the body can be measured. This chapter contains a summary of the most interesting parts for the squatting and the cycling experiment of an article about gait analysis by H.S. Gill [7].

3.1 The variables

Gait analysis is the quantified measurement of movement patterns during walking. The whole analysis consists of measurement, storage of the measurement data, retrieval, processing, analysis and presentation.

There are two types of variables, the kinematic ones, like cadence, speed, length and angles, and the kinetic ones. The kinetic variables can be divided into the directly measured ground reaction forces and the calculated kinetic variables. The ground reaction force is measured with a force platform that provides three orthogonal components of the ground reaction vector, the position in the plate of its centre of pressure and the associated spin moment about a vertical axis. A similar instrument will be needed when performing a cycling experiment. To calculate the other kinetic variables, numerical processing of the kinematic variables is required. The processed data is then combined with the direct kinetic measurements to derive quantities such as joint moments. To do this, a kinetic model is needed.

The motion measurement system determines the relative positions of the limb segments during motion and captures the data at a rate greater than the highest frequency present in motion. The system must be able to sample data at a rate which is at least twice as fast as the highest frequency component of interest. This prevents aliasing errors.

The system must be able to measure kinetic variables and be able to relate limb position to these kinetic measurements. This means that the system must be able to determine positions in space of the separate limb segments relative to the force plate axis system, in addition to the position and direction of the ground reaction vector. A complete kinetic analysis requires that the mass distribution of the subject be known. As said before, a kinetic model is necessary for such analysis.
3.2 The markers

Skin mounted reflective markers are used, placed such that their movements reflect the motion of the separate body segments. They should be well defined and have good contrast against the body. When they have a known shape, it allows automatic digitisation. A minimum of three markers are needed to determine the location and orientation of each segment in three-dimensional gait analysis. They are illuminated by a primary light source close to each measurement camera. The light source is generally a ring of LED’s around the lens, strobed to the field rate of the camera. Monochromatic infra-red light is commonly used to minimise disturbance to the subject.

Array sensors in the cameras measure two independent co-ordinates. They extract measurements from a two-dimensional image. They detect the intensity of a focused image of the observed set of markers and convert it into digital form. The sensors sample marker positions at a fixed rate, the sample rate. This rate depends upon the particular movement being studied.

The system determines the three-dimensional position of an identifiable source of light from the intersection of rays projected from two or more cameras with known positions and orientations.

3.3 Calibration

To explain how the cameras are calibrated, some definitions have to be introduced first. There are two types of parameters, the external photogrammetric parameters, which are position and orientation, and the internal photogrammetric parameter, which is the principal distance. The orientation of a camera is defined by the principle axis of the camera. This axis is the line, that passes through the nodal point normal to the flat sensor. This is the point of a camera through which all light rays pass between source and sensor. The nodal point defines the position of the camera. The principle distance is the distance between the nodal point and the intersection of the principle axis with the sensor.

A two dimensional camera has got seven parameters: three for the position, three for the orientation, and one for the principal distance. Calibration is the process of determining these photogrammetric parameters. There are at least seven independent observations required for calibration of the cameras independently of one another. These observations are provided by measurement of at least four reference markers (two co-ordinate observations per marker in the image) whose three-dimensional positions are known. Using more than four reference markers, the resulting overdetermination optimises the estimation of the parameters.

Multi-camera calibration allows the estimation of external parameters of one camera relative to another from observations of markers whose three-dimensional positions are initially unknown. The external parameters of the cameras and the reconstructed positions of the observed markers are calculated simultaneously. A small number of markers in known positions (this is called a calibration object) are still required to establish the measurement co-ordinate system origin and axis directions.

The three-dimensional reconstruction of marker positions from multi-camera observations from calibrated cameras is relatively straightforward. The reconstruction
equations are based on simple fact that the measured point in the image plane, the
cameras nodal point and the observed marker are always collinear. The rays from a
pair of cameras should intersect at the observed marker.

Rays are initially unidentified. There is a highly efficient strategy for searching for
ray intersections, or near misses. Except for specific singularities, these ray
intersections occur only where physical markers are present and can thus be used to
group a set of rays from the set of observing cameras with a reconstructed marker. In
most experiments, the movement of three-dimensional markers is coherent. Markers
move along trajectories within definable limits of position, velocity and acceleration.
So unidentified markers can be assigned uniquely to a trajectory. The set of
trajectories are also constrained to one another by physical interconnections between
the observed markers. Kinematic models describe these connections.

3.4 Electromyography

Electromyography is the recording of the electrical signals produced by motor units
of muscle during activity. The signal is the summation of the motor unit
actionpotentials in the localised area of the muscle from which the measurement is
being made. There are surface electrodes used who are attached to the skin over the
muscle. One electrode is connected to the inverting input of the amplifier and another
one to the non-inverting input. A ground reference voltage is also needed, it can be
taken from any point of the body. The signals are amplified and filtered. They are
used to indicate phasic (on/off) activity of the muscle.

3.5 The obtained data

The data produced by optical gait systems is positional data and is subject to noise
or instrumentation error. There are random and systematic errors. Techniques used to
reduce the noise and obtain an estimate of the real trajectory are either some form of
regression or some form of spectral analysis. To obtain velocity and acceleration data,
the gathered positional data is numerically differentiated. Doing this, the noise will be
amplified by the differentiation process, so the data has to be smoothed.
4. Geometric and mechanical modelling of the lower limb

T.-W. Lu [12] has developed a two-dimensional mathematical model of the lower limb. The aim of this project is to calculate the internal knee joint forces during rehabilitation exercises. For this, a model of just the knee is not sufficient, as the knee ligaments and the muscles of the whole leg interact. This is why a model of the lower limb is used rather than a model of the knee.

This model was meant to describe the lower limb in a gait cycle. It can also be used for squatting and sitting and rising from a chair, with a few adjustments it can describe the lower limb while cycling. The model is briefly described as Lu has developed it.

