Total knee arthroplasty biomechanical reflections and modelling, based on quantitative movement analysis

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Total knee arthroplasty
Biomechanical reflections and modelling, based on quantitative movement analysis

A dissertation submitted to the
SWISS FEDERAL INSTITUTE OF TECHNOLOGY
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for the degree of
DOCTOR OF TECHNICAL SCIENCE

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This thesis is written in memory of Markus Meier who passed away two months before my thesis presentation. He was a wonderful person, inspiring me with his energy for my own work.
Si tu veux construire un bateau, ne rassemble pas des hommes pour aller chercher du bois, préparer des outils, répartir les tâches, alléger le travail mais enseigne aux gens la nostalgie de l’infini de la mer.

Le petit prince, Antoine de Saint-Exupéry
Abstract

Currently more than 1'000'000 total knee arthroplasties, TKAs, are implanted world wide each year, thereof 40'000 TKAs in Switzerland. Because of higher life expectancies it is expected that the number of total knee implantations will increase in the next few years. The goals of a TKA are longevity of the implant components and high satisfactory patient rates. Today the rate of a good patient outcome is around 85%, which still leaves a significant absolute number of patients needing early revision surgery. The most common causes for revision surgery are polyethylene (PE) wear, loosening, knee instability and infection. PE wear, loosening and instability are factors associated with altered joint biomechanics after total knee replacement. The femoral component rotational alignment profoundly affects the knee joint’s mechanics in flexion as well as in extension, in all six degrees of freedom. A malrotated femoral component could lead to ligament unbalancing causing lateral flexion instability and pain while standing up from a chair or walking down stairs. In order to avoid these problems a comprehensive mechanical understanding of the knee joint as regards to the alignment of the implant component in TKA is important. A three dimensional computer based model visualising the joint’s kinematics during different motion patterns contributes to this understanding and makes it possible to estimate the load at the knee joint. Such a model requires accurate in vivo kinematic and kinetic data to visualise and calculate the load at the joint of different motion patterns of daily activities. Previous investigations in gait analysis used kinematic and kinetic information from skin mounted markers and force plates during level walking. The problem of this measurement technique is the large error in kinematic data acquisition caused by the movement of the
skin and muscles relative to the underlying bone. Video-fluoroscopy enables the measurement of kinematics of implant components more accurately by a three-dimensional numeric reconstruction of the single plane projection view in the fluoroscopic images, thus avoiding skin movement artefacts. However, this technique is limited to the field of view of the fluoroscopic screen. This problem was solved by using a motor driven trolley built in the laboratory to carry the fluoroscopic unit (x-ray source, image intensifier, c-arm). This movable system allows the tracking of the knee joint during level walking, and a sit down task. An intensity based registration algorithm reconstructs the six degrees of freedom of the implant components relative to the focus of the fluoroscope. The three dimensional reconstruction is within a translational accuracy of 3.1 mm and a rotational accuracy of 1.6°.

Video-fluoroscopy only acquires kinematic data. This means that the loads in the knee joint can not be estimated. In this study was force plate data coupled with the moving fluoroscopic system enabling inverse dynamic calculation. The unit mover was optically tracked by VICON in order to transform the fluoroscopic coordinate system into the global coordinate system with its origin on the centre of one force plate, thereby coupling the fluoroscopic system with the force plate. This transformation was performed within an accuracy of ± 1 mm.

This measuring system results in seven times more accurate inverse dynamic calculation than classic instrumented gait analysis would achieve.

The local mechanics of the total knee was visualised in a three dimensional computer model. The model included TKA geometry from CAD software, and bone geometry from CT scans of the individual. It visualised the kinematics of the implant components during the activities mentioned above. Furthermore, femoral component malrotation was simulated to estimate the alternated strain at the posterior cruciate ligament, the medial, and the lateral collateral ligaments. The simulation of five different degrees of femoral component malrotation shows the relationship between the strain and forces produced by the ligaments under these conditions. The simulation shows that an internally rotated femoral component has a more profound effect on the forces in the ligaments than an external rotational malalignment. This might lead to pain on the medial side and condylar lift-off on the lateral side during a sit down task, as observed by clinicians.

This work sets a basis for further investigation of TKA discussing the patient’s outcome
and contributing to a better mechanical understanding of the knee joint. It may help clinicians and implant developers discuss the effects of their implant design or alignment of the implant components under dynamic loading.
Jährlich werden mehr als 1'000'000 totale Kniearthroplastiken, TKAs, weltweit implantiert, davon um die 8'000 in der Schweiz. Aus Folgen der immer älter werdenden Gesellschaft wird diese Zahl vermutlich in Zukunft stark zunehmen. Wird ein Knie durch eine totale Prothese ersetzt sind die Langlebigkeit der Implantatkomponenten sowie eine hohe Patientenzufriedenheit die anzustrebenden Ziele. Heute liegt die Rate von erfolgreichen totalen Knien bei ungefähr 85%, was immer noch eine sehr hohe absolute Anzahl an unzufriedenen Patienten mit sich bringt, die eine Revision der TKA benötigen. In den meisten Fällen sind Abrieb des Polyethylen Läufers, Implantatlockerung, Knieinstabilität und Infektionen die Ursache für eine Revision. Abrieb des Polyethylene Läufers, Lockerung und Knieinstabilität sind Faktoren, die auf eine veränderte Gelenksmechanik nach der Implantation einer totalen Kniearthroplastik zurückzuführen sind. Vor allem eine Rotationsfehlstellung der femoralen Komponenten hat schwerwiegende Auswirkungen in allen sechs Freiheitsgraden. Eine Rotationsfehlstellung kann eine ligamentäre Misbalance verursachen, was wiederum laterale Flexionsinstabilität und Schmerzen beim sich von einem Stuhl Erheben und beim Treppensteigen mit sich bringen kann.

Um diese Probleme zu vermeiden, ist ein grundlegendes Verständnis des Kniegelenks im Hinblick auf die Ausrichtung der Implantat-Komponenten wichtig. Ein dreidimensionales Computermodell, das die Gelenkskinematik während verschiedener Bewegungsmuster visualisieren kann, hilft, dieses Verständnis zu erlangen. Ein solches Modell ist jedoch auf exakte Kinematik Kinetik angewiesen, um die Bewegung und Belastung während alltäglichen Bewegungsmustern im Kniegelenk zu visualisieren und zu berechnen zu können.
Bis anhin basierte die Ganganalyse auf die Bewegungserfassung mit Hautmarkern und Kraftmessplatten, um die Kinematik und Kinetic während zum Beispiel ebenem Gehen zu erfassen. Die Hauptproblematik dieser Messmethode liegt darin, dass sich die Knochenbewegung mit Oberflächenmarkern nicht exakt messen lässt. Die Haut- und Muskelbewegung verunmöglichen es, die genaue Kinematik zu erfassen. Video-Fluoroskopie ermöglicht eine direkte Bestimmung der Implantatbewegung mittels einem Registrationsalgorithmus, der aus der Projektion im Fluoroskopiebild die Lage und Orientierung im dreidimensionalen Raum rekonstruiert. Diese Technik jedoch ist auf das Sichtfenster des Bildverstärkers beschränkt. Wir lösten dieses Problem durch ein Fahrzeug, dass den C-Bogen des Fluoroskopiergerätes mitfahren lässt. Dieses bewegbare System, Unit Mover genannt, erlaubt eine Bewegungserfassung während ebenem Gehen. Die dreidimensionale Rekonstruktion der Implantatbewegung relativ zum Fokus erfolgt mit einer Genauigkeit von 3.1 mm für die Translation und 1.6° für die Rotation.


Preface

This thesis consists of the four stand alone papers listed below.

Chapter 2: State of the art, literature review.
Biomechanical background and clinical observations of rotational malalignment in TKA -
Literature Review and Consequences
MS Zihlmann, A Stacoff, J Romero, I Kramers-de Quervain, E Stüssi
Clinical Biomechanics, 2005,20(7):661-8

Chapter 3: Improvement of knee joint kinematics acquisition by a moving video-fluoroscopic
system.
Three dimensional kinematics of total knee joint during a full gait cycle of level walking
using single plane video-fluoroscopy; MS Zihlmann, H Gerber, K Burckhardt, G Székely,
E Stüssi
Journal of Biomechanical Engineering, under review

Chapter 4: Inverse dynamic calculation by coupling video-fluoroscopy with force plate data.
Three dimensional kinematics and kinetics of total knee arthroplasty during level walking
using single plane video-fluoroscopy and force plates: A pilot study; MS Zihlmann, H
Gerber, A Stacoff, K Burckhardt, G Székely, E Stüssi
Gait and Posture, submitted
Chapter 5: Estimation on ligament load due to femoral component malrotation.

Femoral component malrotation on TKA - Functional implications on ligaments based on computer model and quantitative in vivo motion analysis using single plane video-fluoroscopy and force plate; MS Zihlmann, A Stacoff, I Kramers-de Quervain, E Stüssi

Journal of Orthopaedic Research, submitted
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Introduction

Nearly a million total knee joint replacements are implanted worldwide [24]. Because of higher life expectancies, increasing markets in China or Russia, which push implant development and sales, and aging demography, ever increasing numbers of total knee implantations in the next few years are expected [57]. More sport in young life, injuries to ligament structures and to menisci due to sports activities [62], [95], and excessive weight [64], [54] could lead to gonarthrosis, leading in many cases in total knee replacement. When replacing a knee joint by a TKA, the functionality of the joint and longevity of the implant components are the main goals resulting in reduced health care costs and good patient satisfaction [8].

Today the rate of good patient outcome lies between 80% and 90% [28], [74], [94], [36] depending on the literary sources. However, this still leaves a significant absolute number of patients needing revision surgery [28], [74], [94] making this a worthwhile topic of investigation. Furthermore, considering the annually increasing number of TKA surgeries, the absolute number of failures due to malalignment will also increase. The most common causes for revision surgery are polyethylene wear, loosening, knee instability and infection [85]. PE wear, loosening and instability are factors associated with altered joint biomechanics after total knee replacement [11], [60], [80]. The alignment of the implant components relative to the bones are essential for functionality and longevity of the re-
placed knee joint [47]. Malalignment could produce higher constraint forces and moments at the knee joint affecting strain on the joint’s ligament structures and implant wear. In particular, malrotation of the femoral component is suspected to contribute to implications like diffuse knee pain, knee joint laxity in flexion, and patellar maltracking [15], [59], [80].

Biomechanical studies of total knee joint replacements focus typically on kinematics of the implant components [90], [103] stress on the polyethylene inlay [60], [82] or patellar tracking [42], [43]. Malalignment of a measurable degree occurs in approximately 10% - 30% of patients, depending on the surgical technique and the anatomical landmarks used [13], [71]. However not all of these patients complain of unsatisfactory results. For most patients, a relative internal rotation of the femoral component between 3\degree and 6\degree is tolerable [12], [50], [79], while external rotation of the femoral component can be as high as 8\degree without causing clinical problems [70]. However, little is known about the mechanical implications of malrotation with regard to strain on ligament structures such as the collateral ligaments and the posterior cruciate ligament. A comprehensive mechanical understanding as regards to individual implant components alignment and ligament balancing is fundamental to avoid malalignment. Visualisation and load estimation of the altered kinematics and kinetics due to femoral component malrotation by a model could be fundamental to a better understanding of the complex mechanics [101]. Therefore, accurate in vivo measurements of a total knee arthroplasty’s kinematics are required in order to have reliable data input for such a model.

Non invasive instrumented gait analysis was used to acquire kinematic and kinetic information from skin mounted markers and force plates during gait and other movement pattern of daily activities [52], [93], [7]. The problem of this measurement technique is the large error in kinematic data acquisition caused by the movement of the skin and muscles relative to the underlying bone [77]. Point cluster methods were applied to reduce the errors due to skin movement artefacts [5]. However, Stagni et al. [88] could show that even if using the point cluster method with more than twenty markers on the thigh and shank, the standard deviation of the skin marker trajectory in the corresponding prosthesis-embedded anatomical frame was up to 31 mm. This large error could nullify the meaningfulness of the estimated parameters. Video-fluoroscopy is a well established method for obtaining kinematic information of artificial joints non invasively by a three-dimensional numeric
reconstruction of the single plane projection view in fluoroscopic images [10], [46], [91], [103]. Video-fluoroscopy enables the measurement of the kinematics of implant parts very accurately by avoiding skin movement artifacts. However, this method is limited to the field of view of the fluoroscopic screen and to kinematic data only. Enabling tracking of the joint movement by the fluoroscopic unit and coupling force plate data would allow the assessment of accurate kinematic information as well as providing kinetic information during normal level walking.

The effects on ligament structures due to femoral component malrotation were studied on cadaver knees [53], [79]. However, it is difficult to simulate motion patterns of daily activities on cadaver knees. Effects of femoral component malrotation on ligament structures, such as the posterior cruciate ligament, PCL, the medial collateral ligament, MCL, and the lateral collateral ligament, LCL, based on physiological motion patterns would help to estimate the alternation of the strain in the ligaments affecting inner forces at the knee joint.

The goals of this work are:

1. To get kinematic and kinetic information of a totally replaced knee joint non invasively with a higher accuracy than with skin mounted markers during level walking and a sit down task.

2. To establish a computer based model visualising the kinematics of the totally replaced knee joint including the implant components and the bones of the subject.

3. To estimate resultant forces and moments of the TKA in the sagittal plane during these activities by inverse dynamic calculation. The accuracy of these results will be discussed and compared with other author’s results.

4. To study the effects of malrotation by simulating different degrees of a rotational malalignment of the femoral component. The altered strain of the ligaments will be considered.

An overview of this project is given in Figure 1.1.
1. INTRODUCTION

level walking
sit down task

single plane video-fluoroscopy
VICON
KISTLER force plate

6 DOFs of implant components
coordinates of unit mover
ground reaction force

total knee joint arthroplasty model

implant component’s kinematics
resultant forces and moments

simulation of different degrees of
femoral component malrotation

Figure 1.1: Project overview
Biomechanical background and clinical observations of rotational malalignment in TKA - Literature Review and Consequences

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Clinical Biomechanics, 2005,20(7):661-8
2.1 Abstract

Malalignment, in particular femoral component malrotation, is a commonly accepted failure mode in total knee arthroplasty (TKA). The general objective of this paper is twofold: Firstly, it accentuates clinical observations of the effects of rotational malalignment in TKA. Secondly, it discusses the relevant parameters of existing knee joint models with regards to rotational malalignment and its biomechanical background, thereby setting a basis for future studies. To summarise, when modelling malalignment in TKA, the following aspects should be considered: Friction between the implant components, ligament and capsular structures, deformable body to model the PE inlay, and an in vivo validation of the model. Because of the large variance in anthropometrical data between individuals, future knee joint models should also incorporate individual data.

2.2 Introduction

Currently about 1’000’000 total knee arthroplasties are implanted world wide each year [24]. Because of higher life expectancies and aging demography it is expected that the number of total knee implantations will increase in the next few years [57]. Although patient satisfaction rates are around 90%, this still leaves a significant absolute number of patients needing revision surgery [28], [74], [94] making the causes leading to revision surgery worthwhile topic of investigation. Furthermore, considering the annually increasing number of TKA surgeries, the absolute number of failures due to malalignment will also increase. The most common causes for revision surgery are polyethylene wear, loosening, knee instability and infection [85]. PE wear, loosening and instability are factors associated with altered joint biomechanics after total knee replacement [11], [60], [80]. Malalignment of the implant components may be the underlying reason for these failure modes. In particular, malrotation of the femoral component is speculated to contribute to implications like diffuse knee pain, knee joint laxity in flexion, and patellar maltracking [15], [59], [80].

Biomechanical studies of total knee joint replacements focus typically on kinematics of the implant components [90], [103] stress on the polyethylene inlay [60], [82] or patellar tracking [42], [43]. However, little is known about the mechanical implications of malrota-
tion with regard to strain on ligament structures such as the collateral ligaments and the posterior cruciate ligament, and the joint capsule considering the relevant parameters of the respective implant design. Thus, the goal of the present review paper is to summarise the implications of femoral component malrotation based on (i) current clinical observations as well as (ii) current biomechanical TKA models and their possible predictions with respect to stress on implant components and strain on ligament structures.

The general objective of this paper is: to accentuate clinical observations of the effects of rotational malalignment in total knee arthroplasty, and to discuss the relevant parameters of existing knee joint models with regards to rotational malalignment and its biomechanical background, thereby setting a basis for future studies.

2.3 Clinical observations

In the following chapter we discuss the aspects of normal knee joint functions and of malalignment from a clinical point of view.

2.3.1 Malalignment in total knee replacement

Malalignment is the most frequently discussed complication in total knee replacement [1], [15], [30], [59], [79], [80]. Malalignment of a measurable degree occurs in approximately 10% - 30% of patients, depending on the surgical technique and the anatomical landmarks used [13], [71], but not all of these patients complain of unsatisfactory results. For most patients, a relative internal rotation of the femoral component between 3° and 6° is tolerable [12], [50], [79], while external rotation of the femoral component can be as high as 8° without causing clinical problems [70].

