Mobile-Bearing Total Ankle Arthroplasty

A Fundamental Assessment of the

Clinical, Radiographic and Functional Outcomes

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Le rêve, c’est tout –
la technique, ça s’apprend
Jean Tinguely

Voor Nadia
Voor mijn kinderen
Voor Truus en Rob

Contents

Chapter 1

General Introduction and Aim of this Thesis

1.1 Introduction

End-stage ankle arthritis can lead to significant symptoms, such as considerable pain during standing and walking, and to a reduction of functional capacities, such as a restricted walking distance and difficulty with stair climbing and stair descending. Two surgical treatment options are available for end-stage ankle arthritis: fusion of the ankle joint (ankle arthrodesis) and prosthetic replacement or total ankle arthroplasty (TAA). Ankle arthrodesis is considered the standard surgical treatment for the severely affected ankle joint, also because inferior results have been published after TAA with use of fixed-bearing 2-component prostheses. A successful ankle arthrodesis will produce a painfree and stable joint. However, the result of ankle arthrodesis can be compromised in the event of malunion or non-union. Moreover, ankle arthrodesis might significantly reduce the patients functional capacity, especially in poly-articular disease, such as in patients with rheumatoid arthritis, and even more so in the event of bilateral ankle arthritis. Finally, bilateral ankle arthrodesis leaves the patient with difficulties during walking, rising from a chair, and climbing and descending stairs.

Three-component mobile-bearing total ankle prostheses appear to give a promising biomechanical solution for the difficulties described in the past with constrained fixed-bearing total ankle prostheses. Until recently, literature on TAA with use of mobile-bearing designs was sparse. For that reason, the role of such designs in the armamentarium of the orthopaedic surgeon still needs to be defined more precisely. Also, the indications for this procedure require an in-depth evaluation.

Our personal experience with mobile-bearing TAA began in 1988, when the cementless New Jersey Low Contact Stress total ankle prosthesis (LCS®, DePuy, Warsaw, Indiana, USA) was introduced at the Slotervaartziekenhuis, Amsterdam. Clinical experience was continued and broadened with the use of its successor, the Buechel-Pappas prosthesis (Endotec, South Orange, New Jersey, USA): from 1993 onwards in Amsterdam, and from 1995 onwards at the Leiden University Medical Center. Early clinical results with these designs were promising, and, with growing experience, showed improvement in result to a level that it became a reliable treatment alternative. Furthermore, the good clinical outcome of mobile-bearing TAA initiated the need for research on the functional outcome of this procedure, in order to enable a more thorough comparison with the outcome of ankle arthrodesis.

1.2 Causes of Ankle Joint Degeneration

The ankle joint can develop arthritic degeneration as a result of several pathologic conditions. Posttraumatic arthritis is the most frequent condition leading to deterioration of the ankle joint[1]. It can occur as sequelae of fractures either around the ankle or at a distance (lower leg or hindfoot). Furthermore, posttraumatic arthritis can develop secondary to chronic ligament laxity or as a late result of a single ankle sprain[2]. Inflammatory joint disease, usually rheumatoid arthritis (RA), will also, with longer duration of the disease, lead to destruction of either the joints of the hind- and midfoot, or the ankle, or both in a substantial number of patients[3,4,5]. In this context, it should be realized that rheumafactorpositive rheumatoid arthritis (Rf+RA) is known to produce more radiographic destruction than rheumafactornegative rheumatoid arthritis (Rf-RA)[6]. Other, infrequent causes of ankle arthritis are: osteonecrosis of the talus, osteochondritis, hemophilic arthritis, haemochromatosis, crystal arthropathy and primary osteoarthritis. In contrast to the incidence of primary osteoarthritis of the hip and the knee, primary osteoarthritis of the ankle has a relatively low prevalence[1].

Symptomatic ankle arthritis is known to have an important negative influence on health as perceived by the patient, on his quality of life, and on walking capacity[7,8,9]. Therefore, end-stage ankle arthritis is expected to have an important negative influence on the ability of these patients to perform their normal occupational and recreational activities. It should furthermore be taken into account that both posttraumatic and inflammatory ankle arthritis usually occur at a younger age than the average age of patients with degenerative hip or knee osteoarthritis. Thus, the aim of surgical treatment should be to give a reliable and long-lasting improvement of the clinical symptoms and the functional capacities of patients with end-stage ankle arthritis.

1.3 Surgical treatment Options of the Arthritic Ankle

There are four established surgical treatments for the osteoarthritic ankle: debridement, corrective osteotomy, ankle arthrodesis and total ankle arthroplasty. Synovectomy is a treatment option in therapy-resistant inflammatory disease, and is mainly indicated in ankles without significant cartilage loss[10]. Ankle joint distraction with use of an external fixator appears to give promising results in osteoarthritic patients[11]. However, the use of this technique has not yet gained wide acceptance, perhaps because of the relatively long period of distraction required: up to 3 or 5 months. Debridement, preferably carried out as an arthroscopic procedure[12,13], and supramalleolar[14] or calcaneal osteotomy[15] can be successful in ankles with localized arthritic changes with moderate symptoms. For the severely affected ankle joint, only two reconstructive surgical options are available: arthrodesis and prosthetic replacement. Until recently, arthrodesis was considered the best treatment. However, arthrodesis is known to have certain surgical disadvantages that could compromise both its short-term and long-term outcome, such as a risk of nonunion and malunion[16,17,18,19]. In addition, arthrodesis will significantly compromise gait.[16,20,21] Finally, with longer follow-up, there is a significant risk of deterioration of the joints of the ipsilateral foot, frequently resulting in recurrent symptoms[22,23,24].