4.1 The procedure to make a model

The general procedure that was used for modelling is as follows:
1. The first step is to model the lower limb as a system of rigid links connected by joints.
2. After that geometrical descriptions of each link and joint have to be established, including muscle and ligament origins and insertions. These are based on anatomical measurements.
3. A local co-ordinate system has to be embedded in each link to allow calculation of positions of the links in space at any instance in order to determine uniquely their relative motion and the changing geometry of the structures at the major joints of the lower limb.
4. Position trajectories of the links are collected. The trajectories are usually determined by those of surface markers placed on certain bony landmarks on the segment, using stereophotogrammetry and kinetic data using forceplates. Temporal EMG-data is also collected for qualitative validation.
5. The position trajectories are then smoothed and differentiated to obtain velocity and acceleration information. Using this information along with kinetic data and anthropometric data, the inverse dynamics analysis is performed to yield joint forces and moments.
6. The forces transmitted by each member of the system are determined by solving the force distribution problem.
7. The last step is the model validation.

4.2 The geometric model

The model is a two-dimensional model in the sagittal plane. Four rigid body segments are taken into account: the pelvis, the thigh, the shank and the foot (Figure 4.2.1).
The principle muscles and the cruciate ligaments are modelled as straight lines wrapping around the bones when necessary. There are seven muscles modelled: the iliopsoas (I), the quadriceps (Q), the hamstrings (H), the gluteus maximus (L), the gastrocnemius (G), the soleus (S) and the tibialis anterior (T). Muscles with small cross-sectional areas and/or intrinsic muscle tendon lever-arms are neglected. The origins and insertions of the muscles are approximated by single points. For broad attachment areas they are chosen as the centroids.

The hip and the ankle are modelled as simple hinge joints. They have congruent surfaces and allow rotation only in the sagittal plane. For the hip the inferior surface of the acetabulum and the head of the femur are modelled as circular arcs. The hip model centre of rotation is the geometric centre of the femoral head. For the ankle the distal end of the tibia and the talus are modelled as reciprocally shaped circular cavities. The centre of rotation is the geometric centre of the trochlea. The articular surfaces are assumed to be frictionless. The contact forces are therefore pure compressive and pass through the respective centres of rotation. The talo-tibial contact force and the femoro-pelvic contact force should remain within the embraces of the distal end of the tibia and the acetabulum. The ligaments of the ankle and the hip are not included.
The knee is modelled as shown in Figure 4.2.2.

Figure 4.2.2: The four-bar linkage model of the knee.

The two cruciate ligaments, the femur and the tibia form a four-bar linkage. The movement of the femur on the tibia is controlled by the two inextensible cruciate ligaments. The rolling and sliding movement pattern of the knee is reconstructed. The articulating femoral surface in the sagittal plane is also generated. During passive knee flexion and extension, the cruciate ligaments rotate isometrically about their points of attachment on the bones. The instant centre of joint rotation is the point at which the cruciate ligaments cross. This centre moves relative to each bone. The tibial plateau is assumed to be flat. The shape of the femoral condyle is the conjugate curve of the tibial plateau and can be generated by rotating the tibia with the femoral link fixed and by further imposing a meshing constraint. The meshing constraint requires that the common normal to the articular surfaces at their point of contact passes through the instant centre of joint rotation. Thus interpenetration and separation of the bones are prevented and the point of contact is located.

The knee joint contact force is purely compressive, because the friction coefficient between the articular surfaces of a synovial joint is very low. The contact force lies along the common normal to the articular surfaces at the point of contact and passes therefore through the instant centre of joint rotation.

Geometric information can be expressed in terms of the flexion angle. This is equivalent with the angle between the tibia link and the femoral link. During motion the orientations of the ligaments and the contact force can be determined once the flexion angle is given.

The patellofemoral joint model of H.S. Gill and J.J. O'Connor [8] is used. The patella is represented as a rectangle with two articulating surfaces. The outer surface represents the ridge of the patella, which makes contact with the trochlear groove. The inner articulating surface is an idealised representation of the outer edges of the patella, which contact the femoral condyles at large flexion angles. This is called odd facet contact. It is assumed that:

- The patellar tendon stays at constant length.
- There is a single contact point with common normal to the articulating surfaces.
- The quadriceps tendon is parallel to the long axis of the femur.
- The tibiofemoral movement is controlled by the four-bar linkage.
- The patellar tendon and quadriceps tendon are maintained in tension.
- The line of action of the patellofemoral contact force lies on the common normal to the articulating surfaces at the point of contact.
- There are three forces acting on the patella, the quadriceps tendon tension, the patella tendon tension and the femoral contact force. These are concurrent and coplanar.

Since the orientation of the quadriceps could be affected by both the knee and the hip, it is unlikely that the quadriceps tendon remains parallel to the long axis of the femur during movement. Therefore, this patellofemoral joint model was modified to take account of the effect of the hip flexion/extension before being incorporated into the lower limb model.

4.3 Mechanical analysis

The anatomical model is divided into three separate segments, the foot, the shank and the thigh. Dynamic analysis of the segmental systems was performed using Newton's theory. Doing so the resultant joint forces and moments can be deduced.

For the foot segment the forces and moments working on the free body are as follows:

![Figure 4.3.1: The foot segment with the forces and moments working on it.](image)

The total sum of these forces should be equal to the mass of the foot multiplied by its acceleration, stating this the resultant force can be determined.

\[ \sum F_{O_f} = m_f \ddot{a}_f \]

\[ \vec{F}_i + \vec{R}_a + m_f \vec{g} = m_f \ddot{a}_f \]

\[ \vec{R}_a = m_f \ddot{a}_f - m_f \vec{g} - \vec{F}_i \]

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with: \( \vec{R}_a \) is the resultant force at the ankle,
\( m_f \) is the mass of the foot,
\( \vec{a}_f \) is the acceleration of the foot,
\( \vec{g} \) is the gravitational acceleration,
\( \vec{F}_i \) is the ground reaction force.