2.3.2 Femoral component malrotation

The rotational malalignment of the femoral component is a common cause for revision surgery [1]. Correct rotational positioning of the femoral component is crucial for optimal tracking of the femoropatellar joint and for a balanced flexion gap [53]. Consequences of rotational malalignment of the femoral component include flexion gap asymmetry, patellar maltracking and varus or valgus malalignment of the lower extremity in the flexed position,
2. LITERATURE REVIEW

<table>
<thead>
<tr>
<th>Author</th>
<th>Femoral component rotation [°]</th>
<th>Clinical observations</th>
</tr>
</thead>
<tbody>
<tr>
<td>Akagi et al. [1]</td>
<td>Several</td>
<td>Patellar maltracking</td>
</tr>
<tr>
<td>Anouchi et al. [9]</td>
<td>Several</td>
<td>Flexion gap asymmetry, patellar maltracking</td>
</tr>
<tr>
<td>Barrack et al. [12]</td>
<td>3-6 internal</td>
<td>No clinical problems</td>
</tr>
<tr>
<td>Barrack et al. [12]</td>
<td>More than 6 internal</td>
<td>Knee pain</td>
</tr>
<tr>
<td>Dennis et al. [26]</td>
<td>Several</td>
<td>Flexion gap asymmetry</td>
</tr>
<tr>
<td>Hofmann et al. [47]</td>
<td>More than 6 internal</td>
<td>Knee pain</td>
</tr>
<tr>
<td>Insall et al. [50]</td>
<td>3-6 internal</td>
<td>No clinical problems</td>
</tr>
<tr>
<td>Nagamine et al. [70]</td>
<td>Up to 8 external</td>
<td>No clinical problems</td>
</tr>
<tr>
<td>Olcott and Scott [71]</td>
<td>Several</td>
<td>Flexion gap asymmetry</td>
</tr>
<tr>
<td>Romero et al. [79]</td>
<td>3-6 internal</td>
<td>No clinical problems</td>
</tr>
<tr>
<td>Romero et al. [79]</td>
<td>More than 6 internal</td>
<td>Flexion gap asymmetry</td>
</tr>
<tr>
<td>Stiehl et al. [89]</td>
<td>Several</td>
<td>Flexion gap asymmetry</td>
</tr>
</tbody>
</table>

Table 2.1: Summary of clinical observations of femoral component malrotation

taking the tibia out of the sagittal plane [1], [9], [15]. An excessive external rotation of the femoral component can also cause an abnormally tight popliteus tendon complex inducing loss of rotational laxity of the knee in the late phase of knee flexion [70]. Romero et al. [79] showed that a significant lateral joint opening occurs at 6° internal rotation of the femoral component under axial loading conditions. Other studies documented this joint opening by several degrees of malrotation as well [9], [26], [71], [89]. Therefore, several authors have considered, that a symmetrical space between femur and tibia in flexion as well as in extension is desirable [15], [72], [80].

There are two general concepts in surgical techniques used in implantation of a TKA. Achieving a balanced flexion gap is one, and maintaining the knee joint’s flexion axis by cutting the femur parallel to the epicondylar axis is the other. Each method has advantages and disadvantages. Balancing the flexion gap has the disadvantage that the femoral component is not aligned parallel to the epicondylar axis. The significance of this lies in the well accepted belief that the epicondylar axis approximates the flexion axis [22]. On the other hand, maintaining the flexion axis does not create a symmetric flexion space in all cases [72]. It is not known yet, which concept produces better clinical results [80]. The ideal operation technique to estimate femoral component rotation might be a combination of both methods [47].
2.3.3 Methods to estimate femoral component malrotation

There are different methods for assessing relative rotational alignment of the femoral component [15], [30], [98]. Eckhoff et al. [30] have estimated the rotation of the femoral component by measuring the width of two pegs of the implant component on a lateral radiograph. Berger [15] determined the rotational alignment by estimating the angle between the tangent of the implant’s dorsal condyles and of the clinical epicondylar axis on a CT scan.

Eckhoff’s method is applicable for pegged implant components only.

2.3.4 Clinical observations: Discussion

Correct alignment of the components is critical for satisfactory patient outcome. Rotational alignment of the femoral component is particularly critical because it affects patellar tracking, polyethylene wear and ligament balancing [26], [29], [73], which are the important failure mechanisms in the early postoperative period. It is interesting to note that malrotation of a measurable degree does occur in approximately 10% - 30% of the patients [13], [71]. Why some patients become so strongly symptomatic that they require revision surgery and others do not, is an open question.

Implanting an arthroprostheses into the knee joint always affects anthropometrical features such as the articulating surfaces and the ligament structures. As a consequence knee joint biomechanics is significantly altered. For example, a rotation of the femoral component or a resection of ligament structures can be used to reach a symmetrical flexion gap, but both methods change the knee joint kinematics and, as a consequence, knee joint kinetics.

Clinical methods to determine the rotational malalignment have the inherent disadvantage that no dynamical effects can be estimated. In order to gain more insight into the dynamic effects of malrotation, knee joint models have been developed which allow simulation of knee joint movement, forces, and moments. An overview of these models is given in the following chapters.
2.4 Biomechanical background

The following subchapters show a brief overview of the necessary biomechanical background.

2.4.1 Summary of natural knee joint kinematics and forces

The knee joint is one of the most complex joints in the human body because it moves with three rotational and three translational degrees of freedom. The knee joint contains two articulating surfaces: one between tibia and femur (referred to as femorotibial joint), and the other between the patella and the femur (referred to as femoropatellar joint). Knee joint flexion progresses as a combination of rolling, gliding, and rotation about the longitudinal axis of the femoral condyles over the tibial plateau [55]. The magnitude of each of these motions changes through out the range of flexion. The rolling motion, which predominates early in flexion ($0^\circ - 20^\circ$), produces posterior translation of the tibial-femoral contact point [51]. Sliding becomes dominant at flexion angles beyond $30^\circ$. The cruciate ligaments kinematically interact with the articulating surfaces, maintaining the normal passive movement of the knee. The range of maximum forces in the knee joint during daily activities reaches from 4 times BW (body weight) [68] to 7BW [84]. Morrison [68]
has shown that approximately 70% of the load across the knee joint is sustained by the medial compartment of the knee.

Ligament structures, in particular the cruciate ligaments (anterior cruciate ligament, ACL; posterior cruciate ligament, PCL) and the collateral ligaments (medial collateral ligament, MCL; lateral collateral ligament, LCL), play an important role for the stability of the knee joint. The collateral ligaments are responsible for the transverse stability of the knee during extension. They become taut during extension and slacken during flexion. The cruciate ligaments constrain the knee’s extension and flexion movement in the anterior posterior direction [55]. This profound kinematics (the rolling, gliding and rotating movement of the bones), as well as the kinetics (the stabilising function of the ligament structures and the forces and moments) of the knee joint, should all be considered when replacing parts of the knee with implants [8].

2.4.2 Knee joint models

The related literature contains several knee joint models which deal either with rigid body systems or with deformable bodies. In rigid body systems forces and moments are governed by rigid body motion based on the laws of classical mechanics. In systems with deformable bodies stresses are related to strains through constitutive laws in addition to the former relationships.

The aspects of knee joint modelling, which have to be considered, are ligament structures, contact models, quasi-static or dynamical behaviour, rigid body simulation or finite element analysis, three-dimensional or two dimensional motion. Which aspects need to be considered and which method being used depends on the problem at hand.

A further aspect is the validation of the model to support the simulation results. An example of a modelling process is shown in Figure 2.2.

The various steps of a general modelling process are explained as follows:

(1) Problem: Aspects of knee joint biomechanics. Knee joint kinematics [38], kinematics of the components after TKA [44], patellar tracking [43], stress and wear on the polyethylene inlay [60], [97], stress on patellar surfaces etc.

(2) Model parameter: Material parameters for ligament models, bone and implant geometry, contact model (friction, no friction), muscle models etc.
2. LITERATURE REVIEW

Knee joint model: Solid or rigid modelling (rigid systems, finite element models), quasi static or dynamic calculation, two-dimensional or three-dimensional motion, femorotibial and femoropatellar joint etc.

Model input: Forces and moments (forward dynamics), kinematics and ground reaction force (inverse dynamics)

Model validation: In vivo validation, comparison with a knee joint simulating machine

These aspects of the knee joint model have to be considered by implementing a knee joint simulation. The following chapters illustrate these factors with regard to simulating malalignment in TKA joint models.

2.4.3 TKA models versus natural knee joint models

There are two differences between TKA models and natural knee joint models: First, the geometry of the implant components and its material properties are well known from implant producer, but the bony geometry at the knee joint can only be estimated (i.g. by CT scans) and the material properties of bony structures remain always an assumption. Second, then modelling TKAs, friction between the polyethylene and the femoral component plays an important role [82]. The coefficient of friction ranges between 0.03 and 0.10
whereas in the natural joint, friction will have little effect on kinematics because the
coefficient of friction lies only between 0.002 and 0.03 [69], [82].

2.4.4 Model validation

To support the results of a model simulation, the knee joint model has to be validated. The
validation process comprises the changing of the model parameters iteratively so that
the results match the experimental data as closely as possible. It is often impossible to
match the model results data exactly because of the simplifications made, thus the output
parameters which can be used in the validation process have to be estimated carefully and
should be based on convincing arguments.

The author consider the following: (i) using gait analysis for an in vivo validation [39] [56]
or (ii) using a knee joint simulator [63], [40], [82].

(i) An in vivo analysis of the kinematics and kinetics is the most realistic representation.
Piazza and Delp [75] compared the simulated muscle forces with measured activation
pattern.

(ii) Sathasivam and Walker [82] used a knee joint simulating machine, and compared the
predicted kinematics of their model with the output kinematics of the simulating machine
by applying the same load in both approaches. They concluded that when using data from
knee joint simulating machine it has to be considered that the parameters of the machine
are assumptions. The knee joint model is not more accurate than these assumptions.
However, it is easier to validate a knee joint model with a knee simulator than with
an in vivo study, because the parameters of the simulator are well known and can be
implemented into the knee model.

In summary, to exactly reproduce reality in a model is certainly not possible. Clearly an
effort should be made to improve knee joint models so that they reflect reality as accurately
as possible.

2.4.5 Knee joint models: Discussion

In the following, some of the important aspects of knee models and the process of creating
such a model are summarised and discussed:

- Currently available mathematical models in joint biomechanics typically use either
rigid body systems, in which forces and moments are related to rigid body motions through the laws of classical mechanics, or with deformable bodies, in which stresses are related to strains through constitutive laws [42]. The disadvantage of rigid body models is the interface behaviour of joints which provides only a coarse approximation of the joint’s interface behaviour. Analysing the effects of malrotation, interface reaction of the implant components is of interest. For example knowing the stress on the polyethylene inlay allows one to estimate wear. Deformable bodies can be modelled by finite element analysis, but this has the disadvantage that stress recovery does not include time history [99]. Using finite element analysis models contact elements are characterised by a node-on-node geometry and are therefore restricted to small slips between the contacting bodies [42].

• In some knee models, the tibial plateau has been represented as a deformable surface [60], [27], [31]. It has to be considered that the elastic modulus of polyethylene is 400 times lower than the modulus of the cobalt chrome material of the femoral component (elastic modulus for cobalt chrome: \(2.09 \times 10^5\) MPa; for UHMWPE (ultrahigh-molecular-weight PE): \(5.0 \times 10^2 - 7.0 \times 10^2\) MPa). During passive knee joint motion, the knee is unloaded and so a simplified assumption of rigid-surface in an natural knee is appropriate [18]. It has to be considered that cartilage has an elastic modulus of 5 MPa [18].

• Some of the knee joint models contain the femorotibial as well as the femoropatellar joint [38], [44], [75] others focus on one joint only [21], [40], [43] depending on the given question. A knee joint model which represent a total knee joint in terms of the alignment of the implant components should consider the femorotibial joint as well as the femoropatellar joint. The biomechanics of both joints are affected by malalignment at the tibia as well as at the femur.

• The largest angular motion during knee joint activity takes part in the sagittal plane, thus, many knee joint models simplify knee joint motion to two dimensions [40], [65], [86]. However, in analysing the implications of rotational malalignment it is important to assess femorotibial rotation, thus two-dimensional models do not give enough insight into the problem.

• The friction coefficient between cobalt chrome and polyethylene implies that motion
of the knee after TKA will not be as smooth and continuous, as in a natural knee, where the friction is so small as to be insignificant.

- In some models, ligaments have been modelled as isometric linkage [33], as nonlinear elastic elements [42], [75], [16], or as linear elastic elements [23]. When small elongations are expected, there is a general agreement on the existence of a toe shape in the stress-strain curve of ligaments. For longer elongations of the ligaments, a linear approximation is appropriate [65]. Few models consider the influence of soft tissue restraints (ligament and capsule) as non-linear springs whose stiffness does not vary with the flexion angle [82], [63]. In an intact knee, the stiffness varies significantly with increasing flexion [37]. Godest et al. [40] take this varying stiffness of soft tissue restraints into account in their knee joint model. Furthermore, the ligament-bone interaction of the collateral ligaments near their insertion points is significant [17]. Not many of the models found in the literature take this factor into account.

- Various knee joint models include muscle activation [75], [87]. Because the number of muscles crossing a joint is greater than the number of degrees of freedom specifying the joint movement, the system is over estimated and the force of each muscle cannot be determined individually [87]. All attempts to solve this problem are based on the application of an optimisation criterion. Muscle activation patterns can be estimated by EMG [61]. The difficulty of using muscle activation in knee joint models is that EMG data provides information about the on-off response of the muscle but it is difficult to get information about the muscle forces. Furthermore, the question how representative the nervous stimulation of the muscle force itself is, still remains [45].

- Implant geometries are known from CAD data of the implant producer. The material parameters are well known and can be implemented into the model. On the other hand bone geometry can be estimated from CT scans, and the material parameters of the cortical and cancellous bones remain assumptions.

- There are basically two different approaches to simulate knee joint mechanics. One is the inverse-dynamics approach, in which input data are external forces (e.g. ground
reaction) and the kinematics of the femur and the tibia, and the calculated output is the resultant knee joint forces and moments and inner forces and moments [14], [67]. The second approach is the forward-dynamics method which uses muscle excitations as the input, and the system response gives the knee joint kinematics [87], [100].

- The dynamic load on the knee joint during walking or other daily activities is frequently several times the load in a standing static position [4]. Thus, the dynamic load on the knee joint is an important consideration when attempting to understand the mechanics of malalignment. When modelling knee joint mechanics of walking and other active moving patterns, dynamic aspects such as inertia of bones and implant components play an important role in estimating forces and moments in the joint.

Furthermore, polyethylene wear is a dynamic process affected by many factors including contact stress, sliding distance, frictional behaviour and the cyclic load, and can not be analysed by static or quasi static studies [60]. If only the passive movements are of interest, dynamic aspects and muscle forces can be neglected.

- With respect to validation of the model, different concepts have been applied. One concept is to compare the output with in vivo data [42]. Here, the difficulty is to estimate kinematics and kinetics data of the knee joint in vivo. When using e.g. instrumented gait analysis the source of error due to skin movement artifacts may be large [77]. The other concept is to compare the output data with knee joint simulator results [40]. This concept allows to use the same parameters in the simulator as well as in the model, because these parameters can be estimated well. However, the question of how accurately a knee joint simulator represents a real knee joint in motion, still remains.

2.5 Conclusion

To avoid malalignment, the interaction between ligament structures and implant components should be well understood. A better understanding of the characteristics of joint loading and stress distribution in TKA will define the relevant parameters of knee joint
alignment, particularly regarding the large variety of anthropometric factors in the knee joint. Currently, the exact kinematics and kinetics of the TKA with regards to the effects of malalignment are not well understood.

To summarise, when modelling malalignment in TKA, the following aspects should be considered: Friction between the implant components, ligament and capsule structures, bones and femoral and tibial implant components as rigid models, deformable body to model the PE inlay. Furthermore, the model should consider the dynamical behaviour of the joint in three dimensions and an in vivo validation should be carried out to support the output data as accurate as possible to an actual TKA.

The basic problem in simulating knee joint biomechanics still remains with respect to the large variety of anthropometric data. Hence, a knee joint model which is valid for all various knee joints does not exist. Therefore, a future knee joint model should be adaptable with individual data. This would have the advantage that differences between subjects are omitted which would allow the orthopaedic surgeon to prepare surgery with a model which is based on individual data.
Three dimensional kinematics of total knee joint
arthroplasty during a full gait cycle of level walking using
single plane video-fluoroscopy

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3.1 Abstract

Accurate in vivo measurements of the kinematics of total knee arthroplasty, TKA, are crucial for the improvement of implant designs and the best clinical outcomes. The following study obtains kinematic information of TKA during several full gait cycles via a moving fluoroscopic system, that tracks a moving knee joint. An intensity based registration algorithm was applied to the fluoroscopic image to estimate the three dimensional position and orientation of the implant components relative to the fluoroscopic focus. This measuring method is able to obtain kinematic information of full gait cycles more accurately than the use of skin mounted markers.