1.4 History of Total Ankle Arthroplasty

Potentially, TAA has certain advantages over ankle arthrodesis, both in monoarticular and more specific in polyarticular disease: gait is probably less compromised, and adverse effects on the other joints of the lower extremity are not expected to be of importance[25]. Prosthetic replacement of the ankle started with the introduction of fixed-bearing 2-component designs in the 1970s. Unsatisfactory results with these first-generation 2-component designs have been reported after short-term follow-up[26,27,28,29]. Hence, for the decades to follow, TAA has generally not been seen as an acceptable alternative to arthrodesis. One should realize however, that these results have been achieved with constrained 2-component designs of either a cylindrical or a spherical geometry. The significant intrinsic constraints of these first-generation 2-component designs do not allow an unrestricted ankle motion, and will lead to shear and tensile forces at the bone-prosthesis interface. This easily explains their high incidence of mechanical failure.

New interest in TAA started with the introduction of second-generation semi-constrained 2-component[30] and of unconstrained 3-component designs. Of particular importance for this renewed interest was the development of the unconstrained 3-component mobile-bearing designs. The cementless New Jersey low contact stress prosthesis, introduced in 1981[31], was the first design to apply the mobile-bearing concept to the ankle. Its successor, the Buechel-Pappas mobile-bearing prosthesis[32], and the cementless version of the mobile-bearing Scandinavian total ankle replacement[33] (STAR, Waldemar Link, Hamburg, Germany) were introduced respectively in 1989 and in 1990 in the centers of their designing surgeons. Some years later, mobile-bearing prostheses gained acceptance in several centers in Europe and in North America. Recently, good medium-term results have been described with these two designs (Buechel-Pappas and STAR)[33,34,35,36,37]. In recent years, interest in TAA has raised tremendously. This is reflected by the increasing number of procedures performed in an increasing number of orthopaedic centers, by an increase in scientific publications reporting on the outcome of these implants (see Fig. 1), and by an increasing number of implants that have been developed and were introduced on the market.

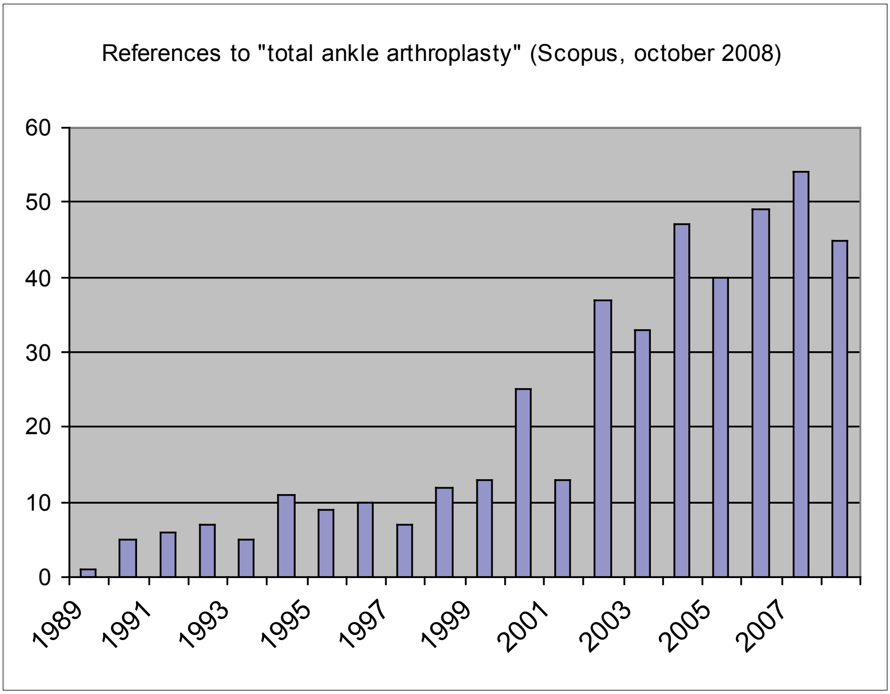

Fig. 1 Graph showing the number of scientific publications since 1989 with "total ankle arthroplasty" as subject (source: www.scopus.com).

1.5 Aim of this Thesis

The aim of this thesis is to assess the long-term clinical and radiographic outcome of mobile-bearing total ankle arthroplasty and to assess its short-term functional outcome by gait analysis and by an analysis of the energy expenditure during level walking.

Additional aims of this thesis are the assessment of the migration pattern of the tibial component, the development of a reliable surgical technique for the endoprosthetic replacement of the arthritic ankle with varus deformity, and the evaluation of the results of salvage arthrodesis after failed TAA.