The total sum of the moments should be equal to the inertia moment round the centre of mass of the foot, the resultant moment can be determined.

\[
\sum M_{O_f} = \vec{H}_f \\
\vec{r}_{df} \times \vec{F}_i + \vec{r}_{pf} \times \vec{R}_a + M_a = \vec{H}_f \\
\vec{M}_a = \vec{H}_f - \vec{r}_{df} \times \vec{F}_i - \vec{r}_{pf} \times \vec{R}_a
\]

with: \( \vec{M}_a \) is the resultant moment at the ankle,
\( \vec{H}_f \) is the inertia moment round the centre of mass of the foot,
\( \vec{r}_{df} \) is the vector between the centre of mass of the foot and the centre of pressure of the ground reaction force on the foot,
\( \vec{r}_{pf} \) is the vector between the centre of mass of the foot and the centre of pressure of the resultant force in the ankle on the foot.

For the shank segment the free body with the forces and moments working on it is as follows:

Figure 4.3.2: The shank segment with the forces and moments working on it.

From the sum of forces the resultant force in the knee can be determined in the same way as is done with the foot.
\[ \vec{R}_k = \vec{R}_a + m_a \vec{a}_a - m_i \vec{g} \]

with: \( \vec{R}_k \) is the resultant force at the ankle,
\( m_a \) is the mass of the shank,
\( \vec{a}_a \) is the acceleration of the shank,

From the sum of moments the resultant moment in the knee can be determined.

\[ \vec{M}_k = \dot{H}_s + \vec{r}_{ds} \times \vec{R}_a - \vec{r}_{ps} \times \vec{R}_k + \vec{M}_a \]

with: \( \vec{M}_k \) is the resultant moment at the knee,
\( \dot{H}_s \) is the inertia moment round the centre of mass of the shank,
\( \vec{r}_{ds} \) is the vector between the centre of mass of the shank and the centre of pressure of the resultant force in the ankle,
\( \vec{r}_{ps} \) is the vector between the centre of mass of the shank and the centre of pressure of the resultant force in the knee.

For the thigh segment the free body with the forces and moments working on it is as follows:

Figure 4.3.3: The thigh segment with the forces and moments working on it.

From the sum of forces the resultant force at the hip is derived.

\[ \vec{R}_h = \vec{R}_k + m_i \vec{a}_i - m_i \vec{g} \]

with: \( \vec{R}_h \) is the resultant force in the hip,
\( m_i \) is the mass of the thigh,
\( \vec{a}_i \) is the acceleration of the thigh.
From the sum of moments the resultant moment can be determined.

\[ \vec{M}_h = \vec{H}_t + \vec{r}_{dt} \times \vec{R}_k - \vec{r}_{pr} \times \vec{R}_h + \vec{M}_k \]

with: \( \vec{M}_h \) is the resultant moment in the hip,
\( \vec{H}_t \) is the inertia moment round the centre of mass in the thigh,
\( \vec{r}_{dt} \) is the vector between the centre of mass in the thigh and the centre of pressure of the resultant force in the knee,
\( \vec{r}_{pr} \) is the vector between the centre of mass in the thigh and the centre of pressure of the resultant force in the hip.

### 4.4 Limiting solutions

There are fourteen model unknowns for twelve structural members. The ankle has five unknowns. The force is transmitted by the gastrocnemius, the soleus and the tibialis anterior, the magnitude of the talotibial contact force and its direction, and it has four structural members. The orientation of other members is determined from the geometric model. The knee has six unknowns, the quadriceps, the hamstrings, the gastrocnemius, the ACL, the PCL and the magnitude of the tibio-femoral contact force, and it has six structural members. The direction of the tibiofemoral contact force is uniquely determined by the geometry of the four-bar linkage model. The hip has got six unknowns: the iliopsoas, the quadriceps, the hamstrings, the gluteus maximus, the magnitude of the femoro-pelvic contact force and its direction, and it has five structural members.

From inverse dynamics the three force components of the resultant forces were obtained at each joint, the horizontal force, the vertical force and the moment. This is explained in paragraph 4.3. Nine equations of force equilibrium were expressed in terms of the fourteen model unknowns. This is an indeterminate set of equations. To solve this problem the limiting solutions approach is used.

Before explaining this method, two more assumptions have to be mentioned. The first is that the tension is the same throughout the length of the muscle. The second is that the rectus femoris contributes one-quarter of the tensile force developed in the quadriceps tendon. The latter assumption is made on the basis of the relative physiological cross sectional area of the rectus femoris.

The problem of indeterminacy is solved by solving a series of reduced determinate problems, which are established by considering only nine unknowns at any one time, setting the values of other forces to zero or the unknown directions of the contact forces at the ankle and the hip to some reasonable value. These problems give limiting solutions. From the standpoint of geometry and mechanics, a significant number of these reduced determinate problems will produce unrealistic limiting solutions. Possible combinations of unknowns were considered for each joint in isolation. Overlapping bi-articular combinations were eliminated. Some could be called inadmissible because muscle and ligament forces must be tensile and joint contact forces compressive. The reduced determinate problem is solved using a Lower-Uppper-decomposition.

Measured data (force data) are forwarded to the system as data files.
4.5 The software

The model was implemented by Lu in C under UNIX. The Vicon system gives two data files, one for the spatial data (marker positions) and one for the analog data (EMG and force plate data). They are transformed from binary- into ASCII-files, and then from DOS into UNIX.

The overall software includes four major modules; (1) the Vicon data pre-processing (2) the dynamic analysis (3) the geometric modelling and (4) a force distribution module (first described by Collins [1]).

(1) During the capturing of data with the Vicon system, it can happen, that the cameras do not spot a certain marker at certain intervals in time, because it is hidden behind a certain part of the subjects body. As a result, the trajectory of this marker is not defined during the whole trial. The first module of the software calculates values for these marker positions by using interpolation. A new spatial data file is thus obtained. When the markers are defined at all instances in time this part of the program can be omitted.

(2) The new spatial data file and the analog file are input to the second module of the software. This program calculates the resultant forces and moments of the joints and the joint angles and stores them in a new data file, the marker positions are stored in a separate file.