Keywords: TKA kinematics, full gait cycle, single plane video-fluoroscopy

3.2 Introduction

Accurate in vivo measurements of the kinematics of total knee arthroplasties (TKA) during normal level walking are fundamental for the investigation of the mechanics of artificial knee joints. Kinematic studies of implant components during daily activities are important to understand the functionality of a replaced knee joint. Typically, in order to get kinematic information, skin mounted markers and opto-electronic devices are used. Instrumented gait analysis is performed in order to obtain kinematic information from skin mounted markers and force plates during level walking and stair climbing at different inclinations [3], [81]. Measurements with skin mounted markers suffer from a large error component due to the movement of the skin and muscle relative to the underlying bone [83]. As a result of skin motion, errors of up to 10 mm of translation and 8 degrees of rotation have been observed [58]. To reduce this error, in vivo studies have recently been made using improved point cluster methods [6]. However, Stagni et al. [88] compared the point cluster method with video-fluoroscopy and showed that the standard deviation of skin marker trajectory in the corresponding prosthesis-embedded anatomical frame was up to 31 mm and up to 21 mm for the two bony parts, respectively. In this study, the ab/adduction and internal/external rotation angles were the most affected by soft tissue artefact propagation. These large errors can nullify the usefulness of these variables in clinical interpretation of gait analysis.
Video-fluoroscopy is a well established method to obtain kinematic information of artificial joints directly via a three-dimensional numeric reconstruction of the single plane projection view of fluoroscopic images [10]. Video-fluoroscopy also provides improved accuracy to detect relative rotations between the implant components even of a few degrees [32] enables the measurement of the kinematics of implant components without skin movement artifacts. Several methods have been proposed to reconstruct the 3D movement of the implant components from the 2D projection of the fluoroscopic image [10], [46], [103]. The disadvantage of this method is the limited field of view of the fluoroscopic screen. To get the kinematic information over a full gait cycle with video-fluoroscopy, some studies have used a treadmill to analyse the kinematic behaviour of the TKA during gait. However, it is known that the gait pattern walking on a treadmill differs from a normal gait [2], and it is known that the knee movement of an entire gait cycle cannot be captured within the field of view. A video-fluoroscopy unit weighs more than 100 kg and all previous studies have avoided moving of this unit. Enabling tracking of the joint movement by the fluoroscopic unit would allow of accurate assessment of kinematic information during an entire gait cycle of level walking. To date, none of the studies found in literature analyse the artificial knee joint kinematic of overground walking during several full gait cycles. The goal of the present study was to achieve more accurate kinematics of the implant components of TKA during level walking of several full gait cycles. This article gives an overview of an innovative measuring approach of the achievement of TKA kinematics.

3.3 Methods

3.3.1 Tracking of several gait cycles by a single plane video-fluoroscope

3.3.1.1 Unit mover

The knee joint moves during level walking in the gait direction, the sagittal plane, with a horizontal velocity of about 1.5 m/s and a maximal acceleration of about ±4 m/s² after toe off. Due to these boundary conditions, it is not possible to track a knee joint movement with a 12 inch fluoroscope (310 mm) during normal walking with a constant velocity, because the knee joint would move out of the field of view. This problem was solved by using a controlled moving system which tracks the knee joint via a sensor.
The C-arm of the fluoroscopic unit (Figure 3.1), together with the x-ray source and the image intensifier, is mounted on a motor driven trolley, called the unit mover. The system accelerates and decelerates thereby keeping the knee joint within the field of view of the fluoroscope [102]. The maximal velocity of the mover is 3.5 m/s, the maximal acceleration is 4.5 m/s².

The stiffness of the C-arm while the system is moving was evaluated. Metal discs with a diameter of about 5 mm were bonded on the bottom of a rigid plastic box, which was fixed on the image intensifier, so that the discs were held in a distance of about 300 mm parallel to the image intensifier. Fluoroscopic images were taken while the unit mover was accelerating and decelerating. An image subtraction of the subsequent images showed no differences. Thus, the C-arm can be assumed to be stiff enough for our application.

This system allows the subject to walk freely between the x-ray source and the image intensifier without any constraints such as the controlled speed of a treadmill. It has the advantage that the knee joint remains within the field of view over several gait cycles without interruption. The knee joint with the total replacement is right next to the image intensifier. The c-arm can be rotated into two positions to film the left as well as the right
The knee joint is tracked by a sensor, which controls the movement of the fluoroscopic unit mover. For safety reasons the system operator, as well as the subject, are able to stop the unit mover immediately by using the emergency buttons. Furthermore, the unit mover stops if an object is in its moving track. The safety system is designed to be double fail safe.

3.3.1.2 Fluoroscopic image capture

Video-fluoroscopy (BV Pulsera, Philips Medical Systems) was performed by acquiring a series of pictures (25 images/second, 8 ms shutter time) with an image intensifier of a diameter of 310 mm. The video sequence with the fluoroscopic images were stored on a hard disk.

The x-ray exposure time of 2 minutes to the subject gives a dose of approximately 0.048 mSv. A standard x-ray image of the knee joint is a dose of 0.06 mSv).

3.3.2 Three dimensional reconstruction of the kinematics of the implant components

3.3.2.1 Distortion correction

In order to correct the distortion of the fluoroscopic images a calibration grid was filmed before each measurement cycle. Due to the rigid connection of the x-ray source and the image intensifier to the c-arm, the same distortion correction can be applied throughout the entire image sequence. The distortion of the images is eliminated by determining the bilinear transformation between the known and measured coordinates of control points on the calibration grid (Figure 3.2). The distortion correction was performed in all images before three dimensional pose reconstruction.

3.3.2.2 Three dimensional reconstruction

A full three dimensional analysis of each fluoroscopic series is achieved by fitting a synthetic x-ray projection of the tibia or the femoral component to the original in each image. In contrast to the common approaches proposed by Banks and Hodge [10], and by Hoff et
al. [46], not only the implants’ contours but also their complete image areas are matched. The underlying algorithm was originally designed to precisely locate hip implants and is described in detail in [19]. It is realised by using CAD models (Mathys Ltd Bettlach, Switzerland) and by simulating the attenuation of the x-rays through the implants. After a rough initial estimation of its pose (position and orientation), the implants’s exact three dimensional parameters in the fluoroscopic system are found by an iterative minimisation (combined Gauss-Newton approach, NAG library www.nag.co.uk) of the difference between the synthetic and the original image. Outputs of the algorithm are the parameters $t_{x,y,z}$ ($x$, $y$ is parallel, in-plane, $z$ perpendicular, out-of-plane, to the image plane) and the angles $xrot$, $yrot$, and $zrot$ relative to the image focus. The latter describes the orientation of the implant via three subsequent rotations around body fixed axes.

### 3.3.2.3 Repeatability of the three dimensional reconstruction of gait cycle images

The repeatability of the reconstruction algorithm was tested in three images of the in vivo collected series during level walking. The images were selected to represent three different types of a full gait cycle: Motion blur (high relative velocity), bone overlay, and no motion (low relative velocity). Motion blur occurs, when the TKA moves relative to the image.

![Standard calibration grid given by the manufacturer.](image-url)
focus and the x-ray source respectively. This occurs during the swing phase and before the stance phase. Bone overlay takes place during the stance phase, during which the contralateral knee joint swings behind the TKA.

The algorithm was run 15 times in each of the three images. Each time the initial estimate was redefined by the user. The standard deviations of $t_{x,y,z}$ and $xrot$, $yrot$, and $zrot$ of the two implant components were calculated using the resulting parameter sets.

3.3.2.4 In vitro validations

The validity of this study was confirmed by a static and dynamic in vitro validations. The implant components were fixed on each other allowing no relative movements between the implant components. The exact position and orientation of the tibial component relative to the femoral component were measured by a three dimensional optical digitiser (Breuckmann OPTO Top HE-200, Germany) that measured surface points within an accuracy of ±0.01 mm. Three nearly orthogonal planes were overlayed on well defined surfaces of the CAD geometries of each implant component (Raindrop Geomagic Qualify 5.0). The same planes were then overlayed on the point clouds of the surface scan. The software fitted the corresponding planes of the CAD geometry into the planes of the point cloud, within a translational deviation of 0.034 mm and a rotational deviation of 0.54°.

The distance and the orientation of the coordinate system of the femur relative to the coordinate system of the tibia was calculated. The experimental results were evaluated by calculating the relative distance vector, $t_{rel}$ (Equation 3.1). The rotation matrices $R_{fem}^0$ and $R_{tib}^0$ represent the orientation of the two components and are calculated inserting $xrot$, $yrot$, and $zrot$ as parameters in the standard definition of a rotation matrix. The translation vectors $t_{fem}^0$ and $t_{tib}^0$ are the translation vectors in the inertial system.

$$t_{rel} = R_{fem}^0 \ast (t_{fem}^0 - t_{tib}^0) \quad (3.1)$$

$$R_{rel} = R_{fem}^0 \ast R_{tib}^0 \quad (3.2)$$

The relative rotation matrix $R_{rel}$ is calculated using Equation (3.2). The screw vector, $u$ and its corresponding angle, $\beta$, are extracted by using the Equations of Rodrigues (see
Appendix D.3. The RMSE, root mean square error, of the calculated distance, the components of the screw vector, and the rotation angle was calculated as described in Equation 3.3, where $x_i$ is the measured value, $n$ the number of measured values, and $\pi$ is the true value.

$$RMSE = \sqrt{\frac{1}{n} \sum (x_i - \pi)^2}$$ (3.3)

**Static in vitro validation:** Eight fluoroscopic images were taken from different points of view relative to the x-ray source, where the distance from the object and the image intensifier was about 100 mm. Thereof four images (series II) were taken, where a bag filled with semolina (thickness of about 50 mm) and a dry tibia were placed between the object and x-ray source simulating the influence of the collateral leg in the fluoroscopic images.

**Dynamic in vitro validation:** One image series (8 ms exposure time, 5 images/s) was taken while moving the implant components fixed on a thread of about 100 mm parallel to the image intensifier with a velocity of about 0.3 m/s. The velocity was estimated by comparing the position of the implant components in the subsequent images.

### 3.4 Results

#### 3.4.1 Tracking of a full gait cycle

##### 3.4.1.1 Relative velocity of the knee joint

Figure 3.3 shows the velocity of the knee joint relative to the image intensifier, optically tracked by markers on the knee and on the unit mover (VICON™, Oxford Metrics Inc., UK). The relative displacement from the centre of the image intensifier in walking direction is $\pm 80$ mm at its maximum. Taking the diameter of the image intensifier of 300 mm, the unit mover is able to track the knee joint while keeping the implant component within the field of view.
Figure 3.3: Horizontal displacement between the knee joint and the centre of the image intensifier during two full gait cycles.

3.4.1.2 Image series of a full gait cycle

The whole gait cycle is within the field of view of the fluoroscope as shown in Figure 3.4. The sharpest images were taken during the very late stance phase, because during this phase the knee joint velocity is low and there is no overlaying bone from the contralateral side. Note that the rotation of the femoral part in the late stance phase is nicely illustrated. (See images 10 - 13 of Figure 3.4).

3.4.1.3 Rotations of the implant components

The rotations of the implant components are shown in Figure 3.5. Even the unfiltered data represents good physiological movement of the three rotations of the components.

3.4.1.4 Repeatability of the three dimensional reconstruction

The standard deviations of the three images are listed in Table 3.1. The results of image 5 can be compared with image 1. Thus, motion blur does affect the standard deviation, especially in the error of the out-of-plane translation. The standard deviations of Image 15 of the in-plane translations, femoral and tibial component, \((t_x, t_y)\) are less than 0.2 mm, the out-of-plane translation \(t_z\) 1.47 mm for the femoral part, and 2.29 mm for the tibial part respectively. The standard deviations of the rotation angles are less than 0.8°. The
### Table 3.1: The standard deviations \((\sigma_{t_x, y, z}, \sigma_{xrot}, \sigma_{yrot}, \sigma_{zrot})\) of the estimated pose parameters. \(\sigma_t [\text{mm}], \sigma_{rot} [^\circ]\)

<table>
<thead>
<tr>
<th>image type</th>
<th>(\sigma_{tx})</th>
<th>(\sigma_{ty})</th>
<th>(\sigma_{tz})</th>
<th>(\sigma_{xrot})</th>
<th>(\sigma_{yrot})</th>
<th>(\sigma_{zrot})</th>
</tr>
</thead>
<tbody>
<tr>
<td>femoral part</td>
<td>0.57</td>
<td>0.72</td>
<td>4.71</td>
<td>0.38</td>
<td>1.00</td>
<td>1.21</td>
</tr>
<tr>
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<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>bone overlay (img 5)</td>
<td>0.70</td>
<td>0.56</td>
<td>8.98</td>
<td>0.63</td>
<td>0.45</td>
<td>1.75</td>
</tr>
<tr>
<td>no motion (img 15)</td>
<td>0.07</td>
<td>0.16</td>
<td>1.47</td>
<td>0.09</td>
<td>0.23</td>
<td>0.34</td>
</tr>
<tr>
<td>tibial part</td>
<td>0.66</td>
<td>0.37</td>
<td>5.64</td>
<td>0.10</td>
<td>1.16</td>
<td>0.37</td>
</tr>
<tr>
<td>motion (img 1)</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>bone overlay (img 5)</td>
<td>0.41</td>
<td>0.35</td>
<td>5.94</td>
<td>0.81</td>
<td>0.30</td>
<td>0.33</td>
</tr>
<tr>
<td>no motion (img 15)</td>
<td>0.11</td>
<td>0.12</td>
<td>2.29</td>
<td>0.77</td>
<td>0.24</td>
<td>0.13</td>
</tr>
</tbody>
</table>

The ratio of the standard deviation of out-of-plane to in-plane translation is about 16. This repeatability study can not distinguish between the effect of motion blur and the effect of noise due to bone overlay.

### 3.4.2 Results of the in vitro validations

**Static in vitro validation:** The results of the static validation test are listed in Table 3.2. The static accuracy test shows a translational RMSE for the series I of 2.03 mm and a RMSE for the series II of 2.36 mm, the rotational errors are 1.77° and 0.75°. No significant difference could be found \((\chi_{\text{emp-trans}}^2 = 1.02 < \chi_{3.5\%}^2 = 7.81; \chi_{\text{emp-angle}}^2 = 0.10 < \chi_{3.5\%}^2 = 7.81)\) between series I and series II. Therefore the mean values of the RMSE are taken, 2.20 mm and 1.26°, respectively.

**Dynamic in vitro validation:** The RMSE of the dynamic validation test (Table 3.3) is 3.25 mm for the translational error and 1.57° for the rotational error. The standard deviations of the three components of the translation vectors are 1.27 mm, 4.81 mm and 4.41 mm relative to the tibial component. Therefore no statements about the in-plane or out-of-plane error relative to the image plane can be made. The standard deviations of the normalised screw vector components are 0.020, 0.023, and 0.064. The standard deviations of the distance between the implant components and the rotation angle are 1.08° and 3.25 mm respectively, the RMSE are 2.17 mm and 0.95° larger than the corresponding standard deviations.

A rough estimate of the error in-plane to out-of-plane is given by the ratio of the standard deviations of the in-plane to out-of-plane errors of the in vivo repeatability study. Hence, the in-plane error is about 0.2 mm.
Table 3.2: Results of the static validation tests series I (plain) and series II (with bone overlay). The translations, $t_{x,y,z}$, are given relative to the coordinate system of the tibial component, $t_{rel}$ is the distance between the coordinate systems of the two implant components. Mean of RMSE is the mean of RMSE of series I and series II respectively.
### 3. 3D KINEMATICS DURING LEVEL WALKING

<table>
<thead>
<tr>
<th>image</th>
<th>$t_x$ [mm]</th>
<th>$t_y$ [mm]</th>
<th>$t_z$ [mm]</th>
<th>$t_{rel}$ [mm]</th>
<th>$u_x$</th>
<th>$u_y$</th>
<th>$u_z$</th>
<th>$\beta$ [°]</th>
</tr>
</thead>
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<td>4.69</td>
<td>39.30</td>
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<td>-0.332</td>
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<td>-4.59</td>
<td>4.34</td>
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<td>-0.329</td>
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<tr>
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<td>-0.325</td>
<td>0.317</td>
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<tr>
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<td>8.43</td>
<td>40.38</td>
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<td>-0.348</td>
<td>0.274</td>
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<td>39.36</td>
<td>-0.904</td>
<td>-0.366</td>
<td>0.224</td>
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<tr>
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<td>7.52</td>
<td>39.61</td>
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<td>-0.342</td>
<td>0.270</td>
<td>103.24</td>
</tr>
<tr>
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<td>9.60</td>
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<td>-5.24</td>
<td>8.47</td>
<td>40.40</td>
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<td>-0.338</td>
<td>0.281</td>
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<tr>
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<td>-37.86</td>
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<td>7.46</td>
<td>38.59</td>
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<td>-0.320</td>
<td>0.263</td>
<td>103.04</td>
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<tr>
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<td>-0.936</td>
<td>-0.275</td>
<td>0.219</td>
<td>102.42</td>
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<td>2.93</td>
<td>7.15</td>
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<td>-0.307</td>
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<tr>
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<td>8.34</td>
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</tr>
<tr>
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<td>40.34</td>
<td>-0.871</td>
<td>-0.301</td>
<td>0.388</td>
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<td>-0.888</td>
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<td>0.353</td>
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<td>std. dev.</td>
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<td>4.41</td>
<td>1.08</td>
<td>0.020</td>
<td>0.023</td>
<td>0.064</td>
<td>0.62</td>
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<td>-37.83</td>
<td>1.19</td>
<td>9.72</td>
<td>39.57</td>
<td>-0.900</td>
<td>-0.318</td>
<td>0.287</td>
<td>103.06</td>
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<tr>
<td>RMSE</td>
<td></td>
<td></td>
<td></td>
<td></td>
<td>3.25</td>
<td></td>
<td></td>
<td>1.57</td>
</tr>
</tbody>
</table>

Table 3.3: Results of the dynamic validation test. The translations, $t_{x,y,z}$, are given relative to the coordinate system of the tibial component, $t_{rel}$ is the distance between the coordinate systems of the two implant components.