1.6 Outline of this Thesis

A prospective clinical study was started at the introduction of the mobile-bearing total ankle prosthesis in the two participating centers. TAA was introduced at the Slotervaartziekenhuis in 1988 and at the Leiden University Medical Center in 1996. This study investigated the long-term clinical and radiographic outcome of TAA in a consecutive cohort of patients suffering from inflammatory arthritis, and evaluated risk factors for failure. Later on, studies were started to investigate the following aspects: the migration pattern of the tibial component, as assessed by radiostereometry; the outcome of a new surgical technique for the correction of varus deformity in the arthritic ankle requiring ankle replacement; and the outcome of salvage of the failed TAA by ankle arthrodesis. In addition to the clinical and radiographic outcome studies, a thorough assessment of the functional outcome after successful TAA was thought to be necessary. Therefore, studies were performed to investigate gait kinematics and kinetics of the replaced ankle in comparison with the normal ankle, followed by a comparative study of the energy expenditure during barefoot walking. These functional outcome studies are the result of a close collaboration with the Faculty of Human Movement Sciences, VU University Amsterdam. Finally, a meta-analysis of the literature was done in order to assess the reported status of mobile-bearing TAA with respect to clinical and radiographic outcome.

Part I of this thesis, the clinical and radiographic outcome part, consists of six chapters:

In ***Chapter 2*** a short description of the anatomy and of the kinematics of the normal ankle joint is given, followed by a description of the pathology of the arthritic ankle.

In ***Chapter 3*** the conservative treatment options and the result of non-endoprosthetic surgical treatment are described.

In ***Chapter 4*** the medium to long-term clinical and radiographic result of mobile-bearing TAA in a prospectively followed consecutive cohort of patients suffering from inflammatory joint disease is presented, and risk factors for failure are evaluated.

In ***Chapter 5*** the stability of the tibial component of the Buechel-Pappas prosthesis is assessed in a radiostereometric analysis study.

In ***Chapter 6*** a new technique for the correction of varus deformity at the time of TAA is described, and the medium-term results of ankles treated by this technique are presented.

In ***Chapter 7*** the results of salvage by ankle arthrodesis of failed TAA are given and are compared with data from the literature.

Part II of this thesis, the functional outcome part,
consists of three chapters:

In ***Chapter 8*** a gait analysis study is presented of patients after successful mobile-bearing TAA in comparison with a healthy control group, matched for age and gender.

In ***Chapter 9*** a kinetic study of the ankle after TAA is presented, based on the same experimental study as described in chapter 8.

In ***Chapter 10*** the energy consumption during barefoot walking on a treadmill in patients after successful mobile-bearing TAA is presented and compared to a healthy control group.

Part III of this thesis consists of a review of the literature on the prosthetic replacement of the arthritic ankle, and of a discussion of the role of TAA in the treatment of the arthritic ankle.

In ***Chapter 11*** an overview of the designs currently available on the market is given, followed by a review of the clinical and radiographic outcome of several fixed and mobile-bearing designs.

In ***Chapter 12*** the preoperative evaluation of the patient with an arthritic ankle and the preferred surgical technique of TAA are discussed.

In ***Chapter 13*** a general discussion is given on the current and future role of TAA, based on the work presented in this thesis.

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Chapter 2

Anatomy and Biomechanics of the Normal Ankle and Pathology of the Arthritic Ankle

2.1 Ankle Anatomy

The ankle or talocrural joint is the junction between the foot and lower leg. Its gross osseous anatomy is relatively simple, as it consists of the ankle mortise, formed by the distal aspects of the tibia and fibula, which articulates with the talus, the uppermost tarsal bone. The ankle mortise encloses the superior aspect of the talus tightly (Fig. 1). The fibula and tibia are strongly connected to each other by the interosseous membrane and by the anterior and posterior tibiofibular ligaments, which together form the tibiofibular syndesmosis. The superior aspect of the talus consists of the talar dome, having a superior articular surface which is wider anteriorly than posteriorly. The radius of the talar dome is smaller medially than laterally[1], thus the talar dome can be seen as the segment of a cone with its base at the lateral side. The upper surface of the talar dome articulates with the tibial plafond. The medial and lateral articular facets of the talar dome articulate with the medial and lateral malleolus respectively. The lateral malleolus lies more posteriorly than the medial malleolus, and it also extends more distally. On the medial side the talus is connected to the medial malleolus by the strong deep and superficial medial or deltoid ligaments. They are the most important passive stabilizing structures of the ankle joint, from a clinical perspective comparable to the posterior cruciate ligament of the knee. On the lateral side the talus is connected to the lateral malleolus by the anterior and posterior talofibular ligaments. These latter two ligaments form, together with the calcaneofibular ligament, the lateral ligament complex. They are the passive elements controlling the rotation in the ankle joint, comparable to the function of the anterior cruciate ligament in the knee. Mechanoreceptors have been identified in both the medial and lateral ankle ligaments, and are important proprioceptive elements[2].

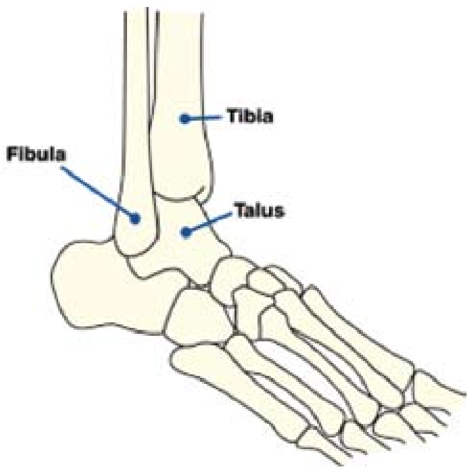

Fig. 1-A

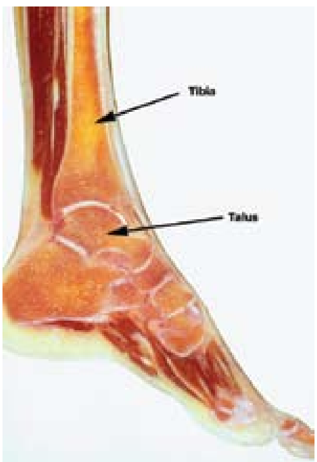

Fig. 1-B

Figs. 1-A through 1-C Osseous anatomy of the ankle and foot. ***Fig. 1-A*** Schematic osseous anatomy of the ankle and foot. ***Fig. 1-B*** Cross-section in the sagittal plane of the lower leg, ankle and foot through the tibia and talus. ***Fig. 1-C*** Drawing of the osseous and ligamentous structures of the ankle and hindfoot, as seen from anteriorly.

Fig. 1-C

The soft tissues around the ankle joint consist of tendons and neurovascular structures. The tendons are located in 3 compartments, all covered by fascial layers having retinacular reinforcements:

1) the anterior compartment, where the tendons of the anterior tibial, extensor hallucis longus, extensor digitorum longus and peroneus tertius muscles are located (Fig. 2), together with the deep peroneal nerve and the anterior tibial vessels. Somewhat proximal to the ankle joint medial and lateral malleolar branches branch off the anterior tibial vessels. These branches supply blood to the anterior, medial and lateral aspects of the ankle joint. The anterior tibial vessels and deep peroneal nerve are at risk during an anterior approach of the ankle joint, and should be protected by opening the ankle joint at the medial side, thus creating a laterally based soft tissue flap.

Fig. 2-A

Fig. 2-B

Fig. 2-C

Fig. 2-D

Figs. 2-A through 2-D The anterior compartment with a demonstration of the anterior approach to the ankle joint. ***Fig. 2-A*** The anterior tibial tendon, the long extensor tendons, and the peroneus tertius tendon, after removal of the extensor retinaculum. ***Fig. 2-B*** Approach between the anterior tibial and extensor hallucis longus tendons, showing the neurovascular bundle overlying the joint capsule. ***Figs. 2-C and 2-D*** Anteromedial arthrotomy to the ankle joint, medial from the neurovascular bundle.

2) the posteromedial compartment, where the tendons of the tibialis posterior, flexor digitorum longus and flexor hallucis longus muscles are located, together with the tibial nerve and the posterior tibial vessels (Fig. 3). The flexor hallucis longus tendon and the posterior tibial artery are at risk for injury during preparation in the posterior part of the ankle joint when the posterior capsule is released from the distal tibia.

Fig. 3-A

Fig. 3-B

Figs. 3-A and 3-B The soft tissues of the posteromedial compartment. ***Fig. 3-A*** The tendon of the flexor digitorum longus muscle overlying the posterior tibial tendon, together with the posterior tibial artery and veins, and the superficial (tibiocalcaneal) medial collateral ligament. ***Fig. 3-B*** After reflection of the superficial medial collateral ligament, the deep (tibiotalar) medial collateral ligament is seen.

3) the lateral, or peroneal compartment, where the tendons of the peroneus brevis and longus muscles are located (Fig. 4), together with the peroneal vessels and sural nerve.

Fig. 4

Fig. 4 Lateral view of the ankle, after removal of the superficial fasciae and retinaculae. The peroneus brevis and longus tendons are seen, running in the peroneal compartment. Posterior to these tendons the short saphenous vein is visible, and anterior to the lateral malleolus are the anterior lateral malleolar vessels. The origin anterior talofibular ligament is clearly visible, the posterior talofibular and the calcaneofibular ligaments are lying under the peroneal tendons.

The following structures are encountered during the anterior surgical approach to the ankle joint (from superficial to deep): the superficial peroneal nerve in the subcutaneous fat, the superior and inferior extensor retinaculum, the interval between the anterior tibial and extensor hallucis tendons, the medial malleolar vessels and the anterior capsule of the ankle joint. The medial dorsal nerve to the foot (a branch of the superficial peroneal nerve) is at risk to be injured during the anterior approach.[3] This branch has a variable course. It normally runs in the vicinity of the wound in the most distal part of the anterior incision, but can have a more medial course, thereby requiring extensile preparation and mobilization (Fig. 5). In order to avoid injury to the anterior tibial vessels the ankle joint should be opened medially. The medial malleolar vessels should be ligated or cauterized in order to prevent bleeding.

Fig. 5 Intraoperative image of the medial dorsal branch (arrow) of the superficial peroneal nerve in relation to the sutured retinaculum.

Fig. 5

2.2 Biomechanics and Kinematics of the Normal Ankle and Hindfoot

The ankle joint is highly loaded during walking, up to five times body weight[4], although it has a relatively small surface contact area of about 350 square mm[5]. Most of the load is distributed through the superior articular surface of the talus[6], while the remaining load is transmitted through the talar facets, with the medial facet accepting twice the load of the lateral. The contact area is the largest with the talus in neutral position and in dorsiflexion, the position of the ankle joint during about half of the stance phase[6]. These forces are transmitted further downwards to the foot, and the medial tarsal bones (navicular, cuneiform bones) play an important role in this force transmission.

Although the range of motion of the tarsal joints is less compared to the ankle, its range of motion is higher than was previously thought. The motion patterns of the individual tarsal joints (talocalcaneal, talonavicular and calcaneocuboid) are coupled, leading to a constraint motion pattern of the hindfoot when it moves from eversion to inversion. This constraint motion pattern is defined by the articular structures (both by the geometry of the joint surfaces and by the articular ligaments). Detailed cadaver experiments of the tarsal kinematics have been done by Van Langelaan[7] with use of roentgen stereophotogrammetry. He found that a motion pattern around helical axes existed of both the talocrural and the individual tarsal joints. Benink[8] confirmed these findings in a continuous motion setup study in vitro and in vivo. Lundberg et al[9,10,11,12] carried out in-vivo kinematic studies of the ankle and hindfoot in healthy volunteers by roentgen stereophotogrammetry. They described that the dorsiflexion-plantarflexion motion at the ankle joint occurs around a horizontal axis in dorsiflexion and that, with increasing plantarflexion, this axis rotates into varus. Furthermore, they found that a large inter-individual variance exists in the inclination of the vertical rotation axis of the ankle joint. Their studies confirmed the cadaver findings by Van Langelaan[7] and Benink[8]. An excellent overview of the functional anatomy of the ankle and foot was written by Huson[13]. He showed that the subtalar joint has a close-packed condition in the neutral position, and that with movement into inversion a loose-packed condition develops. The inversion movement in the subtalar joint is combined with an external rotation movement of the lower leg with respect to the foot. This means that in the loaded situation the talus shifts laterally and the calcaneus medially, so that axial loading remains in the axis of the lower leg. The ankle joint has a close-packed position in dorsiflexion, and rotation around the longitudinal axis becomes possible with plantarflexion.

To summarize, it can be stated that the kinematics of the loaded ankle and hindfoot joints are linked. This means that a rotation of the lower leg will result in motion at the tarsal joints, and furthermore, that a reduced motion of the ankle joint will produce an abnormal kinematic pattern at the hindfoot and vice-versa. Therefore, any disturbance in the normal function of the ankle could have its influence on the foot.

2.3 Pathology of the Arthritic Ankle

Aging has a limited effect on the quality of the articular cartilage of the ankle, in which it differs from the cartilage of the knee and hip[14]. This might explain the low incidence of primary (idiopathic) osteoarthritis of the ankle joint in the elderly, as was seen in epidemiologic studies[15,16]. Compared to other joints, ankle arthritis mostly develops as a late result of either trauma or of inflammatory joint disease. This explains that patients developing ankle arthritis in general are younger than those developing degenerative arthritis of the hip or knee[5].

Post-fracture arthritis can be the result of either a malleolar fracture, a tibial plafond fracture, a talar fracture or a lower leg fracture. If such fractures have healed with correct alignment, arthritis is probably due to cartilage damage at the time of the initial injury. High-energy injury is expected to have a higher risk of such cartilage damage, but ankle arthritis can also occur after low-energy injury. However, symptoms caused by posttraumatic arthritis in well-aligned ankles can remain at a tolerable level for many years until, finally, surgical reconstruction may become necessary. Fractures that have healed with either limited or gross malalignment are of course more prone to develop post-fracture arthritis[17]. If the ankle mortise is abnormal, either due to a shortened and/or externally rotated fibula[18], due to a malunited posterior malleolus[19], or due to a medial ligament injury[20], the joint contact area will be reduced significantly, leading to localized cartilage overload. This cartilage overload could then result in early and rapidly progressive ankle arthritis. A non-anatomically healed tibial plafond fracture will frequently cause the same phenomenon of early and progressive ankle arthritis. Extra-articular malalignment after, for example a lower leg fracture, in general has a less severe impact on the ankle joint, and will mostly not lead to rapidly progressive disease.

Posttraumatic arthritis due to chronic ligament laxity might be the result of cartilage damage at the time of the initial injury, or, more likely, due to recurrent ankle sprains. Inversion injuries frequently lead to symptomatic arthritis. Schaap et al[21] reported that 30 per cent experienced residual symptoms after 9 months of follow-up. At 6.5 years 39 per cent of a subgroup treated non-operatively remained symptomatic[22]. Lateral ligament insufficiency results in a rotatory instability of the ankle joint, giving rise to shear forces on the cartilage, and thereby to a slowly progressive cartilage damage and to ankle arthritis. Thus, if not healed properly, chronic instability may result in arthritis: ***instability arthritis.*** The typical late result of instability arthritis is an ankle with moderate to severe varus deformity, whereby the talus lies tilted in the ankle mortise and cartilage loss at the medial part of the ankle joint exists. Not uncommon, this pathology can occur bilaterally, and thereby might result in substantial functional impairments for the patient (Fig. 6).

Fig. 6

Fig. 6 Anteroposterior weightbearing radiograph of the ankles of a 50-year old man, a former soccer player, suffering from bilateral instability arthritis. Asymmetric joint space narrowing exits at the medial aspects of both talocrural joints, together with a concomitant varus deformity of 15 degrees.

With inflammatory joint disease, arthritis can occur both at the ankle and the foot, however with a changing and not always predictable expression and disease pattern. Over time, most patients suffering from RA will develop symptoms both in the foot and the ankle. Vainio reported a 91 per cent prevalence of foot and ankle symptoms in female rheumatoid patients and 85 per cent in male rheumatoid patients in an in-patient setting[23]. Another study noted a 94 per cent prevalence of foot and ankle symptoms in an outpatient setting[24]. They also reported that at a mean 13.5 years of disease duration 42 per cent of patients thought their ankle and/or hindfoot symptoms were worse than their forefoot symptoms, while 28 per cent thought their forefoot symptoms were worse. Forefoot involvement has been reported to occur more often early in the disease, while hindfoot involvement would be more prevalent with longer disease duration. In the rheumatoid hindfoot, the typical deformity is valgus deformation, as a result of tenosynovitis and insufficiency of the posterior tibial tendon and muscle, and of synovitis of the tarsal joints (the talonavicular joint in particular). Rheumatoid valgus deformity is a relatively early phenomenon and can affect gait significantly[25]. Valgus deformity of the hindfoot can eventually result in eccentric overload of the ankle joint and, especially in combination of synovitis of the ankle joint, in destructive changes of the ankle joint with a limited to moderate degree of valgus deformity at the level of the ankle joint itself. Sometimes valgus deformity of the ankle-hindfoot complex can, in the event of a coexisting osteopenia, result in a spontaneous fracture of the distal fibula, and thereby to a progressive deformity (Fig. 7).

Fig. 7

Fig. 7 Anteroposterior radiograph of the ankle joint of a female patient suffering from longstanding rheumatoid arthritis and severe arthritic changes of the ankle joint. She developed a spontaneous distal fibular fracture, aggravating a pre-existing valgus deformity of the ankle and hindfoot.

Inflammatory joint disease of the ankle without concurrent hindfoot disease occurs infrequently, and, if so, in general has a slowly progressive disease pattern. In such a situation, due to ankle synovitis, the lateral ligaments can become attenuated, and this could then also result in a secondary varus deformation, similar to instability arthritis. Varus deformity in inflammatory joint disease was seen in our two-center study in eight ankles out of ninety-three[26], and apparently is not a rare phenomenon. Inflammatory joint disease is furthermore characterized by the relatively high incidence of osteopenia and of cystic lesions in the juxta-articular osseous structures (Fig. 8). These rheumatoid comorbidities could compromise both the surgical procedure as expressed by a higher rate of intraoperative fractures, and the result at follow-up, as expressed by a higher rate of aseptic loosening and edge-loading due to persistent deformity in the frontal plane.

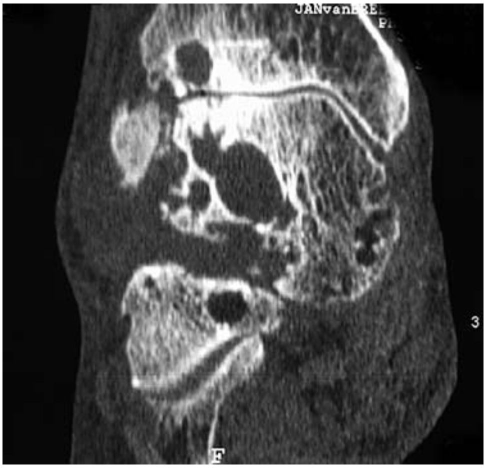

Fig. 8-A

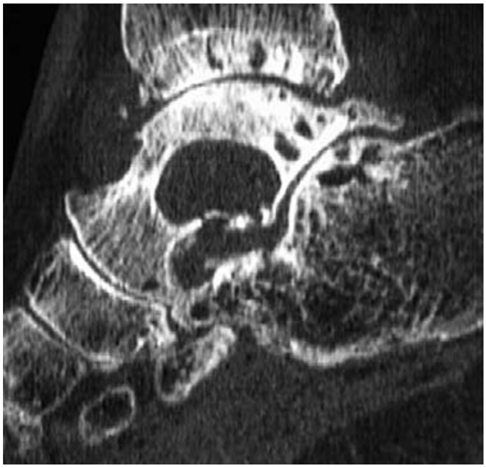

Fig. 8-B

Figs. 8-A and 8-B Computer tomography scan with reconstructions in the frontal ***(Fig. 8-A)*** and sagittal ***(Fig. 8-B)*** plane of the right ankle of a 67-year old female patient with rheumatoid arthritis. There is severe cartilage loss at the ankle joint and large intra-osseous cysts have developed in the talus, together with smaller cysts in the distal tibia and superior part of the calcaneus.

Note: Fig. 1-C has been obtained from Primal Pictures Ltd, ***Fig. 1-B*** has been obtained from the Department of Anatomy, Leiden University, and ***Figs. 2 through to 4*** have been made at the Department of Anatomy, Leiden University.

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Chapter 3

Conservative and Non-Endoprosthetic Surgical Treatment Options for the Arthritic Ankle

3.1 Conservative Treatment

Conservative treatment options for the arthritic ankle consist of the following modalities: shoe adaptations like a rocker sole or a surgical shoe, ankle-foot orthoses, analgesic medication, physical therapy, walking aids, and intra-articular injections with corticosteroids or hyaluronic acid. Few reports exist on the efficacy of such treatment options in patients with ankle or hindfoot disease. Thompson et al[1] showed that foot orthoses were more effective than nonsteroidal anti-inflammatory drugs in the treatment of symptoms due to osteoarthritis of the ankle or foot. Huang et al[2] investigated the effect of different custom-made orthoses (an ankle-foot orthosis, a rigid and an articulated hindfoot orthosis) on the restriction of ankle-hindfoot motion in 13 patients with ankle osteoarthritis. They found that the rigid hindfoot orthosis allowed more forefoot motion and was as efficient as an ankle-hindfoot orthosis in restricting motion at the ankle-hindfoot complex. Woodburn et al[3], in a randomized trial, studied the effectiveness of early foot orthosis intervention for painful correctable valgus deformity of the hindfoot in rheumatoid arthritis. They found that foot orthoses used continuously resulted in a reduction in foot pain, foot disability and functional limitation. March et al[4], in a n-of-1 study comparing the efficacy of nonsteroidal anti-inflammatory drugs and paracetamol for the treatment of osteoarthritis, found that many patients may achieve adequate control with paracetamol alone. Intra-articular injection therapy with hyaluronic acid has been shown to be of value for the treatment of ankle arthritis in a double-blind randomized controlled trial[5].

3.2 Non-Endoprosthetic Surgical Procedures

Besides TAA, the following surgical treatment options are available: ankle joint distraction, debridement, realignment osteotomy and ankle arthrodesis.

Results with ankle joint distraction with use of an external fixator were first described in 1995 by van Valburg et al.[6]. Good results have been described in a prospective study in the majority of a population with end-stage ankle arthritis, and improvement appeared to continue over time[7]. Currently however, few centers have applied this technique, and so it remains unclear whether distraction can safely be used on a wide scale for ankle arthritis.

Arthroscopic removal of osteophytes in the anterior compartment is usually successful if the osteoarthritic changes are localized, but less successful if generalized osteoarthritis is present[8,9].

Supramalleolar osteotomy, either as an isolated distal tibia procedure or as a procedure for both distal tibia and fibula for the treatment of pathologic entities of the adult distal tibia and foot and ankle is technically demanding and requires an extensive and careful preoperative planning. For varus deformities, a medial opening wedge osteotomy has the advantages of an easy-to-make bone cut and of no resultant leg-length discrepancy, but the potential disadvantages of graft morbidity, failure of graft incorporation, delayed healing of the osteotomy, necessity for greater fixation strength, and potentially increase in the medial joint load by tensioning of the medial extrinsic tendons. Lateral closing wedge osteotomies have the advantages of easy fixation, no graft requirement, a reliable and rapid healing, and no possibility of medial joint load increase[10,11]. For instability arthritis combined with a pre-existing cavovarus foot, lateral ligament reconstruction and valgus calcaneal osteotomy (Dwyer type) has been recommended[12].

3.3 Ankle Arthrodesis

Ankle arthrodesis can be considered if the ankle joint shows severe cartilage loss and the patient experiences difficulties in performing normal activities of daily life. Many surgeons still consider fusion as the treatment of choice for the severely affected ankle joint. The optimum position for ankle fusion is considered to be: neutral in the sagittal plane, slight valgus (5 degrees) and slight external rotation of the hindfoot (5 to 10 degrees), and some retroposition of the talus with respect to the tibia[13]. However, ankle arthrodesis is not an easy surgical procedure and therefore has not a predictably good result. In a recent systematic review of the literature on the intermediate to long-term outcome after ankle fusion (mean follow-up time 5.3 years, range 1.9-23) and total ankle arthroplasty (mean follow-up time 4.7 years, range 2.3-9), Haddad et al[14] reported a ten per cent nonunion rate (39 studies dealing with ankle fusion with a total of 1262 patients included; almost all studies were retrospective in nature). Revision of the arthrodesis was done in nine per cent, mainly for nonunion or infection. Furthermore, five per cent of the patients eventually underwent a below-knee amputation. Reasons for the amputations were not specified in this meta-analysis. Posttraumatic arthritis was the primary indication for arthrodesis in 57 per cent. Mean AOFAS ankle score at follow-up was 75.6 (95% confidence interval 71.9 to 84.5) and according to patient assessment 74.1 per cent experienced an excellent or good result. Comparing these results with the pooled data from 10 studies on TAA, they concluded that the intermediate outcome of total ankle arthroplasty appears to be similar to that of ankle arthrodesis and that comparative studies are needed.

The following early complications are not infrequent after ankle fusion: infection, nonunion and malunion. An important late finding is the high radiographic incidence of hindfoot arthritis[16,17,18]. Fortunately, hindfoot arthritis is not always clinically symptomatic. When hindfoot symptoms build up after ankle fusion they can be difficult to treat. Especially in the younger and more active patient a pantalar fusion will then become necessary.

In summary, conservative treatment of the arthritic ankle by an ankle-foot orthosis or corrective shoe wear can give acceptable results. If unsuccessful, for patients with moderate disease and with symptoms from osseous impingement, an arthroscopic debridement is probably the best treatment. In the event of deformity in the frontal plane a corrective osteotomy should be considered. For the ankle with end-stage arthritis two surgical treatment options are available: ankle arthrodesis and total ankle arthroplasty. As both treatments have a somewhat similar outcome, patient characteristics and their preferences should play an important role in decision making.

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Chapter 4

Total Ankle Arthroplasty in Inflammatory Joint Disease with use of Two Mobile-Bearing Designs

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Abstract

Background Interest in mobile-bearing total ankle arthroplasty (TAA) has increased in recent years. However, to our knowledge, no study has focused exclusively on patients with the diagnosis of inflammatory joint disease (IJD) or has provided a detailed analysis of the risk factors for failure.
Methods A prospective observational study of the results of cementless mobile-bearing TAA in patients with IJD (mainly rheumatoid arthritis) was conducted at two centers. Ninety-three total ankle arthroplasties were performed. The LCS (low contact stress) prosthesis was used initially, in nineteen ankles, between 1988 and 1992, and a modification of the LCS prosthesis, the Buechel-Pappas design, was used in seventy-four ankles between 1993 and 1999. Clinical and radiographic follow-up was performed at yearly intervals. Three clinical scoring systems were used and any complication was recorded throughout follow-up. Actuarial survival (with revision as the end point), multivariate analysis, and a competing risk approach were used to describe the long-term outcome.
Results The clinical result at one year after surgery showed a significant improvement in the scores in all three scoring systems ($p < 0.05$). Ankle dorsiflexion (mean, 7º) also improved significantly ($p < 0.05$). The most frequent complication was a malleolar fracture, which occurred in twenty ankles. Only when it occurred in combination with a deformity in the frontal plane did this complication have an adverse effect on the end result. At a mean follow-up of eight years, seventeen patients (twenty-one ankles) had died and fifteen ankles had been revised because of aseptic loosening (six ankles), primary or secondary axial deformity with edge loading (six ankles), deep infection (two ankles), and a severe wound healing problem (one ankle), leaving fifty-seven ankles (61%) that were evaluated. The mean overall survival rate at eight years was 84%. An increased failure rate was encountered in ankles with a preoperative deformity in the frontal plane of >10º ($p = 0.03$) and in ankles in which an undersized tibial component had been implanted ($p = 0.02$).
Conclusions Mobile-bearing TAA is a valid treatment option for the rheumatoid ankle if proper indications are used. Aseptic loosening and persistent deformity are the most important modes of failure.
Level of Evidence: Therapeutic Level IV. See Instructions to Authors for a complete description of levels of evidence.

4.1 Introduction

Arthrodesis is considered to be the standard treatment for the severely arthritic ankle joint, although this procedure is known to have certain disadvantages. Short-term potential problems are the risk of nonunion, which has been reported to occur in 4% to 36% of such ankles[1,2,3]; malunion; and infection. Furthermore, there is always a disturbed gait pattern following ankle arthrodesis[4,5]. In the long-term, there is both a significant ($p < 0.0001$) risk of deterioration of the joints of the ipsilateral foot and overall functional impairment as described by Takakura et al.[6] in 1999, by Coester et al.[7] in 2001 and by Fuchs et al.[8] in 2003 in the treatment of both posttraumatic and primary osteoarthritis.

4

Total ankle arthroplasty (TAA) has certain theoretical advantages over ankle arthrodesis, both in monoarticular and more specifically in polyarticular disease: gait is less compromised and adverse effects on the other joints of the lower extremity are not expected to occur[9]. High failure rates, however, have been reported with several two-component fully constrained TAA designs[10,11,12,13,14], and TAA generally has not been considered an acceptable alter¬na¬tive to arthrodesis. The intrinsic constraints of such implants create high shear and tensile forces at the bone-prosthesis interface, leading to mechanical failure of these designs. In a study with the semiconstrained two-component Agility ankle prosthesis, Pyevich et al., in 1998, reported better results[15].

Unconstrained TAA with use of a three-component mobile-bearing prosthesis has distinct mechanical and kinematical advantages over two-component designs, as there is full congruency of the articulating surfaces without restriction of rotational motion by the prosthesis. The New Jersey Low Contact Stress (LCS) total ankle prosthesis was the first design applying this mobile-bearing principle to the ankle joint[16]. In 1989, the LCS prosthesis was modified to the current Buechel-Pappas prosthesis[17]. Both designs were developed for application without cement. Buechel et al., the designers of this prosthesis, published good long-term results in 2003[18]. Kofoed and Lundberg-Jensen[19] and Wood and Deakin[20] have published good medium-term results with the mobile-bearing version of the Scandinavian total ankle replacement (STAR). Anderson et al.[21] reported less favorable medium-term results with this design but still recommended the procedure, especially in patients with rheumatoid arthritis. None of these studies compared diagnostic groups and no studies were carried out exclusively for the diagnosis of IJD. In recent years other designs applying the same principle of a mobile-bearing prosthesis have been introduced, but they lack long-term follow-up.
The aims of this study were to report the long-term results of mobile-bearing TAA for the treatment of IJD and to evaluate the factors influencing outcome.

4.2 Materials and Methods

4.2.1 Patient Demographics

A prospective observational study was carried out at two centers on all patients receiving a mobile-bearing ankle prosthesis design for the treatment of IJD. