JOINT KINEMATICS OF UNCONSTRAINED ANKLE ARTHROPLASTIES

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Abstract

Early designs of total ankle arthroplasties (TAA) in the 1970s showed a high failure rate. Based on the more promising mid- and long-term outcome results of the 2nd generation TAA, as well as due to the association of a higher risk of osteoarthritis in adjacent joints with arthrodesis, TAA has obtained more and more acceptance and the number of ankle replacements is increasing. One important design criterion of TAA is to recover normal anatomical function to master motion tasks of daily activities accompanied by less need for compensation and thereby a protection of adjacent joints. Thus, a crucial part of the evaluation of the functionality of a TAA is the in vivo analysis of its kinematic behavior. Previous investigations on TAA kinematics include cadaver studies and gait analyses by means of skin marker tracking. Cadaver studies do not simulate in vivo conditions accurately. Skin marker tracking is limited by skin movement artefacts and the difficulty to distinguish between motion at the TAA or at the subtalar joints caused by the inaccessibility of the talus. Recent technological developments, such as videofluoroscopy, enable the in vivo measurements of the three-dimensional (3D) kinematics of implant components more accurately than by means of skin marker tracking. However, up to now videofluoroscopy was only rarely used at the ankle and either limited by a small number of analyzed frames or by a restriction to static assessments.

The overall goal of this PhD thesis was to develop a technical procedure that allows gaining a better understanding of the joint mechanisms of unconstrained TAA during daily activities. Specifically, to evaluate the functionality of an unconstrained TAA (Mobility™ Total Ankle, DePuy) on the basis of the relative motion between the implant components.

Two videofluoroscopic in vivo procedures, one for TAA subjects and one for healthy normal subjects with tantalum markers, were developed and applied to assess the 3D ankle kinematics during five gait conditions (level gait, walking up- and downhill, walking over a side inclined slope, once inclined laterally once medially). Four good outcome TAA subjects having a Mobility™ Total Ankle and two healthy normal subjects with bone embedded tantalum markers were assessed during the stance phase of the five gait conditions.

The internal tracking of the TAA was achieved by videofluoroscopic image capture and a registration algorithm that fits a virtual projection of the CAD models of the TAA into the fluoroscopic images. The respective output was the 3D pose of the implant components with an accuracy of 0.4 mm and 0.2° in plane and 2.1 mm and 1.3° out of plane. Thus, a procedure was developed that enabled to accurately estimate TAA kinematics in vivo, without being limited by skin movement artefacts. Hereby isolating TAA motion from subtalar motion, which is not possible
with skin marker analysis.

The videofluoroscopic approach for the healthy normal subjects allowed the quantification of the 3D tibiotalar and talocalcaneal joint kinematics. To enable a comparison to the TAA subjects, the same measurement set up and conventions as in the approach for the TAA subjects were used. The internal tracking of the bones was accomplished by videofluoroscopic image capture of the bone embedded tantalum markers. The 3D reconstruction was computed as a three point spatial resection on the base of the geometrical tantalum marker arrangements determined by CT segmentation. Due to large out of plane translational inaccuracies, the out of plane motion path had to be restricted. An error analysis revealed the high influence of the tantalum image marker tracking precision and a smaller but not negligible effect of the CT segmentation precision.

Simultaneously to the videofluoroscopic image capturing, joint kinematics was assessed by means of skin marker tracking using a 12 camera VICON system. A whole body marker set including a two-segment foot model and functional approaches for the determination of joint centers, respectively axes was applied. This allowed a comparison between the external and the internal measurement approaches and made compensation mechanisms in the foot and the knee visible. The major motion of the TAA arose during all five gait conditions and for all four subjects around the talar construction axis. This is favorable for the wear characteristics of the implant, but it remains to be seen if it allows a physiological role of the surrounding ligaments. The TAA subjects showed, during level gait, comparable dorsi-/plantarflexion motion characteristics, as well only showed minor limitations in range of motion (ROM), compared to the healthy ankle. Restrictions were mainly seen during walking uphill, caused by a dorsiflexion restriction. The static available sagittal ROM showed generally a shift compared to the ROM that was functionally used during gait. Summarized, any restrictions were caused by a limitation in dorsiflexion, whereas plantarflexion was sufficiently provided. Contrary to expectations concerning the unconstrained design, hardly any motion occurred in the transverse and frontal planes. If this results in an unfavorable strain behavior of surrounding ligaments should be addressed in future investigations. If prevalent restrictions were due to a changed muscular activation pattern, caused by scarred connective tissue or by the implant design itself is unclear. However, since during level gait only minor limitations were found in predominant sagittal plane joint kinematics of the TAA one can conclude that compared to arthrodesis there was less need for compensation in adjacent joints. Summarized, the TAA allowed close to, but not in all parameters physiological motion.
The developed procedures are suitable for the assessment of the in vivo kinematics of the TAA and the healthy ankle during gait. By the videofluoroscopic tracking and the reconstruction methods, the 3D kinematics of the isolated tibiotalar joint is accessible without skin movement artefacts.

This work allowed a first insight into the joint kinematics and in vivo performance of the Mobility™ Total Ankle. With the application of the developed procedure on a larger number of subjects it has the potential to help clinicians and implant developers to evaluate current designs and future design modifications.
Zusammenfassung


Es wurden zwei in vivo Verfahren entwickelt und angewendet, wobei mittels Videofluoroskopie die 3D Kinematik bei Patienten mit einer Sprunggelenksarthroplastik, sowie bei gesunden Probanden mit Tantalummarkern während fünf verschiedenen Gangbedingungen (ebenes Gehen, auf- und abwärts Gehen, Gehen über eine seitengeneigte Ebene, einmal medial abfallend, einmal la-
teral abfallend) gemessen wurde. Hierbei wurden die Gangdaten von vier Patienten mit einer Sprunggelenksarthroplastik sowie von zwei gesunden Probanden, die über in den Knochen integrierte Tanatalummarker verfügten, während der Standphase der fünf obig genannten Gangbedingungen erfasst.

Die interne Messung der Bewegung der Sprunggelenksarthroplastik erfolgte mittels videofluoroskopischer Bildverfolgung und anschliessender 3D Rekonstruktion. Diese wurde mit Hilfe eines Registrierungsalgorithmus berechnet, der eine virtuelle Projektion der CAD Geometrie der Sprunggelenksarthroplastik in die Fluoroskopiebilder einpasste. Hierbei konnte die 3D Position und Orientierung der Implantatkomponenten mit einer Genauigkeit von 0.4 mm und 0.2° in der Bildebene und 2.1 mm und 1.3° senkrecht zur Bildebene ermittelt werden. Es wurde somit ein Verfahren entwickelt, welches es ermöglicht, die 3D Kinematik von Sprunggelenksarthroplastiken in vivo mit hoher Genauigkeit und frei von Hautbewegungsartefakten zu bestimmen. Des Weiteren konnte mit diesem Ansatz die Bewegung der Arthroplastik isoliert, d.h. unabhängig von subtalarer Bewegung, betrachtet werden, was mit Hautmarkermessungen ebenfalls nicht möglich ist.


Simultan zur videofluoroskopischen Bildverfolgung wurde die Gelenkskinematik mittels Hautmarkern und einem 12 Kamera VICON System ermittelt. Die Hautmarkermessung erfolgte mit Hilfe eines Ganzkörpermarkersets, welches ein 2-segmentiges Fussmodell sowie funktionelle Ansätze zur Bestimmung der Gelenkszentren und -achsen beinhaltete. Dies ermöglichte einen Vergleich zwischen extern und intern gewonnenen kinematischen Daten sowie das Sichtbarma-
chen von Kompensationsmechanismen in Fuss und Knie.


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1 Introduction

Ankle osteoarthritis causes severe pain and functional limitations, reduces daily life mobility and quality of life. Arthrodesis or total ankle arthroplasties (TAA) are the two main surgical treatments for osteoarthritis. Arthrodesis provides pain relief, restores joint stability and allows a realigning of deformities (Fragomen et al., 2008; Haddad et al., 2007), whereas TAA gives the theoretical benefit of preservation of movement. The history of TAA goes back to 1970 (Lord and Marotte, 1973), followed by the development of many different implant designs of the mainly cemented two-component first generation TAA (Bolton-Maggs et al., 1985; Hamblen, 1985; Kempson et al., 1975; Kofoed, 1995; Newton, 1979, 1982; Scholz, 1976). Although early results were promising, long-term results demonstrated high failure rates and complications (Bolton-Maggs et al., 1985; Hamblen, 1985; Kitaoka et al., 1994; Kitaoka and Patzer, 1996; Newton, 1982). Over many years arthrodesis continued to be the standard operative procedure for ankle osteoarthritis. However, with the advancement in prosthetic design and the encouraging short- and midterm results of the second generation uncemented three-components TAA (Buechel et al., 1988; Hintermann, 1999; Rippstein, 2002; San Giovanni et al., 2006; Schenk et al., 2004; Stengel et al., 2005; Vienne and Nothdurft, 2004; Wood and Deakin, 2003), TAA revived and obtained over the last years more and more acceptance. The realization of the increased risk of osteoarthritis of adjacent joints associated with ankle arthrodesis (Coester et al., 2001) amplified the growing interest in TAA.

As survival rates are getting satisfactorily (Haddad et al., 2007; Stengel et al., 2005) and TAA become a more common treatment for ankle osteoarthritis, the quantification of the functionality and performance of the TAA are getting of increased importance. To achieve a close to normal anatomical function and motion is one of the main design criterions in today’s TAA design improvements. Functionality can be assessed by clinical ankle scores (Ali et al., 2007; Haddad et al., 2007; Kopp et al., 2006) and more objectively by kinematic analyses. A TAA should aim to restore motion in its range to preserve adjacent joints from compensation motions, as well as in its pattern, such that the function of surrounding structures, as ligaments can be maintained (Leardini and Moschella, 2002).

Compared to other joints like e.g. the knee joint, at the ankle still only little is known about the kinematics during daily activities, neither of the healthy ankle joint nor of the replaced ankle joint. To help clinicians and implant developers to evaluate current TAA designs, a better understanding of the kinematics of the ankle joint complex during daily activities is crucial. The purpose of this work is to develop a technical procedure that allows gaining a better under-
standing of the joint mechanisms of unconstrained TAA during daily activities. Specifically, to evaluate the functionality of unconstrained TAA on the basis of the kinematics of the implant components, which includes the development of appropriate procedures to assess ankle joint kinematics.

Chapter 2 addresses the review of the literature. It includes basic anatomic terminology concerning the ankle and foot complex, discusses today’s treatment options for osteoarthritis of the ankle and gives a short overview over history and outcome aspects of TAA. Furthermore, current measurement methods and corresponding results of the assessment of ankle kinematics of the healthy and the replaced ankle joint are reviewed.

Chapter 3 specifies, on the basis of the literature review, the goal of the PhD thesis. This defines the structure of the following four chapters.

Chapter 4 studies the assessment of the three-dimensional (3D) kinematics of TAA. A procedure was developed and applied to measure the kinematics of TAA during daily activities, thereby enabling a discussion of its functionality on the basis of relative motion.

Chapter 5 focuses on the assessment of the 3D kinematics of the healthy ankle joint complex. A procedure was developed and applied to get a basic understanding of the kinematics of the healthy ankle joint serving as a comparison for the TAA subjects.

Chapter 6 evaluates if the implant allows a physiological path and range of motion.

Chapter 7 compares and combines skin marker tracked with videofluoroscopy gained kinematics. A comparison between internal and external measured data is performed and possible compensation mechanisms in the foot and knee joints are discussed.

Chapter 8 summarizes the most important conclusions and gives an outlook for future work.
2 Literature Review

2.1 The Foot and Ankle Complex

The foot and ankle complex plays an important role in human locomotion, it serves as foundation to the ground, absorber of impact loads, as well as propulsion engine. The foot shows flexibility and hereby adaptation possibilities to different ground surfaces. And it also provides stability as a lever arm during the push-off phase of walking. Its functionality is provided by a very complex anatomical structure, consisting of 26 bones, about 30 inserting muscles and over 100 attached ligaments. The foot can be divided into three main sections: the forefoot, the midfoot and the rearfoot, whereas rearfoot and midfoot bones together are often referred to as tarsus.

A big portion of mobility is provided by the ankle-rearfoot complex, which links the foot to the shank. Considerable is, that the load, which in the foot is distributively borne by several bones, needs all to be transmitted by the articulating surfaces of talus and tibia.

2.1.1 Basic Anatomy

Bones and Joints

The bones of the shank are the tibia and the fibula. The articulations between them is effected by an interrosseus membrane and ligaments which connect both extremities, as a result, the movement permitted in these articulations is very small. The rearfoot consists of the talus above and the calcaneus below. The midfoot comprises five bones, the navicular on the medial side, articulating with the talus, the cuboid articulating with the calcaneus and the three cuneiform bones, which lie in a row, distal to the navicular. All the seven bones of the rearfoot and the midfoot combined are called the tarsus. The foerfoot consists of the five metatarsal bones and the 14 phalanges, there being two in the great toe and three in each of the other toes (see Figure 1).

The ankle joint, also called talocrural joint, is formed by the distal parts of the tibia and the fibula and the upper surface of the talus. It consists of upper, medial and lateral surfaces. The upper surface is formed by the tibial mortise and the trochlea tali. The lateral surfaces are between the lateral facets of the talus and the inner aspect of the distal fibula and tibia bones, namely the lateral and the medial malleoli. The tibial mortise is wider anteriorly and anteroposteriorly concave, medially it continues into the articular surface of the medial malleoli, which articulates with the medial talar surface. The lateral talar surface articulates with the lateral malleolus, precisely the inner triangularly shaped facet of the fibula. The lateral malleolus is bigger than the medial, extends about 1 cm farther distal and is located more posteriorly than the medial one. The trochlea tali is anteroposteriorly convex, slightly concave from side to side.
and wider anteriorly than posteriorly (Gray, 1858). The ankle forms a functional unit with the subtalar joint, together also referred to as ankle joint complex (Wu et al, 2002). The subtalar joint is formed by the inferior surface of the talus and the upper surface of the calcaneus. It consists of three articular facets, with the largest being the posterior one and the smaller middle and anterior one located on the bottom side of the talar neck. The talus is the only bone of the foot articulating with the bones of the shank, on the other side the talus articulates distally with the calcaneus and anteriorly with the navicular. The articulation between the talus and the navicular is one of the two joints of the Chopart joint, with the second being the articulation between calcaneus and cuboid. The distal part of the navicular is articulating with the proximal surfaces of the three cuneiform bones. The Lisfranc joint connects the midfoot to the forefoot. Motion which takes place in the area of the Chopart and Lisfranc joints, is always a combination of the relatively small motion of each individual articulation.

**Ligaments of the Ankle Complex**  The stability of the ankle complex is given by the geometry of the articular surfaces and the surrounding soft tissues, such as ligaments. The soft tissues play a very important role in rearfoot stability. A schematic representation of the liga-

**Figure 1: Bones of the foot and ankle complex.**
Figure 2: Ligaments of the ankle complex of a right foot, left: lateral view, right: medial view. lig.: ligament. To allow a clear representation, the superficial tibiotalar ligament is not illustrated. Deltoid ligament: 1-3. Lateral collateral ligament: 8-11.

ment arrangement at the ankle complex is given in Figure 2.

Tibia and fibula are at their distal end connected over the interosseus membrane and also through the anterior and the posterior tibiofibular ligaments - together also referred as distal tibiofibular syndesmosis. At the ankle, ligaments can be divided into two main groups, the medial and the lateral collateral ligaments. Medial as well as lateral, the ligaments originate at the malleoli and insert on the navicular, talus and calcaneus bones. On the medial side we have the deltoid ligament, consisting of a superficial and a deep layer. The variations in its anatomy are wide, but the following main bands can be located: tibionavicular, tibiocalcaneal, superficial tibiotalar, anterior tibiotalar and posterior tibiotalar ligaments. The lateral collateral ligaments of the ankle joint consist of the calcaneofibular, the anterior and the posterior talofibular ligaments. The subtalar joint is additionally stabilized by the interosseus, the medial, the lateral and the posterior talocalcaneal ligaments.

2.1.2 Axis of Rotation of the Ankle and the Subtalar Joints

Beginning of the 20th century Fick (1911) assumed the ankle and the subtalar joints to be ideal hinge joints. For the ankle joint the axis running horizontally and orthogonal to the sagittal plane, for the subtalar joint an oblique axis running from antero-medio-superior to postero-
latero-inferior. Inman (1976) and Isman and Inman (1969) did an anthropometric study on over 107 specimens and agreed to the previous assumption of a single fixed axis, but stated that the orientation of the ankle joint axis is oblique, precisely in the sagittal plane posteriorly inclined downward and directed anteromedially in the transverse plane. Additionally they deduced that the variation in the positions of the different axes is such that they require individual determination. Inman and Mann (1978) developed a wood model having an oblique oriented axis, which explains the coupled movements at the ankle. When the ankle pronates, it is a coupled movement of dorsiflexion, eversion and tibial internal rotation. When supinating, plantarflexion, inversion and tibial external rotation are linked.

Barnett and Napier (1952) assessed 152 specimens and stated that the ankle joint axis changes orientation, due to the combination of the three different radii of curvature of the trochlea tali (medial posterior radius > lateral radius > medial anterior radius). They showed that in the frontal plane during dorsiflexion the axis is inclined downwards and laterally, but during plantarflexion downwards and medially. One year later this prediction was confirmed by the in vitro excursion test performed by Hicks (1953).

The changing axis of the ankle joint was also in vivo affirmed by the X-ray study of Sammarco et al (1973). The analysis on 26 ankles showed that the locations of the instantaneous centers of rotation are dependent on the ankle flexion position. In vitro, but three dimensionally van Langelaan (1983) confirmed the moving of the axes of the ankle joint as well as of the subtalar joint using roentgen stereophotogrammetry for the determination of the instantaneous axis of rotation of the relative motion between several rear- and midfoot bones. Furthermore Lundberg (Lundberg et al, 1989d; Lundberg, 1989) investigated the axis of rotation in vivo in eight healthy volunteers using roentgen stereophotogrammetry and showed that the ankle joint axes, as well as subtalar joint axes, change continuously during movement.

Summarized, looking at the axis of rotation it could be concluded, that neither the talocrural nor the subtalar joint have a single fixed axis, but rather the axes change their direction continuously throughout the range of motion and differ considerably between individuals.

### 2.1.3 Motion at the Ankle Joint

**Terminology** Ankle joint motion occurs in the sagittal, frontal and transverse planes. The terminology is not standardized, but the following paragraph gives an often used definition, which will be used throughout this work. Rotation of the rearfoot relative to the shank in the sagittal plane, in other words about the mediolateral axis, is termed dorsi-/plantarflexion. Motion in the frontal plane (about the anteroposterior axis) is called inversion/eversion. Motion
in the transverse plane (about the vertical axis) is named tibial internal/external rotation, foot external/internal rotation or rearfoot abduction/adduction depending on reference segment, which can be the proximal or the distal segment. Supination stands for a combination of plantarflexion, inversion and tibial external rotation, whereas pronation refers to dorsiflexion in combination with eversion and tibial internal rotation.

**Range of Motion (ROM)** The range of sagittal motion, precisely the maximal possible ROM when moving from a static maximal plantarflexed to a static maximal dorsiflexed position, has been analyzed by several authors. But the reported values differ markedly, probably due to:


ii) different test set ups: weightbearing - non weightbearing, passive - active, in vivo - in vitro, tibiotalar vs. tibiocalcaneal, thus combined tibiotalar and subtalar motion, vs. tibiopedal motion, which additionally includes midtarsal movements

iii) different concepts to describe joint kinematics: projected angles vs. Euler and Cardan angles vs. helical angles and various definitions of joint coordinate systems

iv) individual variations

The values reported are in the range of 12° to 25° for dorsiflexion, 23° to 55° for plantarflexion and 37° to 70° for total sagittal plane motion. Rotation about the vertical and the anteroposterior axes have been objects of investigation of cadaver excursion tests (Close, 1956; McCullough and Burge, 1980; Rasmussen and Tovborg-Jensen, 1982; Siegler et al, 1988). For internal/external rotation maximal ROM values between 9° and 52° and for frontal plane motion between 11° and 32° were reported. In most of the in vivo studies when reporting ankle joint motion, it actually refers to rearfoot-shank motion, due to the impossibility of externally tracking talus motion. Generally it was found that neither sagittal nor frontal plane movement occur exclu-
sively at a single joint, but result from combined rotations at the ankle and the subtalar joint (Siegler et al, 1988; Wong et al, 2005). During moving from maximal dorsi- to plantarflexion, adduction/abduction and inversion/eversion occurred in almost equal amounts at the ankle and subtalar joints. Though the ankle joint contributes higher to dorsi-/plantarflexion, whereas the subtalar joint shows a higher contribution to inversion/eversion (Siegler et al, 1988; Wong et al, 2005). In case of vertical rotation the contribution of the ankle joint is approximately equal to that of the subtalar joint (Siegler et al, 1988).

**Coupling** The motion transfer between calcaneus and tibia, also called the coupling mechanisms of the ankle complex, was the point of interest of several in vitro (Hintermann et al, 1994b; Hintermann and Nigg, 1995a; Michelson and Helgemo, 1995; Olerud and Rosendahl, 1987; Siegler et al, 1988; Wong et al, 2005) and in vivo (Stacoff et al, 2000) studies. Summarized, dorsiflexion is coupled with eversion and internal tibial rotation, whereas plantarflexion results in inversion and external tibial rotation. The coupling coefficient is dependent on the transfer direction, the loading, as well as on the input motion (Hintermann and Nigg, 1995a; Siegler et al, 1988). Additionally, Siegler et al (1988) investigated translational movements of the calcaneus with respect to the tibia and found that they are negligible for movements of inversion/eversion and internal/external rotation. However, dorsiflexion is associated with an anterior displacement of the calcaneus while plantarflexion is associated with a posterior displacement of the calcaneus. Thus, the assumption of an oblique hinge joint (Inman, 1976; Isman and Inman, 1969) or a universal joint (Olerud and Rosendahl, 1987) is far to simple and fails in modeling the natural ankle joint mechanisms.

**The Ankle in Gait** During the stance phase of level gait the healthy ankle complex shows the following motion pattern: During the first 10% of the stance phase a plantarflexion motion occurs in the sagittal plane. This is followed by dorsiflexion until around 70% of stance, followed by plantarflexion until toe off. In the transverse plane, from heel strike until around 20% stance the foot shows an external rotation relative to the shank, followed by rearfoot internal rotation until toe off. In the frontal plane at the beginning of the stance phase, the ankle complex shows an eversion movement, followed by inversion until shortly before toe off (see also 2.4.3 and 2.4.4).
2.2 Treatment of Osteoarthritis of the Ankle Joints: Arthrodesis vs. Total Ankle Arthroplasties (TAA)

Ankle osteoarthritis causes severe pain and functional limitations, which ends up in a loss of mobility and reduced quality of life. Currently the two main possibilities of surgical treatment for osteoarthritis are arthrodesis or TAA. Traditionally arthrodesis has been the surgical treatment of end-stage osteoarthritis of the ankle joint. It is a well accepted treatment among orthopaedic surgeons and in general it provides pain relief, restores joint stability and allows a realigning of lower extremity deformities (Fragomen et al, 2008; Haddad et al, 2007; Wu et al, 2000). The replacement of the ankle joint gives the theoretical benefit of preservation of movement. This leads to an improved gait pattern including, a reduction of limp, which helps to have less disposition for altered biomechanics of adjacent joints.

One long term effect of ankle arthrodesis is the increased risk of osteoarthritis of the adjacent joints. Coester et al (2001) found in 22 arthrodesis patients with a mean follow-up of 22 years significantly more severe osteoarthritis of the ipsilateral subtalar, talonavicular, calcaneocuboid, naviculocuneiform, tarsometatarsal and first metatarsophalangeal joints, compared to those joints of the contralateral foot. Osteoarthritis in the knee did not develop more frequently (Coester et al, 2001).

Probably one of the first gait analysis study performed on arthrodesis patients, showed that arthrodesis patients only show mild limitations during level walking, provided that they wear appropriate shoes (Mazur et al, 1979). Limitations and therefore compensations occur when walking barefoot and going up and down hills (Mazur et al, 1979). The compensation is mainly present as midtarsal motion, more precise the range of sagittal motion of the Chopart and the Lisfranc joints were found to be very large (Mazur et al, 1979). This is in agreement with the skin marker gait analysis study of Wu et al (2000), that showed a loss of rearfoot sagittal motion which was compensated by an increase in forefoot motion.

In vitro, Gellman et al (1987) found a deficit in tibiopedal dorsiflexion of 50.7% and a deficit of plantarflexion of 70.3%, which is in agreement with the study of Hintermann and Nigg (1995b), that showed that after arthrodesis still about 50% of dorsiflexion and 30% of plantarflexion (tibiopedal motion) occurred. Both studies confirm that remaining motion are compensatory movements in the neighboring small joints of the foot.

Concerning contact pressure, Jung et al (2007) found in a cadaver study, a significant increase in contact pressure in the talonavicular and the calcaneocuboid joints after arthrodesis. Suckel et al (2007) confirms in his cadaver gait simulator study an increase in force and maximal pressure in
the talonavicular joint, but controversially to Jung et al. (2007), a decrease in the calcaneocuboid joint was present after arthrodesis. Suckel et al. (2007) noticed a relocation of force and maximal pressure from the lateral to the medial column after arthrodesis.

Comparing TAA to arthrodesis Piriou et al. (2008) found in a gait analysis study for the TAA group, a greater movement at the ankle, a more symmetrical timing of gait and a better restoration of the ground reaction force pattern compared to the arthrodesis group. Concerning ankle motion it must be added, that the marker set used in the study of Piriou et al. (2008) (see Wu et al. (2000) Figure 6) only allows estimating motion between the rearfoot and the shank and not isolated ankle motion (see also 2.4.4).

To sum up, arthrodesis causes persistent alterations in gait, which appears in a loss of ankle motion that is compensated with an increased motion of the neighboring small joints. This has the long term effect of a higher risk of osteoarthritis in those joints. If TAA allow physiological ankle motion and gait pattern, thus less need for compensation and therefore a protection of adjacent joints, still needs to be determined. However, due to the improved longevity of the second generation TAA (Kofoed, 1995; Pyevich et al, 1998; Rippstein, 2002) and the realization of problems associated with ankle arthrodesis (Coester et al, 2001; Morgan et al, 1985; Muir et al, 2002), it still is a hot debate which treatment should be chosen (Haddad et al, 2007; Piriou et al, 2008; Suckel et al, 2007).

2.3 Total Ankle Arthroplasties (TAA)

Early designs of TAA showed a high failure rate (Giannini et al, 2000; Hamblen, 1985) mainly due to loosening at the bone-to-implant interface. Modern designs enabled more promising mid- and long-term outcome results (Kofoed, 1995; Pyevich et al, 1998; Rippstein, 2002). Due to the improving results, during the last decade TAA has gained more and more acceptance and the number of ankle replacements is increasing. The first TAA were two-component designs mainly functioning as hinge joints, whereas most of the newer designs consist of three components with less articular constraints (Pappas and Buechel, 2005; Rippstein, 2002).

2.3.1 History of Design Concepts

The first TAA was implanted in 1970 by Lord and Marotte (1973). It consisted of a tibial titan component with a long stem, coupled with a polyethylene component that replaced the talar body. Like other following TAA of the first generation (Bolton-Maggs et al, 1985; Hamblen, 1985; Kempson et al, 1975; Kitaoka et al, 1994; Kitaoka and Patzer, 1996; Kofoed, 1995; Kofoed and Sorensen, 1998; Newton, 1979, 1982; Scholz, 1976) it consisted of two components and was
implanted using cement. The early encouraging results were followed by high failure rates and complications at mid- to long-term follow-up (Bolton-Maggs et al, 1985; Hamblen, 1985; Kitaoka et al, 1994; Kitaoka and Patzer, 1996; Newton, 1982). Consequently, arthrodesis was still recommended as the treatment of choice for arthritic ankles (Bolton-Maggs et al, 1985; Newton, 1982).

The second generation implants started with the introduction of the New Jersey LCS TAA (Buechel et al, 1988). The New Jersey LCS and most of the following TAA designs of the second generation consist of three components. These are a congruent ultra high molecular weight polyethylene bearing (also called inlay) which is inserted between the metallic tibial and talar components. With this three-component set up, metal-to-bone interfaces at the level of tibial and talar anchorage and at the same time metal-to-polyethylene interface at the gliding surfaces can be provided. This allows for uncemented fixation and good tribological characteristics. The three-component designs can be distinguished in mobile bearing designs and fixed bearing designs. The mobile bearing design is characterized by a moving bearing. And in the fixed bearing designs, two of three components build a unit, therefore they have only one articulation. With the use of those second generation implants, short- and midterm results of TAA outcome have been more encouraging (Buechel et al, 1988; Hintermann, 1999; Hintermann et al, 2004; Kopp et al, 2006; Patsalis, 2004; Rippstein, 2002; Rudigier et al, 2004; San Giovanni et al, 2006; Schenk et al, 2004; Stengel et al, 2005; Valderrabano and Hintermann, 2004; Vienne and Nothdurft, 2004; Weber et al, 2004; Wood and Deakin, 2003).

Generally there are two mechanical design philosophies of TAA, the constrained prostheses which act more or less as a hinge joint and the unconstrained prostheses which provide free internal/external rotation as well as free translation in the transverse plane (see Figure 3). It is obvious that the constrained prostheses enable more stability and the unconstrained more mobility. Constrained prostheses have the disadvantage of a higher risk of loosening of the components in the bone due to larger shear stresses in the bone-to-implant interface (Dettwyler et al, 2004; Kofoed, 1995; Matejczyk et al, 1979; Newton, 1979, 1982; Pappas and Buechel, 2005; Pyevich et al, 1998). The disadvantage of the unconstrained prostheses is the reduced stability and the absence of the ability of transmitting transverse forces and axial moments by the articulating surfaces of the arthroplasty (Dettwyler et al, 2004; Matejczyk et al, 1979; Procter, 1980). In the absence of axial and transverse constraints between the components, only surrounding soft tissues can contribute to joint stability (Leardini, 2000). It is still not clear if this unconstrained design principle allows restoring physiological motion and if it assures the compatibility to the anatomical role of the surrounding ligaments (see also 2.3.3).
To conclude, due to improving patient outcome, TAA becomes a more acceptable option and since patient’s expectations are increasing, ankle arthrodesis is becoming a less acceptable solution. The development of TAA is continuing and the wide variety of designs which are available today, suggest that there is still a need for a better knowledge of the biomechanical mechanisms of the artificial ankle joint.

2.3.2 Outcome

The description of results concerning TAA patient outcome includes factors like ankle function, pain, revision, implant survival and quality of life. A recently published systematic review showed five-year and ten-year implant survival rates following TAA of 78% and 77%, respectively (Haddad et al, 2007). The most common reason for revision was loosening and/or subsidence (28%) (Haddad et al, 2007). Table 1 pools several 2nd generation TAA outcome studies and shows five-year survival rates between 70% and 94% and ten-year survival rates around 75%.

The outcome instruments used in most studies on TAA are clinical ankle scores, such as AOFAS (American Orthopaedic Foot and Ankle Society) ankle hindfoot score and radiographic assessment of component stability and migration. The AOFAS score is the most commonly used rating system in the studies of TAA (Haddad et al, 2007) and is composed of the parameters pain (max 40 points), function (max 50 points) and alignment (max 10 points). Recent TAA outcome studies showed AOFAS ankle scores around 80 points, compared to preop scores of around 35 points (see Table 2).

The quantification of the functionality and performance of TAA by kinematic analyses is addressed in 2.4.
Table 1: Survival rates of recent 2nd generation TAA outcome studies. n=: number of subjects.

<table>
<thead>
<tr>
<th>Author</th>
<th>Type</th>
<th>n=</th>
<th>Survival Rates 5 year</th>
<th>8 year</th>
<th>10 year</th>
</tr>
</thead>
<tbody>
<tr>
<td>Haddad et al (2007)</td>
<td>systematic review</td>
<td>852</td>
<td>78%</td>
<td>-</td>
<td>77%</td>
</tr>
<tr>
<td>Stengel et al (2005)</td>
<td>systematic review</td>
<td>1086</td>
<td>90.6%</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Anderson and Pandy (2003)</td>
<td>STAR</td>
<td>51</td>
<td>70%</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Carlsson (2006)</td>
<td>STAR</td>
<td>109</td>
<td>94%</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Doets et al (2006)</td>
<td>diverse</td>
<td>57</td>
<td>-</td>
<td>84%</td>
<td>-</td>
</tr>
<tr>
<td>Fevity et al (2007)</td>
<td>diverse</td>
<td>257</td>
<td>89%</td>
<td>-</td>
<td>76%</td>
</tr>
<tr>
<td>Henrikson et al (2007)</td>
<td>diverse</td>
<td>531</td>
<td>78%</td>
<td>62%</td>
<td>-</td>
</tr>
<tr>
<td>Hosman et al (2007)</td>
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<td>202</td>
<td>86%</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Hurowitz et al (2007)</td>
<td>Agility</td>
<td>64</td>
<td>67%</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Rudigier et al (2004)</td>
<td>ESKA</td>
<td>105</td>
<td>82%</td>
<td>-</td>
<td>75%</td>
</tr>
<tr>
<td>San Giovanni et al (2006)</td>
<td>Buechel-Pappas</td>
<td>28</td>
<td>-</td>
<td>93%</td>
<td>-</td>
</tr>
<tr>
<td>Spirt et al (2004)</td>
<td>Agility</td>
<td>306</td>
<td>80%</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Wood and Deakin (2003)</td>
<td>STAR</td>
<td>200</td>
<td>93%</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

To conclude, available literature suggest that 2nd generation TAA are favorable on patient outcomes and show an acceptable benefit-risk ratio (Stengel et al, 2005).

Table 2: AOFAS scores of recent 2nd generation TAA outcome studies. Excellent is defined as 90 to 100 points; good, as 75 to 89 points; fair, as 50 to 74 points; and poor, as < 50 points. n=: number of subjects.

<table>
<thead>
<tr>
<th>Author</th>
<th>Type</th>
<th>n=</th>
<th>AOFAS score preop</th>
<th>postop</th>
</tr>
</thead>
<tbody>
<tr>
<td>Haddad et al (2007)</td>
<td>systematic review</td>
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<td>78.2</td>
</tr>
<tr>
<td>Ali et al (2007)</td>
<td>Buechel-Pappas</td>
<td>34</td>
<td>34.6</td>
<td>73.3</td>
</tr>
<tr>
<td>Kopp et al (2006)</td>
<td>Agility</td>
<td>38</td>
<td>33.6</td>
<td>83.3</td>
</tr>
</tbody>
</table>

2.3.3 Compatibility between Ligaments and Total Ankle Arthroplasties

A TAA aims to provide mobility as well as stability (Leardini and Moschella, 2002). Stability is on the one hand given intrinsically by the geometry of the articular surfaces of the implant and on the other hand extrinsically by the surrounding anatomical structures, namely ligaments, joint capsule and muscles (Pappas and Buechel, 2005). Stormont et al (1985) as well as McCullough and Burge (1980) showed that articular surfaces are the major inversion/eversion stabilizers under load, whereas articular surfaces only account for 30% of internal/external rotational stability (Stormont et al, 1985; Tochigi et al, 2006) and therefore soft tissues are the main torsional stabilizers of the ankle.

In the unconstrained TAA the articular surfaces account for less constraints in the transverse plane, and it can be assumed that the ankle shows a reduced stability, but Burge and Evans (1986) as well as Garde and Kofoed (1996) reported that the ankle is stable after implantation.
of an unconstrained TAA. The only increased laxity was found in an increase of translation from 5.5 mm to 9.7 mm in anteroposterior direction (Burge and Evans, 1986).

Compared to the natural anatomy of the ankle joint, unconstrained TAA show a smaller rotational stiffness, thus it may be expected that the passive structures, such as ligaments and joint capsules need to bear more load (Dettwyler et al, 2004). The latter was considered by Dettwyler (2005), who compared the tibia net torsional moments of patients with an unconstrained TAA to normals during the stance phase of gait, but surprisingly no significant differences could be found. It was concluded that surrounding soft tissues are mainly accountable for the transmission of these loads. Therefore, if due to the unconstrained design surrounding soft tissues need to compensate for the reduced intrinsic stability, this could result in an overstretching of the posteromedial ankle ligaments. This could explain the findings of a clinical short- and mid-term follow-up study, in which it was observed, that TAA patients often suffer pain on the posteromedial side of their ankle joint and in addition develop bone hypertrophy in this area (Hintermann and Valderrabano, 2003). To summarize, less intrinsic stability demands more extrinsic stability, which leads to more stress on the surrounding soft tissues.

Furthermore Leardini et al (2000) found a near isometric pattern for the calcaneofibular and the tibiocalcaneal ligaments in passive motion of the healthy ankle joint and concluded that these two ligaments control and guide passive motion, whereas the other ankle ligaments are slack and only get stretched near the extremes, so that they limit but do not guide motion. There is a close interaction between the geometry of the ligaments and the shapes of the articular surfaces in guiding and stabilizing motion in the ankle joint, which gives the compatibility between the prosthetic components and the ligaments an important role in the design of ankle arthroplasties (Leardini, 2001). In other words, the shape of the TAA must be compatible with the geometry of the surrounding ligaments (Leardini, 2001), and therefore Leardini and Moschella (2002) stated that the current unconstrained mobile bearing TAA are unlikely to restore the original characteristic pattern of ligament tension. Compared to other joints like the knee joint, for the ankle still only little is known about the function of the corresponding ligaments in their role of guiding and limiting motion (Leardini, 2001).

Summarized, ligaments account very strongly for a good outcome of TAA. The aim should be to restore the original pattern of slackening/tightening of the ligaments. It is still not clear if the unconstrained design allows physiological motion and proper ligament balancing.
2.4 Kinematics of the Ankle Complex

One important design criterion of TAA is to achieve normal anatomical function and motion. This said, the functionality of a TAA, or in other words the quality of performance, depends on its behavior in the daily life. One way of quantification of the performance of the TAA are traditional clinical ankle scores, like e.g. the AOFAS ankle score or the Kofoed ankle score (see 2.3.2). These clinical ankle scores enable to quantify the outcome of the TAA. One issue is, that they are not only based on objective, but several subjective parameters. Next to clinical scores the functionality can be quantified in a very objective way by kinematic analyses, which are addressed in the following sections.

2.4.1 Static Radiographs

With the use of a TAA one aims to restore at least some of the loss of motion by osteoarthritis. The maximal possible ROM in the sagittal plane can be estimated externally with a goniometer or much more accurate with static lateral radiographs.

Healthy Ankle Joint  In a fluoroscopic study of TAA Komistek et al (2000) found for the contralateral healthy ankle joint of 10 patients a mean sagittal ROM of $37.4^\circ \pm 22^\circ$ (16.7° to 83.4°). Kadakia et al (2008) looked at 10 healthy subjects and found a much higher mean ROM of about $70^\circ \pm 10^\circ$, with the difference, that Komistek et al (2000) measured tibiotalar and Kadakia et al (2008) tibiopedal motion. In the study of De Asla et al (2006) the sagittal tibiotalar ROM of five healthy ankles was analyzed in a non-weightbearing condition using a 3D reconstruction algorithm (see 2.4.6) and a mean of $47.5 \pm 2.2^\circ$ was found.

TAA Demottaz et al (1979) analyzed 19 patients with various TAA designs with weight-bearing lateral radiographs and found a maximal ROM in the sagittal plane between the foot and the tibia of $30^\circ \pm 9.1^\circ$, whereas in average 73% was contributed by motion of the prosthetic ankle joint. Comparable to this result Coetzee and Castro (2004) found for 61 Agility ankles a mean combined ROM of the foot to the tibia of $31.3^\circ$ (7 to 74) and for the exclusive tibiotalar ROM a mean of $23.4^\circ$ (4° to 57°). Likewise, Rippstein (2002) found for 25 Buechel-Pappas prostheses a mean true prosthesis motion of $22.8^\circ$ (11°-38°), compared to a combined foot motion of $34.1^\circ$. In the study of Zerahn and Kofoed (2004), 16 patients with a STAR ankle were radiographically analyzed and the sagittal plane ROM was, with a mean of $45^\circ$ (30° to 55°), larger as compared to the other studies. However, no detailed explanations of the methods can be found, such that it is not clear if tibiotalar or tibiopedal ROM was measured. In a non-weightbearing condition
Pyevich et al (1998) looked at 98 patients with a Agility TAA and found a mean tibiotalar ROM of 36° (10° to 64°).

**Limitations** The ROM measurements in lateral radiographs without any 3D reconstruction show the disadvantage of projection errors induced by the positioning of the patient relative to the X-ray beam axis and furthermore static radiographs are of course limited by being static and therefore can’t predict the behavior of the ankle during dynamic tasks such as level walking.

### 2.4.2 Cadaver Tests

Numerous investigations have been carried out to analyze the kinematic characteristics of the ankle joint complex in vitro. Most cadavers were tested in excursion tests to investigate e.g.: (i) maximal ROM of the ankle complex (Engsberg, 1987; Richter et al, 2007; Siegler et al, 1988; Tochigi et al, 2005; Valderrabano et al, 2003a; Wong et al, 2005), (ii) the coupling characteristics of the ankle complex (Hintermann et al, 1994b; Hintermann and Nigg, 1995a; Michelson and Helgemo, 1995; Michelson et al, 2000; Olerud and Rosendahl, 1987; Siegler et al, 1988; Valderrabano et al, 2003b,c; Wong et al, 2005), (iii) contribution of ankle and subtalar joint to motion of the foot-shank complex (Leardini et al, 1999c; Siegler et al, 1988; Wong et al, 2005) (iv) effects of loading (Hintermann and Nigg, 1995a; Michelson et al, 2000), (v) tendon excursions (Hintermann et al, 1994a) and (vi) ligament elongations (Allard et al, 1982; Leardini et al, 1999c; Saltzman et al, 2004).

Only a few cadaver studies focused on walking simulations to investigate foot and ankle kinematics (Hamel et al, 2004; Nester et al, 2007b).

**Healthy Ankle Joint** A summary of results of cadaver excursion tests on healthy ankles is given in 2.1.3. For data on ROM and the contribution of ankle and subtalar joint to motion of the foot-shank complex see paragraph *Range of motion* in 2.1.3, concerning coupling see paragraph *Coupling* in 2.1.3.

An in vitro approach to investigate the kinematics of foot and shank bones dynamically, is the use of a dynamic walking cadaver model (Hamel et al, 2004; Nester et al, 2007b). Hamel et al (2004) simulates the stance phase of gait with the input of sagittal plane kinematics of the tibia and loads to the extrinsic tendons of the foot and controls it over foot-ground kinetics, whereas Nester et al (2007b) used as input only muscle forces by motors connected to tendons and controlled it over the rearfoot kinematics. Both studies could accurately quantify motion
of the foot and the ankle, but if in vivo gait could be replicated is unknown.

**TAA** Valderrabano et al (2003a,b,c) investigated the performance of three TAA designs (Agility, Hintegra and STAR) and arthrodesis compared to normal cadavers using excursion tests. Under an axial load of 200 N, a moment of 100 Nm was applied to the footplate and the ROM of the calcaneus relative to the tibia in the sagittal and the frontal plane were measured (Valderrabano et al, 2003a). The ROM in the transverse plane was determined by applying the same torque on the tibia. For the sagittal plane, ankle arthrodesis and STAR showed a significant smaller plantarflexion, whereas Agility and Hintegra did not show a significant difference in plantarflexion (normal 28.2°, arthrodesis 8.1°, Agility 30°, Hintegra 26.2°, STAR 22.7°). The ROM in dorsiflexion was significantly smaller for all ankle joint interventions compared to the normal cadavers (normal 14.7°, arthrodesis 4.4°, Agility 10°, Hintegra 11.1°, STAR 10.6°). In the frontal plane the arthrodesis showed a significant decrease, Agility showed a significant increase and Hintegra and S.T.A.R showed no significant differences compared to the normal condition (eversion: normal 5°, arthrodesis 3.5°, Agility 11.9°, Hintegra 5.3°, STAR 7.5°; inversion: normal 13.8°, arthrodesis 10.9°, Agility 17.4°, Hintegra 10.3°, STAR 7.8°). In the transverse plane the arthrodesis showed a significant decrease, for the Agility and the Hintegra there was no significant difference, but the STAR showed a significant higher ROM than the normal condition (external rotation: normal 8.2°, arthrodesis 2.5°, Agility 10°, Hintegra 7.7°, STAR 15.6°; internal rotation: normal 15.2°, arthrodesis 12.2°, Agility 16.3°, Hintegra 16.4°, STAR 4.9°). Three prostheses changed the ROM less than ankle fusion did, they replicated normal joint ROM closely. Tochigi et al (2005) focused on the effect of different bearing thicknesses and talar positions on the ROM of the STAR in the sagittal plane. Any change in bearing thickness (increase and decrease), as well as an anteriorly implanted talar component resulted in a decreased ROM. Richter et al (2007) found a decreased dorsiflexion and no change for plantarflexion ROM for the Hintegra; for the German Ankle System no difference could be found compared to the specimen without prostheses. Michelson et al (2000) and Valderrabano et al (2003b) both investigated if the TAA show a changed behavior of movement transfer, respectively coupling compared to normal ankles. When moving the foot from plantar- to dorsiflexion, for all prostheses (Hintegra, STAR, Agility) neither the movement transfer to tibial rotation, nor to inversion/eversion changed (Valderrabano et al, 2003b). But a decreased coupling between foot eversion and tibial rotation was found for all ankle prostheses, but more in the Agility and the STAR than in the Hintegra (Valderrabano et al, 2003b). However compared to the arthrodesis, the movement transfer of the TAA were changed much less (Valderrabano et al, 2003b). In contrast, in the study of Michelson et al
(2000) investigating the TAA a much smaller coupling from dorsi/plantarflexion to tibial rotation and inversion/eversion was found for the TAA than for the normal ankle.

To conclude, no consensus could be found for the kinematics of the different TAA designs during in vitro tests. However TAA changes the natural ankle joint kinematics less than ankle arthrodesis.

**Limitations**  Using cadaver tests in vivo gait can only be replicated to some extent, in other words cadaver tests do not simulate in vivo conditions accurately. Cadaver tests are mainly limited by a lack of muscle activity. It is also not known if the cadaver shows changed mechanical properties of biological structures. Additionally, any effects of adaptation processes are missing.

**2.4.3 Gait Analysis Using Bone Pins**

The probably most accurate measurement method to describe skeletal kinematics is the use of bone pins. This internal invasive approach produces high quality in vivo kinematic data, which provides a basis for the understanding of joint mechanisms and the estimation of skin movement artefacts when measuring externally.


With six subjects Lundgren et al (2008) showed the largest number of subjects. During the stance phase of level walking for the tibiotalar joint the mean ± SD total ROM was $15.3^\circ ± 2.0^\circ$, $8.1 ± 3.8^\circ$ and $7.8^\circ ± 2.7^\circ$ in the sagittal, frontal and transverse planes, respectively. For the talocalcaneal motion the mean ± SD total ROM was $6.8^\circ ± 1.4^\circ$, $9.8 ± 1.8^\circ$ and $7.5^\circ ± 2.0^\circ$ in the sagittal, frontal and transverse planes, respectively. And for the motion between calcaneus and tibia the mean ± SD total ROM was $17.0^\circ ± 2.1^\circ$, $11.3 ± 3.5^\circ$ and $7.3^\circ ± 2.4^\circ$ in the sagittal, frontal and transverse planes, respectively. As expected, tibiotalar ROM showed the predominance of sagittal plane motion. Motion in the frontal and transverse planes were as well considerable. For the talocalcaneal joint, which is generally considered to show mainly frontal and transverse plane motion, considerable motion in the sagittal plane was found (Lundgren et al,
Looking at each subjects motion strategy (see Figure 4) Lundgren et al (2008) educe that there is a interdependency between the tibiotalar and the talocalcaneal joint. In other words the extent of motion in the tibiotalar joint depends upon the ability of the talocalcaneal joint to move. Accordingly, relating to TAA patients it can be followed that the talocalcaneal joint may be able to compensate a lack of potential tibiotalar motion.

**Limitations** Bone pin measurements provide very accurate skeletal motion data under dynamic conditions, but the applicability of bone pins is highly limited due to its invasiveness. Additionally, it is not clear if the walking pattern of subjects is altered due to the bone pins and the coupled anesthesia.

**2.4.4 Gait Analysis Using Skin Markers**

Early kinematic analyses were restricted to two-dimensional analyses, but with advances in computerized movement analysis three-dimensional motion analysis systems became a frequently used tool (Andriacchi and Alexander, 2000; Capozzo et al, 2005; Chiari et al, 2005; Leardini et al, 2005). These systems allow for the reconstruction of the 3D position of markers, which are attached to the body segments. Either the attachment of the marker is invasive (bone pins, see 2.4.3) or in case of skin markers the markers are attached on the skin surface. Today, gait analysis using skin markers allows an objective quantification of the functional performance of a prosthesis in vivo and during dynamic daily activities.
Healthy Ankle Joint  The investigation of the in vivo kinematics of the healthy ankle joint during daily activities was addressed in a large number of studies (see Figure 5). However, there are numerous suggestions of biomechanical models and in this case markersets (see Figure 6), modeling the foot from one rigid up to eight rigid segments. Each of them shows their advantages and disadvantages and it is not possible to come up with one best solution. According to the research question and the measurement conditions the best suiting model should be chosen. When looking at the ankle joint it seems to be obvious to use a model, which allows estimating the motion between the isolated rearfoot and the shank, without being influenced by markers on the mid- or forefoot, which would be motion of the foot itself. In other words, when the foot is treated as one rigid segment, motion which is actually present between the forefoot and the rearfoot affects the computation of ankle joint rotations and leads to an overestimation of the motion at the ankle (List et al, 2006, 2008).

The external measured ROM that a healthy ankle joint, specifically the rearfoot relatively to the shank, shows during the stance phase of level gait, were found to be between 12.0° and 22.0° for the sagittal plane, between 7.1° and 14.9° for the frontal plane and between 8.4° and 12.9° for the transverse plane (see Table 3).


<table>
<thead>
<tr>
<th>Author</th>
<th>Motion Task</th>
<th>n=</th>
<th>Analyzed Motion</th>
<th>Mean ROM sag</th>
<th>Mean ROM front</th>
<th>Mean ROM transv</th>
</tr>
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<tr>
<td>Wu et al (2000)</td>
<td>gc</td>
<td>10</td>
<td>rf to shank</td>
<td>16.3° ± 3.7°</td>
<td>7.1° ± 2.3°</td>
<td>10.6° ± 3.8°</td>
</tr>
<tr>
<td>Liu et al (1997)</td>
<td>stance</td>
<td>10</td>
<td>rf to shank</td>
<td>16.6°</td>
<td>14.9°</td>
<td>8.4°</td>
</tr>
<tr>
<td>Leardini et al (1999a)</td>
<td>stance</td>
<td>9</td>
<td>rf to shank</td>
<td>12.0°</td>
<td>9.5°</td>
<td>10.7°</td>
</tr>
<tr>
<td>Rattanaprasert et al (1999)</td>
<td>stance</td>
<td>10</td>
<td>rf to shank</td>
<td>20.2°</td>
<td>13.7°</td>
<td>10.3°</td>
</tr>
<tr>
<td>Hunt et al (2001)</td>
<td>stance</td>
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<td>rf to shank</td>
<td>22.0°</td>
<td>8.0°</td>
<td>10.0°</td>
</tr>
<tr>
<td>Woodburn et al (2004)</td>
<td>stance</td>
<td>5</td>
<td>rf to shank</td>
<td>19.1° ± 2.1°</td>
<td>12.5° ± 2.8°</td>
<td>12.9° ± 2.6°</td>
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</tbody>
</table>


**TAA** Probably the first in vivo 3D gait analysis study on TAA patients was performed by Demottaz et al (1979), who analyzed 19 patients with various TAA designs during level walking using tape and movie data, synchronously ground reaction force and electromyographic data was recorded. Nowadays, several studies recorded gait data of TAA patients (see Table 4 and Figure 7). Most of them compared the level gait data of the TAA patients to a healthy normal control group (Benedetti et al, 2008; Doets et al, 2007; Dyrby et al, 2004; Valderrabano et al, 2007). Muller et al (2006) compared the gait data of the ipsilateral TAA ankle to the contralateral healthy ankle and Piriou et al (2008) compared a TAA group to an arthrodesis group during level walking, double toe raise, stepping over an object and ascending and descending steps.

During a gait cycle the TAA patients showed for whole foot relative to shank motion between 16.4 and 24.7° and for rearfoot to shank motion between 13.0° and 27.3° ROM in the sagittal plane (see Table 4). Compared to the healthy control groups or the contralateral leg, the TAA patients, besides of in the study of Doets et al (2007), showed overall a diminished ROM in the sagittal plane (see Table 4). Though, compared to a preoperative baseline or arthrodesis patients, TAA improved the gait characteristics toward normal values (Dyrby et al, 2004; Piriou et al, 2008; Valderrabano et al, 2007).

Performing a qualitative comparison between the studies looking at TAA patients and the studies investigating healthy normals (see Figure 5 and Figure 7), one can conclude that the TAA patients showed a near normal behavior concerning the gait pattern in the sagittal plane.

However, how much of this external marker measured motion is actually taking place in the TAA and what level is compensated by the adjacent joints, such as e.g. the subtalar joint, is still unknown.
<table>
<thead>
<tr>
<th>Author</th>
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<th>n=</th>
<th>Subjects</th>
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<th>Mean ROM front</th>
<th>Mean ROM transv</th>
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<td>19</td>
<td>various TAA designs</td>
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<td>24.7° ± 4°</td>
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<tr>
<td></td>
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<td>19</td>
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<td>30.6° ± 6.6°</td>
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<td>STAR</td>
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<td>16.4° ± 4.1°</td>
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<tr>
<td></td>
<td>gait cycle</td>
<td>9</td>
<td>healthy controls</td>
<td>whole foot to shank</td>
<td>26.8° ± 3.3°</td>
<td>-</td>
<td>-</td>
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<td>12</td>
<td>Hintegra</td>
<td>rearfoot to shank</td>
<td>~15°</td>
<td>~7°</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>gait cycle</td>
<td>12</td>
<td>contralateral</td>
<td>rearfoot to shank</td>
<td>~19°</td>
<td>~5°</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>gait cycle</td>
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<td>Hintegra</td>
<td>whole foot to shank</td>
<td>19.5°</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>gait cycle</td>
<td>12</td>
<td>contralateral</td>
<td>whole foot to shank</td>
<td>29.7°</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td>Doets et al (2007)</td>
<td>gait cycle</td>
<td>10</td>
<td>Buechel-Papas</td>
<td>rearfoot to shank</td>
<td>21.0° ± 4.5°</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>gait cycle</td>
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<td>healthy controls</td>
<td>rearfoot to shank</td>
<td>23.5° ± 3.0°</td>
<td>-</td>
<td>-</td>
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<tr>
<td>Valderrabano et al (2007)</td>
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<td>15</td>
<td>Hintegra</td>
<td>whole foot to shank</td>
<td>23.2° ± 1.5°</td>
<td>8.4° ± 0.9°</td>
<td>15.4° ± 0.8°</td>
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<td>gait cycle</td>
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<td>contralateral</td>
<td>whole foot to shank</td>
<td>34.0° ± 1.5°</td>
<td>10.7° ± 0.9°</td>
<td>16.9° ± 0.7°</td>
</tr>
<tr>
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<td>15</td>
<td>healthy controls</td>
<td>whole foot to shank</td>
<td>30.2° ± 1.7°</td>
<td>10.2° ± 1.0°</td>
<td>17.2° ± 1.4°</td>
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<tr>
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<td>20</td>
<td>New Jersey LCS, STAR</td>
<td>whole foot to shank</td>
<td>23.1° ± 7.7°</td>
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<td>-</td>
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<td>gait cycle</td>
<td>10</td>
<td>healthy controls</td>
<td>whole foot to shank</td>
<td>36° ± 5.2°</td>
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<td>Piriou et al (2008)</td>
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<td>12</td>
<td>Salto</td>
<td>rearfoot to shank</td>
<td>13.9°</td>
<td>-</td>
<td>-</td>
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<tr>
<td></td>
<td>gait cycle</td>
<td>12</td>
<td>healthy controls</td>
<td>rearfoot to shank</td>
<td>27.3°</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>toe raise</td>
<td>12</td>
<td>Salto</td>
<td>rearfoot to shank</td>
<td>10.0°</td>
<td>-</td>
<td>-</td>
</tr>
<tr>
<td></td>
<td>toe raise</td>
<td>12</td>
<td>healthy controls</td>
<td>rearfoot to shank</td>
<td>21.8°</td>
<td>-</td>
<td>-</td>
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<tr>
<td></td>
<td>stepping over</td>
<td>12</td>
<td>Salto</td>
<td>rearfoot to shank</td>
<td>43.0°</td>
<td>-</td>
<td>-</td>
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<tr>
<td></td>
<td>stepping over</td>
<td>12</td>
<td>healthy controls</td>
<td>rearfoot to shank</td>
<td>9.0°</td>
<td>-</td>
<td>-</td>
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<tr>
<td></td>
<td>stepping over</td>
<td>12</td>
<td>arthrodesis</td>
<td>rearfoot to shank</td>
<td>28.3°</td>
<td>-</td>
<td>-</td>
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<tr>
<td></td>
<td>stair climbing</td>
<td>12</td>
<td>Salto</td>
<td>rearfoot to shank</td>
<td>55.9°</td>
<td>-</td>
<td>-</td>
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<tr>
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<td>stair climbing</td>
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<td>healthy controls</td>
<td>rearfoot to shank</td>
<td>16.7°</td>
<td>-</td>
<td>-</td>
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<tr>
<td></td>
<td>stair climbing</td>
<td>12</td>
<td>arthrodesis</td>
<td>rearfoot to shank</td>
<td>22.3°</td>
<td>-</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 4: State of the art of gait analysis studies using skin markers concerning TAA patients. ROM sag: ROM in the sagittal plane, ROM front: ROM in the frontal plane, ROM transv: ROM in the transverse plane. ~ data estimated out of a graph. n=: number of subjects.
Limitations When describing the kinematics of the foot and ankle it is necessary to define a model including the definition of a suitable marker set. In doing so, the foot with its various bones needs to be divided into rigid segments and therefore it is not possible to measure the motion between each bone. There are many different marker sets and models, some treat the foot as one rigid object, some divide the foot into fore- and rearfoot and others are trying to build three or more segments (List et al, 2008). The consequences are, that ankle joint motion sometimes stands for motion between rearfoot and shank and sometimes between the whole foot and the shank, thus one needs to be cautious when comparing data of different studies.