These results are more precise than the in vivo collected data due to the smaller motion blur in the in vitro images. A higher x-ray dose would also improve the results.
Figure 3.4: X-ray images of a full gait cycle after distortion correction. Image 1: heel strike; Image 17: toe off.
Figure 3.5: Femoral and tibial component rotations relative to the image focus.
51

<table>
<thead>
<tr>
<th>reference</th>
<th>type of validation</th>
<th>rot. accuracy [°]</th>
<th>transl. accuracy [mm]</th>
</tr>
</thead>
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<tr>
<td>Banks and Hodge 1996 [10]</td>
<td>relative pose between two fixed components, moved as single unit</td>
<td>1.1 (average)</td>
<td>0.5 (in-plane), 6.6 (out-of-plane, average)</td>
</tr>
<tr>
<td>Hoff et al. 1998 [46]</td>
<td>relative pose</td>
<td>0.3 (RMS)</td>
<td>0.5 (in-plane, RMS)</td>
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<tr>
<td>Zuffi et al. 1999 [103]</td>
<td>relative pose between two fixed components, moved as single unit</td>
<td>1.5</td>
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</tr>
</tbody>
</table>

Table 3.4: References of validation results.

3.5 Discussion

This measuring technique enables the tracking of several full gait cycles of normal level walking maintaining the implant components within the field of view of the fluoroscopic image. After two trials, the subjects became used to walking in the C-arm while keeping a natural gait pattern. The advantage of our system is that the subject can walk freely without any restriction, e.g. without the fixed gait velocity of a tread mill. The measuring system itself is double fail safe which guarantees absolute safety for the subjects as well as for the operators.

The translational RMSE of the static validation tests are 2.03 mm (series I) and 2.36 mm (series II), the rotational RMSE are 1.77°, and 0.75° respectively. No significant difference could be found. The results of the dynamic validation test are 3.25 mm, and 1.57° respectively. The difference of the standard deviation and the RMSE of the dynamic validation of 2.17 mm and 0.95° show a tendency to affect results due to motion blur. This tendency could not be shown in the static validation tests. Hence, motion blur in the fluoroscopic image does affect the accuracy of the pose reconstruction. Bone overlay does not significantly influence the accuracy, but it has an effect on the standard deviation of the out-of-plane translation. The standard deviations of the in-plane reconstruction are about 16 time less than the standard deviations of the out-of-plane reconstruction due to the ratio of the distance x-ray source to the image intensifier and the pixel resolution.

Taking the RMSE of the translation vector of 3.25 mm and a ratio of the in-plane/out-of-plane standard deviation of 16, a rough estimate of the in-plane error of the dynamic in vitro validation is about 0.2 mm.

Table 3.4 lists the validation results of three authors. Hoff et al. [46] found an RMSE of
0.5 mm of in-plane translation and 0.35° for all rotation angles in their in vitro accuracy test.

The in vitro studies of Banks and Hodge [10] show an average error of 0.5 mm for the in-plane translation and 6.6 mm for the out-of-plane translation respectively. The average RMSE of the estimated rotation angles is 1.1°. Zuffi et al. [103] found, in a static in vitro test, relative orientation and position errors of around 1.5 mm for the translations and 1.5° for the rotations. Furthermore, they made an in vivo test of knee bending in the sagittal plane. The real TKA kinematics were not known, but the sequence of the relative poses of the implant components were described as smooth, and as a reasonable physiological pattern of motion.

Our tests show that the accuracy of our reconstruction algorithm is comparable with the results achieved by other authors.

The use of video-fluoroscopy avoids skin movement artefacts which affects the significance of the measured knee joint kinematics dramatically. Stagni et al. [88] found a standard deviation of skin marker trajectory in the corresponding prosthesis-embedded anatomical frame up to 31 mm. Our measure system with a translational error in space of about 3.25 mm would about nine times more precise.

The limitations of the accuracy lie in the error due to motion blur. Improving the control system of the unit mover would reduce the relative velocity between the knee joint and the image intensifier, therefore reducing motion blur in the fluoroscopic images.

### 3.6 Conclusions

Accuracy tests show that the presented three dimensional reconstruction is comparable with other studies used video-fluoroscopy in TKA kinematic data acquisition. These in vitro validation tests were performed according to other authors ([10], [103], [46]).

This pilot study shows that it is possible to capture knee component kinematics of level walking during a full gait cycle with single plane video-fluoroscopy within a translational accuracy of 3.25 mm (spatial) and a rotational accuracy of 1.57°. Until now, it was not possible to get kinematic data of full gait cycles of normal walking with video-fluoroscopy. Enabling the more accurate tracking of the implant component kinematics during normal level walking due to the avoidance of skin movement artefacts sets a basis for further
investigation on the functionality of totally replaced knee joints. E.g. the effect of the screw home mechanism during normal walking could be studied on different implant designs such as fixed bearing or mobile bearing TKA.

Movable video-fluoroscopy could open a new area in TKA outcome studies, due to the achievement of more accurate implant component kinematics.

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Three dimensional kinematics and kinetics of total knee arthroplasty during level walking using single plane video-fluoroscopy and force plate - a pilot study

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Submitted to Gait and Posture
Abstract

The goal of the study was to simultaneously obtain accurate kinematic and kinetic data from a total knee arthroplasty (TKA) during level walking, by coupling a force plate data with the kinematics of TKA measured by a movable video-fluoroscopic system. Kinematic and kinetic information of a TKA is crucial for the improvement of implant designs and for the increased longevity of the implant components. Instrumented gait analysis, with skin mounted marker tracking and force plates, is a well established method for the acquisition of kinematic and kinetic data of TKA in vivo and for non invasively estimating the functionality of the joint. However, resultant moments at the knee joint is inaccurate with this method, due to skin movement artifacts. Video-fluoroscopy reduces these uncertainties by means of the direct tracking of the implant components with x-ray. However this measuring technique has the disadvantages of only providing kinematic data, and of a limited field of view of the image intensifier. This paper presents a newly developed measuring technique, which enables a more accurate resultant moments calculation for level walking than could be achieved by classical instrumented gait analysis.

**Keywords:** movable video-fluoroscopy, level walking, kinematics and kinetics, total knee arthroplasty, force plate data

4.1 Introduction

Accurate in vivo kinematic and kinetic data of total knee arthroplasties is important to understand the complexity of knee joint mechanics after knee joint surgery. This knowledge is crucial for the prevention of higher stress and strain on the ligament structure as well as on the implant components. A better understanding of knee joint mechanics could lead to better surgical strategies, improved implant designs, and increased longevity of the implant components. These factors improve patient satisfaction and eventually reduce health care costs.

As the functionality determines the load on the joint and the motion of the joint, there is a relationship between the longevity of the implant components and the functionality of the TKA [3]. Thus reliable subject’s kinematic and kinetic data are required. Instrumented gait analysis is a well established method for gathering kinematic and kinetic
data of the knee joint in vivo and for non invasively estimating the functionality of joints [52], [93]. Optical markers are placed on the subjects extremities in order to gain kinematic information from the segments of interest, and force plates fixed on the floor simultaneously measure ground reaction forces.

The main problem with this approach is the sizeable error caused by skin movement artefacts [83], [48]. Point cluster methods were used to improve movement analysis [6]. However, Stagni et al. [88] found a standard deviation of skin marker trajectory of thigh and shank marker cluster of twenty markers each of 31 mm, and 21 mm respectively. They found root mean square errors in internal and external rotations and ab/adduction of 192% and 117% relative to the corresponding range.

Single plane video-fluoroscopy is a method, which enables the reconstruction of the three dimensional position and orientation (pose) of the implant components more accurately by avoiding errors due to skin and muscle movements [83]. A registration algorithm estimates the pose of the implant components from the single plane projection view of the fluoroscopic image series [10], [46], [103] within an accuracy of 1.5 mm.

Standard video-fluoroscopy has the disadvantage of the small field of view of the image intensifier, which is usually in a fixed position with a field of view of 320 mm, making it impossible to obtain kinematic data from the knee during level walking. Zihlmann et al. [102] proposed a movable fluoroscopic system enabling the tracking of the knee joint during level walking, therefore collecting kinematic data of the implant components with video-fluoroscopy during several gait cycles. Video-fluoroscopy is limited to kinematic data only, so that the load at the knee joint can not be calculated.

Some studies on knee joint kinematics and kinetics were performed using force plates and video-fluoroscopy [35], [32], [25]. However, this data was not obtained simultaneously. Simultaneous kinematic and kinetic data acquisition is important in order to have reliable information about the functionality of a knee joint. Until now, none of the studies found in literature were able to achieve simultaneous kinematic and kinetic data during level walking in an accuracy as presented in this work.

The goal of this pilot study was to attain kinematic and kinetic data during level walking more accurately than could be achieved by classical instrumented gait analysis. This was implemented by coupling movable video-fluoroscopy with force plate data.
4.2 Materials and methods

4.2.1 Experimental setup

The subject had to perform several consecutive strides on the laboratory walkway, a distance of approximately 10 m. During this task the subject’s kinematic data was gathered with a movable video-fluoroscopic system (BV Pulsera, Philips Medical Systems, Switzerland), by a pulsed mode of 25 Hz and 8 ms shutter time.

The C-arm of the fluoroscope was fixed on a motor driven trolley, (thereafter the unit mover) enabling the tracking of the knee joint movement by keeping the implant components within the field of view. Three reflective markers (VICON™, Oxford Metrics Inc., Oxford, UK) were placed on the unit mover and the C-arm in order to record the trolley’s movement with optical motion tracking in the laboratory’s coordinate system (subsequent called global coordinate system). The kinetics were measured by five KISTLER™ (KISTLER AG, Switzerland) force plates, which were fixed on the floor. The global coordinate system’s origin was set in the centre of the third force plate, the y-axis was set in gait direction, the z-axis vertical to the floor, and the x-axis medio/lateral.

**KISTLER force plate:** The force plates used for this study were five 400 mm × 600 mm multicomponent force plates based on piezo-electric sensors. All force plates were completely mechanically decoupled from the surrounding floor thus avoiding an interaction with the fluoroscopic system (Figure 4.1). The sampling frequency of the force plates was 1000 Hz. The global coordinate system’s origin was set at the centre of the third force plate.

**Optical tracking system VICON:** The 8 cameras of the optical tracking system VICON (V612 MX40) were calibrated to a space of 2000 mm × 4000 mm × 2000 mm with the coordinate system’s origin at the centre of the third force plate. All series were measured with a sampling frequency of 100 Hz.

**Movable video-fluoroscopy:** The C-arm of the fluoroscopic unit, together with the x-ray source and the image intensifier, is mounted on the unit mover. The system accelerates and decelerates thereby maintaining the knee joint within the field of view of the fluoroscope [102].

**Distortion correction of the fluoroscopic images:** The distortion in the fluoroscopic
images was corrected by a calibration grid which was fixed on the image intensifier and filmed before each measurement cycle. The distortion of the images was corrected by determining the bilinear transformation between the known and measured coordinates of control points on the calibration grid. Due to the rigid connection of the x-ray source with the image intensifier, the calibration was performed statically and was used over all fluoroscopic images. The principal point lies at the image centre, the focus length is 1700.3 mm given by the manufacturer.

**Three dimensional reconstruction of the implant’s pose:** A three dimensional analysis of each fluoroscopic image was achieved by fitting a synthetic x-ray image of the implant component (CAD model) to the original x-ray image [19]. After a rough initial estimation of its pose, the implant’s exact six degrees of freedom were found by an iterative minimisation of the difference between the synthetic and the original image.

The estimated parameters were the position vector \( \mathbf{t} \) of the implant, parallel \( (t_x \text{ and } t_y) \) and perpendicular \( (t_z) \) to the image plane, and the orientation, defined by the angles of out-of-plane rotation \( x_{rot} \) and \( y_{rot} \) and the angle \( z_{rot} \) of the rotation in the image plane. The rotation angles \( x_{rot}, y_{rot}, \text{ and } z_{rot} \) are inserted in the standard definition of the rotation matrices, \( R_{x_{rot}}, R_{y_{rot}}, \text{ and } R_{z_{rot}} \).

The orientation of the implant components is calculated by multiplying the rotation ma-
trices in the order $R_{zrot}$, $R_{xrot}$, $R_{yrot}$. The matrix $R_{fem}$ is the orientation of the femoral component, $R_{tib}$ the orientation of the tibial component. A third order Butterworth low pass, zero-phase forward and reverse filter was applied in order to slightly smooth the pose reconstructions. A normalised cut off frequency of 0.1 rad/s was used.

### 4.2.2 Subject

The subject for this pilot study had a total knee replacement (balanSys™ fixed bearing, Mathys Ltd Bettlach, Switzerland), one year after surgery, without any complication.

### 4.2.3 Coupling the measurement systems

**Transformation of the fluoroscopic image focus into the laboratory’s global coordinate system:** The reconstruction of the implant’s pose was performed relative to the focus of the fluoroscopic image. With the use of force plates, a transformation of the implant’s six degrees of freedom into the global coordinate system was required (Figure 4.2).

![Figure 4.2: Focus coordinate system and global coordinate systems.](image)

In order to perform this transformation, the calibration grid was fixed in a frame, which was screwed onto one force plate (Figure 4.3) in a well defined position relative to the force plate’s origin.

A program written in MATLAB (Matlab, The MathWorks, Inc.) calculated the focus location in the global coordinate system by comparing the location of the grid’s centre and the distance of the grid lines of the fluoroscopic image with the measured true location.
Figure 4.3: The calibration grid was fixed in a frame on a defined position on the force plate. Three reflective markers are fixed on image intensifier, C-arm, and x-ray source.

and size. VICON tracked the markers during all the events thus enabling the estimation of the focus' location over all image frames.

The focus moves in the x-y-plane only \((x_{global}, y_{global})\) because the vertical distance of the C-arm was adjusted to the length of the subject’s lower leg and fixed at this height. The focus in the global coordinate system was calculated by a linear transformation (Equation 4.1) into the x-y-plane, where \(M\) is the transformation matrix containing two unit vectors, \(e_1\) (marker 1 to marker 2) and \(e_2\) (marker 1 to marker 3), in its columns, \(c\) is the vector from focus to marker 1, and \(f_{global}\) is the time dependent vector of the focus' location over all measured frames in the global coordinate system.

\[
f_{global} = M^{-1} \ast c \ast M;
\]

(4.1)
Equation 4.2 is the transformation from the focus’ coordinate system into the global coordinate system, where \( \mathbf{R} \) is the rotation matrix (\( \mathbf{R}_\text{fem} \) or \( \mathbf{R}_\text{tib} \)) which has been multiplied with \( \mathbf{S} \), where \( \mathbf{S} = [0, -1, 0; 0, 0, 1; 1, 0, 0] \). \( \mathbf{t}_{\text{focus}} = (t_x, t_y, t_z) \) is the implant’s position vector in the focus coordinate system, and \( \mathbf{t}_{\text{global}} = (t_x, t_y, t_z) \) the position vector in the global coordinate system.

\[
\mathbf{t}_{\text{global}} = \mathbf{R} \cdot \mathbf{t}_{\text{focus}} + \mathbf{f}_{\text{global}} \tag{4.2}
\]

**Time synchronisation:** A reflective marker was placed at the end of a stick with a length of 500 mm. This stick was introduced into the fluoroscopic image with a short downwards movement at the beginning of each movement task after starting the x-ray beam and the VICON system. This vertical movement was visible both in the VICON’s vertical coordinates and in the corresponding fluoroscopic image, enabling a time synchronisation of the measuring systems.