(3) These two files are input to the third module, the geometric modelling. This part determines the lines of action and the lever arms of the load-bearing structures.

(4) The force distribution module calculates all the possible solutions for distributing the resultant joint forces and moments among nine of all the muscles and ligaments and contact forces present in the joints. This gives 498 possibilities. From this large amount of possibilities, a selection has to be made using the following assumptions, muscle force is not compressive and the contact force is not tensile. An output file is generated containing the limiting solutions. For every solution at least three anatomical structures are active at each joint, and information is given about the magnitude of force in these. After running this program a small amount (generally varying between 5 and 25) of possible solutions will remain.

The EMG data are filtered and plotted using MATLAB. The filter used is a second order Butterworth digital filter, with a cut-off frequency of 6 Hz and performed zero-phase forward and reverse digital filtering once. Using the EMG information a selection can be made between the remaining limiting solutions on ground of muscle activity.

An advantage of this model is, that it can also be used for example for patients with a ruptured ACL. A solution without ACL can be selected.

Lu has also developed a similar two dimensional model for the lower limb that contains eight muscles instead of seven, the quadriceps is divided into two separate muscles: the rectus femoris and the vastus medialis. This gives a larger amount of possible solutions (1179). This model however did not give answers with rectus and vastus activity at the same time during sitting down and rising up from a chair, even though the EMG data showed that they were working simultaneously. With rectus and vastus both being knee extensors, you would also expect them to work together. This is the reason why, after running a few test trials, was decided not to use the eight...
muscle model. Naturally this problem did not occur when using the seven muscle
model, as the rectus and vastus were modelled together as the quadriceps.
5 The squatting and sitting & rising experiment

Squatting and sitting down and rising up from a chair are two of the rehabilitation exercises that patients with knee injuries or ACL-replacements perform. As is said in the introduction, current rehabilitation practice has developed as a result of trial and error. To understand why some fail and why others are successful, could help developing new protocols. This experiment was performed to obtain values for the internal knee joint forces in order to compare these forces for different exercises. In this experiment, half squats were performed, from fully extended hip and knee to flexion of \(\pm 90\) degrees for both hip and knee. The squatting was performed with and without lifting the heel, also squatting on one leg with keeping the heel on the ground. An other performed exercise was sitting down and rising up from a chair. The latter exercise can be stated as less controlled than the former ones, this specific exercise is usually performed in an earlier stage of rehabilitation then squatting. The exercises were performed at a calm tempo. This is just a small variety out of the whole range in which squatting can be performed, a selection had to be made. Knowing the forces in the knee a comparative value can be given to these exercises.

5.1 The experiments

Subjects
The experiments were performed on four male subjects, varying in age from 24 to 35 years old. Their weight varied from 70.0 to 82.7 kg with an average of 77.1 kg, their height varied from 1.76 to 1.84 m with an average of 1.80 m.

Instrumentation
The Vicon system and the force plate, that are described in chapter 3, were used to capture data. The Vicon system was calibrated as is described in paragraph 2.3. Seven cameras were situated around the subject so that every marker was detected by at least two cameras. The force plate was balanced, all outputs set to zero.

Preparation
Marker balls were stuck on the subject, to indicate certain anatomical landmarks, that were needed for the calculations with the two-dimensional mathematical model. The markers indicate the position of the skin, overlaying the bone, at a certain instant. To be able to use this information for the position in time of the bone, there has to be as little skin movement relative to the bone as possible. This fact has been considered in choosing the landmarks. The markers used were the following:
- RTOE: The right toe, the most proximal lateral aspect of the fifth metatarsal.
- RFOO: The right foot, the prominence of the navicular tubercle.
- RHEE: The right heel, the posterior projection where the plantarflexor (Achilles-) tendon inserts on the calcaneus.
- RLMA: The right lateral malleolus.
- RMMA: The right medial malleolus.
- RTT: The right tibial tuberosity.
- RSHA: The right shank, most lateral and distal part of the head of the fibula.
- RLFC: The right lateral femoral condyle.
- RMFC: The right medial femoral condyle.
- RTHI: The right thigh, midpoint of the line joining the lateral condyle and the greater trochanter.
- RTRO: The most lateral projection of the right greater trochanter.
- RASI: The right anterior superior iliac spine.
- LASI: The left anterior superior iliac spine.
- RPSI: The right posterior superior iliac spine.
- LPSI: The left posterior superior iliac spine.

While the subject stood still, a trial was captured without using the force plate. This is called a subject calibration. It is used to calculate the positions of markers relative to each other. As the medial markers (RMFC, and RMMA) are easily disturbed during motion, they were later removed together with the LPSI marker. With the calibration information the program can still calculate values for their positions. Taking a subject calibration is necessary, when a three-dimensional mathematical model of the lower limb is used.

After doing this, electrodes were stuck on the subject, two on each muscle of interest one centimetre apart from each other. EMG signals were taken from the following muscles:
- The rectus femoris (part of the quadriceps).
- The vastus medialis (part of the quadriceps).
- The biceps femoris (part of lateral hamstrings)
- The gastrocnemius medialis.
- The gluteus maximus.
- The tensor fascia latae.
- The tibialis anterior.

In the mathematical model of the lower limb (see paragraph 3.2) the iliopsoas and the soleus were also modelled. As these muscles lie very deep, no EMG-signals could be taken from them. For the sitting and rising experiment, similar EMG data was found in literature ([14] and [15]). Munton et al [15] also collected EMG from the soleus. This information was later used to select answers from the remaining limiting solutions, see paragraph 4.5.

Procedure

Two trials were taken of the subject performing four half squats with lifting the heel from the ground. The same was done with keeping the heel on the ground while squatting. After this the subject sat down on a stool 20.5 cm high and stood up again twice. Five of those trials were recorded. The last exercise was squatting with one leg, keeping the heel on the ground. Three trials of three repetitions each were taken. All the exercises were performed at medium speed (20 sec. for sitting down and rising).