Another limitation is, that the talus is inaccessible for putting markers, thus, it is not possible to distinguish between motion of the subtalar and the ankle joint. Concerning TAA kinematics, due to this inaccessibility of the talus it is not possible to measure the motion, which is taking place in-between the implant components. The motion measured is always at least the combined motion of the TAA and the subtalar joint.

Furthermore, skin markers show the problem of a relative movement between the soft tissue and the underlying bone (skin movement artefacts), which is believed to be the most important error in human movement analysis (Capozzo et al, 2005; Leardini et al, 2005; Reinschmidt et al, 1997b,a). Westblad et al (2002) and Reinschmidt et al (1997b) compared bone pin measured data to simultaneously captured skin marker data. Both found in general a good shape agreement between external and internal measured data. Westblad et al (2002) had three skin markers on the rearfoot and three on the shank, whereas in the study of Reinschmidt et al (1997b), the foot markers were located on the shoe, such that the differences between marker and bone pin captured data do not only stand for skin movement artefacts, but include as well the motion between the shoe and the foot. Westblad et al (2002) found good magnitude agreement for sagittal plane motion (mean root mean square (RMS) difference 1.7°) and more problems with transverse (mean RMS difference 2.8°) and frontal plane motions (mean RMS difference 2.5°) when measuring with skin markers. Controversially, Reinschmidt et al (1997b) found the least differences in the transverse plane (mean RMS difference 2.5°) and a higher amplitude for skin marker data in the sagittal (mean RMS difference 3.4°) and the frontal plane (mean RMS difference 3.1°). Nester et al (2007a) compared bone pin data to skin marker and plate mounted marker data out of three separate data collection sessions and found similar total ROM during stance in all cases (mean differences between bone pin and skin marker data: sagittal plane 2.6°, frontal plane 2.3° and transverse plane 1.9°), although, on average skin markers underestimated the ROM in the frontal and sagittal plane. In contrast, in the study of Reinschmidt et al (1997b) a general overestimation was found when using external skin markers, whereas Westblad
et al (2002) couldn’t formulate a clear conclusion concerning systematic patterns of over- or underestimation of external skin markers relative to bone pins at the ankle joint. It can be concluded, skin marker gait analysis is a powerful tool for full body analyses. However for measuring exclusively motion at the ankle, it doesn’t not provide the required accuracy.

2.4.5 Roentgen Stereophotogrammetry

Healthy Ankle Joint  Lundberg (Lundberg, 1989; Lundberg et al, 1989a,b,c,d) did an in vivo roentgen stereophotogrammetric examination on eight healthy subjects under full body load. Each subject had at least three 0.8 mm tantalum markers introduced into each of tibia, fibula, talus, calcaneus, navicular, medial cuneiform and first metatarsal bones of one foot. The subjects were standing on a platform, which was moved in 10° steps from 30° plantar- to 30° dorsiflexion, from 20° pronation to 20° supination and from 20° internal to 10° external tibial rotation. Exposures were taken in each position of two X-ray tubes, which were placed perpendicular to each other. It was shown, that although the talocrural joint was found to account for most of the rotation around the mediolateral axis, there was a substantial contribution of the joints distal to the talocrural articulation to participate in dorsi-/plantarflexion. This ability to provide a certain flexibility in the joints below the talus can get important when the talocrural joint is immobilized (Lundberg et al, 1989a), precisely in the case of a tibiotalar arthrodesis or a lack of mobility provided by a TAA. Lundberg et al (1989a) additionally showed that motion of the foot in the sagittal plane effects in a coupled tibiotalar rotation around the vertical axis. Moving the foot around the anteroposterior axis resulted in an average of 0.2° of external tibial rotation in the tibiotalar joint (Lundberg et al, 1989b). Looking at the axis of rotation, it could be concluded, that the joint axis of the talocrural joint changes its direction continuously throughout the range of motion and considerably differs between individuals (Lundberg et al, 1989d).

Limitations  Roentgen Stereophotogrammetric examinations are limited due to the dual X-ray measurement set up to analyses of static positions. It is not possible to track the foot during a dynamic condition, such as e.g. level gait. Another irremovable drawback is the exposure to radiation. Additionally the placement of the tantalum markers is an invasive approach and therefore the number of available subjects will always be limited. The measurement procedure itself is a very accurate approach that enables to capture the internal motion of the bones without being distracted by other structures motion.
### 2.4.6 Videofluoroscopy

As a diagnostic tool, fluoroscopy has had extensive usage in the podiatric surgery (Wilner and Lepow, 1983). Interestingly, the use of videofluoroscopy to evaluate the kinematics of the foot and the ankle goes only back to the investigation of Green et al (1975), who described the two-dimensional tarsal motion of four subjects with various biomechanical pathologies during external to internal rotation of their leg. The proven efficacy of videofluoroscopy to identify motion within the foot, in combination with a digitization process was often used to analyze the sagittal motion of the foot and ankle. Several investigations analyzed the sagittal plane motion of the calcaneus and/or the arch height and length during level gait (Gefen et al, 2000; Gefen, 2003; Perlman et al, 1996; Wearing et al, 1998, 1999, 2004, 2005). Besides, Komistek et al (2000) and Conti et al (2006) both evaluated ankle joint sagittal plane motion during a plantarflexion to dorsiflexion movement, respectively level gait. Recently Wrbaskic and Dowling (2007) used fluoroscopic imaging to investigate the two-dimensional movement of the foot bones during several tasks, like a vertical jump or a calf raise. However, all these studies were limited to a two-dimensional analysis procedure.

Recent technological development now provides the opportunity to measure internal kinematics of either implant components or bones using a videofluoroscopic system three-dimensionally. For that purpose an algorithm is used that matches a synthetic projection of an object into the two-dimensional fluoroscopic images and consequently enables to estimate the three-dimensional kinematics out of the two-dimensional images. Due to this technological advance, videofluoroscopy now enables to estimate the three-dimensional internal kinematics in vivo, without invasive procedure, and with the advantages of not being limited by skin movement artefacts. In addition it enables to distinguish between motion at a certain joint from motion of the adjacent joints.

Single plane videofluoroscopic analysis (Banks and Hodge, 1996; Banks et al, 2005; Dennis et al, 1998; Foresti et al, 2008; Hoff et al, 1998; Stiehl et al, 1995; Tsai et al, 2006; Zihlmann et al, 2006), as well as dual orthogonal fluoroscopy (Li et al, 2004) has effectively been used to provide valuable information in the study of total knee arthroplasties. In the field of ankle and foot kinematics three-dimensional videofluoroscopic analysis of daily motion tasks were only rarely used up to now (see Table 5).

#### Healthy Ankle Joint
looked at three positions of a simulated stance phase, whereas the subject had to pause at heel strike, midstance and toe off. In case of De Asla et al (2006) the three-dimensional ankle joint kinematics, in case of Wan et al (2006) the articular contact areas of the talocrural joint and in case of Wan et al (2008) the cartilage contact deformation were analyzed. Li et al (2008) coupled this kinematic approach with a force plate measurement and analyzed contact deformation in the ankle joint in a weightbearing standing position under different loading conditions. Yamaguchi et al (2008) analyzed the three-dimensional talocrural and subtalar joint kinematics of seven healthy subjects during nonweightbearing and weightbearing movement from dorsiflexion to plantarflexion using a single plane fluoroscopy measurement set up with a frame rate of 7.5 Hz. Furthermore Tung-Wu et al (2008), as well as Blankevoort et al (2008) recently presented validation studies of single plane three-dimensional bone matching approaches, but up to now only cadaver and no in vivo data were presented.

De Asla et al (2006) presented the three-dimensional gait data of five healthy normal ankles. For the talocrural joint, from heelstrike to midstance, the average ROM for the sagittal plane was 9.1° ± 5.3°, for the transverse plane -1.6° ± 5.9° and for the frontal plane 3.8° ± 8.2°. From midstance to toe off 4.4° ± 13.0°, -1.6° ± 5.9°, -1.7° ± 2.7° average ROM were found in the sagittal, the transverse and the frontal plane, respectively. The average contribution of the subtalar joint from heelstrike to midstance were -0.9° ± 1.2° (sagittal plane), -1.5° ± 9.9° (transverse plane), 1.7° ± 2.7° (frontal plane) and from midstance to toe off 8.5° ± 2.9° (sagittal plane), 12.3° ± 8.3° (transverse plane), -10.7° ± 3.8° (frontal plane). From heelstrike to midstance the sagittal plane motion contribution from the talocrural joint was much higher than that of the subtalar joint, whereas from midstance to toe off the subtalar joint showed a significantly larger amount of internal rotation and inversion compared to the talocrural joint. To conclude, talocrural joint motion was dominant during the early stance phase while subtalar joint motion was prevailing during the end of the stance phase.

TAA  Looking at the kinematics of ankle arthroplasties, previous studies using an in vivo single plane videofluoroscopic approach, determined successfully the relative motions between the implant components under dynamic conditions, but are limited to three or four analyzed positions. Komistek et al (2000) looked at 10 subjects with a Buechel-Pappas TAA at three weightbearing positions when moving from maximum dorsiflexion to maximum plantarflexion. Conti et al (2006) analyzed 10 Agility patients and Leszko et al (2008) 20 Salto patients. Both described the relative motion of the implant components in four positions of the stance phase of gait and a step-up motion (heel strike, 22% and 66% stance and toe off). The latter three
investigations all used the three-dimensional matching algorithm by Hoff et al (1998). The average ROM moving from maximum dorsiflexion to plantarflexion was $32.3 \pm 24.3^\circ$ for the Buechel-Pappas TAA (Komistek et al, 2000). In the study of Leszko et al (2008) the average ROM of the talar relative to the tibial component during the stance phase was found to be $9.2^\circ$ (level gait) and $8.0^\circ$ (step-up) in the sagittal plane and $2.7^\circ$ (level gait) and $2.5^\circ$ (step-up) in the frontal plane. In the transverse plane, an average internal rotation from $0.5^\circ$ to $2.1^\circ$ was observed for gait, whereas for step-up an average of $0.5^\circ$ to $2.0^\circ$ external rotation was shown (Leszko et al, 2008). Furthermore the tibia translated anteriorly $1.5$ mm during gait and $2.3$ mm during step-up (Leszko et al, 2008). Conti et al (2006) found an average ROM in the frontal plane of $1.3^\circ$, whereas from subject to subject the measurements deviated significantly, such that the patient with the maximal amount of frontal plane ROM showed $14.1^\circ$ eversion and in contrast two other patients showed $11.3^\circ$ inversion. Similarly, the average ROM in the transverse plane was $2.8^\circ$ external rotation, whereas 8 patients showed internal and two external rotation (Conti et al, 2006). The anteroposterior translation was in average $0.3$ mm and for all subjects less than $3.5$ mm (Conti et al, 2006). Thus, it was not possible to find a motion characteristic that was comparable in-between subjects.

Recently Petrolo et al (2008) presented an approach that combines single plane fluoroscopy with simultaneous measurement of ground reaction forces, but unfortunately no data has been presented.

**Accuracy** As can be expected due to the set up, the dual-orthogonal approach shows the best accuracy with $0.1$ mm and $0.1^\circ$ (Li et al, 2004). For the single plane fluoroscopic approaches the in plane accuracy varies between $0.3^\circ$-$1.4^\circ$ for rotation and $0.5$ mm up to $0.8$ mm for translations. When looking at the out of plane accuracy, the accuracy gets worse, translational errors between $2.2$ mm up to $6.6$ mm, and rotational up to $1.5^\circ$ were found (see Table 5). But one should keep in mind, that the accuracy is not only given by the measurement system and the analysis procedure, but as well dependent on the geometry and the material of the tracked object. A surrounding of the implant by cement additionally complicates the matching. It can be assumed, that implants can be matched with higher accuracy than bone.

**Limitations** It should be kept in mind, that the video fluoroscopic approach shows the disadvantage of an irremovable radiation exposure. Furthermore the tracking frequency is smaller than what is possible with today’s video photogrammetry systems. The single plane approach shows a limitation in the out of plane accuracy.
To conclude, videofluoroscopy is an in vivo approach to quantify the three-dimensional kinematics of TAA or foot bones during daily activities more accurately than with skin markers. It is therefore suited for research questions which require high accuracy and aim for a better understanding of a single joint and do not ask for a full body analysis. Previous studies showed the capability of the videofluoroscopic approach, but most of them are limited to a small number of analyzed frames and/or the restricted measurement set up, which does not allow measuring the stance phase of daily motion tasks, such as free level walking.
<table>
<thead>
<tr>
<th>Author</th>
<th>Subjects</th>
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<th>Analyzed Frames</th>
<th>Measurement System</th>
<th>Matching Algorithm</th>
<th>Accuracy</th>
</tr>
</thead>
</table>
| Komistek et al (2000) | 10 Buechel-Pappas | pf to df weightbearing          | 3 pf, MS, df    | single plane fluoroscopy image size 640x480 pixels | 3D matching (Hoff et al, 1998)      | IP: 0.5 mm, 0.3°  
|                   |                  |                                 |                 |                                      |                                      | OP: 2.2 mm, 1.5° (Hoff et al, 1998) |
| Conti et al (2006) | 10 Agility       | stance phase of gait            | 4 HS, 33% and 66% stance phase, TO |                                      |                                      |                 |
| Leszko et al (2008) | 20 Salto         | gait and step-up                | 4 HS, 33% and 66% stance phase, TO |                                      |                                      |                 |
| Petrolo et al (2008) | -                | -                               | -               | single plane fluoroscopy (10Hz) force plate | 3D matching (Banks and Hodge, 1996) | IP: 0.5 mm, 1.1°  
|                   |                  |                                 |                 |                                      |                                      | OP: 6.6 mm (Banks and Hodge, 1996) |
| Yamaguchi et al (2008) | 7 healthy normals | pf to df non- and weightbearing | -               | single plane fluoroscopy (7.5Hz) oblique lateral | 3D bone matching (Banks and Hodge, 1996) |                 |
| De Asla et al (2006) | 5 healthy normals | simulated stance phase of gait (paused at HS, midstance, TO) | 3 HS, MS, TO    | dual fluoroscopy image size 1000x1000 pixels | 3D bone matching (Li et al, 2004)  | 0.1 mm, 0.3° (Li et al, 2004) |
| Li et al (2008) | 4 healthy normals |                                 | -               |                                      |                                      |                 |
| Tung-Wu et al (2008) | cadaver          |                                 | -               |                                      |                                      |                 |
| Blankevoort et al (2008) | cadaver         |                                 | 2 images        |                                      | 3D bone matching (tomography base bone contour method) (Beimers et al, 2008) | 0.1 mm, 0.3° (Beimers et al, 2008) |

Table 5: State of the art of investigations using videofluoroscopy concerning the ankle. pf: plantarflexion, df: dorsiflexion, HS: heel strike, TO: toe off, MS: mid stance, -: no information available
2.5 Conclusions for this Project

The two main approaches for the treatment of osteoarthritis are arthrodesis or TAA. Both provide pain relief, but arthrodesis causes persistent alterations in gait, which is seen in a loss of ankle motion compensated with an increased motion of the neighboring small joints, the long term effect being a higher risk of osteoarthritis in those joints. Limitations and therefore compensations concerning gait of arthrodesis patients occur especially when walking barefoot and going up and down hills (Mazur et al, 1979).

Overall, 2nd generation TAA are effective on patient outcomes (Stengel et al, 2005). They show a five year survival rate between 70-94% and an increase in the AOFAS score from preop around 35 points to postop around 80 points. Cadaver studies showed that TAA changed ROM less than arthrodesis; they replicated normal ROM closely, but those studies are limited by the in vitro approach. Skin marker tracked data of TAA patients showed close to normal gait patterns, but it can’t be answered how much motion actually took place in the TAA and how much was compensated by adjacent joints. The tendency in design concept development is going in the direction of the unconstrained design. But it is still not clear if the unconstrained design principle allows restoring physiological motion and a close to normal gait pattern. Consequently, it provides less need for compensation and therefore a protection of adjacent joints. Furthermore it is not clarified if today’s TAA designs assure proper ligament balancing.

For better outcome results the TAA’s performance needs to be evaluated in terms of a kinematic analysis of the implant components during daily motion tasks, such as level walking. For the ankle joint more challenging ones, such as walking up- and downhill or on a side inclined slope should be analyzed. Since for the ankle joint performance only the stance phase is crucial and talocrural joint motion is dominant during the early stance phase, there is especially a need for a better knowledge of the 3D kinematics of the TAA during the early stance phase in before mentioned daily activities.

Previous investigations in gait analysis analyzed ankle kinematics using the following measurement approaches:

- Static radiographic measurements, which are limited by projection errors and can’t predict the dynamic behavior of the TAA.
- Cadaver studies, which do not simulate in vivo conditions accurately.
- Bone pin tracking, which is highly invasive, therefore only applicable to a small number of subjects and very unlikely for the use on patients.
- Skin marker tracking, which show the limitations of skin movement artefacts and the inaccessibility of the talus, thus the impossibility to distinguish tibiotalar from subtalar motion. To conclude, skin marker tracking is very powerful for full body analyses but less appropriate for single joint analyses that require high accuracy.

- Roentgen stereophotogrammetry, which is limited due to the dual X-ray set up, that only allows the measurement of static positions and no unrestricted level walking. Furthermore the use of tantalum markers is highly invasive and subjects experience a certain radiation exposure.

- Videofluoroscopy, which is up to now only rarely used at the ankle and most studies are either limited by a small number of analyzed frames or by a restricted measurement set up that did not allow free gait. One not negligible limitation is the radiation exposure. To conclude, it is suited for single joint research questions that demand high accuracy and in vivo conditions, but do not ask for full body analyses.

The wide variety of TAA designs which are available today, suggest that there is still a need for a better knowledge of their in vivo performance. One important design criterion of a TAA is to recover some of the normal anatomical function to master motion tasks of daily activities. To evaluate the functionality of an existing TAA design, the knowledge of the accurate 3D in vivo kinematics of its components are crucial. Since this analyzes requires high accuracy and is focused on a single joint, videofluoroscopy is the most appropriate measurement technique.
3  Goal of the Project

The overall purpose of this PhD thesis is to develop a technical procedure that allows gaining a better understanding of the joint mechanisms of unconstrained TAA during daily activities, such as level walking, walking uphill, walking downhill and walking on a side slope. Specifically, the goal is to evaluate the functionality of an unconstrained TAA on the basis of the relative motion between the implant components during the stance phase of daily activities. The following four chapters highlight each a different aspect concerning ankle joint kinematics. The combination of the four allows getting a better understanding of the joint mechanisms of unconstrained TAA.

Chapter 4 assesses the 3D kinematics of the implant components of an unconstrained TAA (tibial and talar component) during the stance phase of level walking, walking uphill, walking downhill and walking over a side inclined slope using videofluoroscopy.

Chapter 5 assesses the 3D kinematics of calcaneus, talus and tibia of a healthy ankle using the same measurement set up as for the implants.

Chapter 6 compares the joint kinematics of the implant components to the one of the healthy ankle. In other words to clarify the question if the implant allows a physiological path and range of motion.

Chapter 7 combines and compares information on ankle kinematics measured externally and internally, thus fluoroscopy gained vs. skin marker gained data.
4 3D Kinematics of Total Ankle Arthroplasties Using Videofluoroscopy

4.1 Introduction

One important design criterion of a TAA is to recover some of the normal anatomical function to master motion tasks in daily life and therewith improve the patients’ quality of life. Therefore, a crucial part of the evaluation of the functionality of a TAA, is the in vivo analysis of its kinematic behavior. Previous investigations on TAA kinematics include:

(i) Cadaver studies, which do not simulate in vivo conditions accurately (see also 2.4.2).

(ii) Skin marker tracking, which are limited by skin movement artefacts and the difficulty to distinguish between motion at the TAA from motion at the adjacent joints, as e.g. the subtalar joint (see also 2.4.4).

(iii) Videofluoroscopic investigations, up to now only rarely used on the kinematics of TAA patients and mostly restricted by their measurement set up and or limited in the number of analyzed frames (see also 2.4.6).

From these studies, the kinematics of TAA components during walking are not yet fully understood.

This chapter aims to assess the 3D kinematics of an unconstrained TAA (tibial and talar component) during the stance phase of level walking, walking uphill, walking downhill and walking over a side inclined slope using videofluoroscopy. For that purpose a procedure was developed and applied, the following section describes the approach.

4.2 Methods

This section gives information about the participating subjects (4.2.1), the measurement set up (4.2.2), the test procedure (4.2.3) and describes the following steps of the analysis procedure: namely distortion correction and estimation of projection parameters (4.2.4), the CAD model preparation and the 3D reconstruction (4.2.5), as well as the estimation and applied conventions concerning the relative rotations and translations (4.2.6). Overall a videofluoroscopic procedure for the kinematic analyses of TAA, in the following called TAA approach, was developed and applied. A schematic overview of the main steps of the whole procedure is given in Figure 8.
The whole videofluoroscopic procedure
TAA approach

- Test procedure
- Fluoroscopic images
- Distortion correction
- CAD model preparation
- 3D reconstruction
- 3D pose of implant components
- Motion of talar relative to tibial component

Figure 8: The main steps of the whole videofluoroscopic procedure of the TAA approach.

4.2.1 Subjects

Four male volunteers, all good outcome patients, having on one side an unconstrained TAA (Mobility™ Total Ankle, DePuy (Figure 9)) participated in this study. Further information on the subjects is given in Table 6. Each subject signed an informed consent in accordance to the research ethics committee of the ETH Zurich.

![Mobility™ Total Ankle, DePuy](image)

Figure 9: Mobility™ Total Ankle, DePuy

<table>
<thead>
<tr>
<th>Subject</th>
<th>Age [Years]</th>
<th>Height [m]</th>
<th>Weight [kg]</th>
<th>Years postop</th>
<th>Assigned color</th>
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<td>1.82</td>
<td>74</td>
<td>3.3</td>
<td>green</td>
</tr>
</tbody>
</table>

Table 6: Information on TAA subjects
4.2.2 Measurement Set Up

The measurement set up consisted of a videofluoroscopy system integrated in a walkway and the time synchronized optical tracking system VICON (Figures 10, 11, 12).

Videofluoroscopy System  The videofluoroscopic image capture was performed using the videofluoroscopy system BV Pulsera (Philips Medical Systems, Switzerland) with a pulsed mode of 25 Hz and 1 ms shutter time. The videofluoroscopy system consists of a c-arm that includes the X-ray source and the image intensifier. The field of view of the image intensifier has a diameter of 30.5 cm. The included CCD camera allows capturing images with the resolution of 1000 x 1000 pixels with a grey scale resolution of 12bit (Foresti, 2009).

The fluoroscopy coordinate system is defined as follows: x- and y-axes lie in the image plane, z-axis is directed perpendicular to the image plane (Figure 12).

The radiation exposure dose for 60 sec pulsed test time is 0.12 mSv. The whole measurement procedure including 33 trials and 12 single pictures equates an exposure dose of approximately 0.086 mSv, which is 1.7% of the boundary level of 5 mSv/year. Comparatively a standard X-ray of the foot results in a dose of 0.06 mSv.

Walkway  The videofluoroscopy system is integrated in a walkway of 10 m length and 1 m width (see Figure 11). The height of the walking level is 50 cm above the room floor. The c-arm is inclined, with the X-ray source lower than the image intensifier, such that the bottom ray is parallel to the floor, resulting in a minimized shade of the floor and maximized usable field of view. Additionally, the c-arm is installed with an inclination of around 25° around its long axis to allow the walkway to pass above the centerpiece. The height of the c-arm is defined by the distance of the center of the image intensifier being 11 cm above the walkway (see Figure 10).

For the test procedures with an inclined surface, a ramp with an inclination angle of 10° was used. For those test conditions the inclination and height of the c-arm was accordingly adapted (see Figure 12).

Optical Tracking System  The 3D motion analysis system used, is a VICON MX system (Oxford Metrics Group, UK). It consists of eight fixed and four movable MX40 motion-capture cameras with a resolution of 2352 x 1728 pixels and a capture frequency of 100 Hz. The size of the capture volume was 300 cm x 500 cm x 200 cm. The instrumental error of marker position
Figure 10: Videofluoroscopy system integrated in walkway. C-arm inclination and height of the measurement set up for level gait.

detection is $\leq 1$ mm (root mean squared error (RMSE)). Additional information concerning the VICON measurements are given in 7.2.3.

The VICON laboratory coordinate system is defined using the standard calibration procedure implemented in the VICON software. The y-axis is in line with gait direction, the x-axis is directed towards the side and the z-axis vertical to the walkway (Figure 11).

Using an input trigger sent by the X-ray generator of the videofluoroscopy system to an analog channel of the VICON system, the time of each generated videofluoroscopic image is recorded as an analog signal in the VICON system (Foresti, 2009).

4.2.3 Test Procedure

The test procedure consisted of four main parts, namely calibration, basic motion tasks, static and dynamic trials.

- **Calibration:** The calibration procedure had to be performed for the four positions of the c-arm (set up level gait, set up up- and downhill, set up side slope bottom and set up side slope top). For calculation reference, six images were captured during the calibration procedure: one image of the calibration grid (see Figure 13) used for image distortion correction (4.2.4) and registration of the fluoroscopy coordinate system in the VICON laboratory coordinate system (7.2.5) and five images of the calibration tube to estimate projection parameters of the videofluoroscopic system (see Figure 14).
• **Basic motion tasks (BMT):** The BMT consist of four basic motions to determine ankle, knee and hip joint centers/axes of the skin marker model. Basic motion tasks are only captured by the VICON system (for further information see 7.2.3).

• **Static trials:** The subject had to perform six static trials in different positions (each repeated twice). These positions were two standing trials in an anatomic upright position once captured from the side, once from the front, as well as a maximal plantarflexed, a maximal dorsiflexed, a maximal everted and a maximal inverted static loaded position.

• **Dynamic trials:** The kinematics of the ankle of each subject was tracked using videofluoroscopy and the VICON system simultaneously during five different gait conditions. Namely, level gait (gt), walking downhill (dnh), walking uphill (uph), and walking over a side inclined slope, once inclined to the medial (sst), once to the lateral side (ssb) of the ankle (see Figures 11 and 12). For each of the five gait conditions the subjects had to perform five valid trials. The videofluoroscopy system captured the stance phase, whereas the VICON system captured at least two consecutive steps including the stance phase captured by the videofluoroscopy system.
Figure 11: Measurement set up for level gait. C-arm including X-ray source and image intensifier integrated in a 10 m walkway. VICON laboratory coordinate system: y-axis in gait direction, x-axis towards the side, z-axis vertical to the walkway.

Figure 12: Measurement set up for side slope bottom (ssb) (top left), side slope top (sst) (top right), uphill (uph) (down left) and downhill (dnh) (down right). Fluoroscopy coordinate system: x- and y-axes lie in the image plane, z-axis is directed perpendicular to the image plane.
4.2.4 Distortion Correction and Estimation of Projection Parameters

In order to correct the distortion of the videofluoroscopic images, a well defined calibration grid, placed perpendicularly to the X-ray beam, was captured for each measurement set up. According to Foresti (2009) image distortion was then eliminated by determining the bilinear transformation between the known and the measured coordinates of the beads of the calibration grid (see Figure 13).

The projection parameters were estimated by the use of a calibration tube and the exact knowledge of the embedded 25 marker positions (see Figure 14). For each measurement set up, the tube was filmed in five different positions on the image intensifier. Since X-ray imaging can be described as a central projection, the projection parameters (focal distance and the location of the principle point in the image plane) could then be determined by a least-squares optimization (Foresti, 2009).

![Figure 13: Calibration grid (top) used for image distortion correction. Before (left) and after (right) distortion correction.](image)

4.2.5 CAD Model Preparation and 3D Reconstruction

3D reconstruction was performed using the CAD models of the implant components and a registration algorithm, initially designed for the location of hip implants in X-ray images (Bur-
Figure 14: Tube used for determination of the projection parameters. The tube has a length of 30 cm and comprises 25 beads on 12 and 13 well defined positions on both ends.

For each of the implant components, a coordinate system (implant coordinate system) is defined in the respective CAD data using the software UGS NX4.0 (Siemens PLM Software, Texas). The x-axis and the y-axis were chosen to lie in the symmetry plane of the component, whereas the x-axis is directed anteroposteriorly and the y-axis vertically. The z-axis is directed perpendicular to the symmetry plane. For the talar component the origin was chosen to lie in the symmetry plane on the anterior of the two mediolateral directed construction curvatures of the lower surface of the talar component. For the tibial component the origin was defined in the center of the stem on the upper surface of the flat part of the tibial component (Figure 16).

The used registration algorithm (Burckhardt, 2002) computes a virtual projection of the volume information of the implant components using the CAD models, and then locates their pose by an optimization, in which the virtual projection is sequentially compared to the actual fluoroscopic image. Respective output is the 3D pose of the tibial and talar components (corresponding orientation matrices and position vectors $R_{\text{tib}}, t_{\text{tib}}, R_{\text{tal}}, t_{\text{tal}}$).

Error analysis for the 3D Reconstruction For the analysis of the error of the 3D reconstruction, the tibial and the talar component of the Mobility™ Total Ankle, both of size three, were fixed together and embedded into polyurethane foam, such that components couldn’t move relative to each other. Using a positioning set up, composed of a cross table and a marking unit, the TAA components were captured in 63 positions (Foresti, 2009) (see Figure 15). Each image was 3D reconstructed using the CAD data of the size three parts of the manufacturer, and the translational, as well as the rotational error were estimated for both components. Results are presented in Table 7. The rotational error ($\sigma_{\text{Rot}}$) was found to be smaller than 0.2° in plane and 1.3° out of plane. The translational error ($\sigma_{\text{Trans}}$) was smaller than 0.4 mm in plane and 2.1 mm out of plane. The larger error in out of plane direction is caused by the single plane approach.
Figure 15: Positions used for the error analysis for the 3D reconstruction. x- and y-axes are parallel, z-axis is perpendicular to the image plane.

This issue is explained in the following example of a central projection with a typical set up. An object of 30 mm size in a focus to object distance of 890 mm results in a projection of 30.3 mm object size on the image plane in a focus to image distance of 900 mm. A shrinking of the corresponding image size of the object by 0.1 mm is equivalent to a out of plane translation of the object of 3 mm.

The error of the 3D reconstruction is strongly dependent on the manufacturing accuracy of the implanted TAA. A discrepancy between the surface of the implant components and the respective CAD data results in an error of the 3D reconstruction (Foresti, 2009). However, the present accuracy is in the range of or even better than the 3D reconstruction algorithms presented by Banks and Hodge (1996) and Hoff et al (1998) (see also Table 5).

\[
\text{Table 7: SD of the 3D reconstruction for the talar and the tibial components.}
\]

<table>
<thead>
<tr>
<th>Component</th>
<th>(\sigma_{\text{Rot}})</th>
<th>(\sigma_{\text{Trans}})</th>
</tr>
</thead>
<tbody>
<tr>
<td>(x)</td>
<td>(y)</td>
<td>(z)</td>
</tr>
<tr>
<td>Tibial</td>
<td>0.16</td>
<td>1.31</td>
</tr>
<tr>
<td>Talar</td>
<td>0.32</td>
<td>0.41</td>
</tr>
</tbody>
</table>

4.2.6 Estimation of Relative Rotations and Translations

Rotations and translations of the talar component are described relative to the tibial component. Thus, the relative motion \(Q_{\text{rel}}\) of the talar to the tibial component was calculated as follows:

\[
Q_{\text{rel}} = Q_{\text{tal}}^{-1} \cdot Q_{\text{tib}}
\]
with

\[
Q_i = \begin{bmatrix}
R_i & t_i \\
0 & 0 & 0 & 1
\end{bmatrix}
\]

for \( i = \text{tib, tal, rel} \)

Any finite movement can be described in terms of a translation along and a rotation about a screw vector. Thus, from the rotation matrix \( R_{\text{rel}} \), rotation angle \( \beta \) and screw vector \( \mathbf{u} (\| \mathbf{u} = 1 \|) \) were extracted using quaternion calculations (Dettwyler, 2005).

**Clinical Rotations**  To describe joint rotation with clinical terminology the attitude vector \( \theta = \beta \cdot \mathbf{u} \) is decomposed along the axes of the implant coordinate system (Woltring, 1994). In other words, the three rotations described with respect to the implant coordinate system were estimated by the direction cosines of the screw vector to the axes of the implant coordinate system. It follows that dorsi-/plantarflexion motion is described as the rotation around the z-axis, inversion/eversion motion is described with respect to the x-axis and adduction/abduction is equivalent to rotation around the y-axis (Figure 16). Neutral, thus 0°, is defined by the initial position of the implant components determined by the implant coordinate system.

**Motion of the Talar Construction Axis**  The talar construction axis is defined as the cylinder axis of the talar component (Figures 16 and 17). To analyze the talar motion, the location and orientation of the construction axis relative to the tibial component was calculated. Anteroposterior translation of the talar relative to the tibial component was analyzed as translation of the construction axis with respect to the x-axis of the tibial coordinate system.

**Stance Phase**  In all gait trials, the stance phase, thus the data during the time frame from heel strike (HS) to toe off (TO), was evaluated and normalized to 0-100% stance. HS and TO were set in the VICON software. The corresponding fluoroscopic images were determined using the analog trigger signal of the videofluoroscopic system (see 4.2.2).

**ROM**  For the static trials, a maximal ROM around the z-axis (\( ROM_{\text{stat pfdf}} \)) between the maximal plantarflexed (\( \text{max}_{\text{stat pf}} \)) and the maximal dorsiflexed (\( \text{max}_{\text{stat df}} \)) static loaded positions, as well as the ROM around the x-axis (\( ROM_{\text{stat invev}} \)) between a maximal everted and a maximal inverted static loaded position were calculated. Thus, the static maximal ROM was denoted as the relative change in the ankle angle between the two maximal static positions. For the dynamic trials, neither adduction/abduction, inversion/eversion, nor the anteroposte-
rior translation of the construction axis showed a characteristic motion pattern, following that a ROM corresponding to a specific motion direction and time phase during the stance phase of gait could only be defined for dorsiflexion ($ROM_{dyn\ df}$) and plantarflexion ($ROM_{dyn\ pf}$).

The corresponding definitions of $ROM_{dyn\ pf}$ and $ROM_{dyn\ df}$, as well as the peak values of plantarflexion ($max_{dyn\ pf}$) and dorsiflexion ($max_{dyn\ df}$) are shown in Figure 18.

For adduction/abduction, inversion/eversion and the anteroposterior translation a maximal ROM ($ROM_{dyn\ adab}, ROM_{dyn\ invev}, ROM_{dyn\ t_{ap}}$), thus the range between the maximal and the minimal reached values at any time during stance were evaluated.

Figure 18: Definition of ROM for ankle dorsi-/plantarflexion. $ROM_{dyn\ pf}$: ROM between HS and first peak of plantarflexion ($max_{dyn\ pf}$), $ROM_{dyn\ df}$: ROM between first peak of plantarflexion ($max_{dyn\ pf}$) and peak of dorsiflexion ($max_{dyn\ df}$).
4.3 Results

4.3.1 Static Maximal ROM

A mean static maximal ROM around the z-axis under weightbearing of $\text{ROM}_{\text{stat pf df}} = 27.0° \pm 11.9°$ was found (see Table 8).

In the standing trials all the TAA subjects showed, relative to the neutral position of the implant components, a plantarflexed position of the talar relative to the tibial component (see Table 8). Starting from the ankle in position of the standing trial, sub 1, 2 and 3 showed at least $9°$ of functional available ROM in direction of dorsiflexion. Whereas sub 4 has only $1°$ more to go.

In direction of plantarflexion all subjects showed more than $10°$ of functional available ROM compared to the standing trial position.

The translation of the construction axis moved for all subjects less than $2.3 \text{ mm}$ in anteroposterior direction, when moving from a maximal plantarflexed to a maximal dorsiflexed position (see Figure 19).

For the static maximal ROM around the x-axis $\text{ROM}_{\text{stat inv ev}}$ the TAA subjects all showed less than $5°$ of rotation (see Table 8).

<table>
<thead>
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<th>Static Trials</th>
<th>Rotation around z-axis</th>
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<tr>
<td></td>
<td>$\text{max}_{\text{stat df}}$</td>
</tr>
<tr>
<td>sub 1</td>
<td>1.9°</td>
</tr>
<tr>
<td>sub 2</td>
<td>3.5°</td>
</tr>
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<td>sub 3</td>
<td>5.6°</td>
</tr>
<tr>
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<tr>
<td>mean ± SD</td>
<td>1.1° ± 5.4°</td>
</tr>
</tbody>
</table>

Table 8: $\text{ROM}_{\text{stat pf df}}$ around the z-axis between the maximal plantarflexed $\text{max}_{\text{stat pf}}$ and the maximal dorsiflexed $\text{max}_{\text{stat df}}$ static loaded positions. $\text{StandingTrial}$ shows the position of the implant components in the standing trial. $\text{ROM}_{\text{stat inv ev}}$ around the x-axis. Neutral is given by the initial position determined by definition of the implant coordinate system.

4.3.2 Pose of the Implant Components during Gait - Image Sequences

Three sequences of 3D reconstructed videofluoroscopic images are exemplarily shown in Figures 20, 21 and 22. Figure 20 shows the pose of the implant components of sub 3 during level gait, Figure 21 sub 2 during walking uphill and Figure 22 sub 1 during walking downhill.
Figure 19: Maximal static dorsiflexed (grey) and maximal static plantarflexed (red) position of the talar component and construct axis relative to the tibial component. Sagittal view. From left to right: sub 1, sub 2, sub 3, sub 4.
Figure 20: Sub 3 during the stance phase of level gait. Eight 3D reconstructed videofluoroscopic images. Frames 1 (HS), 2, 5, 8, 10, 12, 15 and 17.

Figure 21: Sub 2 during the stance phase of walking uphill. Eight 3D reconstructed videofluoroscopic images. Frames 1 (HS), 3, 6, 12, 15, 19, 22 and 25.

Figure 22: Sub 1 during the stance phase of walking downhill. Eight 3D reconstructed videofluoroscopic images. Frames 1 (HS), 3, 6, 11, 15, 18, 21 and 23.
4.3.3 Clinical Rotations and Motion of Talar Construction Axis during Gait

Clinical rotations as well as the anteroposterior translation of the talar construction axis of the whole stance phase of all four TAA subjects are shown in Figures 23 (gt), 25 (uph), 26 (dnh), 27 (ssb) and 28 (sst). Generally, the relative rotations between the talar and the tibial component showed large interindividual differences. The interindividual differences are not only present in the magnitude of rotation, but also in the motion pattern, especially concerning temporal variation.

During all gait tasks, main motion occurred around the z-axis, thus dorsi-/plantarflexion dominated. The TAA experienced only minimal inversion/eversion and adduction/abduction motion. And for neither of them a characteristic motion pattern was recognizable. Only the definitions of the maximal ROM $\text{ROM}_{\text{dyn adab}}$ and $\text{ROM}_{\text{dyn invev}}$ were possible, which were defined as the differences between the maximal and the minimal reached values occurring at any time point during stance. The mean $\text{ROM}_{\text{dyn adab}}$ and $\text{ROM}_{\text{dyn invev}}$ were $7.2^\circ$ and $3.1^\circ$ (sub 1), $4.4$ and $2.0^\circ$ (sub 2), $6.6^\circ$ and $2.2^\circ$ (sub 3) and $7.8^\circ$ and $3.1^\circ$ (sub 4), exemplarily provided for level gait.

For all other gait conditions, values are provided in Table 9. As well exemplarily for level gait, the variability showed magnitudes up to $\pm 4.1^\circ$ for adduction/abduction and up to $\pm 1.6^\circ$ for inversion/eversion during the stance phase. The magnitude of rotation around the x- and the y-axes were in the range of the variability. This proportion of small ROM to large variability around the x- and the y-axes was also valid for the other four gait conditions and can quite clearly be seen in Figures 23 to 28.

Adduction/abduction showed shifts between the subjects, which are due to the differences in transverse rotation of how the TAA components were implanted. In other words, this shifts show the misalignment of the anteroposterior axes of the talar and the tibial components. This was likewise seen in the standing trials and very nicely seen when looking at the orientation of the talar construction axes during level gait (Figure 24). E.g. sub 3 shows an internally rotated orientation of the talar component relative to the tibial component of around $12^\circ$.

The average translation of the talar construction axis in anteroposterior direction was for all gait conditions less than $3.4$ mm (sub 1), $4.6$ mm (sub 2), $2.8$ mm (sub 3), $1.7$ mm (sub 4) (see Table 9). The variability of the anteroposterior translation showed magnitudes during stance up to $\pm 1.0$ mm (sub 1), $\pm 1.7$ mm (sub 2), $\pm 0.8$ mm (sub 3), $\pm 0.7$ mm (sub 4). Again, the magnitudes were small and in the range of variability.

In Figure 24, the motion of the talar construction axis is shown during one typical level gait trial of each subject. The construction axis stays more or less parallel and as well shows only
little translation.

The predominant motion at the TAA was for all gait tasks dorsi-/plantarflexion. The motion graphs of all four subjects show the following characteristic motion pattern: after HS the TAA experiences a plantarflexion motion, followed by dorsiflexion and another shift in direction plantarflexion. Concerning the timing of the minimum, respectively maximum, sub 3 showed a different behavior. The minimum was for all subjects around 10% of stance, whereas sub 1, 2 and 4 had the maximum between 70-90% of stance, but sub 3 already around 40%.

Mean and SD of $ROM_{dyn\, pf}$ and $ROM_{dyn\, df}$ over all trials of each subject, as well as the mean values over all subjects are provided in Table 10. The interindividual differences are large showing values between 0.3° - 10.4° and 2.8° - 17.1° for $ROM_{dyn\, pf}$ and $ROM_{dyn\, df}$, respectively. Sub 4 showed the smallest, sub 3 the largest $ROM_{dyn\, pf}$ and $ROM_{dyn\, df}$, which is in accordance to the static maximal ROM $ROM_{stat\, pfdf}$. Comparing walking uphill to level gait, all subjects showed less $ROM_{dyn\, pf}$, as well as less $ROM_{dyn\, df}$. Walking downhill shows the tendency of an increase in $ROM_{dyn\, pf}$ compared to level gait. Neither for ssb, nor for sst a systematic change in ROM is present compared to level gait.

Interestingly, none of the subjects exploited the actual available static $ROM_{stat\, pfdf}$ (see Table 8) during the stance phase of all gait tasks. But sub 1 about reached its static maximal dorsiflexed position $max_{stat\, df}$ during uph, ssb and sst ($max_{dyn\, df} \approx max_{stat\, df}$). And sub 4 even exploits its $max_{stat\, df}$ in all conditions including level gait (see Table 8 and Figures 23 to 28).
Table 9: Mean and SD of all trials of each subject of $ROM_{dyn}$ $ abduction$, $ROM_{dyn}$ $ in dorsiflexion$ and $ROM_{dyn}$ $ t_{an} $ during the stance phase of all five gait tasks. Last row: mean and SD over all four subjects.

<table>
<thead>
<tr>
<th>$ROM_{dyn}$</th>
<th>$gt$</th>
<th>$m$</th>
<th>$df$</th>
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<th>$m$</th>
<th>$df$</th>
<th>$dnh$</th>
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<td>2.3 mm</td>
<td>5.0°</td>
<td>2.7°</td>
<td>1.7 mm</td>
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<td>2.5°</td>
<td>3.4 mm</td>
<td>8.8°</td>
<td>2.8°</td>
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<td>2.6°</td>
<td>3.2 mm</td>
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<td>± 0.5°</td>
<td>± 0.4 mm</td>
<td>± 1.2°</td>
<td>± 0.6°</td>
<td>± 0.4 mm</td>
<td>± 1.3°</td>
<td>± 0.2°</td>
<td>± 0.6 mm</td>
<td>± 4.7°</td>
<td>± 0.2°</td>
<td>± 0.8 mm</td>
<td>± 0.8°</td>
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<td>± 0.6 mm</td>
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<td>± 1.0°</td>
<td>± 1.1°</td>
<td>± 1.0 mm</td>
<td>± 4.8°</td>
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<td>± 0.7 mm</td>
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</tr>
<tr>
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<td>± 1.2°</td>
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<td>± 0.6°</td>
<td>± 0.3 mm</td>
<td>± 0.5°</td>
<td>± 0.8°</td>
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<td></td>
</tr>
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<td>2.0 mm</td>
<td>6.5°</td>
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<td>3.1 mm</td>
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<td>2.9 mm</td>
</tr>
<tr>
<td>± 1.5°</td>
<td>± 0.6°</td>
<td>± 1.2 mm</td>
<td>± 2.0°</td>
<td>± 0.5°</td>
<td>± 0.8 mm</td>
<td>± 1.0°</td>
<td>± 0.4°</td>
<td>± 1.1 mm</td>
<td>± 2.5°</td>
<td>± 0.4°</td>
<td>± 0.8 mm</td>
<td>± 1.1°</td>
<td>± 0.2°</td>
<td>± 1.3 mm</td>
<td></td>
</tr>
</tbody>
</table>

Table 10: Mean and SD of all trials of each subject of $ROM_{dyn}$ $ pf $ and $ ROM_{dyn}$ $ df $ during the stance phase of all five gait tasks. Last row: mean and SD over all four subjects.
Figure 23: Ankle kinematics during the stance phase of level gait. Mean and SD over all trials of each subject. sub 1 (black), sub 2 (blue), sub 3 (red), sub 4 (green). rel: relative to.

Figure 24: Motion of the talar construction axis relative to the tibial component during one typical level gait trial of each subject. From left to right: sub 1, sub 2, sub 3, sub 4. rel: relative to.
Figure 25: Ankle kinematics during the stance phase of walking uphill. Mean and SD over all trials of each subject. sub 1 (black), sub 2 (blue), sub 3 (red), sub 4 (green). rel: relative to.

Figure 26: Ankle kinematics during the stance phase of walking downhill. Mean and SD over all trials of each subject. sub 1 (black), sub 2 (blue), sub 3 (red), sub 4 (green). rel: relative to.
Figure 27: Ankle kinematics during the stance phase of side slope bottom. Mean and SD over 5 trials of each subject. sub 1 (black), sub 2 (blue), sub 3 (red), sub 4 (green). rel: relative to.

Figure 28: Ankle kinematics during the stance phase of side slope top. Mean and SD over all trials of each subject. sub 1 (black), sub 2 (blue), sub 3 (red), sub 4 (green). rel: relative to.
4.4 Discussion

4.4.1 Static Maximal ROM

The four TAA subjects showed large interindividual differences in their static maximal ROM around the z-axis \( \text{ROM}_{\text{stat}} \text{pfdf} \). However, all four subjects showed values that are comparable to radiographically assessed tibiotalar ROM of the Agility TAA (Coetzee and Castro, 2004) and of the STAR prosthesis (Zerahn and Kofoed, 2004), as well as to the fluoroscopy gained ROM of the Buechel-Pappas TAA (Komistek et al, 2000).

Due to the small magnitude of translation of the construction axis in anteroposterior direction one can conclude, that dorsi-/plantarflexion motion takes mainly place around the construction axis. The occurring motion was mainly rotation and only very little sliding.

The static maximal ROM around the x-axis \( \text{ROM}_{\text{stat}} \text{invev} \) was marginal. And since the polyethylene inlay and the talar component both show a double curved shape, it can be assumed that during this small amount of inversion/eversion motion the congruency can more or less be maintained and no increased risk of wear should arise.

4.4.2 Pose of the Implant Components during Gait - Image Sequences

Due to the inclination of the c-arm of the videofluoroscopy system, the view of the TAA is slightly from beneath. Only in the gait condition ssb, the orientation of the TAA was more or less perpendicular to the principle axis of the X-ray beam. Combined with the plane symmetric shape of the TAA components, the symmetry plane of the tibia was then close to parallel to the image plane, thereby especially the measurement of the tibial out of plane rotation around the y-axis was affected. In other words in this orientation, the projection shows a poor geometry for the 3D reconstruction, which appears in a bouncing over the symmetry plane.

The location of the TAA in the field of view of the image intensifier is approximately in the vertical center. On the horizontal axis, it moves from one side to the other. Dependent on the foot size, the TAA leaves the field of view. This limitation allowed only for sub 1 the whole stance phase to be tracked. TO was missing for sub 2, 3 and 4. Motion blur was only present around TO, but did not show a substantial impact on the 3D reconstruction.

4.4.3 Clinical Rotations and Motion of Talar Construction Axis during Gait

Dorsi-/plantarflexion was for all gait conditions and all subjects the dominating rotation occurring at the TAA. This is in agreement with the study of Leszko et al (2008), who investigated the Salto TAA during level gait and a step up task. Likewise matchable is the mean \( \text{ROM}_{\text{dyn}} \text{df} \)
during level gait of 10.4° in this study vs. 9.2° in the study of Leszko et al (2008). In this study a motion pattern consisting of two direction shifts during the stance phase was found, which is not in agreement to the study of Leszko et al (2008), in which a close to linear motion pattern for dorsi-/plantarflexion was seen. But, since in the study of Leszko et al (2008) only the four images (HS, 33%, 66%, TO) were analyzed, it can be assumed that this information got lost due to the insufficient sampling frequency.

Inversion/eversion as well as adduction/abduction motion was small but showed large variability, which has also been stated by Conti et al (2006). The small ROM were in the range of the variability. The very large variability of adduction/abduction in the ssb condition, especially for sub 2, can be explained by the unfavorable orientation of the tibial component relative to the principle axis of the X-ray beam: This orientation resulted in a poor condition for the 3D reconstruction, and therefore a poor accuracy. For future studies looking at TAA of a plane symmetric geometry, it need to be assured that the principle axis of the X-ray beam is not perpendicular to the symmetry plane of the TAA components. Thus, in future studies the c-arm inclination should be increased for the gait condition ssb.

Even though the analyzed TAA is of an unconstrained design, that would allow free translation in the transverse plane, the measured translation was small. This is in agreement with the findings of Conti et al (2006) and Leszko et al (2008). Furthermore, since the construction axis stayed more or less parallel and as well only showed little translation, one can conclude, that the main rotation occurred around the construction axis itself.

None of the subjects exploited the actual available $\text{ROM}_{\text{stat}}\ pfd\ f$ during the stance phase of all gait tasks. Thus, it could be assumed that the TAA had sufficient ROM and therefore did generally not limit gait. Though, sub 1 and 4 exploited their static maximal dorsiflexed position $\text{max}_{\text{stat}}\ df$ in several of the gait tasks. This could explain the flat motion graph of sub 1 during uph, as well as of sub 4 in all gait conditions. Here the assumption can be made that missing ROM in direction of dorsiflexion, thus a small $\text{max}_{\text{stat}}\ df$, partly limited sub 1 and in most cases limited sub 4. And since dnh compared to gt only showed an increase in $\text{max}_{\text{dyn}}\ pf$ and not $\text{max}_{\text{dyn}}\ df$, meaning requirement for a larger ROM in direction of plantarflexion, but not into direction of dorsiflexion, sub 1 did not show any limitation during walking downhill. This leads to the conclusion, that not only the magnitude of ROM determines if the TAA restricts gait, but also the region where it is situated. In other words, a large ROM does not give an advantage if it is restricted to one direction only.

Contrary to expectations, the ssb and the sst gait conditions did not provoke an instable behavior of the TAA. However, walking uphill seems to present a bigger challenge for the TAA,
especially concerning restriction due to limited ROM in direction of dorsiflexion (small $max_{stat \; df}$). Generally, it can be concluded, that the available amount of maximal dorsiflexion $max_{stat \; df}$ is crucial for the question if the TAA is restrictive to gait, whereas the available amount of maximal plantarflexion $max_{stat \; pf}$ seems to be sufficient for any of the analyzed gait tasks. The question which part, the TAA itself or the surrounding soft tissues, account for the restriction in dorsiflexion can not be answered and should be part of further investigations. Another research should be done investigating if the area of available ROM could e.g. be shifted in direction of dorsiflexion by implantation of the TAA in a more plantarflexed position of the talar relative to the tibial component.
4.5 Conclusions

A procedure was developed that enables estimation of TAA kinematics in vivo, without being limited by skin movement artefacts. Hereby separating motion at the TAA from motion at adjacent joints, namely the subtalar joint. This is not possible with skin marker tracking by reason of inaccessibility of the talus. The quantification of $ROM_{stat \ pfd\ df}$ is, due to the 3D reconstruction, not limited by the projection errors, which would be present in 2D radiographic assessments.

One limitation is the out of plane accuracy given by the single plane approach. An irremovable disadvantage of the videofluoroscopic approach is the radiation exposure of the test subject. This said, the whole test procedure only results in an exposure dose of 1.7% of the annual boundary level, which nevertheless should not be neglected. Another disadvantage is the limitation to the restricted field of view. For large foot sizes TO could not be evaluated. Not to be neglected is the fact that the whole analysis procedure is very time consuming.

In total, the presented procedure allows an objective analysis of the performance of TAA, which provides a better understanding of its functionality. With a larger number of analyzed TAA subjects it has the potential to help clinicians and implant developers to evaluate current TAA designs and future design modifications.

Two of the four subjects exploited their static maximal dorsiflexed position $max_{stat\ df}$ during some of the gait tasks, but none of the subjects got close to their static maximal plantarflexed position $max_{stat\ pf}$. It can be concluded, that the unavailability of ROM in direction of dorsiflexion ($max_{stat\ df}$) is the main problem resulting in a restriction during gait. The available ROM in direction of plantarflexion ($max_{stat\ pf}$) provided by the TAA seems to be sufficient.

The interindividual differences between the four subjects were large, however the TAA seemed not to restrict level gait. During walking uphill, the TAA was closer to possibilities and for two subjects restrictions were noticeable.

It was observed that dorsi-/plantarflexion dominated the motion at the ankle. Translation of the construction axis, as well as inversion/eversion and adduction/abduction were small and in the range of their variability. The construction axis showed only minor translation and change in its orientation. Thus, it could be concluded that the major rotation arises around the construction axis itself. This fact is favorable for the wear characteristics of the TAA, but if it allows a physiological role of the surrounding ligaments remains to be seen.
5 3D Kinematics of Healthy Ankles Using Tantalum Markers and Videofluoroscopy

5.1 Introduction

The healthy ankle joint has been intensively investigated in cadaver studies (see 2.4.2). In vivo and during gait, ankle kinematics was mainly analyzed using skin marker tracking, being limited by skin movement artefacts and the inaccessibility of the talus (see 2.4.4). Accurate in vivo gained dynamic gait data is very rare due to the invasiveness of the possible measurement approaches. In addition, due to different conventions for the description of ankle kinematics, comparisons between different studies are only limited possible.

This chapter aims to assess the 3D kinematics of calcaneus (calc), talus (tal) and tibia (tib) of a healthy ankle using the same measurement set up as for the TAA (see 4). For that purpose a procedure, in the following named tantalum approach, was developed and applied.

5.2 Methods

The tantalum approach analyzes internal foot bone movements by tracking bone embedded tantalum (Ta) markers using videofluoroscopy with the objective of improving the basic understanding of foot kinematics. The section methods includes 5.2.1 giving subject information, 5.2.2 concerning the measurement set up and the test procedure, 5.2.3 to 5.2.7 explaining the analysis procedure and 5.2.8 dealing with the corresponding error estimation. An overview of the main steps of the videofluoroscopic procedure of the tantalum approach is given in Figure 29.

![The whole videofluoroscopic procedure tantalum approach](image)

Figure 29: The main steps of the whole videofluoroscopic procedure of the tantalum approach.
5.2.1 Subjects

The two male subjects are healthy normals having Ta markers of a diameter of 0.8 mm implanted. Each bone has at least three embedded Ta markers. The implantation of the Ta markers was already performed for a previous study using roentgen stereophotogrammetry. Further information on the two subjects is given in Table 11.

<table>
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<tr>
<th>Subject</th>
<th>Age [Years]</th>
<th>Height [m]</th>
<th>Weight [kg]</th>
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<td>90</td>
<td>light-grey</td>
</tr>
<tr>
<td>sub 6</td>
<td>39</td>
<td>1.78</td>
<td>80</td>
<td>dark-grey</td>
</tr>
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</table>

Table 11: Information on healthy normal subjects

5.2.2 Measurement Set Up and Test Procedure

The measurement set up and the test procedure including calibration and the measurement itself is identical to the set up in the TAA approach and is described in 4.2.2 and 4.2.3.

5.2.3 Distortion Correction and Estimation of Projection Parameters

The estimation of the projection parameters of the fluoroscopic system, as well as the distortion correction of the fluoroscopic images is performed analogue to the TAA approach described in 4.2.4. A sequence of fluoroscopic images before distortion correction of sub 6 during the stance phase of walking uphill is provided in Figure 30. An example of a sequence of fluoroscopic images after distortion correction of sub 5 during the stance phase of level walking is given in Figure 31. The elimination of the distortion is best seen when comparing the distorted shape of the floor in Figure 30 to the straight floor in Figure 31.

5.2.4 Reconstruction of Ta Marker Locations - Segmentation of CT Images

For each bone (tibia, fibula, talus and calcaneus) the arrangement of the Ta markers, thus their relative position to each other, is 3D reconstructed using a CT data segmentation process in the software AMIRA (Mercury Computer Systems, Berlin, Germany). The appropriate image pixels were assigned by manual selection. The spatial image resolution of the CT images was 0.49 mm x 0.43 mm x 0.43 mm. Further processing included the data format conversion from .wrl to .stl using the software Geomagic (Geomagic Inc., North Carolina). Then, surface point clouds of the Ta markers were read into MATLAB (MathWorks, Massachusetts), where their centers and for each bone the distances between the centers of the Ta markers were calculated. The precision for the determination of the location of the centers of the Ta markers ($\sigma_{segm}$) was
Figure 30: Before distortion correction. Sub 6 during walking uphill. Six videofluoroscopic images of the stance phase. Frames 1 (HS), 3, 6, 11, 13, 16, 19, 22 and 25.

Figure 31: After distortion correction. Sub 5 during level walking. Six videofluoroscopic images of the stance phase. Frames 1 (HS), 3, 7, 13, 16 and 19.

estimated by repeating the whole procedure five times for all the 13 markers of one subject and resulted in a standard deviation of 0.07 mm.

For the definition of the joint coordinate system, as well as for visual purposes the geometry of the bones was also 3D reconstructed using AMIRA. The build in functions for intensity threshold and unconstrained smoothing were used.

Bone as well as marker arrangements of both subjects can be seen in Figure 32.

Figure 32: Arrangement of Ta markers sub 5 (left) and sub 6 (right), in each case side view and front view.
5.2.5 Ta Marker Tracking

The fluoroscopic images were imported into MATLAB, where the marker locations in the images were detected semi-automatically. The marker had to be selected manually and the center of each marker was then calculated by grey scale weighting of the surrounding pixels. The marker shows a diameter of about three to four pixels in the images (1 pixel is about 0.28 mm) (see Figure 33). The precision for the detection of the centers of the markers in the images ($\sigma_{\text{track}}$) was evaluated by repeated tracking (five times) of all 13 markers during all frames of one trial and resulted in a SD of 0.14 mm in x as well as in y direction.

![Figure 33: Ta marker in fluoroscopic image.](image)

5.2.6 3D Reconstruction - Three Point Spatial Resection Problem

**Analytic Solution**  With a minimum of three markers per bone, the pose of each bone is 3D defined. Using the distances between each bone’s markers provided by the CT image segmentation, the 3D position of the markers and thereby of the bones was reconstructed. The reconstruction algorithm, written in MATLAB, is numerically determining the analytic solution assuming a central projection and therefore solving a three point spatial resection problem (see Figure 34).

The origin of the fluoroscopic coordinate system is set in the focus of the fluoroscopic system. The image plane is oriented perpendicular to the z-axis and positioned with a focus distance
such that the image center has the coordinates \((0,0,f)\). Three marker points \(P_1, P_2\) and \(P_3\) are central projected on to the image plane \((P'_1, P'_2\) and \(P'_3)\). The image marker points \(P'_1, P'_2\) and \(P'_3\), as well as the distances between the three markers \(S_{12}, S_{13}\) and \(S_{23}\) are known from the Ta marker tracking and the reconstruction of the Ta marker locations, respectively (see 5.2.4 and 5.2.5). The unknown positions of the three marker points \(P_1, P_2\) and \(P_3\) relative to the fluoroscopic coordinate system were determined using following approach. \(P_i\) is located on the ray between the origin and \(P'_i\), thus for \(i = 1, 2, 3\)

\[
P_i = \lambda_i P'_i
\]

with

\[
P_i = \begin{pmatrix} x_i \\ y_i \\ z_i \end{pmatrix}, \quad P'_i = \begin{pmatrix} u_i \\ v_i \\ f \end{pmatrix}, \quad 0 < \lambda_i < 1
\]

thus

\[
\lambda_i = \frac{z_i}{f}
\]

The known side lengths \(S_{12}, S_{13}\) and \(S_{23}\) satisfy

\[
S^2_{12} = (x_2 - x_1)^2 + (y_2 - y_1)^2 + (z_2 - z_1)^2
\]

\[
S^2_{13} = (x_3 - x_1)^2 + (y_3 - y_1)^2 + (z_3 - z_1)^2
\]

\[
S^2_{23} = (x_3 - x_2)^2 + (y_3 - y_2)^2 + (z_3 - z_2)^2
\]

by substituting for \(x_i\) and \(y_i\) the following quadratic and multivariable system of equations with the unknowns \(z_1, z_2\) and \(z_3\) is obtained

\[
(fS_{12})^2 = A z_1^2 + 2D z_1 z_2 + B z_2^2
\]

\[
(fS_{13})^2 = A z_1^2 + 2E z_1 z_3 + C z_3^2
\]

\[
(fS_{23})^2 = B z_2^2 + 2F z_2 z_3 + C z_3^2
\]
with

\[
A = u_1^2 + v_1^2 + f^2 \\
B = u_2^2 + v_2^2 + f^2 \\
C = u_3^2 + v_3^2 + f^2 \\
D = u_1 u_2 + v_1 v_2 + f^2 \\
E = u_1 u_3 + v_1 v_3 + f^2 \\
F = u_2 u_3 + v_2 v_3 + f^2
\]

An analytic approach for solving this system of equations is found in Finsterwalder (1937). In this work a numeric approach was used (see Figure 35). A start value for $\lambda_1$ is set for $P_1$. $P_2$ and $P_3$ are then located on spheres with center point $P_1$ and radius $S_{12}$, respectively $S_{13}$. Also, the position has to be on the rays between $P_2$ and $P_2'$ or $P_3$ and $P_3'$, respectively. Thus $P_2$ and $P_3$ satisfy

\[
P_2 = \lambda_2 P_2' \\
P_3 = \lambda_3 P_3'
\]

and as well the ball equations.

\[
(P_2 - P_1)^2 = S_{12}^2 \\
(P_3 - P_1)^2 = S_{13}^2
\]

This leads to the following equation system:

\[
(\lambda_2 u_2 - \lambda_1 u_1)^2 + (\lambda_2 v_2 - \lambda_1 v_1)^2 + (\lambda_2 f - \lambda_1 f)^2 = S_{12}^2 \tag{1} \\
(\lambda_3 u_3 - \lambda_1 u_1)^2 + (\lambda_3 v_3 - \lambda_1 v_1)^2 + (\lambda_3 f - \lambda_1 f)^2 = S_{13}^2 \tag{2}
\]

The two quadratic equations 1 and 2 lead to two solutions for each factor $\lambda_2$ and $\lambda_3$, which gives four possible solutions of the distance in-between ($S_{23}(\lambda_1)$) (see Figure 35).

The four possible solutions computed for $\lambda_1 = 0...1$ for an example projection are shown in Figure 36. Two geometrical possible solutions for $\lambda_1$ are obtained that solve the known distance $S_{23}$.

With the use of additional constraints, given by the anatomical possible positions of the bones
Figure 34: Three point spatial resection. $P_x$ marker points, $P'_x$ image marker points, $f$ focus distance, $S_{xy}$ distance between markers.

relative to each other, the two solutions were cut down to one unique analytic solution.

**Redundant Solution** For bones, that include more than three Ta markers, thus a redundant marker point cloud, the location of the marker points were optimized using the information of all available markers. The analytic solution was used as the start value for the least-squares optimization. The three rotations of the marker point cloud, as well as the position of the center of the marker point cloud were optimized, such that the distances between the projected marker points and the tracked marker points were minimized. This interim solution is named redundant solution.

**Out of Plane Fitted Solution** Due to the single plane approach, the existing errors induced by the process of Ta marker tracking (see 5.2.5) and the segmentation process (see 5.2.4) the out of plane translational error was large and had to be corrected (see 5.2.8). Consequently, the out of plane location of the marker points were forced on a motion path (out of plane shift $\Delta z$). The motion path was determined by a linear regression fit of the $z$-coordinates of the centers of the marker point clouds during the whole stance phase.

Using the out of plane shifted marker point cloud as start value, the three rotations of the marker point cloud and the in plane translations of the center of the marker point cloud were again
Figure 35: Numeric approach to solve the three point spatial resection problem, illustrated in 2D. Location of $P_1$ is numerically searched, such that one of the four possible distances between the intersection points of the two spheres with radius $S_{12}$ and $S_{13}$ and the respective rays of the projection, equals the distance $S_{23}$.

Figure 36: Two solutions of the three point spatial resection problem.
optimized in a least-squares sense. The translational boundaries of the optimization routine were set, such that the projection of the maximal shift of the center of the point cloud was not bigger than $\sigma_{\text{track}}$.

**Pose of the Bones** With the information of the out of plane fitted solution, the pose of the marker point clouds and the pose of the bones is determined. A sequence of images of sub 5 during walking uphill is exemplarily provided in Figure 37.

### 5.2.7 Estimation of Relative Rotations and Translations

Corresponding orientation matrices and position vectors ($R_{\text{tib}}, t_{\text{tib}}, R_{\text{tal}}, t_{\text{tal}}, R_{\text{calc}}, t_{\text{calc}}$) were determined using a least-squares fit of point clouds (Gander and Hrebicek, 1997). The used reference point clouds were determined by the standing trial, performed in a natural upright position. Hence, the neutral position (0° rotation) was defined by the standing trial.

The relative motion between calcaneus and talus (talocalcaneal motion), as well as between talus and tibia (tibiotalar motion) were estimated from the distal relative to the proximal segment. The computation was performed according to the TAA approach as described in 4.2.6.

**Clinical Rotations** For the description of the clinical rotations the attitude vector is decomposed along the axes of the joint coordinate system (see 4.2.6). The same coordinate system is used for tibiotalar and for talocalcaneal motion.
The joint coordinate system is anatomically defined using the geometry of the bones 3D reconstructed in the CT images (see 5.2.4) and transformed into their pose in the standing trial. The mediolateral axis ($e_2$) is defined as the cylinder axis of the cylinder, which is fitted to the upper surface of the talus. The vertical axis $e_1$ lies in the plane spanned by the lateral and the medial malleoli and the center of the top part of the tibia. It is directed perpendicular to the mediolateral axis. The posteroanterior axis $e_3$ is perpendicular to $e_1$ and $e_2$. Dorsi-/plantarflexion is described as the rotation around the $e_2$-axis, inversion/eversion is described with respect to the $e_3$-axis and adduction/abduction is equivalent to rotation around the $e_1$-axis (Figure 38).
Motion of the Talar Cylinder Axis  Anteroposterior translation of the talus relative to the tibia was analyzed as translation of the cylinder axis (see Figure 38) with respect to the $e_3$-axis of the joint coordinate system (analogous to the TAA approach in 4.2.6).

Stance Phase  Analogous to the TAA approach, the stance phase, thus the time frame from HS to TO, was evaluated and normalized to 0-100% of stance (see 4.2.6).

ROM  For the static trials, a maximal ROM around the $e_2$-axis ($ROM_{stat \, pfdf}$) between the maximal plantarflexed ($max_{stat \, pf}$) and the maximal dorsiflexed ($max_{stat \, df}$) static loaded position was estimated.

The definitions of the ROM during gait for plantarflexion ($ROM_{dyn \, pf}$) and dorsiflexion ($ROM_{dyn \, df}$), as well as the maximal values of plantarflexion ($max_{dyn \, pf}$) and dorsiflexion ($max_{dyn \, df}$) are defined analogous to the TAA approach and are shown in Figure 18.

Also identical to the TAA approach, for tibiotalar adduction/abduction, inversion/eversion and anteroposterior translation the maximal ROM (tibiotalar $ROM_{dyn \, adab}$, tibiotalar $ROM_{dyn \, invev}$, tibiotalar $ROM_{dyn \, tap}$), thus the range between the maximal and the minimal reached values at any time during stance were estimated.

Likewise for talocalcaneal motion, for dorsi-/plantarflexion, adduction/abduction and inversion/eversion the maximal ROM (talocalcaneal $ROM_{dyn \, pfdf}$, talocalcaneal $ROM_{dyn \, adab}$, talocalcaneal $ROM_{dyn \, invev}$) between the maximal and the minimal reached values at any time during stance were estimated.

Mean and SD  Mean and SD over all trials was only computed in those time frames that showed at least two data points.

5.2.8 Error Analysis

There are two sources of error influencing the 3D reconstruction, namely the error induced by the tracking precision ($\sigma_{track}$) and the one induced by the precision of segmentation ($\sigma_{segm}$) (see Tables 12 and 16).

Methods of the Error Estimation  For the error analysis of both sources of error, one frame of a gait trial (around midstance) of sub 5 was used as start value of the marker points. By computing a virtual projection of those marker points, the start values of the image marker positions were determined.

For the analysis of the error induced by $\sigma_{track}$, one SD of the tracking precision $\sigma_{track}$ (Table
12) was applied to each image marker position resulting in 30 randomly distributed noisy image marker positions for each marker.

For the analysis of the error induced by $\sigma_{\text{segm}}$, one SD of the respective segmentation precision $\sigma_{\text{segm}}$ (Table 16) was applied in x, y, and z-direction to each marker point resulting in 30 randomly distributed distances between the markers of each bone.

In both cases, the noisy data was computed using the algorithm described in 5.2.6 resulting in 30 solutions for each bone. The resulting error was then computed as the SD over these 30 solutions. For both sources of errors the following errors were estimated:

- For the analytic and the redundant solution of the algorithm, the orientation and position errors ($\sigma_{\text{Rot}}, \sigma_{\text{Trans}}$) of each bone with respect to the fluoroscopy coordinate system (in plane: x, y, out of plane: z).
- For the analytic and the redundant solution of the algorithm, the rotational errors of the tibiotalar, as well as talocalcaneal motion with respect to the joint coordinate system ($e_1, e_2, e_3$). And the translational error of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.
- The orientation and position errors of each bone induced by the out of plane shift $\Delta z$. Thus, these errors show, how much the shifting of the marker points in out of plane direction influences the orientation and the in plane positions of the marker points.
- For the out of plane fitted solution of the algorithm, the orientation and position errors of each bone with respect to the fluoroscopy coordinate system.
- For the out of plane fitted solution of the algorithm, the rotational errors of the tibiotalar, as well as talocalcaneal motion with respect to the joint coordinate system. And the translational error of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.

**Results of the Error Estimation - Error Induced by $\sigma_{\text{track}}$**

- The resulting SD for the analytic solution of each bone is provided in Table 13. The rotational error $\sigma_{\text{Rot}}(\sigma_{\text{track}})$ is smaller than 0.5° for in plane and 1.1° for out of plane. The translational error $\sigma_{\text{Trans}}(\sigma_{\text{track}})$ is smaller than 0.6 mm for in plane and 8.6 mm for out of plane. The translational out of plane error is very large, which is also evident in Figure 39.

The joint rotations show rotational errors smaller than 0.6° in plane and 1.4° out of plane.
Figure 39: Analytic solution influenced by $\sigma_{track}$. Positions of the centers of each bones marker point cloud for the 30 3D reconstructed noisy images. From left to right: x-, y-, z-coordinate.

The anteroposterior translation of the cylinder axis shows a SD of 1.4 mm.

For the tibia, consisting of a redundant marker point cloud, the redundant solution is as well given. However the improvement in accuracy due to the redundancy is small.

- The errors induced by the out of plane shift $\Delta z$ are provided in Table 14. The applied out of plane shift resulted in a rotational error smaller than 0.6° and in a translational error smaller than 0.2 mm.

- The resulting SD for the out of plane fitted solution of each bone is provided in Table 15. The rotational error $\sigma_{Rot}(\sigma_{track})$ is smaller than 0.5° for in plane and 0.9° for out of plane. The translational error $\sigma_{Trans}(\sigma_{track})$ is smaller than 0.5 mm for in plane and 1.7 mm for out of plane.

The joint rotations show rotational errors smaller than 0.6° in plane and 1.2° out of plane. The anteroposterior translation of the cylinder axis shows an error of 0.4 mm.

**Results of the Error Estimation - Error Induced by $\sigma_{segm}$**

- The resulting SD for the analytic solution of each bone is provided in Table 17. The rotational error $\sigma_{Rot}(\sigma_{segm})$ is smaller than 0.3°. The translational error $\sigma_{Trans}(\sigma_{segm})$ is smaller than 0.2 mm for in plane and 3.2 mm for out of plane. The translational out of plane error is the largest, which is also evident in Figure 40. However, compared to the error induced by $\sigma_{track}$ it is of minor importance.

The joint rotations show rotational errors smaller than 0.7°. The anteroposterior translation of the cylinder axis shows an error of 0.3 mm.

For the tibia, consisting of a redundant marker point cloud, the redundant solution shows a clear improvement of the accuracy compared to the analytic solution.
Figure 40: Analytic solution influenced by $\sigma_{\text{segm}}$. Positions of the centers of each bone marker point cloud for the 30 noisy input distances between the marker points. From left to right: x-, y-, z-coordinate.

- The errors induced by the out of plane shift $\Delta z$ are provided in Table 18. The applied out of plane shift resulted in a rotational error smaller than 0.2° and in a translational error smaller than 0.1 mm.

- The resulting errors for the out of plane fitted solution of each bone is provided in Table 19. The rotational error $\sigma_{\text{Rot}}(\sigma_{\text{segm}})$ is smaller than 0.2°. The translational error $\sigma_{\text{Trans}}(\sigma_{\text{segm}})$ is smaller than 0.1 mm for in plane and 0.8 mm for out of plane.

The joint rotations show rotational errors smaller than 0.3°. The anteroposterior translation of the cylinder axis shows an error of 0.1 mm.
Table 12: Tracking precision $\sigma_{\text{track}}$ evaluated by repeated tracking (five times) of all markers in each bone during all frames of one trial.

<table>
<thead>
<tr>
<th>Bone</th>
<th>$\sigma_{\text{track}}$ [mm]</th>
<th>x direction</th>
<th>y direction</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>0.11</td>
<td>0.09</td>
<td></td>
</tr>
<tr>
<td>Talus</td>
<td>0.19</td>
<td>0.24</td>
<td></td>
</tr>
<tr>
<td>Calcaneus</td>
<td>0.11</td>
<td>0.09</td>
<td></td>
</tr>
</tbody>
</table>

Table 13: Error analysis for $\sigma_{\text{track}}$. SD for the analytic/redundant solutions. For each bone orientation and position errors ($\sigma_{\text{Rot}}, \sigma_{\text{Trans}}$) with respect to the fluoroscopy coordinate system (in plane: x, y, out of plane: z). For each joint, rotational errors ($\sigma_{\text{Rot}}$) with respect to the joint coordinate system ($e_1, e_2, e_3$) and the translational error ($\sigma_{\text{Trans}}$) of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.

<table>
<thead>
<tr>
<th>Bone</th>
<th>$\sigma_{\text{Rot}}(\Delta z)$ [°]</th>
<th>$\sigma_{\text{Trans}}(\Delta z)$ [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>0.10/0.10</td>
<td>0.16/0.10</td>
</tr>
<tr>
<td>Talus</td>
<td>1.12</td>
<td>0.28/0.15</td>
</tr>
<tr>
<td>Calcaneus</td>
<td>0.71</td>
<td>0.25/0.22</td>
</tr>
<tr>
<td>Joint</td>
<td>$e_1$</td>
<td>$e_2$</td>
</tr>
<tr>
<td>Tibiotalar</td>
<td>1.22/1.15</td>
<td>0.56/0.54</td>
</tr>
<tr>
<td>Talocalcaneal</td>
<td>1.38</td>
<td>0.57</td>
</tr>
</tbody>
</table>

Table 14: Error analysis for $\sigma_{\text{track}}$. Errors induced by the out of plane shift $\Delta z$.

<table>
<thead>
<tr>
<th>Bone</th>
<th>$\sigma_{\text{Rot}}(\Delta z)$ [°]</th>
<th>$\sigma_{\text{Trans}}(\Delta z)$ [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibia</td>
<td>0.04</td>
<td>0.01</td>
</tr>
<tr>
<td>Talus</td>
<td>0.52</td>
<td>0.02</td>
</tr>
<tr>
<td>Calcaneus</td>
<td>0.59</td>
<td>0.07</td>
</tr>
</tbody>
</table>

Table 15: Error analysis for $\sigma_{\text{track}}$. SD for the out of plane fitted solution. For each bone orientation and position errors ($\sigma_{\text{Rot}}, \sigma_{\text{Trans}}$) with respect to the fluoroscopy coordinate system (in plane: x, y, out of plane: z). For each joint, rotational errors ($\sigma_{\text{Rot}}$) with respect to the joint coordinate system ($e_1, e_2, e_3$) and the translational error ($\sigma_{\text{Trans}}$) of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.
Table 16: CT segmentation $\sigma_{segm}$ precision evaluated by repeated segmentation (five times) of all 13 markers of one subject.

<table>
<thead>
<tr>
<th>Bone</th>
<th>$\sigma_{Rot}(\sigma_{segm})$ $\sigma_{Trans}(\sigma_{segm})$</th>
<th>$\sigma_{Rot}(\sigma_{segm})$ $\sigma_{Trans}(\sigma_{segm})$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>x</td>
<td>y</td>
</tr>
<tr>
<td>Tibia</td>
<td>0.13/0.06</td>
<td>0.12/0.06</td>
</tr>
<tr>
<td>Talus</td>
<td>0.13</td>
<td>0.14</td>
</tr>
<tr>
<td>Calcaneus</td>
<td>0.29</td>
<td>0.10</td>
</tr>
</tbody>
</table>

Table 17: Error analysis for $\sigma_{segm}$. SD for the analytic/redundant solutions. For each bone orientation and position errors ($\sigma_{Rot}$, $\sigma_{Trans}$) with respect to the fluoroscopy coordinate system (in plane: x, y, out of plane: z). For each joint, rotational errors ($\sigma_{Rot}$) with respect to the joint coordinate system ($e_1$, $e_2$, $e_3$) and the translational error ($\sigma_{Trans}$) of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.

<table>
<thead>
<tr>
<th>Bone</th>
<th>$\sigma_{Rot}(\Delta z)$ $\sigma_{Trans}(\Delta z)$</th>
<th>$\sigma_{Rot}(\Delta z)$ $\sigma_{Trans}(\Delta z)$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>x</td>
<td>y</td>
</tr>
<tr>
<td>Tibia</td>
<td>0.02</td>
<td>0.01</td>
</tr>
<tr>
<td>Talus</td>
<td>0.11</td>
<td>0.13</td>
</tr>
<tr>
<td>Calcaneus</td>
<td>0.22</td>
<td>0.08</td>
</tr>
</tbody>
</table>

Table 18: Error analysis for $\sigma_{segm}$. Errors induced by the out of plane shift $\Delta z$.

<table>
<thead>
<tr>
<th>Bone</th>
<th>$\sigma_{Rot}(\sigma_{segm})$ $\sigma_{Trans}(\sigma_{segm})$</th>
<th>$\sigma_{Rot}(\sigma_{segm})$ $\sigma_{Trans}(\sigma_{segm})$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>x</td>
<td>y</td>
</tr>
<tr>
<td>Tibia</td>
<td>0.13</td>
<td>0.14</td>
</tr>
<tr>
<td>Talus</td>
<td>0.14</td>
<td>0.09</td>
</tr>
<tr>
<td>Calcaneus</td>
<td>0.27</td>
<td>0.34</td>
</tr>
</tbody>
</table>

Table 19: Error analysis for $\sigma_{segm}$. SD for the out of plane fitted solution. For each bone orientation and position errors ($\sigma_{Rot}$, $\sigma_{Trans}$) with respect to the fluoroscopy coordinate system (in plane: x, y, out of plane: z). For each joint, rotational errors ($\sigma_{Rot}$) with respect to the joint coordinate system ($e_1$, $e_2$, $e_3$) and the translational error ($\sigma_{Trans}$) of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.
**Total Error**  The total error for the clinical rotations ($\sigma_{\text{Rot}}(\sigma_{\text{track}}, \sigma_{\text{segm}})$) and the anteroposterior translation of the cylinder axis ($\sigma_{\text{Trans}}(\sigma_{\text{track}}, \sigma_{\text{segm}})$) is derived according to Gauss as follows:

$$\sigma_{\text{Rot}}(\sigma_{\text{track}}, \sigma_{\text{segm}}) = \sqrt{\sigma_{\text{Rot}}^2(\sigma_{\text{track}}) + \sigma_{\text{Rot}}^2(\sigma_{\text{segm}})}$$

$$\sigma_{\text{Trans}}(\sigma_{\text{track}}, \sigma_{\text{segm}}) = \sqrt{\sigma_{\text{Trans}}^2(\sigma_{\text{track}}) + \sigma_{\text{Trans}}^2(\sigma_{\text{segm}})}$$

The resulting total error for the clinical rotations was smaller than 0.6° for in plane and 1.3° for out of plane, respectively. The error of the translation of the cylinder axis in anteroposterior direction was about 0.4 mm (see Table 20).

Summarized, the translational out of plane error was very large and therefore out of plane translation had to be restricted. The influence of the error induced by the tracking process is larger than the error induced by the segmentation process. $\sigma_{\text{track}}$ is in the subpixel range. Thus, to improve the tracking precision, a higher resolution of the fluoroscopic image would be required.

<table>
<thead>
<tr>
<th>Joint</th>
<th>$\sigma_{\text{Rot}}(\sigma_{\text{track}}, \sigma_{\text{segm}})$ [°]</th>
<th>$\sigma_{\text{Trans}}(\sigma_{\text{track}}, \sigma_{\text{segm}})$ [mm]</th>
</tr>
</thead>
<tbody>
<tr>
<td>Tibiotalar</td>
<td>$e_1$ 0.76, $e_2$ 0.62, $e_3$ 0.97</td>
<td>0.36</td>
</tr>
<tr>
<td>Talocalcaneal</td>
<td>$e_1$ 0.89, $e_2$ 0.59, $e_3$ 1.31</td>
<td>-</td>
</tr>
</tbody>
</table>

Table 20: Error analysis including the influence of $\sigma_{\text{track}}$ and $\sigma_{\text{segm}}$. SD for the out of plane fitted solution. For each joint, rotational errors ($\sigma_{\text{Rot}}$) with respect to the joint coordinate system ($e_1$, $e_2$, $e_3$) and the translational error ($\sigma_{\text{Trans}}$) of the translation of the cylinder axis in anteroposterior direction of the joint coordinate system.
5.3 Results

5.3.1 Static Maximal ROM

The static maximal ROM $\text{ROM}_{\text{stat}}$ $\text{pf} df$ around the $e_2$-axis of sub 5 and sub 6 was $43.2^\circ$ and $62.0^\circ$, respectively (see Table 21).

<table>
<thead>
<tr>
<th>Static Trials</th>
<th>Rotation around $e_2$-axis</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>$\text{max}_{\text{stat}}$ $\text{df}$</td>
</tr>
<tr>
<td>sub 5</td>
<td>$25.9^\circ$</td>
</tr>
<tr>
<td>sub 6</td>
<td>$33.6^\circ$</td>
</tr>
</tbody>
</table>

Table 21: ROM $\text{ROM}_{\text{stat}}$ $\text{pf} df$ around the $e_2$-axis between the maximal plantarflexed $\text{max}_{\text{stat}}$ $\text{pf}$ and the maximal dorsiflexed $\text{max}_{\text{stat}}$ $\text{df}$ static loaded positions. Neutral is given by the standing trial.

5.3.2 Clinical Rotations and Motion of the Cylinder Axis during Gait

Since the out of plane location of the bones had to be restricted and the errors were still considerable, the following data had to be analyzed with caution. But tendencies could still be determined and are presented and discussed in the following paragraphs.

Tibiotalar and talocalcaneal rotations as well as the anteroposterior translation of the cylinder axis of the whole stance phase of both healthy normal subjects are shown in Figures 41 and 42 (gt), 43 and 44 (uph), 45 and 46 (dnh), 47 and 48 (ssb) and 49 and 50 (sst).

Tibiotalar motion showed the expected predominance of dorsi-/plantarflexion during all gait conditions. The characteristics of the dorsi-/plantarflexion motion pattern were comparable across the two subjects. An initial plantarflexion until about 10% stance was followed by a dorsiflexion movement and then again plantarflexion. This pattern was consistent for all gait conditions for sub 5. Whereas sub 6 showed during gt, ssb and sst a discontinuity in dorsiflexion approximately around 40% stance with a small change of movement direction to plantarflexion and then again dorsiflexion. The timing of changes in the movement direction differed between the gait conditions. Especially during uphill both subjects reached the maximum in dorsiflexion $\text{max}_{\text{dyn}}$ $\text{df}$ earlier compared to the other gait conditions. Compared to level gait, during uphill a larger $\text{max}_{\text{dyn}}$ $\text{df}$, but a smaller $\text{max}_{\text{dyn}}$ $\text{pf}$ occurred. During downhill both subjects showed an increase in $\text{max}_{\text{dyn}}$ $\text{df}$ and $\text{max}_{\text{dyn}}$ $\text{pf}$ compared to level gait. And also during ssb and sst both subjects showed an increased $\text{max}_{\text{dyn}}$ $\text{df}$.

The mean tibiotalar $\text{ROM}_{\text{dyn}}$ $\text{df}$ and $\text{ROM}_{\text{dyn}}$ $\text{pf}$ around the mediolateral axis is presented in Table 22. Compared to level gait, except for uph, in all gait conditions $\text{ROM}_{\text{dyn}}$ $\text{pf}$ as well as $\text{ROM}_{\text{dyn}}$ $\text{df}$ were increased.
The maximal rotations \( \text{max}_{\text{dyn}} \, df \) and \( \text{max}_{\text{dyn}} \, pf \) were in all gait conditions considerable smaller than the static maximal rotations \( \text{max}_{\text{stat}} \, df \) and \( \text{max}_{\text{stat}} \, pf \). Thus, neither the static maximal reached dorsiflexion \( \text{max}_{\text{stat}} \, df \), nor the static maximal reached plantarflexion \( \text{max}_{\text{stat}} \, pf \) were exploited during any of the gait conditions.

Tibiotalar adduction/abduction, inversion/eversion and anteroposterior translation of the cylinder axis showed ROM between maximal and minimal reached values (see Table 23) that were in the range of their variability. E.g. during level gait mean ranges of rotation for adduction/abduction were 5.4° and 3.1°, for inversion/eversion 4.4° and 2.7°, for anteroposterior translation 3.5 mm and 2.3 mm for sub 5 and 6, respectively. And the corresponding variabilities during stance were up to ± 2.7° and ± 3.8°, ± 2.6° and ± 3.5°, ± 3.1 mm and ± 0.9 mm, respectively.

Talocalcaneal motion showed for all gait conditions and both subjects, except inversion/eversion for sub 6, mean ROM below 7° (see Table 24) and maximal variabilities during stance in the range of ± 1.0° up to ± 8.0°. For sub 6 a mean talocalcaneal \( ROM_{\text{dyn}} \, \text{invev} \) of up to 11.5° occurred during walking uphill. Overall sub 6 showed more or less in all gait conditions, but mainly pronounced during gt and uph, gradual inversion at the talocalcaneal joint throughout stance. However, in general talocalcaneal motion was small and in the range of its variability. Dorsi-/plantarflexion at the talocalcaneal joint was small. To sum up, the tendency is apparent, that the subtalar joint rotates more during the second half of the stance phase and inversion/eversion is slightly prevailing.

<table>
<thead>
<tr>
<th>( ROM_{\text{dyn}} )</th>
<th>( \text{gt} )</th>
<th>( \text{uph} )</th>
<th>( \text{dnh} )</th>
<th>( \text{ssb} )</th>
<th>( \text{sst} )</th>
</tr>
</thead>
<tbody>
<tr>
<td>( pf )</td>
<td>( df )</td>
<td>( pf )</td>
<td>( df )</td>
<td>( pf )</td>
<td>( df )</td>
</tr>
<tr>
<td>sub 5</td>
<td>3.4°</td>
<td>± 3.0°</td>
<td>12.8°</td>
<td>± 3.6°</td>
<td>2.0°</td>
</tr>
<tr>
<td>sub 6</td>
<td>7.3°</td>
<td>± 4.1</td>
<td>10.3°</td>
<td>± 0.4</td>
<td>9.8°</td>
</tr>
</tbody>
</table>

Table 22: Mean and SD of \( ROM_{\text{dyn}} \, pf \) and \( ROM_{\text{dyn}} \, df \) of all trials of the two healthy normal subjects during the stance phase of all five gait tasks. *1 only for one trial data available. - : no data available.
<table>
<thead>
<tr>
<th>Tibiotalar ROM&lt;sub&gt;dyn&lt;/sub&gt;</th>
<th>gt adab</th>
<th>innev</th>
<th>t&lt;sub&gt;ap&lt;/sub&gt;</th>
<th>uph adab</th>
<th>innev</th>
<th>t&lt;sub&gt;ap&lt;/sub&gt;</th>
<th>dnh adab</th>
<th>innev</th>
<th>t&lt;sub&gt;ap&lt;/sub&gt;</th>
<th>ssb adab</th>
<th>innev</th>
<th>t&lt;sub&gt;ap&lt;/sub&gt;</th>
<th>sst adab</th>
<th>innev</th>
<th>t&lt;sub&gt;ap&lt;/sub&gt;</th>
</tr>
</thead>
<tbody>
<tr>
<td>sub 5</td>
<td>5.4°</td>
<td>± 1.1°</td>
<td>4.4° ± 1.5°</td>
<td>3.5 mm</td>
<td>± 2.8°</td>
<td>4.4° ± 1.1°</td>
<td>3.7 mm</td>
<td>± 0.9°</td>
<td>4.0° ± 0.6°</td>
<td>± 0.9 mm</td>
<td>± 3.1°</td>
<td>3.9° ± 1.8°</td>
<td>± 1.6 mm</td>
<td>± 2.1°</td>
<td>4.9° ± 0.9°</td>
</tr>
<tr>
<td>sub 6</td>
<td>3.1°</td>
<td>± 1.1°</td>
<td>2.7° ± 1.3°</td>
<td>2.3 mm</td>
<td>± 1.2°</td>
<td>3.8° ± 3.1°</td>
<td>3.6 mm</td>
<td>± 1.4°</td>
<td>4.8° ± 1.3°</td>
<td>± 2.6 mm</td>
<td>± 1.4°</td>
<td>1.4° ± 0.9°</td>
<td>± 1.0 mm</td>
<td>± 2.0°</td>
<td>5.1° ± 1.2°</td>
</tr>
</tbody>
</table>

Table 23: Mean and SD of all trials of each subject of tibiotalar ROM<sub>dy</sub> adab, tibiotalar ROM<sub>dy</sub> innev and tibiotalar ROM<sub>dy</sub> t<sub>ap</sub> during the stance phase of all five gait tasks.

<table>
<thead>
<tr>
<th>Talocalcaneal ROM&lt;sub&gt;dy&lt;/sub&gt;</th>
<th>pf df</th>
<th>adab</th>
<th>innev</th>
<th>uph pf df</th>
<th>adab</th>
<th>innev</th>
<th>dnh pf df</th>
<th>adab</th>
<th>innev</th>
<th>ssb pf df</th>
<th>adab</th>
<th>innev</th>
<th>sst pf df</th>
<th>adab</th>
<th>innev</th>
</tr>
</thead>
<tbody>
<tr>
<td>sub 5</td>
<td>3.7°</td>
<td>± 0.9°</td>
<td>5.0° ± 2.0°</td>
<td>4.9°</td>
<td>± 0.8°</td>
<td>4.5° ± 0.7°</td>
<td>4.4°</td>
<td>± 1.5°</td>
<td>5.5° ± 1.2°</td>
<td>± 2.2°</td>
<td>± 1.4°</td>
<td>5.4°</td>
<td>± 2.5°</td>
<td>± 1.0°</td>
<td>4.9° ± 0.9°</td>
</tr>
<tr>
<td>sub 6</td>
<td>4.7°</td>
<td>± 2.5°</td>
<td>7.9° ± 2.1°</td>
<td>3.6°</td>
<td>± 4.2°</td>
<td>5.3° ± 0.6°</td>
<td>11.5°</td>
<td>± 2.3°</td>
<td>6.1° ± 3.1°</td>
<td>± 3.7°</td>
<td>± 4.2°</td>
<td>7.5°</td>
<td>± 1.6°</td>
<td>± 3.1°</td>
<td>4.9° ± 0.7°</td>
</tr>
</tbody>
</table>

Table 24: Mean and SD of all trials of each subject of talocalcaneal ROM<sub>dy</sub> pf df, talocalcaneal ROM<sub>dy</sub> adab and talocalcaneal ROM<sub>dy</sub> innev during the stance phase of all five gait tasks.
Figure 41: Tibiotalar kinematics during the stance phase of level gait. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.

Figure 42: Talocalcaneal kinematics during the stance phase of level gait. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.
Figure 43: Tibiotalar kinematics during the stance phase of walking uphill. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.

Figure 44: Talocalcaneal kinematics during the stance phase of walking uphill. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.
Figure 45: Tibiotalar kinematics during the stance phase of walking downhill. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.

Figure 46: Talocalcaneal kinematics during the stance phase of walking downhill. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.
Figure 47: Tibiotalar kinematics during the stance phase of side slope bottom. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.

Figure 48: Talocalcaneal kinematics during the stance phase of side slope bottom. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.
Figure 49: Tibiotalar kinematics during the stance phase of side slope top. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.

Figure 50: Talocalcaneal kinematics during the stance phase of side slope top. Mean and SD over all trials of each healthy subject. sub 5 (light-grey), sub 6 (dark-grey). rel: relative to.
5.4 Discussion

5.4.1 Static Maximal ROM

The difference between $ROM_{stat \, pf \, df}$ of the two subjects was pretty large. This is in agreement to other authors, stating that intersubject variations at the talocrural joint are large (De Asla et al, 2006; Lundberg et al, 1989a). However, both subjects showed $ROM_{stat \, pf \, df}$ that are comparable to 2D fluoroscopic data presented by Komistek et al (2000) and the 3D dual plane fluoroscopic assessed data by De Asla et al (2006). The study of Kadakia et al (2008) showed a larger mean ROM, but due to the fact that tibiopedal motion was measured, this is explicable. Comparing $max_{stat \, pf}$ and $max_{stat \, df}$ to the position of the ankle in the standing trial, both subjects showed a larger available ROM in direction of dorsiflexion, than in direction of plantarflexion. However this is strongly dependent on the position of the standing trial.

5.4.2 Clinical Rotations and Motion of the Cylinder Axis during Gait

Generally the motion graphs showed several gaps. The causes are the following: (i) Due to the restricted field of view of the image intensifier the respective Ta markers could not be tracked over the whole stance phase, precisely the ankle ran out of the field of view shortly before TO. (ii) In some frames some markers could not be pinpointed because they were covered by a skin marker. (iii) Due to the small size of the Ta markers, already a small amount of motion blur made it impossible to locate the marker. (iv) In midstance markers were not always visible because of the overlapping of the contralateral leg.

Dorsi-/plantarflexion was for both subjects and all gait conditions the dominating rotation and occurred mainly at the talocrural joint, which is in agreement with De Asla et al (2006) and Lundberg et al (1989a). A direct comparison of results to data of other studies is difficult, especially because of different measurement techniques, test set ups and concepts to describe joint kinematics. To the best of the authors knowledge, there are no comparable studies for the gait conditions uph, dnh, ssb and sst. However, for level gait, the general motion pattern of tibiotalar dorsi-/plantarflexion is comparable to the bone pin studies of Arndt et al (2004) and Lundgren et al (2008). The $ROM_{dyn \, df}$ in level gait (see Table 22) is rather at the lower end, but more or less in line with previously presented data of Arndt et al (2004) (10.6° - 18.7°), De Asla et al (2006) (~ 13.5°) and Lundgren et al (2008) (15.3° ± 2.0°).

The other gait conditions, except uph, produced all an increase in $ROM_{dyn \, pf}$ as well as $ROM_{dyn \, df}$ compared to gt. In uph both subjects showed a smaller ROM compared to gt, whereas they experienced a higher level of $max_{dyn \, df}$, but a lower level of $max_{dyn \, pf}$. Thus, it
can be concluded that compared to gt the four gait conditions uph, dnh, ssb and sst were more challenging for the talocrural joint. However, since \( max_{dyn} df \) and \( max_{dyn} pf \) were considerably smaller than the static maximal values \( max_{stat} df \) and \( max_{stat} pf \), it can be concluded, that none of the gait conditions posed a big challenge for the healthy ankle joint.

Tibiotalar adduction/abduction, as well as inversion/eversion were circumstantial and showed magnitudes in the range of their variabilities. During level gait the tibiotalar \( ROM_{dyn} \ adab \) (see Table 23) is in line with the data presented by Arndt et al (2004) (4.0° - 5.0°), De Asla et al (2006) (\( \sim 3.2° \)) and slightly less than in the study of Lundgren et al (2008) (7.8° ± 2.7°).

Inversion/eversion showed mean tibiotalar \( ROM_{dyn} \ invev \) (see Table 23) which match the data of Arndt et al (2004) (4.5° - 6.3°) and De Asla et al (2006) (\( \sim 3.8° \)) pretty well, whereas the ranges presented by Lundgren et al (2008) (8.1° ± 3.8°) are higher. Differences could be due to different concepts of describing joint kinematics, since Lundgren et al (2008) were using Euler angles and technical frames, whereas Arndt et al (2004) as well used technical frames, but likewise to the present study, described clinical rotations with the use of helical component rotation. Besides in the study of De Asla et al (2006) a simulated stance phase with pausing at HS, midstance and TO was investigated, thus, the dynamic factors were missing. Anyway, since intersubject variabilities are large and the here presented data is additionally effected by a significant portion of inaccuracy, one should be cautious with comparisons of such small magnitudes.

Anteroposterior translation of the cylinder axis showed for sub 5 a maximal \( ROM_{dyn} \ t_{ap} \) of up to 3.7 mm and for sub 6 up to 5.2 mm (see Table 23) with, besides for the gait condition ssb, maximal variabilities during stance up to ± 3.5 mm. The variability of sub 5 during ssb showed a maximal range of ± 8.5 mm. This large variability is probably due to the out of plane inaccuracy of the used method. Since the direction of \( e_3 \) is not exactly parallel to the image plane, the accuracy of the anteroposterior translation of the cylinder axis is effected by the out of plane error. If due to the different motion task, the ankle was positioned in a larger angle relative to the image plane, the translational accuracy in direction of \( e_3 \) got worse due to the larger influence by the out of plane error. Next to that, again magnitudes are small and in the range of their variability. In comparison, De Asla et al (2006) found an anteroposterior tibiotalar translation of only 0.3 mm ± 1.2 mm.

Talocalcaneal motion was in the range of its variability, but showed the tendency of higher involvement in the second half of the stance phase with inversion/eversion as the predominant rotation. During gt, talocalcaneal \( ROM_{dyn} \ pfdf \), \( ROM_{dyn} \ adab \), \( ROM_{dyn} \ invev \) (see Table 24) were as well in line with the study of Arndt et al (2004) (dorsi-/plantarflexion: 2.8° - 3.7°, adduction/abduction: 2.0° - 6.1°, inversion/eversion: 4.6° - 8.3°).
Summarized, dorsi-/plantarflexion was the dominating rotation and occurred mainly at the talocrural joint. Tibiotalar rotations around the other two axes ($e_1$, $e_3$) were small and in their range of variability. Talocalcaneal motion was in the range of its variability, but showed the tendency of higher involvement in the second half of the stance phase with inversion/eversion as the predominant rotation.
5.5 Conclusions

The presented procedure enabled to quantify the 3D kinematics of the tibiotalar and the talocalcaneal joints using single plane videofluoroscopic tracking of bone embedded Ta markers of a size of 0.8 mm. However, data had to be analyzed with caution, since the translational out of plane errors were large (up to a SD of ± 8.6 mm) and the out of plane location of each bone had to be restricted. The restriction was performed using a linear regression fit of the z-trajectory of the center of each marker point cloud over the whole stance phase. The sources of the error were twofold. Though the error induced by the tracking of the Ta markers in the fluoroscopic images affected the accuracy more than the error induced by the segmentation of the CT images to reconstruct the geometric arrangement of the marker point clouds. Since the precision of marker tracking was in the subpixel range, an improvement of it would involve a further improvement of the resolution of the fluoroscopic images. Another limitation of the presented procedure were the missing data points due to the impossibility of tracking of some markers in certain time frames due to overlapping of the contralateral leg, restricted field of view of the image intensifier, motion blur and covering by skin markers. A possible way of improving the data accuracy would be to combine the described method with a matching of the bone contours, alike to the TAA procedure. At the ankle there is always an overlapping by certain bones that would effect the contour matching, but in combination with the Ta markers the prospects of high quality 3D kinematic data is given. Of course, also the dual plane approach would diminish the out of plane error, but creates other problems, such as the measurement set up allowing dynamic gait tasks.

Concerning the kinematics, for all gait conditions dorsi-/plantarflexion was the predominant rotation, mainly occurring at the talocrural joint. Tibiotalar inversion/eversion as well as adduction/abduction were small and in the range of variability. This applies also to the talocalcaneal rotations, however the tendency of a higher involvement in the second half of the stance phase with inversion/eversion as the predominant rotation was present at the subtalar joint.

The static maximal possible magnitudes of rotation were in none of the gait conditions, not even closely, reached. Yet walking uphill needed the highest range of maximal dorsiflexion $max_{df}^\text{dyn}$, whereas downhill generally needed the highest range of maximal plantarflexion $max_{pf}^\text{dyn}$ as well as $ROM_{df}^\text{dyn}$ and $ROM_{pf}^\text{dyn}$. Thus, uphill is the task to reveal problems of a restriction in dorsiflexion, whereas downhill is suited to detect limitations in plantarflexion and overall ROM.

Summarized, even though the presented procedure is afflicted by certain limitations, tendencies
of the kinematic characteristics of the ankle joint complex could be visualized. But certainly there are still many open questions, that call for improved methods to accurately quantify the 3D kinematics of the ankle joint complex in vivo, especially also during more challenging motion tasks alike the ones presented here.
6 Functionality of Total Ankle Arthroplasties - Comparison to Healthy Ankle Joints

6.1 Introduction

One important design criterion of a TAA is to achieve close to normal anatomical function and motion. In other words, the TAA should replicate the motion of the healthy ankle as close as possible. If a TAA restores a close to normal gait pattern, it provides less need for compensation and therefore a protection of adjacent joints. Additionally the original pattern of slackening/tightening of the ligaments could be restored.

This chapter of the thesis aims to compare the joint kinematics of the implant components to the one of the healthy ankle. The discussion leads to clarify the question if the implant allows a physiological path of motion and/or if it shows any restrictions during gait.

6.2 Methods

The kinematics of the TAA subjects (sub 1 - sub 4) assessed in chapter 4 were on the one hand compared to the kinematics of the healthy ankles (sub 5 - sub 6) provided by chapter 5 and on the other hand to bone pin data given by the studies of Arndt et al (2004) and Lundgren et al (2008).

Due to the limited number of subjects, the comparisons can only be performed qualitatively. The following comparisons were performed:

- Static $\text{ROM}_{\text{stat}}^{pf df}$, $\text{max}_{\text{stat}}^{pf}$ and $\text{max}_{\text{stat}}^{df}$. The respective definitions are provided in 4.2.6.

- Motion patterns during all gait conditions.

- Dynamic $\text{ROM}_{\text{dyn}}^{pf}$ and $\text{ROM}_{\text{dyn}}^{df}$ during all five gait conditions. The respective definitions are provided in 4.2.6, and shown in Figure 18.

- Comparison between the static available $\text{ROM}_{\text{stat}}^{pf df}$ (see 4.2.6) and the dynamic maximal used $\text{ROM}_{\text{dyn}}^{pf df}$. The definitions for $\text{ROM}_{\text{dyn}}^{pf df}$ are shown in Figure 51.
Figure 51: Definition of ROM\textsubscript{dyn} pf/df for ankle dorsi-/plantarflexion graphs. \textit{max\textsubscript{dyn}}: maximum of dorsi-/plantarflexion graph, depending on motion pattern this is equal to \textit{max\textsubscript{dyn}} df (left and middle) or to dorsiflexion at HS (right). \textit{min\textsubscript{dyn}}: minimum of dorsi-/plantarflexion graph, depending on motion pattern this is equal to \textit{max\textsubscript{dyn}} pf (left and right) or to dorsiflexion around TO (middle). \textit{ROM\textsubscript{dyn} pf/df}: maximal ROM, thus between \textit{min\textsubscript{dyn}} and \textit{max\textsubscript{dyn}}, depending on motion pattern this can equal \textit{ROM\textsubscript{dyn} df} (left) or \textit{ROM\textsubscript{dyn} pf} (right).

6.3 Results and Discussion

Since this chapter of the thesis is based on the results of chapters 4 and 5, the author has preferred to join the sections results and discussion.

6.3.1 Static ROM

Static \textit{ROM\textsubscript{stat} pf/df}, \textit{max\textsubscript{stat} pf} and \textit{max\textsubscript{stat} df} of the TAA subjects are provided in Table 8. Respective values for the healthy normal subjects are given in Table 21. Except sub 2, all TAA subjects showed a diminished \textit{ROM\textsubscript{stat} pf/df} compared to the two healthy normal subjects. However, individual differences are already very large within healthy normal ankles (see 2.1.3). In consensus, a significantly smaller \textit{ROM\textsubscript{stat} pf/df} for TAA than for healthy ankles was found in the cadaver studies of Richter et al (2007) and Valderrabano et al (2003a), as well as in several skin marker studies (Doets et al, 2006; Muller et al, 2006; Piriou et al, 2008).

Comparing the functional available ROM in direction of dorsiflexion and plantarflexion with the standing trial considered as neutral, the TAA subjects showed between 1° and 15° of available motion in direction of dorsiflexion, whereas the two healthy normal subjects possessed between 26° and 34°. Thus, the TAA subjects showed a diminished available static range of dorsiflexion. Concerning plantarflexion, the subjects did not manifest a restriction (TAA: 12° - 27° vs. healthy normals: 17° - 28°). It has to be kept in mind, that neutral is dependent on the standing trial, thus is afflicted by a certain amount of variation. However, this one-sided restriction in dorsiflexion has also been observed in previous skin marker studies (Doets et al, 2006; Muller et al, 2006) as well as cadaver studies (Richter et al, 2007; Valderrabano et al, 2003a).
Coetzee and Castro (2004) stated, that the main factor determining the postoperative ROM is given by the preoperative ROM. On the basis of these findings, the restrictions seem to be due to other factors than the design of the prostheses itself. Hintermann et al (2004) e.g. ascribed the major reason for the decreased ROM to a scarred capsule and soft tissues. Most important is how effectively the static available ROM is used and if it is sufficient for a physiological behavior of the TAA ankle during dynamic activities, such as gait and therefore if the TAA may preserve adjacent joints from compensation movements and therefore increased stresses.

6.3.2 Motion Patterns

Clinical ankle rotation graphs of the TAA subjects are provided in Figures 23 to 28, and for the healthy normal subjects see Figures 41, 43, 45, 47, 49. Likewise to the healthy normal subjects, all TAA subjects showed dorsi-/plantarflexion as the predominant rotation in all gait conditions. During level gait the motion pattern showed beside for sub 3 a comparable timing for the TAA and the healthy normal subjects, with a plantarflexion peak around 10% and a dorsiflexion peak around 80% stance. This is as well in agreement to the bone pin studies of Arndt et al (2004) and Lundgren et al (2008). Only sub 3 reached the dorsiflexion peak already before midstance and therefore showed a deviant behavior concerning the motion pattern. Magnitudes for adduction/abduction and inversion/eversion were for the TAA subjects, as well as for the healthy normal subjects small and in the range of variability. In this project, neither for the TAA subjects, nor for the healthy normal ankles a characteristic motion pattern could be seen. Thus, the latter two rotations seem to show no apparent differences between the TAA and the two healthy subjects of this project. Arndt et al (2004) described for the three bone pin assessed subjects consistent motion characteristic for the frontal as well as transverse planes. An initial inversion up to 20% stance was followed by eversion, followed by more or less a static behavior and a final strong inversion during the last 10% stance, which is in agreement to the data presented by Lundgren et al (2008). For the transverse plane only Arndt et al (2004) could describe a characteristic motion pattern which was consistent for all subjects. Namely, an adduction movement followed an initial abduction movement. Thus, taking this bone pin data as a base, the TAA subjects did not replicate transverse and frontal plane motion characteristics of the healthy ankle, even though the maximal ranges were in line.

Concerning anteroposterior translation of the cylinder axis relative to the tibia, respectively tibial component, likewise only minimal magnitudes, but large variability were present for the TAA subjects as per the healthy normal subjects. In the study of De Asla et al (2006) a mean anteroposterior translation of 1 mm ± 2.5 mm was found. Thus, the intersubject variations
were as well large and a comparison therefore even more difficult. However, altogether the TAA subjects did not show large differences concerning anteroposterior translation compared to the little data available of healthy ankles.

Summarized, sagittal plane motion characteristics of the TAA subjects were in line with the healthy ankles, whereas the motion characteristics of healthy ankles, as described in literature for the transverse and the frontal planes, were not replicated by the TAA subjects. If this differences in motion pattern, whereupon it has to be added, that we talk about differences in the range of smaller than 5°, do have a relevant impact on the strain behavior of the surrounding soft tissues remains to be determined. Though, involving the large intersubject variabilities, one could probably only answer this question by an intrasubject comparison to the unaffected contralateral leg.

### 6.3.3 Dynamic ROM

Mean and SD over all trials of each subject of $\text{ROM}_{\text{dyn d f}}$ and $\text{ROM}_{\text{dyn p f}}$ are for the TAA subjects sub 1 - sub 4 provided in Table 10, and for the healthy normal subjects sub 5 and sub 6 in Table 22.

Before all else, compared to the two bone pin studies (Arndt et al, 2004; Lundgren et al, 2008), providing sagittal plane ROM data, but only during level gait, the $\text{ROM}_{\text{dyn d f}}$ of the two healthy normal subjects of this project were similarly in level gait. Comparing $\text{ROM}_{\text{dyn d f}}$ between the TAA subjects and the healthy normal subjects, the tendency was seen that sub 1 and sub 2, except for sub 2 during uph, more or less were in line with the healthy normal subjects, whereas sub 3 and sub 4 showed during all gait conditions smaller $\text{ROM}_{\text{dyn d f}}$. Concerning $\text{ROM}_{\text{dyn p f}}$ mainly sub 4 showed a limited behavior, whereas the other three TAA subjects reached ranges comparable to the healthy normal ankles.

Both healthy normal subjects showed during ssb and sst larger $\text{ROM}_{\text{dyn p f}}$ and $\text{ROM}_{\text{dyn d f}}$ compared to gt, whereas all TAA subjects showed rather smaller values compared to gt. This could be explained by a higher muscular activity with the aim of joint stabilization and therefore larger restraints following in smaller flexibility.

Summarized, in terms of ROM compared to healthy normal ankles, minor limitations for sub 3 and sub 4 were seen concerning $\text{ROM}_{\text{dyn d f}}$. Furthermore sub 4 showed restrictions with respect to $\text{ROM}_{\text{dyn p f}}$. If the limitations are caused by a limitation in static available ROM, thus a total exploitation of the static available ROM or if other factors inhibit a effective utilization is discussed in the next paragraph.
6.3.4 Static versus Dynamic ROM

A comparison between the static available ROM $\text{ROM}_{\text{stat}}$ and the $\text{ROM}_{\text{dyn}}$, which was actually used during the dynamic motion tasks is graphically provided in the five barplots in Figure 52. The unexploited motion ranges $\Delta f$ and $\Delta df$ give a measure of how much of $\text{ROM}_{\text{stat}}$ was used during the dynamic gait tasks. The centerlines $\text{cl } \text{ROM}_{\text{stat}}$ and $\text{cl } \text{ROM}_{\text{dyn}}$ allow an estimate of how well situated the range of available motion is. In other words, if $\text{ROM}_{\text{stat}}$ is located, such that the dynamic gait tasks are enabled without restrictions, or if the availability of motion is distributed one-sided, recognizable with an asymmetric vertical bar pattern.

First of all, besides sub 2, all TAA subjects showed a smaller $\text{ROM}_{\text{stat}}$. Furthermore, it is seen that in uph $\Delta df$ is smaller than in gt, thus uph demanded a higher $\text{max}_{\text{dyn}}$. Whereas in dnh $\Delta f$ is smaller, thus motion was more needed in direction of dorsiflexion.

A possible restriction during gait due to a limitation in the available $\text{ROM}_{\text{stat}}$ is seen when $\Delta df$ is missing or marginal, which was the case for sub 4 during all gait tasks and for sub 1 during uph, ssb and sst. All those restrictions concern the direction of dorsiflexion, thus $\text{max}_{\text{dyn}}$. Contrary, $\text{max}_{\text{stat}}$ was for none of the subjects and none of the gait conditions even closely exploited.

Interestingly, sub 4 showed during all dynamic gait tasks, a higher maximal reached dorsiflexion than in the static positions ($\text{max}_{\text{dyn}} > \text{max}_{\text{stat}} df$). Either he did not perform the static positions maximally or due to the larger moments during the dynamic tasks, a larger rotation could be enforced.

In terms of distribution of the unexploited motion ranges $\Delta f$ and $\Delta df$, the healthy subjects showed during gt a more or less symmetric distribution ($\Delta f \approx \Delta df$), whereas all the TAA subjects showed an asymmetry with a larger range in direction of plantarflexion than in direction of dorsiflexion ($\Delta f > \Delta df$). This is also seen by the shift of $\text{cl } \text{ROM}_{\text{stat}}$ in direction of plantarflexion compared to $\text{cl } \text{ROM}_{\text{dyn}}$. One can conclude, that the TAA subjects do not only have a restricted $\text{ROM}_{\text{stat}}$, but additionally it can not be used effectively, since it is not distributed functionally in terms of requirements for gait. And since gait is the main function of the ankle, it should be the aim to adjust the motion possibilities of the TAA to comply optimally in its function during gait. If it is possible to shift the range of available motion, by e.g. a different positioning in the process of implantation, should be part of further investigations. Interestingly, the decreased ROM in direction of dorsiflexion was as well observed in a cadaver study of Richter et al (2007) looking at the Hintegra TAA, whereas no differences
were found for the German Ankle System. Thus, on the base of the study of Richter et al (2007) the design of the components showed an influence. However the design of the Mobility™ Total Ankle would allow same amounts of dorsiflexion and plantarflexion, thus this factor can be excluded. Hintermann et al (2004) sees the main reason of the restriction in the scarred capsule and soft tissues. Doets et al (2006) found a higher electromyography activity pattern for the gastrocnemius in midstance and for the anterior tibial muscle in late stance for the TAA group than in the control group. He concluded, that there was more co-contraction in TAA subjects than in controls. Following, that the decreased $ROM_{df}$ could be explained by the restraints of the surrounding muscular co-contraction. Alike, Benedetti et al (2008) found for TAA subjects a pre-activation in initial stance of the gastrocnemius and peroneus longus muscles together with tibialis anterior. They explained this co-contraction in pain-free patients with a protection mechanism due to a preoperatively structured motor scheme related to pain or due to a distorted proprioception of surrounding structures associated to the modified biomechanics of the TAA aiming at joint stabilization. If the main factors for the restrictions in ROM in pain-free subjects are stiffness of the surrounding soft tissues, the changed muscular activity, the implant design or if its of multifactorial nature, needs further investigation.

Summarized, the TAA subjects showed a restriction in the available ROM in direction of dorsiflexion, or in other words, a shift between the for dynamic gait function needed and the actual available distribution of ROM. This does not really restrict the TAA subjects during level gait, but since during walking uphill a larger amount of dorsiflexion is needed, uph rather posed a challenge for the TAA ankles. Possibilities to enable a symmetric distribution of the ROM should be part of further studies.
Figure 52: Static versus dynamic ROM. Whole bar (black plus red plus grey) refers to $\text{ROM}_{\text{stat}}$ \(pfd\). The red part of the bar shows the functional used $\text{ROM}_{\text{dyn}}$ \(pfd\) during gait. The grey part stands for $\Delta pf$, which is the difference between $\max_{\text{stat}} pf$ and $\min_{\text{dyn}}$. The black part shows $\Delta df$, which is the difference between $\max_{\text{stat}} df$ and $\max_{\text{dyn}}$. The black line stands for $\text{cl ROM}_{\text{stat}}$ \(pfd\) and shows the mid of $\text{ROM}_{\text{stat}}$ \(pfd\). The red line stands for $\text{cl ROM}_{\text{dyn}}$ \(pfd\) and shows the mid of $\text{ROM}_{\text{dyn}}$ \(pfd\). $\Delta df$ and $\Delta pf$ give a measure of how much of $\text{ROM}_{\text{stat}}$ \(pfd\) was used during the dynamic gait tasks. $\text{cl ROM}_{\text{stat}}$ \(pfd\) and $\text{cl ROM}_{\text{dyn}}$ \(pfd\) allow an estimate of how well situated the range of available motion is. Sub 1 during sst and sub 4 during all gait conditions showed a larger $\text{ROM}_{\text{dyn}}$ \(pfd\) than $\text{ROM}_{\text{stat}}$ \(pfd\), in these cases the dotted line shows $\text{ROM}_{\text{stat}}$ \(pfd\).
6.4 Conclusions

The TAA subjects showed main motion characteristics which were similar to the one of the healthy ankle. The predominant rotation was dorsi-/plantarflexion, with a motion pattern likewise to healthy ankles. Differences were found in the transverse and frontal planes, where the TAA subjects, in contrast to literature data on healthy ankles, did not show characteristic motion patterns. Some TAA subjects showed smaller static as well as dynamic ROM. A restriction in ROM was mainly seen in direction of dorsiflexion, which manifested itself mainly in walking uphill.

The main rotation of the TAA, namely dorsi-/plantarflexion was comparable to the healthy ankle in its characteristic as well as only showed minor limitations in its magnitude during level gait. One can assume that the compensation motions observed in arthrodeses patients, that lead to an increase in stress and following degenerative changes in those joints, are much smaller in TAA subjects. The possibility of maintaining motion in the ankle joint by the use of a TAA could be shown and is a clear advantage to arthrodesis in terms of preserving adjacent joints from degenerative changes.

A further improvement of the functionality of the TAA could be achieved if it was possible to shift the ROM more in direction of dorsiflexion, such that the available ROM fits the functional used ROM. Since the design itself allows the same amount of motion in both directions, one needs to assume that the surrounding structures play an important role in the restricted behavior. A change in implant positioning in the process of the surgery or an intensified physiotherapy during the rehabilitation could be successful.

If a restriction in ROM is caused by a change in muscular activity pattern, if the passive surrounding soft tissues account for the limitation in flexibility or if its a multifactorial problem remains an open question.

Next to the restriction in ROM, it is not yet clear if the role of the surrounding ligaments can be maintained by the TAA. Specifically, if e.g. the changed behavior of rotation in the frontal and transverse planes show an unfavorable influence on the strain behavior of the ligaments. A possible tool to evaluate this, would be to simulate different motion patterns in a model and estimate their effects on the strain behavior of the ligaments.

To conclude, the TAA allowed close, but not in all cases, physiological motion. Some restrictions could be revealed, but overall the functionality of the analyzed TAA showed a satisfying kinematic behavior. Great care should be taken to implant the TAA at the right position, which might be in a more plantarflexed position.
7 Simultaneously Measured Skin Marker and Videofluoroscopy Joint Kinematics

7.1 Introduction

Previous in vivo studies on TAA kinematics were mainly performed using skin marker tracking. Relating to this external measured data, the TAA groups showed a close to normal gait pattern for level walking and only minor reductions in sagittal ROM were found compared to healthy ankles (see 2.4.4).

One disadvantage of skin marker tracking are skin movement artefacts. Comparing bone pin to skin marker tracked healthy tibiocalcaneal kinematics (Nester et al, 2007a; Reinschmidt et al, 1997b; Westblad et al, 2002), it was not possible to formulate a clear conclusion concerning systematic patterns of over- or underestimation of the external relative to the internal approach (see 2.4.4). Thus, due to the individuality, externally measured data is always affected by an uncertainty of how much motion is actually taking place between the bones, respectively implant components.

The second drawback of skin marker tracking is the limitation of not being able to assess the motion of the talus due to its inaccessibility. Thus, it is not possible to analyze the actual motion between the talar and the tibial component. Externally, it is only possible to track motion of the whole rearfoot.

However, a big advantage of skin marker tracking is the convenience of application and the possibility to perform a full body analysis, thus to evaluate possible compensation mechanisms occurring in e.g. the knee or in the foot itself.

This chapter aims (i) to compare ankle kinematics measured externally via skin marker tracking to internally, thus fluoroscopy gained data and (ii) the additional data gained by skin marker tracking concerning foot and knee kinematics shall be combined to analyze potential compensation strategies.

7.2 Methods

Skin marker tracking and videofluoroscopic measurements were performed simultaneously. The method concerning the videofluoroscopic procedure of TAA subjects is described in 4.2, respectively for the healthy normal subjects in 5.2. The following sections explain the data collection, including subject information (7.2.1) and measurement set up (7.2.2), the processing of the skin marker data (7.2.3), which is based on Dettwyler (2005), List (2005), Unternaehrer (2005) and...
List et al (2008), the adaptation of the videofluoroscopic approaches (7.2.4), as well as location registration (7.2.5) and time synchronization (7.2.6) of the two systems to enable a comparison between the videofluoroscopy and the skin marker data. 7.2.7 provides information about the comparison between the internal and the external assessed data and 7.2.8 about the assessment of compensation strategies.

7.2.1 Subjects

The kinematic data of the four TAA subjects (sub 1 - sub 4) and of one healthy normal subject (sub 5) was evaluated (for information on the subjects see Tables 6 and 11). Sub 6 had to be excluded due to missing skin marker data of the rear- and forefoot.

7.2.2 Measurement Set Up and Test Procedure

The measurement set up, including the videofluoroscopy and the VICON systems is described in 4.2.2. The test procedure is specified in 4.2.3. Additional information concerning the skin marker processing is given in the following section.

7.2.3 Skin Marker Data Analysis

Markerset The used markerset consisting of 53 skin markers with a diameter of 9 mm is shown in Figure 53. The marker set includes a two segment foot model. The foot is divided into fore- and rearfoot. Because if the foot is treated as one rigid segment, motion which is present between fore- and rearfoot affects the computation of ankle joint rotations and leads to an overestimation of them (List et al, 2006, 2008). Each segment is defined by a redundant marker cluster, based on the following principles:

- Marker visibility: All markers should be visible by at least two VICON cameras during the whole gait cycles.

- Marker cluster distribution: The distance between markers and offset from the line joining the other markers should be as large as possible. Thus, the mean cluster radius is maximized (Soderkvist and Wedin, 1993).

- Number of markers: Each segment cluster consists of a redundant number of markers, since the increase from three up to four or five markers shows the trend of improved estimation of orientation accuracy (Challis, 1995).

- Markers are located on body positions, that show minimal skin movement artefacts.
Figure 53: Markerset and marker based joint coordinate systems. Functionally estimated joint centers (yellow markers). The whole markerset consists of 53 markers (due to a shortage of space foot markers are only shown on the left leg of the skeleton). Red markers are next to segment tracking additionally used for the definition of the marker based joint coordinate systems, grey markers are only used for segment tracking. The HJC is used as a virtual marker for the thigh segment.
Estimation of Relative Rotations and Translations  Corresponding orientation matrices and position vectors of each segment ($R_{\text{thigh}}$, $t_{\text{thigh}}$, $R_{\text{shank}}$, $t_{\text{shank}}$, $R_{\text{rearfoot}}$, $t_{\text{rearfoot}}$, $R_{\text{forefoot}}$, $t_{\text{forefoot}}$) were determined using a least-squares fit of point clouds (Gander and Hrebicek, 1997). The used reference point clouds were determined by the standing trial, performed in a natural upright position. The neutral position (0° rotation) was defined by the standing trial.

The relative motion between forefoot and rearfoot (foot motion), rearfoot and shank (ankle joint motion), as well as shank and thigh (knee joint motion) was described from the distal relative to the proximal segment. The computation was performed according to the TAA approach as described in 4.2.6.

Clinical Rotations  For the description of the clinical rotations the attitude vector is decomposed along the axes of the marker based joint coordinate system (see also 4.2.6).

Joint centers, respectively axes were estimated via functional approaches. Therefore the subjects had to perform four basic motion tasks (BMT) described in Table 25.

<table>
<thead>
<tr>
<th>BMT</th>
<th>Description</th>
</tr>
</thead>
<tbody>
<tr>
<td>Hip BMT</td>
<td>Three circumduction motions in the hip, standing on one leg, with keeping the leg straight. Support on the contralateral side was allowed.</td>
</tr>
<tr>
<td>Knee BMT</td>
<td>Loaded knee flexion/extension motions (three repetitions). In other words, bending the loaded knee with keeping the leg in the sagittal plane.</td>
</tr>
<tr>
<td>Ankle BMT 1</td>
<td>Loaded inversion/eversion motion (three repetitions). In other words, movement of the shank in the frontal plane with maximal range of motion relative to the standing foot.</td>
</tr>
<tr>
<td>Ankle BMT 2</td>
<td>Loaded dorsi-/plantarflexion motion (three repetitions). In other words, movement of the shank in the sagittal plane with maximal range of motion relative to the standing foot.</td>
</tr>
</tbody>
</table>

Table 25: Basic motion tasks (BMT) performed for the functional estimation of the ankle and hip joint centers (AJC, HJC), as well as for the knee joint axis ($u_k$).

The idea behind the functional approach is to prevent, at least to a certain amount, the dependency on the location of anatomical landmarks and allow hereby a higher accuracy than prediction approaches (Leardini et al, 1999b). For the determination of the joint centers, respectively axes (hip joint center (HJC), ankle joint center (AJC), knee joint axis ($u_k$)), the hip joint as well as the ankle joint were modeled as ball-and-socket joints, the knee joint was modeled as hinge joint. Using the above described corresponding BMT the joint centers, respectively axes were determined by minimizing the sum of the differences between the modeled and the measured locations of the marker points.

The marker based joint coordinate systems are orthogonal right-handed coordinate systems built
with the use of those functional estimated joint centers. The definition were performed as follows and illustrated in Figure 53:

- Marker based foot joint coordinate system: The posteroanterior axis \( e_{ff} \), connection HL and TO is the leading axis. The direction of the mediolateral axis \( e_{fs} \) is perpendicular to \( e_{ff} \) and parallel to the ground. The vertical axis \( e_{ff} \) is perpendicular to the latter two.

- Marker based ankle joint coordinate system: The vertical axis \( e_{at} \), connecting line between AJC and KJC, is the leading axis. The direction of the mediolateral axis \( e_{as} \) is perpendicular to \( e_{at} \) and lies in the plane spanned by the malleoli and the KJC. The posteroanterior axis \( e_{af} \) is perpendicular to the latter two.

- Marker based knee joint coordinate system: The mediolateral axis \( e_{ks} \) is defined by the functional estimated knee joint axis \( u_k \). The direction of the vertical axis \( e_{kt} \) is perpendicular to \( e_{ks} \) and lies in the plane spanned by \( e_{ks} \) and the HJC. The posteroanterior axis \( e_{kf} \) is perpendicular to the latter two.

Hence, clinical rotations are described as provided in Table 26.

![Table 26](image)

**Joint** | **Relative movement** | **Rotation axis** | **Clinical Rotation**
--- | --- | --- | ---
Foot | forefoot rel rearfoot | \( e_{fs} \) | dorsi-/plantarflexion
 | | \( e_{ft} \) | adduction/abduction
 | | \( e_{ff} \) | inversion/eversion

Ankle | rearfoot rel shank | \( e_{as} \) | dorsi-/plantarflexion
 | | \( e_{at} \) | adduction/abduction
 | | \( e_{af} \) | inversion/eversion

Knee | shank rel thigh | \( e_{ks} \) | flexion/extension
 | | \( e_{kt} \) | internal/external rotation
 | | \( e_{kf} \) | adduction/abduction

Table 26: Description of clinical rotations for the foot, the ankle and the knee based on the marker based joint coordinate systems shown in Figure 53. rel: relative to.

**Stance Phase**  Analogous to the TAA and the tantalum approaches, the stance phase, thus the time frame from HS to TO, was evaluated (see also 7.2.6).

**ROM**  For the static trials, a maximal ankle ROM \((ROM_{stat} \, pf df)\) around \( e_{as} \) between the maximal plantarflexed and the maximal dorsiflexed static loaded position was estimated. The definition of the ankle ROM during the stance phase of the gait tasks for ankle plantarflexion \((ROM_{dyn} \, pf)\) and dorsiflexion \((ROM_{dyn} \, df)\) is analogous to the TAA approach and shown in Figure 18.
7.2.4 Adaptation of the Videofluoroscopic Approaches

The data analysis of the fluoroscopic gained data is described in 4.2 and 5.2. To enable the comparison between external and internal measured data without being influenced by the choice of different conventions for the description of ankle kinematics, the marker based joint coordinate system defined in the skin marker data analysis (see 7.2.3) was as well used to describe the kinematics of the fluoroscopic gained data. Skin marker as well as videofluoroscopic gained kinematics were analyzed using the same concept to describe joint rotations and the same joint coordinate system.

To allow a transformation of the fluoroscopic gained position and orientation of the implant components, respectively bones, into the VICON laboratory coordinate system (see Figure 11), the fluoroscopy coordinate system (see Figure 12) had to be registered within the VICON laboratory coordinate system. The underlying procedure is explained in 7.2.5. Figure 55 shows exemplarily for sub 4 the implant components including the implant coordinate system, as well as the skin markers and the marker based ankle joint coordinate system of the standing trial in the VICON laboratory coordinate system.
In chapter 4 the kinematics of the implant components were described relative to the initial position defined by the implant coordinate system. To enable a comparison to the skin marker analysis, for this chapter videofluoroscopic gained TAA kinematics was described relative to the standing trial. The neutral position (0° rotation), defined by the standing trial, was the same for the videofluoroscopic as for the skin marker analysis.

Since skin marker of the rearfoot were located on the calcaneus, for the healthy normal subject (sub 5) next to tibiotalar motion, also tibiocalcaneal motion was estimated.
7.2.5 Registration of the Fluoroscopy Coordinate System in the VICON Laboratory Coordinate System

The orientation and position of the fluoroscopy coordinate system is registered in the VICON laboratory coordinate system by the use of the grid used for image distortion correction (see 4.2.4). Six skin markers were attached to the grid at predefined positions. The grid including the skin makers was assessed simultaneously by the fluoroscopic and the VICON system. Consequently the positions of the skin markers were identified in both coordinate systems. Accordingly, the transformation between the two coordinate systems was determined using a least-squares fit of point clouds (Gander and Hrebicek, 1997).

7.2.6 Time Synchronization of the Two Systems

Using an input trigger sent by the X-ray generator of the videofluoroscopy system to an analog channel of the VICON system, the time of each generated videofluoroscopic image was recorded as an analog signal in the VICON system (Foresti, 2009). HS and TO were set in the VICON software and the corresponding fluoroscopic images were determined using the analog trigger signal.

7.2.7 Comparison External vs. Internal Measured Joint Kinematics

To describe differences between the external and the internal measured joint kinematics the following comparisons were performed.

- Motion patterns of the clinical rotation graphs, as well as $\text{ROM}_{\text{stat}} pf df$ were qualitatively compared.
- The mean and the maximum root mean square differences (RMS diff, max diff) between the external and the internal measured motion graphs were calculated over the complete stance phase as a descriptor of differentiation.
- $\text{ROM}_{\text{dyn}} pf$ as well as $\text{ROM}_{\text{dyn}} df$ once estimated by videofluoroscopy, once by skin marker tracking were compared using a Mann-Whitney U-Test and a 5% significance level.

7.2.8 Compensation Strategies

For level gait, skin marker joint kinematics were analyzed for two consecutive steps. Thus, the joint kinematics of the ipsi- and the contralateral leg were analyzed during the corresponding
consecutive stance phases. Possible compensation strategies performing between fore- and rear-foot, or between shank and knee were qualitatively assessed. Moreover, $ROM_{dympf}$, $ROM_{dyn df}$, $ROM_{dyn df1}$, $ROM_{dyn ex}$, $ROM_{dyn fl1}$ were compared between the ipsi- and the contralateral leg using a Mann-Whitney U-Test and a 5% significance level.
7.3 Results

7.3.1 External vs. Internal Measured Ankle Joint Kinematics

Static Maximal ROM \( ROM_{stat \, pfdf} \) of sub 1 to sub 5, once estimated by the video-fluoroscopic procedure, once by skin marker tracking are provided in Table 27. For sub 5 tibiotalar (tt), as well as tibiocalcaneal (tc) rotation is given. The external measured \( ROM_{stat \, pfdf} \) does not significantly differ from the internal estimated \( ROM_{stat \, pfdf} \). Sub 1, 2 and 5 show a smaller \( ROM_{stat \, pfdf} \) by the skin marker measurements, whereas in contrast for sub 3 and 4 skin markers provide larger values. Sub 5 shows a difference of about 9° between videofluoroscopy gained tibiotalar and tibiocalcaneal rotation.

<table>
<thead>
<tr>
<th></th>
<th>( ROM_{stat , pfdf} )</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>fluoroscopic gained</td>
</tr>
<tr>
<td>sub 1</td>
<td>21.3°</td>
</tr>
<tr>
<td>sub 2</td>
<td>41.9°</td>
</tr>
<tr>
<td>sub 3</td>
<td>29.7°</td>
</tr>
<tr>
<td>sub 4</td>
<td>14.5°</td>
</tr>
<tr>
<td>sub 5</td>
<td>tt: 41.6° / tc: 50.8°</td>
</tr>
</tbody>
</table>

Table 27: \( ROM_{stat \, pfdf} \) between the maximal plantarflexed and the maximal dorsiflexed static loaded positions. Neutral is given by the standing trial. For healthy normal subject sub 5 tibiotalar (tt) and tibiocalcaneal (tc) rotation is provided.

Clinical Ankle Rotations during Gait  Clinical ankle rotations for all gait conditions measured via skin marker tracking as well as simultaneously measured by videofluoroscopy are shown in Figures 56 (sub 1), 57 (sub 2), 58 (sub 3), 59 (sub 4), 60 (sub 5). For the healthy normal subject (sub 5) videofluoroscopy measured tibiotalar as well as tibiocalcaneal kinematics are provided.

Sub 1 showed for all gait conditions generally similar motion patterns for the two measurement approaches. For dors-/plantarflexion, as well as for adduction/abduction the graphs show similar motion patterns, but are shifted relative to each other. For inversion/eversion differences were found during all gait conditions, but at various points in time, mainly in higher amplitudes for the skin marker data.

Sub 2 showed high agreement for almost all gait conditions and rotations. The only minor differences between internal and external measured data were found for the first part of the stance phase for inversion/eversion during ssb and sst. The videofluoroscopy measured adduction/abduction showed large variability due to the unfavorable positioning of the tibial component relative to the image plane (see 4.4.2).

Sub 3 showed minor differences between the videofluoroscopy measured and the external mea-
sured dorsi-/plantarflexion graphs, especially in the first part of the stance phase. In adduction/abduction a general shift was present, whereas inversion/eversion differed mainly at the end of the stance phase during gt, uph and dnh and over the whole stance phase during ssb and sst.

Sub 4 showed no significant differences for dorsi-/plantarflexion. In adduction/abduction, the differences were in the range of the variability. But skin marker showed more eversion during the second phase of the stance phase for gt, uph, dnh and more inversion during the first part of the stance phase in ssb. During sst there was a shift between the inversion/eversion graphs.

Sub 5 showed for dorsi-/plantarflexion good conformity between the skin marker tracked ankle kinematics and videofluoroscopy gained tibiotalar, as well as tibiocalcaneal kinematics. The differences were in the range of the variabilities. For adduction/abduction, skin marker and tibiocalcaneal motion agreed pretty well, whereas tibiotalar motion showed differences over the whole stance phase. In contrast, for inversion/eversion videofluoroscopy gained tibiotalar and tibiocalcaneal motion showed high conformity, whereas skin marker data showed higher deflections in direction of inversion, as well as eversion.

Summarized, videofluoroscopy gained and skin marker tracked data showed, besides the apparent shift for sub 1, high conformity for dorsi-/plantarflexion. Adduction/abduction between the two measurement approaches showed, in addition to a shift, minor differences for sub 3, and major differences for tibiotalar motion of sub 5. However, the largest differences between the internal and the external measured data were found for the inversion/eversion motion graphs of all subjects. Generally, for inversion/eversion larger movement amplitudes were found for the skin marker data.

Root mean square differences (RMS diff) and maximal differences (max diff) between the skin marker and the videofluoroscopy gained joint kinematics during stance are provided in Table 28. The mean ± SD of the RMS diff over all subjects and all gait conditions was for dorsi/plantarflexion 1.7° ± 1.0°, for adduction/abduction 2.9° ± 1.9° and for inversion/eversion 2.3° ± 1.4°.

$\text{ROM}_{\text{dy}n \text{ pf}}$ as well as $\text{ROM}_{\text{dy}n \text{ df}}$ once estimated by videofluoroscopy, once by skin marker tracking are visualized for each subject in Figure 61. $\text{ROM}_{\text{dy}n \text{ pf}}$ as well as $\text{ROM}_{\text{dy}n \text{ df}}$ only showed a few significant differences (marked with *) between external and internal measurements, generally the differences were in the range of the variability. Sub 1, 3 and 4 showed a tendency of overestimation by skin marker, whereas no clear tendency was seen for sub 2 and sub 5.
Table 28: Root mean square difference (RMS diff) and maximal difference (max diff) between the skin marker and the videofluoroscopy gained joint kinematics during the stance phase. \( e_{as} \): Rotation around \( e_{as} \), thus dorsi-/plantarflexion, \( e_{at} \): Rotation around \( e_{at} \), thus adduction/abduction, \( e_{af} \): Rotation around \( e_{af} \), thus inversion/eversion. tt: tibiotalar, tc: tibiocalcaneal. Note that all values were calculated from the mean curves of the corresponding subjects.
Figure 56: Sub 1. Videofluoroscopy measured (black) vs. skin marker measured (red) ankle kinematics. Mean and SD over all trials.
Figure 57: Sub 2. Videofluoroscopy measured (black) vs. skin marker measured (red) ankle kinematics. Mean and SD over all trials.
Figure 58: Sub 3. Videofluoroscopy measured (black) vs. skin marker measured (red) ankle kinematics. Mean and SD over all trials.
Figure 59: Sub 4. Video fluoroscopy measured (black) vs. skin marker measured (red) ankle kinematics. Mean and SD over all trials.
Figure 60: Sub 5. Videofluoroscopy measured tibiotalar (black) and tibiocalcaneal (grey) vs. skin marker measured (red) ankle kinematics. Mean and SD over all trials. Gap in midstance caused by overlapping of the contralateral leg.
Figure 61: $\text{ROM}_{\text{dyn}} \ pf$ and $\text{ROM}_{\text{dyn}} \ df$ of videofluoroscopy measured (black) and skin marker measured (red) ankle kinematics. Mean and SD over all trials. For sub 5 next to videofluoroscopy measured tibiotalar (black), also videofluoroscopy measured tibiocalcaneal (grey) ROM is provided. pf: plantarflexion, thus $\text{ROM}_{\text{dyn}} \ pf$, df: dorsiflexion, thus $\text{ROM}_{\text{dyn}} \ df$. * significant with $p < 0.05$. 
7.3.2 Compensation Strategies

The skin marker tracked ankle, foot and knee rotation graphs of the ipsi- and the contralateral leg of sub 1 - 5 are shown in Figures 62 (ankle), 63 (foot) and 64 (knee).

The qualitative visual inspection of the motion graphs revealed the following differences.

Sub 1: Compared to the contralateral ankle the TAA ankle showed less dorsiflexion in the second part of the stance phase. Around the same time frames sagittal plane motion between the fore- and the rearfoot was larger for the TAA side.

Sub 2: The TAA ankle showed a decreased first plantarflexion peak (at around 10% stance) and as well around 10% stance an increase in foot plantarflexion compared to the contralateral leg. Furthermore ipsilateral knee extension was increased during the second half of the stance phase.

Sub 3: During the second half of the stance phase less dorsiflexion occurred at the TAA ankle compared to the contralateral ankle. Foot dorsi-/plantarflexion showed different motion patterns for the ipsi- and the contralateral side, especially in the second half of the stance phase.

Sub 4: The TAA ankle showed compared to the contralateral ankle in all planes different motion patterns. Generally the contralateral ankle showed higher magnitudes in all motion directions. Likewise the ipsilateral foot showed less motion than the contralateral. But knee flexion/extension of the ipsilateral side showed a more dynamic motion pattern than the contralateral knee.

Sub 5, the healthy normal subject: No large differences were seen for the ankle and foot rotations, but the ipsilateral knee joint showed a decreased first flexion peak compared to the contralateral leg.

The corresponding ROM (ankle: $ROM_{dyn \, pf}$, $ROM_{dyn \, df}$, foot: $ROM_{dyn \, dff}$, knee: $ROM_{dyn \, fl1}$, $ROM_{dyn \, ex}$, $ROM_{dyn \, fl2}$) of the ipsilateral (ipsi) and the contralateral (contra) leg are presented in Table 29. At the ankle joint two of the four TAA subjects (sub 2 and 4) showed a significantly decreased $ROM_{dyn \, pf}$ and three out of four (sub 1, 3 and 4) a significantly decreased $ROM_{dyn \, df}$ of the TAA ankle compared to the contralateral. At the foot two TAA subjects (sub 1 and 2) showed a significantly increased $ROM_{dyn \, dff}$ at the TAA side. At the knee joint two TAA subjects (sub 2 and 4) showed a significantly increased $ROM_{dyn \, ex}$ of the ipsilateral leg. Likewise for two TAA subjects (sub 1 and 4) $ROM_{dyn \, fl2}$ of the ipsilateral leg was significantly increased. The healthy subject 5 showed a significant difference between the left and the right leg for $ROM_{dyn \, ex}$. 
Table 29: Mean and SD of the skin marker measured ankle, foot and knee ROM of all trials of each subject during the stance phase of level gait. ipsi: TAA, respectively tantalum marker side, contra: contralateral side.
Figure 62: Left-right differences in ankle kinematics. Videofluoroscopy measured ankle kinematics (black), TAA, respectively tantalum marker side skin marker tracked (red) and contralateral side skin marker tracked (green). Mean and SD over all trials of each subject.
Figure 63: Compensation strategies in the foot. Skin marker tracked forefoot relative to (rel) rearfoot kinematics. TAA side, respectively tantalum marker side (red) vs. contralateral side (green). Mean and SD over all trials of each subject.
Figure 64: Compensation strategies in the knee. Skin marker tracked shank relative to (rel) thigh kinematics. TAA side, respectively tantalum marker side (red) vs. contralateral side (green). Mean and SD over all trials of each subject.
7.4 Discussion

7.4.1 External vs. Internal Measured Ankle Joint Kinematics

**Static Maximal ROM** $ROM_{stat \ pfd}$ did not differ significantly between the external and the internal measured approaches. Since in the external approach skin markers were attached to the calcaneus, therefore estimated motion included rotation between the calcaneus and the talus, one would assume larger values for the external approach. Contrary to the expectation, for sub 5, for which additionally tibiocalcaneal rotation was analyzed, the tendency is present that external measured $ROM_{stat \ pfd}$ underestimated actual tibiocalcaneal rotation, but reflected pretty well tibiotalar rotation during static conditions. This is likewise valid for the tibiotalar $ROM_{stat \ pfd}$ of the TAA subjects, where the difference between the external and internal approaches was for all subjects smaller than 6.4°. Thus, one can conclude that tibiotalar rotation was pretty well reflected by the external skin marker measurements. However, since external skin markers actually measure tibiocalcaneal rotation, and tibiocalcaneal $ROM_{stat \ pfd}$ is generally combined of tibiotalar and talocalcaneal motion (Siegler et al, 1988; Wong et al, 2005), the assumption can be made that skin markers show the tendency of underestimating the actual rotation of the underlying bones during those static measurements.

**Clinical Ankle Rotations during Gait** A general shift between the skin marker and the videofluoroscopy tracked gait data was found for sub 1 for dorsi-/plantarflexion, as well as for sub 1 and 3 for adduction/abduction. This is very well seen in the motion graphs, but also reflected in the corresponding larger RMS diff. Since neutral is defined by the standing trial, this shift probably results, at least to a certain extent, from the differences in the estimated neutral position of the standing trial.

Besides this shift, dorsi-/plantarflexion showed for all subjects and during all conditions high conformity between the internal and the external approach. Thus, it can be concluded, that tibiotalar dorsi-/plantarflexion motion was during all gait conditions well reflected by the external skin marker measurements. Looking at the healthy normal subject (sub 5) only minor differences were found between tibiotalar and tibiocalcaneal dorsi-/plantarflexion motion graphs. Based on the available data of only one subject, the hypothesis could be made, that during the gait tasks only little sagittal plane motion was present in the subtalar joint, such that it was more or less possible to estimate tibiotalar motion by externally tracking the calcaneus and the shank. Furthermore it seems, that skin marker artefacts were small or did not substantially effect sagittal plane ankle joint kinematics.
The motion graphs of adduction/abduction and especially of inversion/eversion showed larger discrepancies between the external and the internal approaches. Some of the adduction/abduction motion and the major part of inversion/eversion motion is assigned to the subtalar joint (Siegler et al, 1988; Wong et al, 2005), which could explain the apparent differences between the external measured tibiocalcaneal and the internal measured tibiotalar motion. But, looking at sub 5, showing a good agreement between internal and external measured tibiocalcaneal kinematics for adduction/abduction but not for inversion/eversion, this can only be supported for adduction/abduction, but not for inversion/eversion. No general conclusion should be drawn on the base of one subject and especially not with the knowledge of existing inaccuracies (see 5.2.8) concerning the kinematics of sub 5. What level of the differences are accounted to skin marker artefacts and how much is contributed by subtalar motion, could not reliably be assigned.

The subjective assessment of curves shape agreement was generally good, which is in line with Westblad et al (2002) and Reinschmidt et al (1997a).

The mean RMS diff compare very well with the mean RMS diff between bone pin and skin marker measurements presented by Westblad et al (2002) (mean RMS diff dorsi-/plantarflexion 1.7\(^\circ\), adduction/abduction 2.8\(^\circ\), inversion/eversion 2.5\(^\circ\)) and are as well comparable to the studies of Nester et al (2007a) (mean RMS diff dorsi-/plantarflexion 2.6° ± 2.0°, adduction/abduction 2.3° ± 2.3°, inversion/eversion 3.6° ± 2.6°) and Reinschmidt et al (1997a) (mean RMS diff dorsi-/plantarflexion 3.1°, adduction/abduction 2.5°, inversion/eversion 3.4°).

The smallest RMS diff between external and internal measurement approaches was found for dorsi-/plantarflexion, which is in agreement to the study of Westblad et al (2002). Especially in relation to the amplitude of the movement, magnitudes of sagittal plane motion were reflected very well with skin markers, whereas the other two planes caused more problems.

Neither \(ROM_{dyn\ pf}\), nor \(ROM_{dyn\ df}\) showed a trend of over- or underestimation. This is contrary to the study of Reinschmidt et al (1997a), where a general overestimation by the skin marker measurements was found. It is as well in contrast to Nester et al (2007a), who found an underestimation of frontal and sagittal plane motion, but in agreement to Westblad et al (2002) who could not state a clear conclusion of over- or underestimation by the external skin marker measurements.

Summarized, the curve shape agreement as well as the ROM of dorsi-/plantarflexion were very well reflected by the skin marker measurements, whereas adduction/abduction and especially inversion/eversion showed larger differences, especially in relation to the actual amplitude of the movement. If this differences can be accounted to subtalar motion or if they result from skin movement artefacts remains to be further evaluated. However, no systematic pattern of over-
or underestimation of the external relative to the internal approach was found.

### 7.4.2 Compensation Strategies

The comparison of the skin marker tracked ankle kinematics to corresponding previous studies was performed visually by an overplot of the data of sub 1 to sub 4 into the overplot of kinematic data of previous studies using skin markers presented in 2.4.4 (see Figures 5 and 7). Skin marker tracked ankle kinematics of the TAA ankle was in line with similar previous studies on TAA kinematics (see Figure 65). And kinematics of the contralateral ankle of sub 1 to sub 4 was in the range of previous studies on healthy normal ankles (see Figure 66).

Except for sub 4, the major differences concerning motion pattern were seen for sagittal plane motion of the ankle, the foot and the knee. Sub 1 showed compared to the contralateral ankle a smaller dorsiflexion peak in the second half of the stance phase and coincident a significantly decreased $ROM_{dyn\,df}$, which seemed to be compensated by an increase in dorsiflexion between fore- and rearfoot. Sub 2 showed in the TAA ankle a decreased initial plantarflexion peak, which seemed to be compensated by an initial supplementary plantarflexion peak in the foot, which was not seen in the contralateral foot. Sub 3 showed a decrease in dorsiflexion of the TAA ankle during the second half of the stance phase. This goes along with a general change in motion pattern, such that the plantarflexion peak is already reached earlier during stance (see also 4.3.3). Likewise, sub 3 showed a change in foot dorsi-/plantarflexion motion pattern. Compared to the contralateral leg, sub 3 experienced at the TAA ankle not only a change in ROM, but also an asymmetry of timing, in other words a typical limp characteristic. For sub 4 decreased ROM at the TAA compared to the contralateral ankle was found for all planes. Since sub 4 showed a much more dynamic flexion/extension motion pattern at the ipsilateral knee, which was manifested by the significantly larger $ROM_{dyn\,ex}$, as well as $ROM_{dyn\,fl}$, it can be assumed that sub 4 was compensating in the ipsilateral knee. Likewise, Piriou et al (2008) noticed an increase in ipsilateral knee motion in arthrodesis patients.

Summarized, all four TAA subjects showed significant differences between ipsilateral and contralateral sagittal plane motion, which is in agreement to previous skin marker studies on TAA kinematics (Muller et al, 2006; Valderrabano et al, 2007). Compensation movements were either seen in the foot or the knee.
Figure 65: Ankle joint sagittal plane motion of TAA subjects during level gait. Mean and SD of three studies (Doets et al, 2007; Dyrby et al, 2004; Muller et al, 2006) gathered and normalized in one plot (color graphs) overlayed with the TAA ankle kinematics of sub 1, 2, 3 and 4 (black graphs).

Figure 66: Ankle joint sagittal plane motion during level gait. Mean and SD of 10 studies (Hunt et al, 2001; Leardini et al, 2007; List et al, 2008; Liu et al, 1997; MacWilliams et al, 2003; Moseley et al, 1996; Rattanaprasert et al, 1999; Simon et al, 2006; Stebbins et al, 2006; Wu et al, 2000) on healthy normal subjects gathered and normalized in one plot (color graphs) overlayed with the kinematics of the contralateral ankle of sub 1, 2, 3 and 4 (black graphs).
7.5 Conclusions

For dorsi-/plantarflexion the motion pattern, as well as the static and dynamic ROM agreed very well between the videofluoroscopy gained and the skin marker tracked joint kinematics. Skin marker measurements estimated tibiocalcaneal motion, videofluoroscopy assessed tibiotalar motion. Since the two approaches agreed very well, two conclusions are possible. Either hardly any subtalar sagittal motion occurred during gait or, provided that the subtalar joint showed sagittal plane motion, skin marker data underestimated sagittal plane tibiocalcaneal motion. Rotations in the frontal and the transverse planes showed larger differences, mainly seen as a tendency in higher magnitudes for the skin marker measurements. If these differences may be accounted to subtalar motion or if they result from skin movement artefacts, remained still unanswered. Future investigations, especially on a higher number of subjects, should include the internal estimation of tibiotalar as well as subtalar motion.

Skin markers gave a good estimate of tibiotalar dorsi-/plantarflexion. Whereas differences for adduction/abduction and inversion/eversion, especially in relation to their actual magnitudes during gait, were not negligible. Especially due to inconsistency between subjects, no general tendency of over- or underestimation was seen.

Comparisons between the ankle, foot and knee kinematics of the ipsilateral to the contralateral leg using skin marker tracking during level gait, revealed for all subjects a deficiency in the TAA sagittal plane ankle motion compared to the contralateral leg. Compensation mechanisms were on the one hand seen between rear- and forefoot and on the other hand at the ipsilateral knee. In general, skin marker tracking is a very powerful tool for an approximate estimation of ankle joint motion, but especially concerning TAA motion, it does not allow the assessment of talar motion. Thus, skin marker measurements are always influenced by a combination of unknown amount of highly individual subtalar participation and the interference by skin movement artefacts.

Skin marker tracking allows full body analyses and the description of joint kinematics of the ipsi- and the collateral leg during consecutive steps, thus during the same trial, which is very powerful for left-right comparisons and therefore to discuss compensation mechanisms. According to the five subjects of this project, tibiotalar sagittal plane motion was very well reflected by the skin marker measurements, transverse and frontal plane tibiotalar motion showed, especially in relation to magnitudes, larger discrepancies.
8 Conclusions & Outlook

The overall goal of this thesis was to develop a technical procedure that allows getting a better understanding of the joint mechanisms of an unconstrained TAA during daily activities, such as level walking, walking uphill, walking downhill and walking on a side slope. The study was performed related to the evaluation of the functionality of an unconstrained TAA (Mobility™ Total Ankle, DePuy) on the basis of the relative motion between the implant components during the stance phase of daily activities. Therefore two in vivo procedures were developed to assess the 3D ankle kinematics during the stance phase of daily motion tasks using videofluoroscopy.

The TAA approach enabled to estimate TAA kinematics, without being limited by skin movement artefacts. This procedure also allowed the separation of the motion which was actually taking place between the talar and the tibial component from motion at adjacent joints, namely the subtalar joint. With skin marker tracking this is by reason of inaccessibility of the talus not possible. The internal tracking of the TAA was achieved by videofluoroscopic image capture and a registration algorithm, that fits a virtual projection of the CAD models of the TAA into the fluoroscopic images. The respective output was the 3D pose of the implant components with an accuracy of 0.4 mm and 0.2° in plane and 2.1 mm and 1.3° out of plane. By the reconstruction in 3D, projection errors are no longer of a problem, compared to classic radiographic assessments. Limitations were found in a restricted field of view and a small but not negligible exposure to radiation. The developed procedure allows an objective analysis of the performance of a TAA with high accuracy. This provides a basis for a better understanding of its functionality.

The tantalum approach enabled to quantify the 3D tibiotalar and talocalcaneal joint kinematics using the same measurement set up and conventions as for the TAA approach, therefore enabling a basis for the comparison of the TAA kinematics to the one of healthy ankles. The internal tracking of the bones was accomplished by the videofluoroscopic image capture of bone embedded tantalum markers. The 3D kinematic information could then be extracted by a three point spatial resection on the base of the geometrical marker arrangements determined by CT segmentation. A large limitation of the tantalum approach was found in the out of plane accuracy. Consequently, the out of plane coordinate of the motion path of the bones had to be restricted. Error sources were twofold, whereas the influence of the tantalum marker tracking precision, though in subpixel range, was affecting the 3D spatial resection more than the CT segmentation precision. The tantalum approach showed further disadvantage of missing data points during stance, caused by an overlapping of the contralateral leg, a restricted field of view.
of the image intensifier, some motion blur, as well as the covering by skin markers. Despite these limitations, it was possible to visualize tendencies of ankle joint complex kinematics and the kinematic data was comparable to literature data covering level gait. To the best of the authors knowledge no comparable studies were found that investigated walking uphill, downhill nor over side inclined slopes. The healthy ankle joints showed dorsi-/plantarflexion as the predominant motion, mainly occurring at the talocrural joint. Adduction/abduction and inversion/eversion motion was small and in the range of variability. The talocalcaneal joint showed higher involvement in the second half of the stance phase with inversion/eversion as the predominant movement. Since walking uphill needed the highest range of maximal dorsiflexion it is suited to reveal problems of restrictions in dorsiflexion. Downhill walking reached the highest range of maximal plantarflexion and overall ROM, which qualified it to detect limitations in the latter two.

The kinematic data gained by the videofluoroscopic procedures was compared to simultaneously tracked skin marker tracked data. The same joint coordinate systems and conventions were used for both approaches. Generally, skin markers gave a good estimate of tibiotalar dorsi-/plantarflexion motion pattern and ROM. Whereas differences for adduction/abduction and inversion/eversion, especially in relation to their actual magnitudes during gait, were not negligible. If this differences may be accounted to subtalar motion or if they resulted from skin movement artefacts remains uncertain. No general tendency of over- or underestimation of tibiotalar joint kinematics by skin marker tracking could be deduced. Though tibiotalar sagittal plane kinematics was well reflected by the skin marker tracked rearfoot to shank kinematics, skin marker measurements are always influenced by an unknown amount of subtalar participation and the interference by skin movement artefacts.

The simultaneously tracked skin marker data enabled the discussion of compensation mechanisms between fore- and rearfoot and in the knee of the the ipsi- and contralateral leg. In terms of the latter comparisons, all subjects showed minor deficiencies in the TAA rearfoot to shank motion compared to the contralateral leg. Compensation mechanisms were seen in the foot and the knee.

Skin marker tracking is a powerful tool for full body analysis and less appropriate for single joint analysis. However, in combination with simultaneous videofluoroscopic assessment of the motion of the TAA itself, skin marker tracking broadened the functionality assessment of the TAA by providing an additional information about compensation mechanisms in the foot and knee.

The in vivo ankle kinematics of four TAA subjects during five different gait conditions was very
accurately assessed by applying the TAA approach. The TAA subjects showed large interindividual differences in static as well as dynamic ROM during all five gait conditions. For all subjects the major motion arose around the talar construction axis, which is definitely favorable for the wear characteristics of the implant, but if it allows a physiological role of surrounding ligaments remains to be seen. Compared to the healthy ankle, during level gait dorsi-/plantarflexion was comparable in its characteristics as well as only showed minor limitations in its magnitudes. Restrictions were mainly seen during walking uphill, where the TAA ankle was more challenged due to restrictions of the ROM in direction of dorsiflexion. Generally, the TAA subjects showed a shift between the static available ROM and the during gait functionally needed ROM in direction of plantarflexion. In other words, all restrictions were caused by a limitation in dorsiflexion, whereas plantarflexion was sufficiently provided. If the restriction in sagittal plane ROM is due to a change in muscular activation resulting in a more pronounced co-contraction, due to a limitation by surrounding passive soft tissues, maybe also due to cicatization, caused by the implant itself or if its multifactorial should be part of further investigations. To improve the functionality of the TAA one should aim to balance the static available to the during gait functionally needed ROM. One possible approach to achieve such a shift could be reached by a change in implant positioning during surgery. Interestingly, contrary to the expectations concerning the unconstrained design, hardly any motion occurred in the transverse and frontal planes. If the latter differences to healthy ankle kinematics show an unfavorable influence on the strain behavior of ligaments requires further investigation.

Summarized, during level gait only minor limitations in sagittal plane motion were found for the TAA subjects, such that one can assume that compensation motions as observed in arthrodeses patients are much smaller in TAA subjects, leading to a smaller disposition of degenerative changes in adjacent joints compared to arthrodesis patients. The TAA showed a close to, but not in all parameters physiological motion.

One improvement of the functionality of the TAA could be achieved by a balancing of the static available to the during gait functionally needed ROM by a change in implant positioning during surgery.

The developed procedure allowed a first insight into the joint kinematics, thus functionality, of an unconstrained TAA during daily motion tasks. With its appliance on a larger number of subjects it has the potential to help clinicians and implant developers to evaluate current TAA designs and future design modifications.
Outlook

The presented procedures show room for technical enhancements and future investigations. In the following some ideas are collected:

Technical Enhancements

- To overcome the restriction in the field of view of the image intensifier, the c-arm could be movable, likewise to what has been done by Foresti (2009) to track the knee joint. A translation of about 150 mm in 400 ms in gait direction and simultaneous translation of about 60 mm in 300 ms in vertical direction should allow tracking the whole stance phase including toe off. The motion of the c-arm could be triggered by a change in pixel grey values by the toes entering the field of view.

- The 3D reconstruction of the tantalum approach could be improved by an additional matching of the bone contours similar to the TAA approach. With the start values provided by the Ta markers, the bone contours, or part of it could be fitted and thereby the pose improved.

- The accuracy of the 3D reconstruction in terms of out of plane accuracy could be improved by a dual plane approach. Though, to allow the capturing of dynamic motion tasks, the arrangement of the two c-arms would need to be further evaluated and tested.

- Valuable information could be gathered by simultaneously assessing ground reaction forces using force plates.

- The procedures can be adapted and used for the investigation of other joints and implants.

Future Investigations

- To assess a possible connectivity between pain and ROM, a retrospective study on a larger patient sample, classified into the following four groups would be of large clinical interest: (i) good postoperative clinical ROM and no pain (ii) small postoperative clinical ROM and no pain (iii) small postoperative clinical ROM and pain (iv) good postoperative clinical ROM and pain.

- Further information concerning the functionality of the TAA could be achieved by a model estimation of the effects of a change in motion pattern on ligament strains.

- The influence on implant positioning during surgery could be studied in a cadaver study.
Abbreviations and Terminology

ankle abduction rotation about the vertical axis, such that the foot rotates externally relative to the shank

ankle adduction rotation about the vertical axis, such that the foot rotates internally relative to the shank

ankle dorsiflexion rotation about mediolateral axis, such that toes are brought closer to the shin

ankle joint articulation formed by tibia, fibula and talus

ankle joint complex functional unit formed by ankle and subtalar joint

ankle plantarflexion rotation about mediolateral axis, such that toes are brought closer to the calf

eversion rotation about the posteroanterior axis, such that the lateral edge of the foot is lifted

inversion rotation about the posteroanterior axis, such that the medial edge of the foot is lifted

subtalar joint articulation formed by talus and calcaneus

talocrural joint synonym to ankle joint
	
tibiocalcaneal motion combined tibiotalar and subtalar motion

tibiopedal motion combined tibiotalar, subtalar and midtarsal motion; motion between foot and shank

tibiotalar motion motion between talus and tibia

Agility Agility (DePuy, Warsaw Indiana)

Buechel-Pappas Buechel-Pappas (Endotec, South Orange, New Jersey)

German Ankle System German Ankle System (R-Innovation, Coburg, Germany)

Hintegra HINTEGRA (Integra LifeSciences/Newdeal, Lyon, France)

IAA Irvine Ankle Arthroplasty (Howmedica, Rutherford, NJ, USA)

Mobility™ Total Ankle Mobility (DePuy International, Leeds, UK)

New Jersey LCS New Jersey LCS (DePuy, Warsaw Indiana)

Salto Salto Talaris (Tornier, Saint Ismier, France)

STAR STAR (Waldemar Link, Hamburg, Germany)

2D two-dimensional

3D three-dimensional

adab adduction/abduction

AJC ankle joint center

AOFAS American Orthopaedic Foot and Ankle Score

BMT basic motion task

df dorsiflexion

dnh downhill

dyn dynamic

HJC hip joint center
HS  heel strike
invev  inversion/eversion
KJC  knee joint center
lig.  ligament
max  maximum
min  minimum
pf  plantarflexion
pfdf  plantarflexion/dorsiflexion
rel  relative to
RMS  root mean square
ROM  range of motion
SD  standard deviation
ssb  side slope bottom
sst  side slope top
stat  static
TO  toe off
Ta  tantalum
TAA  total ankle arthroplasty/arthroplasties
$u_k$  functional estimated knee joint axis
uph  uphill
vs.  versus
References


Barnett CH, Napier JR (1952) The axis of rotation at the ankle joint in man; its influence upon the form of the talus and the mobility of the fibula. J Anat 86(1):1–9


Fick R (1911) Handbuch der Anatomie und Mechanik der Gelenke. Gustav Fischer Verlag, Jena


Gray H (1858) Anatomy - Descriptive and surgical. John W. Parker and Son, West Strand, London


Inman V, Mann R (1978) Biomechanics of the foot and ankle. DuVries’ Surgery of the Foot, Edited by Mann, RA Mosby, Saint Louis pp 3–21


List R (2005) A hybrid marker set for future basic research and instrumented gait analysis at the Laboratory for Biomechanics. Diploma thesis, ETH Zurich


Michelson JD, Helgemo J S L (1995) Kinematics of the axially loaded ankle. Foot Ankle Int 16(9):577–82

Michelson JD, Schmidt GR, Mizel MS (2000) Kinematics of a total arthroplasty of the ankle: comparison to normal ankle motion. Foot Ankle Int 21(4):278–84


Pappas MJ, Buechel FF (2005) Biomechanics and design rationale: the Buechel-Pappas\textsuperscript{TM} ankle replacement system. URL www.endotec.com


Tsai T, Lu T, Kuo M, Hsu H (2006) In vivo measurement of the kinematics of normal and acl-deficient knees using fluoroscopy with voxel-based bone models. In: 5th World Congress of Biomechanics, Munich, Germany


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