### 4.2.4 Finite axis of rotation

The finite axis of rotation and the corresponding rotation angle of each time step, \( \Delta t = 40 \text{ ms} \), between the femoral and tibial implant component were calculated by extracting vector, \( \mathbf{u}_{x,y,z} \) and angle, \( \beta \) of the relative rotation matrix \( \mathbf{R}_{rel} \) (Equation 4.3). The relationship between the rotation matrix and the screw vector and angle is given in Equation 4.4 (Equation of Rodrigues). There \( \mathbf{u} \circ \mathbf{u} \) is the diadic product, \( \mathbf{I} \) the identity matrix, and \( \mathbf{u} \times \) is the skew symmetrical vector product of \( \mathbf{u} \). The components of \( \mathbf{R} \) as function of \( x_{rot}, y_{rot}, \) and \( z_{rot} \) are compared with the entries of \( \mathbf{R} \) as function of \( \mathbf{u} \) and \( \beta \).

\[
\mathbf{R}_{rel} = \mathbf{R}_\text{fem} \cdot \mathbf{R}_\text{tib}' \tag{4.3}
\]
\[ \mathbf{R}(\mathbf{u}, \beta) = \cos \beta \mathbf{I} + (1 - \cos \beta) \mathbf{u} \circ \mathbf{u} + \sin \beta \mathbf{u} \times \mathbf{u} \]  

(4.4)

### 4.2.5 Error of resultant moments estimation

An error analysis was performed to estimate the errors affecting resultant forces and moments calculation in the knee joint. Resultant forces were calculated by using the measured ground reaction forces of the force plate. Resultant moments were calculated by reducing the moments into the finite centre of rotation, COR, of the knee joint (see Figure 4.4). Thus, the resultant moments estimation is a function of the location vector of COR, the point of force application, POFA, and the ground reaction force components, \( F_x \), \( F_y \) and \( F_z \). Assuming quasi static behaviour and reducing the movement into the sagittal plane, an uncertainty estimation of the resultant moments of the knee joint was performed, i.e. the error in resultant moments remains a function of \( F_z \), \( F_y \) \( \text{COR}_y \), \( \text{COR}_z \) and \( \text{POFA}_y \) (Equation 4.5), and the error components listed in Table 4.1. The uncertainty of the resultant moments of the knee joint considering the condition listed above can be estimated by using differential calculus (Equation 4.6).

\[
\begin{bmatrix}
M_{\text{res}} \\
0 \\
0 \\
\end{bmatrix} =
\begin{bmatrix}
0 \\
(COR_y - \text{POFA}_y) \\
\text{COR}_z \\
\end{bmatrix}
\times
\begin{bmatrix}
0 \\
F_y \\
F_z \\
\end{bmatrix}
\]

(4.5)

\[
\Delta M_{\text{res}} = \left[ (-F_z \Delta \text{POFA}_y)^2 + (F_z \Delta \text{COR}_y)^2 + (-F_y \Delta \text{COR}_z)^2 + ((\text{COR}_y - \text{POFA}_y) \Delta F_z)^2 + (-\text{COR}_z \Delta F_y)^2 \right]^{\frac{1}{2}}
\]

(4.6)

The errors affecting the resultant moments calculation are listed below.

**Error of the force vector:** The error in ground reaction forces components while using KISTLER force plates was found to be about \( \pm 4 \) N for \( F_z \) and \( F_y \) [14]. The point of force application was found to have an error of 3 mm in a diameter of about 100 mm around
the centre of the force plate.

**Error of the fluoroscopic focus location in the global coordinate system:** Four factors affect the uncertainty of the location estimation of the fluoroscope’s focus. These are:

i) The calibration grid’s well defined fixation to the force plate.

ii) The spatial coordinate estimation of the three markers of the fluoroscopic system.

iii) The uncertainty of the estimation of the focus in the fluoroscopic image.

iv) The stiffness of the C-arm.

The calibration grid’s exact position could be measured with an uncertainty of 0.1 mm. The spatial coordinate estimation of the three markers of the fluoroscopic system was measured to have an absolute uncertainty of 1 mm. The uncertainty of the focus location in the fluoroscopic image was estimated to be 0.1 mm.

To evaluate the stiffness of the C-arm while the system is moving, metal discs with a diameter of about 5 mm were bonded on the bottom of a rigid plastic box, which was fixed on the image intensifier, so that the discs were held in a distance of 300 mm parallel to the image intensifier. Fluoroscopic images were taken while the unit mover was accelerating...
and decelerating. The position of the discs were compared in every taken image by image subtraction (gray scale). No difference in the images could be found. Hence, we assume that the C-arm is stiff enough for our application.

In conclusion, the uncertainty in fluoroscopic focus location in the global coordinate system is 1 mm.

**Error of implant’s pose reconstruction:** A static and dynamic in vitro validation study was performed to estimate the uncertainty of the implant’s pose reconstruction. The three implant components were fixed, allowing no relative movements between the implant components. The exact position and orientation of the tibial component relative to the femoral component was measured by a three dimensional optical digitiser (Breuckmann OPTO Top HE-200, Germany) which measured surface points within an accuracy of $\pm 0.01$ mm. The CAD geometries of the implant components were then fitted in the measured surface point cloud (Raindrop Geomagic Studio 6.0), while the relative position and orientation of the component’s coordinate system’s origin were calculated. Four images of the implant components were taken from different point of view. One series was taken by moving the implant components with a velocity of 0.3 m/s relative to the image focus. The matching algorithm calculated the implants pose in all frames relative to the image focus.

The RMSE of the pose (distance $t$, and spatial rotation angle $\beta$) of the femoral part relative to the tibial part were calculated. The dynamic validation study found an RMSE of the distance of 3.25 mm and a RMSE of the rotation angle of about 1.57°. A repeatability study was performed while using the matching algorithm 15 times on the same images. The ratio of the standard deviations of the in-plane to out-of-plane translation was found to be 16. Thus, a rough estimate of the accuracy of the in-plane translation can be 3.25 mm/16 = 0.20 mm.

### 4.3 Results

#### 4.3.1 Error of resultant moments estimation

The preceding results show that the error of the direction of the axis of rotation can be neglected. Thus, the error of the in-plane translation relative to the focus is a factor,
Table 4.1: RMSE of the components influencing resultant moments estimation of the knee joint.

<table>
<thead>
<tr>
<th>influence factor</th>
<th>method of estimation</th>
<th>RMSE</th>
</tr>
</thead>
<tbody>
<tr>
<td>three dimensional reconstruction</td>
<td>in-plane translation, $\text{COR}_y, \text{COR}_z$</td>
<td>0.20 mm</td>
</tr>
<tr>
<td></td>
<td>spatial rotation angle $\beta$</td>
<td>1.57°</td>
</tr>
<tr>
<td>focus location, global</td>
<td>VICON marker tracking</td>
<td>1 mm</td>
</tr>
<tr>
<td>axis of rotation estimation</td>
<td></td>
<td>1.45 mm</td>
</tr>
<tr>
<td>global</td>
<td></td>
<td></td>
</tr>
<tr>
<td>$F_x$</td>
<td>KISTLER force plate</td>
<td>±4 N</td>
</tr>
<tr>
<td>$F_y$</td>
<td>KISTLER force plate</td>
<td>±5 N</td>
</tr>
<tr>
<td>POFA calculation</td>
<td>KISTLER force plate</td>
<td>±3 mm</td>
</tr>
</tbody>
</table>

which has to be taken account of when estimating the finite axis of rotation. The in-plane translation of the fluoroscopic system equals the y-z-plane of the global coordinate system, where the y-axis was the subject’s gait direction. The uncertainty in estimating the focus in the globale coordinate system has to be taken into account when calculating the centre of rotation. These two errors are independent and therefore their variance can be added, equals $\sqrt{0.2 \text{mm}} + \sqrt{1 \text{mm}} = 1.45 \text{ mm}$.

The axis of rotation was mainly parallel to the x-axis of the global coordinate system, since the largest rotation between the implant components during level walking occurs in the y-z-plane (knee joint flexion). Therefore, the worst case error of the finite axis of rotation location in the global coordinate system is the sum of the error of the in-plane reconstruction and the error of the focus location in the global coordinate system. The sources of errors affecting resultant moments estimation are summarised in Table 4.1.

Assuming $F_y = 50 \text{ N}$, $F_z = 800 \text{ N}$, $\text{COR}_y - \text{POFA}_y = 50 \text{ mm}$, and $\text{COR}_z = 450 \text{ mm}$, and taking the values of the uncertainties as listed in Table 4.1, the sensitivity of $\Delta M_{res}$ is as shown in Figures 4.5, and 4.6. In Figure 4.5 $F_y$ and $F_z$ vary from 1 to 800 N while the other parameters are fixed. In Figure 4.6 $\text{COR}_y$ and $\text{POFA}_y$ vary from 1 to 200 mm. The blue lines are the plots where $\Delta \text{COR}$ is 0.2 mm, as found in our validation, red represents $\Delta \text{COR} = 1.5 \text{ mm}$, and black represents $\Delta \text{COR} = 2 \text{ mm}$. $M_{res}$ varies from 2.26 Nm, 3.26 Nm (blue), 2.26 Nm, 2.95 Nm (red), and 3.44 Nm, 3.93 Nm (black) in Figure 4.5, and from 3.29 Nm to 3.38 Nm (blue), 3.50 Nm to 3.58 Nm (red), and 3.66 Nm to 3.74 Nm (black) in Figure 4.6. For small values of $F_y$ and $F_z$ the uncertainty of $M_{res}$ remains.
Figure 4.5: Sensitivity of $\Delta M_{res}$ by varying $F_y$ and $F_z$. blue: $\Delta COR = 0.2$ mm; red: $\Delta COR = 1.45$ mm; black: $\Delta COR = 2$ mm

Higher $F_z$ do affect $M_{res}$ more profoundly than higher $F_y$ values. While varying $POFA$ and $COR$, i.e. the lever arm of the moment, the uncertainty in $COR$ estimation has an
increasing effect on the resultant moment.

4.3.2 Resultant moments at the knee joint

![Graph showing resultant moments at the knee joint](image)

Figure 4.7: Finite axis of rotation between the implant components and force vectors of one force plate. $\Delta t = 40$ ms.

The results of coupling force plate data with the reconstructed video-fluoroscopic sequences are illustrated in the Figure 4.7. This Figure shows the finite axis of rotation between the femoral and the tibial implant component during level walking in a four dimensional plot (Stüssi et al. [92]). Ground reaction force vectors, where the TKA leg is in contact with the force plate, are plotted in the same Figures. The resultant moments at the knee joint during the stance phase of gait were calculated by reducing the moments in the finite axis of rotation in the sagittal plane (Figure 4.8). The resultant moment at the knee joint reaches 50 Nm. Taking the best and worst results of $\Delta M_{res}$ with $\Delta COR = 1.45$ mm related to the uncertainty in pose reconstruction and the uncertainty in focus location in the global coordinate system, 2.26 Nm, and 3.58 Nm respectively, the ratio of $\Delta M_{res}/M_{res}$ is 4.5% and 7.16%. Assuming an uncertainty in $COR$ estimation of about 30 mm, which is comparable with the error due to skin movement artefact, $\Delta M_{res}$ would be about 24 Nm.
4.4 Discussion

Classic instrumented gait analysis with skin mounted markers has the disadvantage of measuring artefacts due to skin movement relative to the underlying bones. Video-fluoroscopy is a well established method of more accurately measuring knee joint kinematics by avoiding the use of skin markers. The small field of view of the image intensifier and the possibility to only gain kinematic data are the two main disadvantages of this system. The goal of the present study was to acquire kinematic and kinetic data during level walking more accurately than could be achieved by classical instrumented gait analysis. The presented measuring technique enables kinematic and kinetic data acquisition of TKA during the stance phase of level walking more accurately than could be achieved by a tracking system with the use of skin mounted markers. A sensitivity analysis of each estimation and calculation step showed that the uncertainty in calculating the axis of rotation of the knee joint was a translational error in space of 3.1 mm, and an error of the rotation angle $\beta$ of 1.57° which is comparable with the accuracy achieved by other authors [46], [103]). Kinematic and kinetic information, with the accuracy mentioned above, enable a more reliable estimation of the resultant forces and moments, by reducing ground reaction forces into the finite axis of rotation of each time step, than classical instrumented gait analysis. Stagni et al. [88] found a RMS difference of 17.2% and 23.4% respectively of the motion.
range during a sit down tasks. The standard deviations of internal and external rotations and ab/adduction were much larger. They show that these large errors can nullify the usefulness of these variables in the clinical interpretation of gait analysis. In particular, when calculating resultant moments at the knee joint, there are errors due to skin movement. Assuming an uncertainty of about 24 Nm due to the error in COR estimation of about 30 mm, this would lead to an uncertainty of about 50% of the resultant moment. This quantity of the uncertainty of the resultant moment due to skin movement artefacts was found also by Barth [14]. Barth made his error calculation by comparing computed resultant moments and measured ones via a strain gage instrumented lower leg prosthesis. The error calculation in this study shows that the error of the resultant forces at the knee joint is about 3.38 N using this measuring system. This is about seven times less than classic instrumented gait analysis with skin mounted markers would achieve. This study shows that the movable video-fluoroscopy system coupled with force plates allows the subject to walk freely without any restrictions, thus obtaining the subject’s normal gait pattern while obtaining kinematic as well as kinetic data.

4.5 Conclusion

This pilot study showed that it is feasible to measure ground reaction forces and kinematic data simultaneously with a moving fluoroscopic system. More accurate in vivo kinematic information of the TKA during the stance phase of level walking can be obtained than that afforded by instrumented gait analysis, due to the avoidance of skin movement artefacts. Therefore the resultant moments in the knee joint can be estimated up to seven times more accurately than classical instrumented gait analysis. This work presents an improvement in measurement accuracy for the analysis of the stance phase of gait, thereby providing a basis for better inverse dynamic calculation of the load.

Acknowledgement

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Femoral component malrotation on TKA - Functional implications on ligaments based on a computer model and quantitative movement analysis

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5. Summary

Femoral component malrotation has a profound effect on the ligament balancing and is therefore an often discussed problem in modern total knee replacements. Clinicians found that an internally rotated femoral component of $5^\circ$ can cause lateral knee joint laxity and tightness on the medial side, whereas an external rotation up to $8^\circ$ seems tolerable for the patients. The goal of this study was to simulate different degrees of femoral component malrotation to estimate the effect on ligament loading. This simulation was based on accurate quantitative movement analysis performed with coupled video-fluoroscopy and force plate data, and a computer model. The results state that during a sit down task (sitting down and standing up) a femoral component internal rotation of $5^\circ$ shows more load on the anterior bundle of the medial collateral ligament, while the lateral collateral ligament remains lax; However, when the femoral component is rotated externally up to $8^\circ$, the collateral ligaments remained balanced. These effects correspond to the observations of clinicians, who noticed lateral joint laxity and medial tightness due to a femoral component internal rotation. The presented findings might be an explanation why internal rotation of the femoral component is less tolerated than external rotation. This simulation is a basis for further investigations on TKA mechanics based on accurate in vivo kinematics and kinetics and individual anthropometric data leading to a better understanding of the mechanics of the totally replaced knee joint.

Keywords: total knee arthroplasty, knee joint ligaments, femoral component malrotation, video-fluoroscopy, knee joint kinematics and kinetics

5.2 Introduction

Femoral component malrotation in total knee replacements is an often discussed problem in modern total knee replacement surgery [47]. Malrotation may lead to flexion gap asymmetry, polyethylene inlay wear, patellar maltracking [66], and may contribute to implications like knee pain and knee joint laxity [15], [59], [80], [41]. These failures often necessitate revision surgery.

The normal knee joint is one of the most complex joints of the human body. Two bones are held together by ligament structures, namely, the collateral ligaments, the cruciate
ligaments, and the knee joint capsule. The patella which is connected by the patellar ligament, tracks in the femoral groove enabling the rotational stability of the knee joint during flexion. By replacing a knee joint with a total knee arthroplasty, the component’s alignment to the bones affects the mechanical behaviour of the knee joint considerably. This is due to the mechanical constraints of the implant components. The femoral component rotational alignment especially affects the knee joint mechanics in flexion and in extension profoundly in all six degrees of freedom. Changing the rotational alignment could lead to ligament unbalancing causing lateral flexion instability. This typically occurs when getting out of a chair. Other effects are medial tightness possibly leading to secondary arthrofibrosis [80], higher stress on the polyethylene inlay [26], [60] and pain [79].

Malalignment of a measurable degree occurs in approximately 10% - 30% of patients, depending on the surgical technique and the anatomical landmarks used [13], [71], but not all of these patients complain of unsatisfactory results. For most patients, a relative internal rotation of the femoral component between 3° and 6° is tolerable [12], [50], [79], while external rotation of the femoral component can be as high as 8° without causing clinical problems [70].

Anatomic variability makes it often impossible to successfully use statistical conclusions for surgical strategies. Awareness of the mechanical behaviour of the individual knee joint mechanics is essential to reach the best patient outcome. Thus, visualisation of the altered kinematics due to femoral component malrotation is crucial to the understanding of the complex mechanics [101].