After the exercises the height and weight of each subject were measured.

5.2 Results

Data processing

The data obtained from these experiments was input for the software of the two dimensional model of Lu (see paragraph 4.5). As it takes a lot of calculation time and manual effort to run the software, only the fifteen trials for sitting down and rising up
from a chair for three subjects were processed. Although this amount of data is too small to draw conclusions about the sitting and standing that are statistically well grounded, this study still has relevance as a pilot study for testing the lower limb model and for research in rehabilitation exercises.

**Selecting answers**

From the possible limited solutions, answers were selected using the EMG-data. Only those solutions were selected, that contain muscle activity that is consistent with the EMG-data. The EMG signals for one representable trial from subject 1 are given as a fraction, of the largest amplitude recorded during that trial of the filtered absolute signal, minus a threshold value [23].

![EMG signals representing muscle activity](image1.png)

![EMG signals representing muscle activity](image2.png)
All five the muscles are continuously active, except for the tibialis anterior. This muscle is not active at the beginning of the sitting down and at the ending of the rising up.

Just the action of sitting down and rising up was examined. During sitting the model can not be used, as there is a contact force between the subject and the chair, that has not been taken into account in the model.

Five trials from three subjects each were studied. The fifteen trials had an average of 435 frames each. For every frame an average amount of 20 limiting solutions was given. For every trial one sitting down and one rising up action was studied more closely. In Figure 5.2.2 an example of the list of possible limiting solutions for one frame is given.
**Figure 5.2.2:** An example of the limited solutions for one frame during sitting down and one frame during rising up.

The first three set of active muscles and/or contact force (C) and/or direction of contact force (D) represents the forces in the ankle joint. The second set of active muscles and/or ligaments and/or contact force represents the forces in the knee. The third set of active muscles and/or contact force and/or direction of contact force represents the forces in the hip.

No. of solution predicted: 25

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**Figure 5.2.2:** An example of the limited solutions for one frame during sitting down and one frame during rising up.

The first three set of active muscles and/or contact force (C) and/or direction of contact force (D) represents the forces in the ankle joint. The second set of active muscles and/or ligaments and/or contact force represents the forces in the knee. The third set of active muscles and/or contact force and/or direction of contact force represents the forces in the hip. The knee can also be influenced by active muscles that are listed in the hip set, for example the hamstring strap, that can catch the hip and at the same time can flex the knee. The nine columns with numbers are the values of the forces in respective order. In literature [5] muscle forces of the quadriceps, the hamstrings, and the gastrocnemius during rising from a chair were found of the same order of magnitude as the forces calculated by the model.
For thirteen of the fifteen trials during these intervals more than one (and less than seven) answers could be selected this way. For the other two trials during a minor part of the action, only one answer was possible.

**Validation**

The EMG-data serve as validation of the limiting solutions. The EMG-data was consistent with other EMG-data found in literature [14], [15].

Figure 5.2.3: Leg muscle activity when rising from a chair [15].

The only difference was, in this study, the gastrocnemius worked during sitting down and rising from a chair, whereas in [15] was stated that it was not active. In the next part, it will be shown that it is reasonable to expect gastrocnemius activity.

Looking at the line of action of the ground reaction force relative to the posture of the lower limb during sitting and rising gives an expectation of the muscles that will be working at that certain instant. The information about the line of action of the ground reaction force is provided by the Vicon system. In Figure 5.2.4 five frames are given, representing the right leg of one of the subjects, from standing to sitting position. For the other trials/subjects similar pictures were obtained. The force plates (from which only one was used), and the ground reaction force are also shown. The small circles represent the marker balls. The lines connecting them are just there for clarity, they do not represent bones.
Figure 5.2.4: Vicon images, representing the lower limb, from standing to sitting position.
As the exercise was performed slowly (sitting down and rising up in ± 20 seconds) the dynamic influence will be small. This allows the following analysis:

When the line of action passes in front of the ankle joint, the external moment will try and dorsiflex the ankle, the internal moment that will try and compensate for this is provided by the action of the soleus and the gastrocnemius. When the line of action passes behind the ankle the tibialis anterior will start working. When the line of action passes in front of the knee the knee will be extended, the hamstrings and the gastrocnemius will try and compensate for this by flexing it. When the line of action passes behind the knee the quadriceps will become active. When the line of action passes in front of the hip the hip will have an external flexing moment. The hip extensors will become active, namely the hamstrings and the gluteus maximus. When it passes behind the hip the hip flexors, the rectus femoris and the iliopsoas, will try to compensate.

During sitting down and rising up from a chair, the line of action of the ground reaction force passes in front of the hip, behind the knee and usually in front of the ankle. Occasionally the line of action passes behind the ankle. In this experiment it passed in front of the ankle during twelve of the trials, during the other three trials it varied. The ankle joint was always close to the centre of pressure of the ground reaction force and therefore the moment arm of the force was very small. The EMG-data showed that both the ankle dorsiflexor (the tibialis anterior) and the plantarflexor (the gastrocnemius) were working. No EMG-data from the soleus was taken. Even though these antagonistic muscle groups are working at the same time, it may be interesting to see whether the placement of the foot relative to the line of action of the contact force is of considerable influence on the forces in the lower limb. This was not done.

A prediction of the active elements in the knee can also be made, using the knee model from Figure 3.2.2. When the knee is flexed and the hamstrings are active, there is a horizontal force that tries to pull the tibia posteriorly. To counterbalance this a horizontal force is needed that will pull the tibia anteriorly. This force can be provided by the PCL. The more the knee is flexed, the more vertical the PCL, the smaller the horizontal component of the PCL force, the bigger this force needs to be. When the quadriceps are working, the patella tendon will pull the tibia anteriorly. To compensate for this, the ACL will try and pull the tibia posteriorly. The more the knee is extended, the more vertical the ACL, the smaller the horizontal component of the ACL force, the bigger this force needs to be. When the quadriceps and the hamstrings are working together, so called cocontraction, they will counterbalance each other partly or completely. Depending on the forces in these muscle groups the PCL or ACL will work to compensate. The EMG-data show that, during both sitting and rising, cocontraction at the knee occurs.

As is mentioned in paragraph 4.3 the model assumes the ligaments to be inextensible. If the model were to contain an extensible ligament model, like is described by Zavatsky and O'Connor in [24], the forces in the ligaments would be smaller. For example when the hamstrings are active they will pull the tibia further posteriorly. The PCL stretches and will then be more horizontal. To counterbalance for the same force in the hamstrings a smaller force in the PCL is sufficient. The smaller PCL force will have the same horizontal component as a bigger PCL force in the inextensible ligament model. The same goes for the ACL: when the quadriceps are active, the tibia is pulled further anteriorly, the ACL is in a more horizontal position.
and therefore to have the same horizontal component as in the inextensible ligament model, the force in the ACL will be smaller.

Another factor that adds to this effect, is that because of the tibial displacement, the muscles will be directed in a different position. For example when the hamstrings are active and the tibia is pulled posteriorly, the hamstrings will be more vertical and the patella tendon more horizontal. This means that the horizontal component of the hamstring force will be smaller and that of the patella tendon will be bigger, therefore to counterbalance the hamstrings, a smaller force in the PCL will be sufficient. When the quadriceps are active, the tibia will be pulled anteriorly. As a result the patella tendon will be more vertical and its horizontal force component will be smaller, the hamstrings will be more horizontal and its horizontal force component will be bigger. As a result the force in the ACL will be smaller. When interpreting the results one has to keep in mind that because of these effects the actual forces in the ligaments will be smaller.

From the literature [18], it was found that the mean strength in the anterior PCL (aPCL) bundle is 1620 ± 500 N and in the posterior PCL (pPCL) bundle is 258 ± 83 N. These values apply for subjects over 50. For younger subjects (under 26 years) these values can be multiplied by 2.5, so that the strength in the aPCL becomes 4050 N and in the pPCL 645 N.

Results

After answers were selected for all the trials it was found that for one of the subjects, one of the five trials differed considerably from the others. This trial will not be taken into account when giving a list of the general answers. Among the subjects the answers differed, with respect to the active muscles, and/or ligaments, and/or (direction of) contact force, only in so far that the tibialis anterior started and stopped being active at different angles of knee flexion. In general the selected answers were divided in groups, starting being active at certain knee flexion angles for sitting down:

* at 10 degrees GCD-QHC-LIC,
    GCD-QHC-LCD,

* at 40 degrees GCD-QHC-LIC,
    GCD-QHC-LCD,
    GCD-QPC-HLC,

* at 50 degrees TGC-QHC-LIC,
    TGC-QHC-LCD,
    TGC-QPC-HLC,

* at 90 degrees TGC-QHC-LIC,
    TGC-QHC-LCD,
and for rising:
* at 95 degrees TGC-QHC-LIC,
TGC-QHC-LCD,

* at 90 degrees TGC-QHC-LIC,
TGC-QHC-LCD,
TGC-QPC-HLC,

* at 50 degrees GCD-QHC-LIC,
GCD-QHC-LCD,
GCD-QPC-HLC,

* at 40 degrees GCD-QHC-LIC,
GCD-QHC-LCD.

Four of the trials of one subject also contained possible answers with soleus activity at the beginning of sitting down and at the end of rising. [15] Shows that the soleus is active during standing.

The moments around the ankle joint, the knee joint, and the hip joint for sitting down and rising up from a chair are plotted against knee flexion angle. One representable trial for each subject is given.

![Joint moments for subject 1. (S = sitting down, R = rising up)](image)

Figure 5.2.5: Joint moments for subject 1. (S = sitting down, R = rising up)
Joint moments during sitting down and rising from a chair against knee flexion angle (subject 2).

Figure 5.2.6: Joint moments for subject 2. (S = sitting down, R = rising up)

Joint moments during sitting down and rising from a chair against knee flexion angle (subject 3).

Figure 5.2.7: Joint moments for subject 3. (S = sitting down, R = rising up)

For the five trials from each subject, the ACL- and PCL forces during sitting down and rising up from a chair, are plotted against knee flexion angle. As is mentioned before, there is for the mayor part of the trials, more than one possible solution. There are solutions containing ACL- or PCL force, and solutions without ACL- or PCL force. Only the possibilities are plotted with ACL and/or PCL force.
ACL and PCL forces divided by body weight during sitting down and rising up from a chair (subject 1)

Figure 5.2.8: ACL and PCL force for subject 1, including the considerably different result.

ACL and PCL forces divided by body weight during sitting down and rising up from a chair (subject 1)

Figure 5.2.9: ACL and PCL force for subject 1, excluding the considerably different result.
Figure 5.2.10: ACL and PCL force for subject 2.

Figure 5.2.11: ACL and PCL force for subject 3.

The maximum PCL force for subject 1 varies from 1.5 to 2.4 times body weight. For subject 2 it varies from 0.8 to 1.7 body weight, and for subject 3 from 2.5 to 3.6 times body weight.
Source of error

After running the model for all the trials, it was discovered that the origin of the forceplate coordinate system was slightly different, than was programmed in the software. Lu then altered the origin data in the second part of the software, that calculates the forces moments and joint angles. With the altered software some trials were calculated again, and the results were compared with the previous results. The new results are consistent with the Vicon pictures from figure 5.2.4. The moments clearly changed: the maximum knee moment for the altered software was smaller than the previous. It now varies from 20 to 30 Nm for the three subjects. The maximum absolute value of the hip moment was bigger and now varies from 90 to 140 Nm for the three subjects. The maximum absolute ankle moment also changed and varies from 20 to 50 Nm. The PCL forces however were still of the same order of magnitude. Considering this, and the amount of time involved to recalculate all the trials, it was decided to use the old limiting solutions. The new moment values were studied to get an idea of the moments that are involved. For each subject the moments, calculated with the new origin coordinates are plotted. The same trials as in the Figures 5.2.5, 5.2.6, and 5.2.7 are given.

Figure 5.2.12: New joint moments for subject 1. (S = sitting down, R = rising up)
Figure 5.2.13: New joint moments for subject 2. (S = sitting down, R = rising up)

Figure 5.2.14: New joint moments for subject 3. (S = sitting down, R = rising up)

From all these results can be concluded, that there is hardly any difference between sitting down and rising up.
Conclusion

The expectations that are mentioned in the part validation, are that L and H are active at the hip, and that Q is active at the knee, and that S and G are active at the ankle. But when H is active at the hip, it will also be active at the knee. Therefore the force in Q has to be bigger than the force in H to counterbalance the external flexing moment. In [5] was found that during rising from a chair the force in the quadriceps was bigger than the force in the hamstrings at any degree of knee flexion. This rules out the solution which includes PCL activity, as according to the calculations, the force in H is bigger than that in Q. An other reason to doubt the solutions with the PCL, is that the force in the PCL is extremely big compared to the maximum values found in literature [18].

This will lead to the suggestion that in the knee cocontraction counterbalances the muscle- and external forces.

Cocontraction is also found to take place in the ankle and in the hip according to the EMG signals. The solutions that remain contain the possibility of iliopsoas action at the hip and no iliopsoas action at the hip.

J. Perry [17] found that in level walking cocontraction rarely took place. The EMG signals Collins and O’Connor [2] recorded during gait, showed there was no cocontraction in the ankle, and the hip. In the knee cocontraction only took place in the beginning of the early stance phase, and in the last part of the mid swing phase, and during the late swing phase. Expressed in percentages, cocontraction at the knee took place during 35% of the gait cycle.

Lu [12] also investigated gait patterns by using the same Vicon system and the same two-dimensional model, as are used in this study. The solutions calculated with the model, showed ACL force ranging from 0.1 to 1.2 body weight (BW) and a maximum of 1.5 BW in the early stance phase. The maximum PCL force was 0.4 BW at 50% walking cycle (no cocontraction at that moment). Comparable values were found by Collins [2]. Compared with sitting down and rising up from a chair, in walking hardly any cocontraction occurs, and there are ligament forces present.

5.3 Discussion

In the two-dimensional model only seven muscles are modelled. Although the muscles, that are not modelled, have small cross-sectional areas and/or intrinsic muscle tendon lever-arms, they will generate (small) forces and therefore slightly alter the force distribution results that are found.

The muscles, tendons and ligaments are modelled as being inextensible. In reality muscles contract eccentrically and concentrically. Ligaments are in reality also extensible. The PCL and ACL forces are therefore smaller, as is explained in the part validation of paragraph 5.4.

The problem of indeterminacy was solved by considering only nine of the fourteen unknowns at any instant, setting the values of the other unknowns to zero. In reality all the unknowns can have values, although small.

During the experiment, the trajectories of the markers are captured. However careful the markers are placed on the subject’s lower limb, a small amount of skin movement cannot be avoided. This will also have influence on the results.
No EMG signals could be taken from the soleus and the iliopsoas as these muscles lie to deep in the lower limb. For the soleus, EMG data were found in literature [15], for the iliopsoas this was not the case.

Another aspect of the EMG data, that has to be mentioned, is that it is subject to the way the signals are filtered and the chosen threshold. If these two variables are changed, the muscle activity curves alter slightly. The curves were compared to EMG results from literature.

The fact that the origin coordinates programmed in the software differed slightly from the actual coordinates, as mentioned in paragraph 5.2, did not seem to influence the PCL and ACL values much. It did influence the moment values, therefore the moments were calculated again with the actual coordinates.

As is mentioned in the introduction of paragraph 5.2, the number of subjects and trials are too small to draw statistically grounded conclusions. This is why, in presenting the results, no statistical analysis is performed.

When looking at the results; one must bear these imperfections in mind.

Further study

It is interesting to understand why the eight muscle model could not be used for these experiments. An other interesting project is to compare the results of the sitting and rising experiment with the results of the different squatting experiments. As is mentioned in the part validation, the influence of the ankle position, relative to the line of action of the ground reaction force, on the results for the sitting down and rising up can be studied, as well as the influence of performing speed and chair height.
6. Literature review about pedals

Several researchers have taken an interest in cycling. Articles about different aspects of cycling have been published.

6.1 Pedal dynamometers

In 1981 M.L. Hull and R.R. Davis [9] developed a pedal dynamometer that measures six-axis pedal loading during bicycling. In 1988 J. Newmiller and M.L. Hull [16] presented a pedal that measured two components of the pedal loading, the force normal to the pedal and the force in the plane of the pedal in the direction of the (imaginary) bicycle motion. These components are sometimes referred to as the driving loads. The other four forces are called the out of plane loads.

Figure 6.1.1: The definition of foot-pedal reactive load components, crank-arm angle $\theta_1$, and relative pedal angle $\theta_2$. 

\[ \text{CRANK ARM} \]
\[ \text{PEDAL} \]
The complexity of the two different pedals is about the same. The calibration of the two pedals differs. To calibrate the six-component pedal a calibration apparatus is described in [9]. Calibrating the two-component pedal should be less complex, although in [16] the same calibration apparatus is used. This is done to see if the out of plane loads are of influence to the measurement of the driving forces.

In [16] is stated that only the driving loads develop torque about the crank spindle. The two-component pedal could be used in a broad spectrum of projects in cycling biomechanic research. In the same article it is said that the six-component pedal is relevant in research on overuse knee injuries from cycling. In [9] is stated that the out of plane loads are not insignificant. P. Ruby, M.L. Hull and D. Hawkins (1992) [19] state in another article that when the pure moments, developed between the foot and the pedal during cycling, are neglected this leads to a substantial error in computing axial and varus / valgus moments in the knee. In their study, they used the six-component pedal dynamometer from Hull and Davis [9].

In the research project of D. Toutoungi [22] a two-dimensional model of the lower limb will be used to calculate the ligament forces. This means that three load components have to be measured, Fz, Fx and My. The anterior cruciate ligament resists internal rotation and anterior draw. That is why the ligament will be affected by out of plane loads. This together with the findings of Ruby, Hull and Hawkins leads to the idea that it is interesting to use a three-dimensional model. In doing so, all six load components have to be measured.

Bearing this in mind, we choose to use the six-component pedal dynamometer.

6.2 The foot-pedal connection

D.J. Sanderson [20] looks into the pressure distribution in the foot depending on the type of shoe one wears during cycling, hard-soled cycling shoes or soft-soled running shoes. He reports that the peak pressures in the forefoot region are significantly higher for the hard-soled cycling shoe then for the soft-soled running shoe. When the position of the foot on the pedal can be changed during cycling it affects the path of motion of the knee joint. The loading pattern of the tissues within and around the knee joint change as well. Davis and Hull [4] also did a foot-pedal connection study. Their outcomes were the following: (i) a cleated shoe improves the pedalling efficiency, (ii) with toeclips additional efficiency is achieved early in the pedalling cycle, (iii) the maximum torque level decreases with the addition of toeclips and cleats, (iv) torque history changes dramatically with the addition of cleats.

The purpose of this research project is to determine the forces in the knee during rehabilitation exercises. In the experiment that will be done to calculate these forces, the rehabilitation exercise situation has to be approximated. The type of shoe the test subject or patient wears will therefore be a soft-soled shoe that is not fixed to the pedal.

6.3 Indoor and outdoor experiments

P.D. Soden and B.A. Adeyefa [21] made a distinction between indoor and outdoor experiments. In the indoor experiment the subject was mounted on a stationary bicycle with the rear wheel removed and replaced by a dynamometer to simulate resistance to motion. In the outdoor experiment the rear wheel was refitted to the
bicycle. The wires of the instrumented pedal were attached to the riders leg. Hull and Davis [9] do an indoor experiment, where the subject is seated on a normal bicycle that is placed on rollers. There is significant frictional resistance in the rollers. This simulates actual cycling because both wheels revolve.

It is our purpose to approximate the rehabilitation situation instead of the real-life cycling outside situation. We choose therefore an exercise bicycle that is often used in exercises by physiotherapists.
7 Instrumentation

7.1 The six-component pedal dynamometer

The pedal described by Hull and Davis [9], that measures the six load components on the pedal, consists of four subsystems: the dynamometer, the pedal position measurement subsystem, signal conditioning electronics and a minicomputer.

The dynamometer is contained within a modified pedal body, underneath the pedal spindle. Thirty-two strain gauges are connected into eight fully temperature compensated Wheatstone bridge circuits. A strain gauge is a calibrated metal wire or foil, that undergoes a very small change (strain) in one of its dimensions. This mechanical deflection, usually a fraction of 1 \%, causes a change in resistances connected as a bridge circuit, resulting in an unbalance of voltages proportional to the strain. The strain ring orientation, the strain gauge location and the Wheatstone bridge circuits are given in Figure 7.1.1.

\begin{figure}[h]
\centering
\includegraphics[width=\textwidth]{figure711.png}
\caption{Figure 7.1.1: a. The strain ring orientation. b. The strain gauge location. c. The Wheatstone bridge circuits.}
\end{figure}

When a vertical load \( F_v \) is applied, there will be a strain in the strain gauges 1 to 4, no strain will be detected in 5 to 8. When a horizontal load \( F_h \) is applied, the strain gauges 5 to 8 will be strained, the strain gauges 1 to 4 will not be strained. Transverse loads will give no output.

The pedal position measurement subsystem measures the crank arm angle \( \theta_1 \) and the pedal angle relative to the crank arm \( \theta_2 \) with single turn, continuous rotation potentiometers. For the definition of these angles see Figure 4.1.1. For the relative pedal angle the potentiometer body is fixed to the modified pedal housing and the
The wiper is connected to the pedal spindle. For the crank arm angle the potentiometer body is mounted to the bicycle frame and a gear is attached to the wiper. The wiper gear meshes with an identical gear, which attaches to the chainring and revolves with the crank about the bottom bracket spindle axis. The gear ratio is 1 : 1, so the relation between the potentiometer output voltage and the crank arm position is unambiguous.

The signal conditioning electronics is based on carrier amplification. The carrier oscillator signal is input to the transducer bridges. Each Wheatstone bridge output is amplified by a high gain differential carrier amplifier. The signal is then demodulated to a DC signal and the output is amplified by low gain and filtered with a second order anti-aliasing filter.

The minicomputer digitises, stores and analyses the transducer data. For our purpose it is not necessary that the minicomputer analyses the data, just the values of the forces are needed.

### 7.2 The exercise bike

The pedal will be mounted on an exercise bicycle. As is mentioned in paragraph 4.3 we choose to use a simple exercise bicycle of the type: Tunturi family. The bicycle has got a saddle and a handlebar that are both adjustable in height. Resistance is provided by a belt going round the single wheel. The belt can be tightened by a spring, that has got five different settings.

To do a cycling experiment the resistance has to be adjustable to a certain amount of resistance. The data you will obtain from the experiment has to be reproducible. To measure whether the resistance is fixed for every different setting a loadcell is used. One side of the loadcell is horizontally attached to the frame and the other is attached to the resistance belt (see Figure 7.2.1).

![The exercise bicycle with the loadcell.](image-url)
The loadcell is connected to a straingauge instrument (type 581 DNH), that
supplies a micro-voltage output. The output has to be conversed into Newtons, this
can be done by using a conversion graphic, which can be obtained by the following
loadcell calibration. One side of the loadcell is vertically attached to a fixed frame,
while on the other side a tray is hanging with a known weight. Weights are placed on
the tray, so the load on the loadcell is known. The voltage output belonging to the
load can be read from the straingauge instrument.

7.3 Remark

I looked into the feasibility of making the instrumented pedal, however the pedal
could not be realised in the amount of time given for this project. Therefore the
calibration and connection to the Vicon system could not be done.
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