Effects on ligament structures due to femoral component malrotation have been studied on cadaver knees [53], [79]. However, it is difficult to simulate physiological motion patterns using cadaver knees. These effects should be studied on physiological motion patterns. Therefore a knee joint model based on accurate kinematic and kinetic data of subjects during activities is necessary for studying the effects of malrotation.

Single plane video-fluoroscopy is a well established method of measuring the kinematics of the implant components directly while avoiding the large error due to skin movement relative to the underlying bone [88]. Registration algorithms have been used to reconstruct the three dimensional position and orientation of the implant components from the projection view of the x-ray images [10], [46], [103], [19]. Video-fluoroscopy is limited to kinematic
data acquisition, therefore the load at the knee joint can not be estimated. When coupling video-fluoroscopy with force plate measurements, inverse dynamic calculation can be performed. This information provides a basis for a knee joint model simulating the strain on ligament structures with different rotational alignments of the femoral component.

The goal of this study is to show the effects of femoral component malrotation on ligament structures, based on the kinematic and kinetic data of a subject performing a sit down task. Thereby, the posterior cruciate ligament, PCL, the medial collateral ligament, MCL, and the lateral collateral ligament, LCL, were analysed for the alteration of strain during the sit down task. In addition, the forces produced by increased strain in the ligament structures were estimated.

5.3 Material and Methods

Video-fluoroscopy and a force plate were used to collect the subjects kinematic and kinetic data during a sit down task. A knee joint model including the implant components and the bones visualises the kinematics and calculates the strain on ligament structures. Effects on strain on ligaments were studied by simulating different degrees of femoral component malrotation. Further, the resultant moment at the knee joint was calculated to see exactly when the subject is seated on the chair and when standing up. See below for more details.

5.3.1 Experimental setup

A KISTLER force plate (KISTLER AG, Switzerland) and Video-fluoroscopy (BV Pulsera, Philips Medical Systems, Switzerland) were used for the data acquisition of the kinematics and kinetics during a sit down task (sitting down and getting up). The chair and the collateral leg of the subject were not in contact with the force plate. The height of the chair was about 70% of the height of the shank thereby flexing the knee joint 90° while seated on the chair.

**Force plate:** The force plate used for this study was a 400 mm × 600 mm multicomponent force plate based on piezo-electric sensors. All force plates were completely mechanically
decoupled from the surrounding floor avoiding an interaction with the fluoroscopic system. The sampling frequency of the force plates was 1000 Hz. The origin of the global coordinate system was set at the centre of the force plate.

**Single plane video-fluoroscopy and three dimensional pose reconstruction:** A calibration grid fixed on the image intensifier was filmed before each measurement series in order to correct the distortion in each fluoroscopic image. The distortion of the images is eliminated by determining the bilinear transformation between the known and measured coordinates of control points on the calibration grid. The distortion correction was performed in all images before the three dimensional pose reconstruction.

A three dimensional analysis of each fluoroscopic image was achieved by fitting a synthetic x-ray image of the implant component (CAD model) to the original x-ray image after Burckhardt [19]. The estimated parameters were the position vector \( \mathbf{t} \) of the implant, parallel \( (t_x, t_y) \) and perpendicular \( t_z \) to the image plane, and the orientation, defined by the angles of out-of-plane rotation \( zrot \) and \( yrot \) and the angle \( zrot \) of the rotation in the image plane. A third order Butterworth low pass, zero-phase forward and reverse filter was applied to slightly smooth the pose reconstructions.

**Coupling kinematic and kinetic data:** The fluoroscope’s position in the global co-
ordinate system was optically tracked by VICON\textsuperscript{TM} (VICON Motion Systems Ltd.). A calibration frame was fixed onto the force plate in a measured position relative to the global coordinate system’s origin. The calibration frame was filmed by the fluoroscope. A MATLAB\textsuperscript{TM} (Matlab, The MathWorks, Inc.) routine calculated the focus’ location in the global coordinate system by comparing the location of the grid’s centre and the distance of the grid lines of the fluoroscopic image of the calibration frame with the measured true location and size. The focus location in the global coordinate system was defined with this procedure. Before each measurement series a time trigger was set coupling the fluoroscopic images with the ground reaction force data.

### 5.3.2 Subject

The subject (male, 65 years old) for this pilot study was free from additional pathologies of the locomotor system or from other conditions that might affect the motion pattern, one year after total knee replacement (balanSys\textsuperscript{TM} fixed bearing, Mathys Ltd Bettlach, Switzerland). The CT scan of the knee joint shows a neutral alignment of the femoral component.

### 5.3.3 Computer model

The visualisation and simulation of the movement patterns were performed by a computer model created and calculated by MSC.ADAMS 12.0.0 (MSC Software Corporation). CT scans of the subject’s femoral head, the knee joint, and the ankle joint as well as an ap (ante-posterior) scan of the whole leg were performed. The three dimensional geometry of the bones, namely tibia, femur, and fibula were taken from a bone database. The contour of the bony structures in the CT scans were exported as splines (Unigraphics Solutions). The bone’s geometry (database) was fitted to the splines by changing its shape (Raindrop Geomagic Studio 6.0). The CAD geometries of the implant components were aligned relative to the bones as found in the CT scans. The insertion points of the ligament structures, PCL, LCL, and the two bundles of the MCL on femur, tibia and the head of the fibula were identified on the bony prominence. The displacement relative to the corresponding insertion points of the ligaments was es-
timated during the simulation of the sit down task. The forces caused by a change in displacement due to the femoral component malrotation were calculated by using an estimate of the material properties of the ligament structures.

**Kinematics of implant components:** The kinematics of the implant components calculated by the matching algorithm were the input data for the knee joint model. A spline interpolation of the three translational ($t_x, t_y, t_z$) and three rotational ($x_{rot}, y_{rot}, z_{rot}$) degrees of freedom was performed to calculate 300 time steps.

**Ligament models:** The distances between the corresponding insertion points of each ligament, PCL, LCL, and two bundles of MCL were calculated during the simulations. The medial collateral ligament was modelled as two bundles due to the wide structure including the superficial medial collateral ligament added to the deep medical collateral ligament (referred to here as MCL ant) and the posteromedial capsule (referred to here as MCL post) [78].

Assuming that the ligament structure was relaxed at time $t = 0$ s, when the subject was in an upright position (flexion angle $0^\circ$), the distance of the insertion points at this position was taken as reference length, $l_0$. For this purpose the mean value of the distance over 20 frames was taken. The knee joint flexion angle was defined as the angle between the centre of gravity of the femur, the centre of gravity of the femoral component and the centre of gravity of the tibia.

The material properties of the ligaments mentioned above, published by Butler [20] and Weiss [76] were used to calculate the load in the ligaments. The numbers taken for the calculation of the force are listed in Table 5.1. The ligaments were modelled as purely elastic tensile springs with a linear force-strain relation as shown in Equation 5.1, where $F$ is the force, $\varepsilon$ the strain, $A$ the cross-section area and $E$ the elasticity modulus.

\[
F = A \cdot E \cdot \varepsilon; \quad \text{for } l \geq l_0 \tag{5.1}
\]
\[
F = 0; \quad \text{for } l < 0 \tag{5.2}
\]

with $\varepsilon = (l - l_0)/l_0$

**Resultant forces and moments calculation:** The knee joint model shown in Figure 5.2 was used to calculate the resultant forces and moments at the knee joint in the sagittal
Table 5.1: Material properties of ligament structures.

<table>
<thead>
<tr>
<th>ligament</th>
<th>tangent modulus E [MPa]</th>
<th>cross-sectional area [mm²]</th>
</tr>
</thead>
<tbody>
<tr>
<td>PCL</td>
<td>345.0±22.4 [20]</td>
<td>1.81 [20]</td>
</tr>
<tr>
<td>MCL post</td>
<td>332.2±4.8 [76]</td>
<td>2.75</td>
</tr>
<tr>
<td>MCL ant</td>
<td>332.2±4.8 [76]</td>
<td>2.75</td>
</tr>
<tr>
<td>LCL</td>
<td>345.0±22.4 [20]</td>
<td>2.75 [20]</td>
</tr>
</tbody>
</table>

plane by reducing the forces into the finite axis of rotation of the two implant components (see Equation 5.3). A.2 $F_x$ and $F_z$ are the components of the ground reaction force measured by the force plate, $POFA$ is the point of force application on the surface of the force plate, $COR_y$ and $COR_z$ are the corresponding components of the location vector COR.

$$ \begin{bmatrix} M_{res} \\ 0 \\ 0 \end{bmatrix} = \begin{bmatrix} 0 \\ (COR_y - POF_A_y) \\ COR_z \end{bmatrix} \times \begin{bmatrix} 0 \\ F_y \\ F_z \end{bmatrix} $$  \hspace{1cm} (5.3)

Figure 5.2: Knee joint model for resultant moments calculation.

**Femoral component malrotation:** A series of different degrees of femoral component malrotation relative to the femur was applied to the knee joint model, external 5°, external 8°, and internal 5°, internal 8°, and 0° i.e. neutral position. Neutral was defined as the measured position and orientation of the implant components relative to the bones of the subject’s CT scans.
5.4 Results

5.4.1 Resultant forces and moments at the knee joint

Figure 5.3: Vertical component of the ground reaction force, the resultant moments reduced in the finite centre of rotation and the quadriceps force during the sit down task

Figure 5.3 shows the vertical component, $F_z$, of the ground reaction force and the resultant moment at the knee joint during the motion task. The ground reaction force curve illustrates when the subject was sitting down (from 0.9 s to 2.9 s), was seated on the chair (from 2.9 s to 5.2 s), and getting up (from 5.2 s to 7.1 s). The resultant moment increased during knee joint flexion. While the subject was making contact with the chair the resultant moment decreased and remained basically zero while the subject was seated on the chair. The resultant moment at the knee joint reaches about 50 Nm, where $F_z$ is decreasing while the subject is in contact with the chair. During the time the subject was seated, the ground reaction force measured basically the weight of the leg.
5.4.2 Femoral component malrotation

5.4.2.1 Strain on the ligament structures

The strain of the ligaments PCL, MCL ant, MCL post and LCL in the five different femoral components alignments are shown in Figure 5.4. Negative strain signifies a laxity in the ligament structure.

At the very beginning of the sit down task, the strain of the PCL is positive in all cases. The strain of the LCL remains basically zero for neutral and 5° internal rotation. At 8° internal rotation the strain of the LCL is negative, i.e. the ligament is lax during flexion. At 5° external rotation the strain of the LCL increases during knee joint flexion. At 8° external rotation the strain increase to 0.2. The strain of the posterior bundle of the MCL is about 0 for the neutral alignment, increases for femoral component internal rotation and is lax during flexion for femoral component external rotation. The difference between 5° and 8° is marginal. The anterior bundle of the MCL is tight during flexion at neutral alignment. The strain increases more at 5° internal rotation than at 0° femoral component rotation. There is a slight increase of the strain at 8° component internal rotation compared to 5° internal rotation. The strain of the MCL is smaller at femoral component external rotation than at neutral alignment.

5.4.2.2 Load at the ligaments

The results of the load estimation on the ligament structures are shown in Figure 5.5. The PCL is loaded at the very beginning of the sit down task. The LCL is loaded at neutral alignment and at femoral component external rotation with a slight increase of the load at 8° rotation compared to 5°. A femoral component internal rotation at 5° causes higher loads in the MCL ant, this increase remains from 5° to 8°. The LCL is not loaded while the femoral component is internally rotated. Thus, with femoral component internal rotation the MCL is loaded and the LCL is lax, whereas with femoral component external rotation the collateral ligaments are balanced.
Figure 5.4: The strain $\varepsilon$ of the ligaments PCL, MCL, LCL during the sit down task at different degrees of femoral component rotation ($\varepsilon = (l - l_0)/l_0$).

5.5 Discussion

The purpose of the study was to show the effects of the knee joint ligaments at different degrees of femoral component malrotation during a sit down task. These effects were simulated with a knee joint model based on a quantitative motion analysis.

The motion analysis was performed by video-fluoroscopy coupled with force plate data. This measuring system enables accurate kinematics and kinetics acquisition in vivo avoiding skin movement artefacts.

The limitations of this work are that the strain was estimated relative to a defined neutral position of the knee joint at $0^\circ$ flexion angle. The question still remains how lax the ligaments are in this neutral position. Thus, the presented load is to be understood more qualitatively than quantitatively. It is assumed that the kinematics of the implant components remain unchanged while modelling different rotational alignments. A lax LCL during a sit down task may open the joint and change therefore the kinematics of the implant components.
Figure 5.5: Load of the ligaments at different degrees of femoral component malrotation.

The results show that, at neutral alignment of the femoral component, the anterior bundle of the MCL remains taught during flexion whereas the posterior bundle of the MCL relaxes. These effects were also observed by Warren [96] and Horwitz [49]. At femoral component internal rotation of 5° the MCL ant is much more loaded than the MCL post and the LCL is lax during the sit down task. This effect does not increase up to 8° of internal rotation. This constellation might lead to lateral joint opening i.e. lift-off and medial tightness even at femoral component internal rotation of 5°. The lift-off on the lateral side and a tightness on the medial side have been observed by Stiehl et al. [89], Dennis et al. [26], and Insall et al. [50].

At femoral component external rotation the load on the LCL and MCL is small compared to the load on the MCL ant with femoral component internal rotation. Additionally, the difference between the load of the LCL and the MCL ant is small, therefore these structures are balanced. This is one possible explanation as to why, for most patients, a relative
external rotated femoral component can be as high as $8^\circ$ without causing clinical problems, whereas a femoral component internal rotation of $5^\circ$ may already be symptomatic [70], [12], [50], [79]. The PCL produces forces at the very beginning of the sit down task in all evaluated cases.

This work sets a basis for further investigation discussing the alignment of the implant components.

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Philips Medical Systems, Switzerland
Conclusion

The goals of this thesis were to acquire accurate kinematic and kinetic information of TKA during motion patterns of daily activities like level walking and sitting down a chair. A computer based model visualises the knee joint movement and calculated resultant forces and moments. Validation tests were performed to estimate the accuracy of this measuring system. Different degrees of femoral component malrotation were simulated based on the kinematic data of a sit down task.

This computer model was based on anthropometric data of one subject including bones, implant components and the insertion points of the ligaments. The subject had no complications and a femoral component neutral alignment. The input of this model were the subject’s kinematic and kinetic data of two different motion patterns, namely level walking and a sit down task.

In many studies on joint kinematics skin mounted markers were used. The problem with this measurement technique is the large error in kinematic data acquisition caused by the movement of the skin and muscles relative to the underlying bone.

Video-fluoroscopy is a well established method to get kinematic information of artificial joints non invasively by a three-dimensional numeric reconstruction of the single plane projection view in fluoroscopic images. However, this method is limited to the field of
view of the fluoroscopic screen and to kinematic data only. The Fluoroscopic Integrated Motion Acquisition proposed in this thesis, allows the assessment of accurate kinematic information as well as kinetic information during normal level walking. This was achieved by tracking the knee joint during several gait cycles with the fluoroscopic unit and by coupling force plate data with the fluoroscopic system. This system enabled kinematic data acquisition of the implant components within an accuracy of 3.1 mm for the spatial translation and 1.57° for the rotation angle relative to the image focus.

An error estimation showed that when using video-fluoroscopy with force plates, the resultant moment at the knee can be estimated seven times more accurately than with classical instrumented gait analysis using skin mounted markers. The data sets of a sit down task were taken to simulate different degrees of femoral component malrotation, therefore enabling the observation of physiological motion patterns. The knee joint model showed that a femoral component internal rotation of 5° has a more profound effect on the ligament structures than up to 8° external rotation of the femoral component, i.e. the load at the medial collateral ligament at internal rotation is higher than the load at the lateral collateral ligament at external rotation, in which the collateral ligaments are balanced. This might be the reason for lateral joint laxity and medial tightness at the knee joint at 5° internal rotation.

It was assumed that the kinematics of the knee joint do not change while simulating different degrees of malrotation. Thus, higher strain of the ligaments would have an effect on the kinematics and kinetics of the knee joint. A planned outcome study will investigate femoral component malrotation on different implant designs using Fluoroscopic Integrated Motion Acquisition and the three-dimensional knee joint. These results will be compared with the results of the present study.

The resolution of the three-dimensional reconstruction was not high enough to make it worthwhile having the polyethylene inlay modelled as flexible body. In addition, due to the multiple over determined system of a knee joint, only a rough estimate of the load distribution could have been made. The material properties of the ligaments and their cross-sectional areas were taken from literature. Therefore, the force estimation of the ligaments due to femoral component malrotation was a rough estimation only. However, the results of this study correspond to the observations made by clinicians. This knee joint model sets a basis for further investigations on the effects of the alignment of the implant
components. Further steps in knee joint modelling could be made by a simulation of the sit down task by changing the alignment of the implant components with the resultant moment as input. An implemented contact model of the knee joint would simulate the kinematics of the knee joint at different alignments.

To summarise, this work sets a basis for further investigations on TKA mechanics. The fluoroscopic integrated motion acquisition might open a new world in movement analysis on subjects with totally replaced knee joints because more accurate knee joint kinematics can be achieved. The knee joint model sets a basis for further studies on the alignment of the implant components, leading to a better understanding of the functionality of the knee joint. This may help clinicians as well as to engineers to develop new knee joint replacements.
A

Error analysis of resultant moments estimation at the knee joint

A.1 Introduction

Resultant forces, $F_{\text{res}}$ and resultant moments, $M_{\text{res}}$ are not measured directly, they are functions of several measurement values. $F_{\text{res}}$ and $M_{\text{res}}$ are the forces and moments transmitted by the knee joint.

Figure A.1 shows the particular error sources influencing the calculation of resultant forces and moments. More details of these error sources can be found in the following appendices. The pose estimation of the implant components relative to the focus of the fluoroscope was performed. In order to couple kinematic data with the pose reconstruction, the focus was transformed into the global coordinate system with its origin at the centre of the force plate. This transformation was made by tracking the focus of the fluoroscope with VICON. Table A.1 contains the factors and their corresponding values affecting the uncertainty in resultant moments calculation.

Assuming quasi static behaviour and all bodies of the knee joint model to be rigid, the sum of all moments is zero (Equation A.1)
pose estimation of implant components \rightarrow \text{axis of rotation estimation}

\text{spatial marker coordinates reconstruction} \rightarrow \text{focus location in global coordinate system}

\text{estimation of calibration grid’s location} \rightarrow \text{transformation into global coordinate system}

\text{resultant forces and moments calculation}

\text{force vector calculation}

\text{POFA estimation}

Figure A.1: Error sources which affect resultant forces and moments calculations

\[ 0 = \Sigma M_i \] \hspace{1cm} (A.1)

\[
\begin{bmatrix}
M_{\text{res}} \\
0 \\
0
\end{bmatrix} =
\begin{bmatrix}
0 \\
COR_y - POFA_y \\
COR_z
\end{bmatrix}
\times
\begin{bmatrix}
0 \\
F_y \\
F_z
\end{bmatrix} \hspace{1cm} (A.2)

Figure A.2 illustrates the knee joint model reduced into the sagittal plane used for resultant moments calculations. Equation A.2 shows the reduction of the resultant moment in the finite centre of rotation, COR. \( F_{\text{quad}} \) is the force generated by the quadriceps in direction of the femur. \( a \) is the distance between the centre of rotation (assumed to be constant) and \( F_{\text{quad}} \), and \text{COR} is the location vector of the centre of rotation in the global coordinate system. These calculations suppose the activity of the quadriceps muscle only, thus the quadriceps produces the force to control the movement.
Table A.1: Factors affecting resultant moments estimation

<table>
<thead>
<tr>
<th>influence factor</th>
<th>method of estimation</th>
<th>RMSE</th>
<th>reference</th>
</tr>
</thead>
<tbody>
<tr>
<td>axis of rotation estimation (relative)</td>
<td>spatial rotation, $\beta$</td>
<td>1.57°</td>
<td>app D</td>
</tr>
<tr>
<td></td>
<td>translation, $t$</td>
<td>3.25 mm</td>
<td>app D</td>
</tr>
<tr>
<td></td>
<td>translation, in-plane</td>
<td>0.2 mm</td>
<td>app D</td>
</tr>
<tr>
<td></td>
<td>normalised axis of rotation</td>
<td>$u_x = 0.020$</td>
<td>app D</td>
</tr>
<tr>
<td></td>
<td></td>
<td>$u_y = 0.023$</td>
<td></td>
</tr>
<tr>
<td></td>
<td></td>
<td>$u_z = 0.064$</td>
<td></td>
</tr>
<tr>
<td>focus location, global</td>
<td>VICON marker tracking</td>
<td>1 mm</td>
<td>app C</td>
</tr>
<tr>
<td>axis of rotation estimation,</td>
<td></td>
<td>0.2 mm + 1 mm (worst case)</td>
<td>app D</td>
</tr>
<tr>
<td>global</td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>$F_x$</td>
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<td>± 4N</td>
<td>manufacturer</td>
</tr>
<tr>
<td>$F_y$</td>
<td></td>
<td>± 5N</td>
<td>dito</td>
</tr>
<tr>
<td>POFA calculation</td>
<td>KISTLER force plate</td>
<td>3 mm</td>
<td>app B</td>
</tr>
</tbody>
</table>

Figure A.2: Knee joint model

The error of resultant moment calculation was estimated by applying differential calculus. The error of the resultant moment $\Delta M_{res}$ of the knee joint was calculated by using Equation A.3, which is the partial differential of Equation A.2.
\[ \Delta M_{\text{res}} = \left[ (F_z \Delta POF A_y)^2 + (F_z \Delta COR_y)^2 + (-F_y \Delta COR_z)^2 \right. \\
\left. + ((COR_y - POF A_y) \Delta F_z)^2 + (-COR_z \Delta F_y)^2 \right]^{\frac{1}{2}} \]  

(A.3)

A.2 Results

![Graph showing sensitivity of \( \Delta M_{\text{res}} \) by varying \( F_y \) and \( F_z \). Blue: \( \Delta COR = 0.2 \) mm; red: \( \Delta COR = 1.45 \) mm; black: \( \Delta COR = 2 \) mm.]

Assuming \( F_y = 50 \) N, \( F_z = 800 \) N, \( COR_y - POF A_y = 50 \) mm, and \( COR_z = 450 \) mm, and taking the values of the uncertainties as listed in Table A.1, the sensitivity of \( \Delta M_{\text{res}} \) is as shown in Figures A.3 and A.4. In Figure A.3, \( F_y \) and \( F_z \) vary from 1 to 800 N while the other parameters are constant. In Figure A.4, \( COR_y \) and \( POF A_y \) vary from 1 to 200 mm. The blue lines are the plots where \( \Delta COR = 0.2 \) mm, as found in our validation, red represents \( \Delta COR = 1.5 \) mm, and black represents \( \Delta COR = 2 \) mm. \( M_{\text{res}} \) varies from 2.26 Nm to 3.26 Nm (blue), 2.26 Nm to 2.95 Nm (red), and 3.44 Nm to 3.93 Nm (black) in Figure A.3, and from 3.29 Nm to 3.38 Nm (blue), 3.50 Nm to 3.58 Nm (red), and 3.66 Nm to 3.74 Nm (black) in Figure A.4. For small values of \( F_y \) and \( F_z \) the uncertainty of \( M_{\text{res}} \)
Figure A.4: Sensitivity of $\Delta M_{\text{res}}$ by varying CORy and POFAy. blue: $\Delta COR = 0.2$ mm; red: $\Delta COR = 1.45$ mm; black: $\Delta COR = 2$ mm

remains. Higher $F_z$ do affect $M_{\text{res}}$ more profoundly than higher $F_y$ values. With varying $POFA$ and $COR$, i.e. the lever arm of the moment, the uncertainty in $COR$ estimation has an increasing effect on the resultant moment.

A.3 Discussion and Conclusion

The uncertainty of the spatial axis of rotation location in the global coordinate system has its source of errors in the reconstruction of the implant component’s pose estimation. As described in Appendix D, the RMSE of the relative rotation angle between the femoral and tibial component and an estimation of the RMSE of the in-plane translation are $1.57^\circ$ and 0.2 mm respectively. The factors associated with the axis of rotation estimation in the global coordinate system are: i) the uncertainty of the rotation angle, ii) the uncertainty of the in-plane translation due to the almost parallel movement of the knee joint relative to the fluoroscopic image intensifier, and iii) the uncertainty in the focus location in the global coordinate system. The uncertainty in resultant forces calculation is dependent on the uncertainty in the location vector of COR.

Taking the uncertainty of the resultant moment of about 3 Nm and a resultant moment
A. SENSITIVITY OF RESULTANT MOMENTS

at the knee joint of about 40 Nm, which occurs during e.g. a sit down task, the error of the resultant moment is about 7.5%.
Estimation of the force vector and the point of force application

The force plate used for this thesis was a 400 mm × 600 mm multicomponent force plate based on piezo-electric sensors (KISTLER AG, Switzerland). The force plate consists of a top plate mounted on four three-component force sensors, illustrated in Figure B.1. The force and torque applied to the top plate is distributed among the four sensors and separated into three orthogonal force components. The load cells produce signals in a total of 8 channels. The three components of the force vector are given in Equations B.1, B.2, and B.3.

\[
F_x = f_{x12} + f_{x34} \quad \text{(B.1)}
\]

\[
F_y = f_{y14} + f_{y23} \quad \text{(B.2)}
\]

\[
F_z = f_{z1} + f_{z2} + f_{z3} + f_{z4} \quad \text{(B.3)}
\]

where the signal \(f_{x12}\) is the sum of the forces in x direction of the load cells 1 and 2, \(f_{x34}\) is the sum of the forces in x direction of the load cells 3 and 4 and so on. The coordinates of the point of force application, POFA, \(a_x\) and \(a_y\), are calculated as follows:
where $M_x$ and $M_y$ are the components of the resultant moment reduced into the coordinate’s origin in the centre of the force plate.

\[
M_x = b (F_{z1} + F_{z2} - F_{z3} - F_{z4}) \quad \text{(B.6)}
\]

\[
M_y = a (-F_{z1} + F_{z2} + F_{z3} - F_{z4}) \quad \text{(B.7)}
\]
The uncertainty given by the manufacturer is 3 mm for the POFA and ± 5 Nm for $F_x$ and $F_y$ and ± 4 Nm for $F_z$.

### B.0.1 Method

An experimental error estimation was performed to analyse the error of the force vector’s direction and the POFA. The experiment was designed according to the load produced by the subject’s motion pattern analysed in this work (level walking, sit-down task). Therefore force values with a relation of $F_x/F_z$, $F_y/F_z$ respectively of $< 0.2$, and values of $F_z > 300\, \text{N}$ were used for these calculations.

![Figure B.2: Cylinder with fixed reflective markers pointing on defined locations at the force plate](image)

Eight reflective marker points were fixed on a cylinder with a diameter of 60 mm and a height of 1500 mm. A metallic cone point was fixed at the end of the cylinder (Figure B.2). A force of about 300 N was applied on defined points on the force plate (see Table B.1) avoiding shear forces. The marker points were optically tracked by VICON during the force application. The cylinder’s direction was found by an optimisation algorithm (non

<table>
<thead>
<tr>
<th>point number</th>
<th>coordinates [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>one</td>
<td>[-60,40]</td>
</tr>
<tr>
<td>two</td>
<td>[60,40]</td>
</tr>
<tr>
<td>three</td>
<td>[0, 0]</td>
</tr>
<tr>
<td>four</td>
<td>[-60, -40]</td>
</tr>
<tr>
<td>five</td>
<td>[-60,40]</td>
</tr>
</tbody>
</table>

Table B.1: POFA coordinates
linear least square solver, MATLAB) which fitted the same distance between the centre of the cylinder to the markers. After a rough estimate of the cylinder’s direction and POFA, the optimisation algorithm solved the non linear least square problem by minimising the function $F$. $F$ is the difference between the cylinder’s radius and the calculated distance between the centre of the cylinder to the markers (see Equation B.8), where $v$ is the cylinder’s direction, $p_i$ the coordinates of the marker point $i$, $q$ the coordinates of the contact point, and $d$ the radius of the cylinder. The estimated vector $v$ was taken for further calculations if the solver terminated successfully within a value of $r \leq 2$ mm, where $r$ is the squared 2nd norm of the residual. The vectors calculated by the force plate data were compared with the optically tracked direction of the cylinder and the contact point.

\[
F = |v_i \times (p_i - q)| - d; \quad i = 1, ..., 8 \tag{B.8}
\]

### B.1 Results

The error of the POFA coordinates is listed in Table B.1. The standard deviation and root mean square error, RMSE, were calculated over 150 measured frames (corresponds to 1.5 s measuring time). The mean of the RMSE of all force application points are 2.86 mm for the x coordinates, 3.08 mm for the y coordinates.

Table B.3 lists the difference between the calculated force vector measured by the force plate and the estimated cylinder direction. The standard deviation $\sigma$ of the force vector’s difference is about $2^\circ$ (estimate of the error).
Table B.3: Mean and standard deviation of the difference between force vector calculated by the force plate and the estimated direction of the cylinder.

B.2 Conclusion

The results above show that when using the KISTLER force plate in a manner comparable to these experiments, the expected force vector’s direction error is about 2°, and the error in POFA calculation is about 3 mm respectively. These results correspond to the errors given by the manufacturer. Therefore these errors were taken for further calculations.
Error estimate of the fluoroscope’s focus location

C.1 Method

The transformation of the fluoroscopic coordinate system into the global coordinate system was performed by a calibration grid. This grid was fixed on one force plate in a frame, where the position of the grid’s centre relative to the global coordinate’s origin was well defined. This grid was filmed by the fluoroscope, whereby the position of the fluoroscope was optically tracked (VICON) by three markers fixed on the C-arm. After distortion correction of this fluoroscopic image, a MATLAB routine calculated the focus location in the global coordinate system by comparing the location of the grid’s centre and the distance of the grid lines of the fluoroscopic image with the real location and size.

Four factors affect the uncertainty of the location estimation of the fluoroscope’s focus. These are:

i) The fixation of the calibration grid relative to the force plate.
ii) The spatial coordinate estimation of the three markers of the fluoroscopic system.
iii) The uncertainty of the estimation of the focus in the fluoroscopic image.
iv) The stiffness of the C-arm.

These sources of errors are discussed below.
C. ERROR OF FLUOROSCOPE’S FOCUS LOCATION

i) Fixation of the calibration grid: The calibration grid’s exact position was measured five times. The standard deviation was calculated.

ii) Spatial coordinate estimation by VICON™: To estimate the error of the marker’s spatial coordinates by VICON the distances between marker 1 and marker 2, and marker 1 and marker 3 respectively were used. The standard deviation of the calculated distances between the markers of 300 measured frames was calculated. The distances between the markers were assumed to remain the same in all measured frames.

iii) Focus location in the fluoroscopic image: The algorithm to calculate the focus in the global coordinate system was based on an interface, where the user had to click onto the calibration grid lines in the fluoroscopic image. This clicking was performed 20 times on the same image to estimate the uncertainty in the focus calculation.

iv) Stiffness of the C-arm: In order to study the stiffness of the rigid connection x-ray source to C-arm, metal discs with a diameter of about 5 mm were glued on the bottom of a rigid plastic box. This box was fixed on the image intensifier, so that the discs were held in a distance of 300 mm parallel to the image intensifier. 75 fluoroscopic images were taken while the unit mover was accelerating and decelerating. The positions of the discs were compared in every taken image using image subtraction (gray scale).

C.2 Results

i) Fixation of the calibration grid: The uncertainty in the estimation of the calibration grid’s location in the global coordinate system was found to be 0.1 mm.

ii) Spatial coordinate estimation by VICON™: The results are shown in Table C.1. The factor $\sqrt{\frac{1}{2}}$ was included due to the addition of the standard deviations of the two markers based on the assumption that the noise in the coordinates of one marker is independent of the noise in the coordinates of the other marker. This error estimation technique does not include an absolute shift of the marker positions relative to the origin of the coordinate system.

iii) Focus location in the fluoroscopic image: The standard deviation of the focus location in the global coordinate system was found to be 0.05 mm.

iv) Stiffness of the C-arm: No difference could be found by subtracting the subsequent images. Therefore, the C-arm can be assumed to be rigid enough for our application.
<table>
<thead>
<tr>
<th></th>
<th>standard deviation $\sigma/\sqrt{2} \text{ [mm]}$</th>
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<tbody>
<tr>
<td>marker 1 to marker 2</td>
<td>0.52</td>
</tr>
<tr>
<td>marker 1 to marker 3</td>
<td>1.40</td>
</tr>
<tr>
<td>mean</td>
<td>0.96</td>
</tr>
</tbody>
</table>

Table C.1: Standard deviation of marker’s distance over 300 frames

C.3 Conclusion

The relevant error of the estimation of the focus in the global coordinate system comes from the measuring system VICON. Assuming a negligible systematic error the uncertainty focus location in the global coordinate system is 1 mm. This quantity corresponds to the error given by the manufacturer. This estimate is valid to a space of 2000 mm $\times$ 4000 mm $\times$ 2000 mm. The other sources of errors were small enough to be neglected.
Error estimation of the three dimensional reconstruction of the implant components

D.1 Method

The factors influencing the accuracy of the implant components pose reconstruction are motion blur and a reduction of the signal to noise ratio due to bone overlay. In order to estimate the accuracy of the implant’s pose reconstruction relative to the image focus, static and dynamic validation tests were performed.

The three implant components, tibial part, femoral part and polyethylene inlay, were glued to each other allowing no relative motion. The exact position and orientation of the tibial component relative to the femoral component were measured by a three dimensional optical digitiser (Breuckmann OPTO Top HE-200, Germany), which measured surface points within an accuracy of ±0.01 mm. 245 000 points were registered. Three planes were placed on well defined surfaces (planes and not collinear) of the CAD geometries of each implant component (Raindrop Geomagic Qualify 5.0). The same planes were fitted into the point clouds of the surface scan. The software laid the corresponding planes of the CAD geometry into the planes of the point cloud, within a translational deviation of 0.034 mm and a rotational deviation of 0.54°. The deviation is an average of the distance of the points in the point cloud that correspond to the surface that they represent. The degrees
deviation is based on an axis, which is generated by an approximation of the points that
correspond to the CAD surface. The relative position and orientation of the components’
coordinate system’s origin were measured; 42.6 mm for the distance \( t \) (norm of \( t \)) of the
tibial coordinate system’s origin to the femoral coordinate system’s origin, and 104.5° for
the relative rotation angle \( \beta \).
The three dimensional reconstruction algorithm estimated the position vector \( t = t_{x,y,z} \)
and the rotation angles \( x_{rot}, y_{rot}, \) and \( z_{rot} \) of the tibial and femoral component relative
to the image focus. The position and orientation of the femoral part relative to the tibial
part were calculated in each fluoroscopic image. The rotation was represented by the
screw vector, \( u_{x,y,z} \) and the corresponding angle \( \beta \) of the relative rotation matrix \( R_{rel} \)
(Equation D.1). The rotation angles \( x_{rot}, y_{rot}, \) and \( z_{rot} \) were inserted in the standard
definition of the rotation matrices \( R_{x_{rot}}, R_{y_{rot}}, \) and \( R_{z_{rot}} \). The orientation of the implant
components was calculated by multiplying the rotation matrices in the order \( R_{z_{rot}}, R_{x_{rot}}, \)
\( R_{y_{rot}} \). The matrix \( R_{fem} \) is the orientation of the femoral component, \( R_{tib} \) the orientation
of the tibial component. The relationship between rotation matrix and the screw vector
and the corresponding angle is given in Equation D.3 (Equation of Rodrigues), where \( u \circ u \) is the dyadic product, \( I \) the identity matrix, and \( u \times \) is the skew symmetrical vector
product of \( u \). The components of \( R_{rel} \) as function of \( x_{rot}, y_{rot}, \) and \( z_{rot} \) (Equation D.2)
are compared with the entries of \( R_{rel} \) as function of \( u \) and \( \beta \) (Equation D.3).

\[
R = R_{fem} * R_{tib}^T
\]  
\[
R = \begin{bmatrix}
\cos(x_{rot}) \cos(y_{rot}) \\
- \sin(x_{rot}) \cos(z_{rot}) + \cos(x_{rot}) \sin(y_{rot}) \sin(z_{rot}) \\
\sin(x_{rot}) \sin(z_{rot}) + \cos(x_{rot}) \sin(y_{rot}) \cos(z_{rot})
\end{bmatrix}
\]  
\[
\sin(x_{rot}) \cos(y_{rot}) \\
\cos(x_{rot}) \cos(z_{rot}) + \sin(x_{rot}) \sin(y_{rot}) \sin(z_{rot})
\]
\[
\cos(y_{rot}) \sin(z_{rot}) \\
- \cos(x_{rot}) \sin(z_{rot}) + \sin(x_{rot}) \sin(y_{rot}) \cos(z_{rot})
\]
\[
\cos(y_{rot}) \cos(z_{rot})
\]
\[
R = \cos \beta \mathbf{I} + (1 - \cos \beta) \mathbf{u} \circ \mathbf{u} + \sin \beta \mathbf{u} \times \mathbf{u} \tag{D.3}
\]

Repeatability of the three dimensional reconstruction of gait cycle images:
The reconstruction algorithm was tested with regard to the repeatability in three images. These images were selected to represent different types during a full gait cycle: Motion blur (mean relative velocity about 0.1 m/s), bone overlay and motion blur (mean relative velocity about 0.15 m/s), and no motion (mean relative velocity nearly 0 m/s). The velocity in-plane relative to the image focus was estimated by comparing the position of the subsequent images. Unfortunately there was no image of the gait cycle found, where no motion relative to the focus would occur.
The algorithm was run 15 times in each of the four images. Each time the initial estimate was newly defined by the user. The standard deviations of \( t_{x,y,z} \) and \( x\rot, y\rot, \) and \( z\rot \) of the two implant components were calculated using the resulting parameter sets. The body’s own coordinate systems are shown in Figure D.1.

\begin{figure}
\centering
\includegraphics[width=\textwidth]{figure_d_1}
\caption{Body fixed coordinate systems of the tibial and the femoral component.}
\end{figure}

Static in vitro validation: Eight x-ray images from different views, where the distance
between the object and the image intensifier was about 100 mm, were taken. These views were chosen to represent real orientation of the implant components relative to the image intensifier as they occur during level walking or standing up from a chair. Four images (image v - viii) were taken, where a bag filled with semolina (thickness of about 50 mm) and a dry tibia were placed between object and x-ray source simulating the influence of the collateral leg in the fluoroscopic images. After distortion correction and implant’s pose estimation by the reconstruction algorithm the relative rotation and translation between the femoral and the tibial component were calculated.

![Figure D.2: In vitro static validation.](image)

**Dynamic in vitro validation:** One image series (8 ms exposure time, 5 images/s) of 17 images were taken while moving the implant components about 100 mm parallel to the image intensifier with a velocity of about 0.3 m/s. Half of the images were taken while the dry tibia was held between the implant components and the x-ray source, thus simulating the collateral leg during a gait cycle.

**D.2 Results**

**D.2.1 Repeatability**

The standard deviations of the three images (see Figure D.3) are listed in Table D.1. The three dimensional reconstruction of the implant components of the image 15 shows the best results. The results of image 5 can be compared with the results of image 1, except for the out-of-plane translation. The standard deviations of the in-plane translations, femoral and tibial component, \((t_x, t_y)\) in image 15 are less than 0.2 mm. The out-of-plane translation
Figure D.3: Fluoroscopic images used for the repeatability study. Image 1: motion; Image 5: bone overlay; Image 15: no motion

$(t_z)$ is 1.47 mm for the femoral part, and 2.29 mm for the tibial part respectively. The standard deviations of the rotation angles range from $0.09^\circ$ for $x_{rot}$ (no motion) of the tibial component to $1.75^\circ$ for $z_{rot}$ (bone overlay) of the femoral part. The standard deviations of the rotations depend on the geometry of the implant component. The best results are given by the tibial component’s rotation around the axis $z_{rot}$. The worst results are given by the femoral component’s rotation around the axis $z_{rot}$ (see Figure D.1). Table D.2 lists the ratio between the standard deviation of the out-of-plane translation to the in-plane translations. The mean value of these ratios are 12.03 for $\sigma_{t_z}/\sigma_{t_x}$, and $\sigma_{t_z}/\sigma_{t_y}$ 10.08 for the femoral part, 21.88, and 19.84 for the tibial part respectively. The mean of these values is 16.

<table>
<thead>
<tr>
<th></th>
<th>image type</th>
<th>$\sigma_{tx}$</th>
<th>$\sigma_{ty}$</th>
<th>$\sigma_{tz}$</th>
<th>$\sigma_{xrot}$</th>
<th>$\sigma_{yrot}$</th>
<th>$\sigma_{zrot}$</th>
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<tbody>
<tr>
<td>femoral part</td>
<td>motion (img 1)</td>
<td>0.57</td>
<td>0.72</td>
<td>4.71</td>
<td>0.38</td>
<td>1.00</td>
<td>1.21</td>
</tr>
<tr>
<td></td>
<td>bone overlay (img 5)</td>
<td>0.70</td>
<td>0.56</td>
<td>8.98</td>
<td>0.63</td>
<td>0.45</td>
<td>1.75</td>
</tr>
<tr>
<td></td>
<td>no motion (img 15)</td>
<td>0.07</td>
<td>0.16</td>
<td>1.47</td>
<td>0.09</td>
<td>0.23</td>
<td>0.34</td>
</tr>
<tr>
<td>tibial part</td>
<td>motion (img 1)</td>
<td>0.66</td>
<td>0.37</td>
<td>5.64</td>
<td>0.10</td>
<td>1.16</td>
<td>0.37</td>
</tr>
<tr>
<td></td>
<td>bone overlay (img 5)</td>
<td>0.41</td>
<td>0.35</td>
<td>5.94</td>
<td>0.81</td>
<td>0.30</td>
<td>0.33</td>
</tr>
<tr>
<td></td>
<td>no motion (img 15)</td>
<td>0.11</td>
<td>0.12</td>
<td>2.29</td>
<td>0.77</td>
<td>0.24</td>
<td>0.13</td>
</tr>
</tbody>
</table>

Table D.1: Repeatability. The standard deviations $\sigma_{tx,y,z}$ in [mm], $\sigma_{xrot}$, $\sigma_{yrot}$, $\sigma_{zrot}$ in [$^\circ$] of the estimated pose parameters.

D.2.2 Accuracy of three dimensional reconstruction

The three components of the translation vector of the tibial coordinate system’s origin to the femoral system’s origin, $t_x$, $t_y$, $t_z$, the norm of $t$, the components of the screw vector $u$
Table D.2: Ratio of the standard deviations of the out-of-plane to the in-plane translations of the repeatability study

<table>
<thead>
<tr>
<th>image type</th>
<th>ratio $\sigma_{t_z}/\sigma_{t_x}$</th>
<th>ratio $\sigma_{t_z}/\sigma_{t_y}$</th>
</tr>
</thead>
<tbody>
<tr>
<td>femoral part</td>
<td>motion (img 1)</td>
<td>8.26</td>
</tr>
<tr>
<td></td>
<td>bone overlay (img 5)</td>
<td>12.83</td>
</tr>
<tr>
<td></td>
<td>no motion (img 15)</td>
<td>21.00</td>
</tr>
<tr>
<td>mean</td>
<td></td>
<td>12.03</td>
</tr>
<tr>
<td>tibial part</td>
<td>motion (img 1)</td>
<td>8.55</td>
</tr>
<tr>
<td></td>
<td>bone overlay (img 5)</td>
<td>14.49</td>
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<tr>
<td></td>
<td>no motion (img 15)</td>
<td>20.82</td>
</tr>
<tr>
<td>mean</td>
<td></td>
<td>21.88</td>
</tr>
</tbody>
</table>

and the corresponding angle $\beta$ relative to the tibial coordinate system are listed in Table D.3 and Table D.4. The standard deviation as well as the RMSE of the listed values were calculated.

The static accuracy test (see Table D.3) shows a translational RMSE for the series I of 2.03 mm and a RMSE for the series II of 2.69 mm, the rotational errors are 1.77°, and 0.75°. No significant difference could be found ($\chi^2_{emp\text{-}trans} = 1.02 < \chi^2_{3.5\%} = 7.81$; $\chi^2_{emp\text{-}angle} = 0.10 < \chi^2_{3.5\%} = 7.81$) between series I and series II. Therefore the mean values of the RMSE are taken, 2.36 mm and 1.21°, respectively.
Table D.3: Static validation results; Series I without bone overlay and Series II with bone overlay.

<table>
<thead>
<tr>
<th>series I</th>
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<td>6.18</td>
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<td>42.07</td>
<td>-0.503</td>
<td>0.729</td>
<td>0.464</td>
</tr>
<tr>
<td>image ii</td>
<td>13.22</td>
<td>-9.12</td>
<td>37.77</td>
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<td>0.733</td>
<td>0.274</td>
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<tr>
<td>image iii</td>
<td>9.40</td>
<td>-9.55</td>
<td>41.97</td>
<td>44.06</td>
<td>-0.430</td>
<td>0.735</td>
<td>0.524</td>
</tr>
<tr>
<td>image iv</td>
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<td>-3.29</td>
<td>38.65</td>
<td>39.18</td>
<td>-0.504</td>
<td>0.821</td>
<td>0.269</td>
</tr>
<tr>
<td>standard deviation</td>
<td>3.53</td>
<td>3.25</td>
<td>1.85</td>
<td>2.03</td>
<td>0.080</td>
<td>0.044</td>
<td>0.13</td>
</tr>
<tr>
<td>mean</td>
<td>8.58</td>
<td>-8.10</td>
<td>39.67</td>
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Table D.4 lists the results of the translational and the rotational reconstruction relative to the tibial coordinate system. The components, $t_x^0$, $t_y^0$, $t_z^0$, of the distance vector between the tibial and the femoral component of the global coordinate systems are also listed. The RMSE of the dynamic validation test (Table D.4) is 3.25 mm for the translational error and 1.57° for the rotational error. The standard deviations of the three components of the translation vectors are 1.27 mm, 4.81 mm and 4.41 mm relative to the tibial component. The standard deviations of the normalised screw vector components are 0.020, 0.023 and 0.064. The standard deviations of the distance between the implant components and the rotation angle are 1.08 mm and 0.62° respectively.

The $t_y$ values, 11.96 mm and 5.86 mm, of image 4 and image 6 can be characterised as outliers. It is shown that these bad results are caused by the error in $t_z^0$ reconstruction, where the values are 7.76 mm and 2.11 mm which do not correspond to the values of the subsequent images.

A rough estimate of the error in-plane to out-of-plane is given by the ratio of the standard deviations of the in-plane/out-of-plane of the in vivo repeatability study. Hence, the in-plane error is about 0.2 mm.
D. ERROR OF 3D RECONSTRUCTION

Table D.4: Dynamic validation results, standard deviation, mean value and RMSE, the translations $t_x$, $t_y$, $t_z$, $t^0_x$, $t^0_y$, $t^0_z$, $t$ in [mm], and the rotation angle $\beta$ in [$^\circ$]

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The repeatability test could not distinguish between the effect of motion blur and the effect of noise due to bone overlay. Thus, it is shown that motion in the fluoroscopic image does affect the out-of-plane translational error profoundly.

The translational RMSE of the static validation tests is 2.36 mm, the rotational RMSE is 1.21 $^\circ$. The results of the dynamic validation test are 3.25 mm, and 1.57 $^\circ$ respectively. The difference of the standard deviation and the RMSE of the dynamic validation of 2.17 mm and 0.95 $^\circ$ show a tendency negatively affect the results due to motion blur. This tendency could not be shown in the static validation tests. Hence, motion blur in the fluoroscopic image does affect the accuracy of the pose reconstruction. Bone overlay does not significantly influence the accuracy, but it has an effect on the standard deviation of the out-of-plane translation. The standard deviations of the in-plane reconstruction is
about 16 times less than the standard deviations of the out-of-plane reconstruction due to the ratio of the distance x-ray source to the image intensifier and the pixel resolution. Taking the RMSE of the translation vector of 3.25 mm and a ratio of the in-plane/out-of-plane standard deviation of 16, a rough estimate of the in-plane error of the dynamic in vitro validation is about 0.2 mm.

The standard deviations of the rotation angle range from 0.09° to 1.75°. The RMSE of the spatial rotation angle of the dynamic validation test is 1.57° and lies in the range of the standard deviations.

Noticeable is the difference between the RMSE and the standard deviation of the dynamic validation test. There is an offset of about 2.17 mm. The distance between the origin of the coordinate systems of the implant components is smaller than the measured distance. This offset in one direction cannot be seen in the results of the static validation test. It is assumed that this bias is caused by the error of the out-of-plane reconstruction, which might be dependent on the implant geometry. The implant components were not rotated relative to the image plane while they were moved parallel to the image intensifier. Due to the choice of the origin of the coordinate system of the components (see Figure D.1) the distance vector between the origins did not lie parallel to the image plane, therefore the out-of-plane error bias the distance. The error of a parallel distance vector would only be dependent on the in-plane reconstruction error. The translational error could be much smaller if the out-of-plane error due to motion blur would be reduced. Further investigation will be done to study motion blur effects on the implant geometry to improve the three dimensional reconstruction.
Bibliography


Curriculum Vitae

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Studies

2002 - 2005 PhD Thesis project at Laboratory for Biomechanics, Swiss Federal Institute of Technology Zurich, ETH
1997-2002 Master in Mechanical Engineering at ETH Zurich, Switzerland; Majors: Biomedical Engineering and Product Development
1993-1997 Kantonsschule Baden, Switzerland, Matura in ancient languages (Latin)

Practical work

2000-2001 ETH juniors, the junior enterprise of the ETH, Zurich
2000/2001 Part time math teacher at Kantonsschule Baden and Forum 44, private school, Baden, Switzerland
1999 Assistant at Institute of Mechanical Systems, ETH Zurich, Switzerland
1999 Practical work at Zihlmann Systembau AG, Würenlos, Switzerland; Product Engineering
1998 Internship at alpha real ag, Zurich in alternative energy technologies
1997 Basic workshop course at Automation Tooling Systems, ATS, Cambridge, Canada

Grants

Dr. h.c. Robert-Mathys-Foundation, Bettlach, Switzerland; PhD project

Others

Member of ETH seniors (alumni club of ETH juniors)
Tutor assistant of diploma thesis, assistant lecturer at ETH Zurich
Student representative of the department committee at the Department of Mechanical and Process Engineering of ETH Zurich (2000/01)
Student representative of the work group of Bachelor/Master Program at ETH Zurich (2002)

Computer skills

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Hobbies

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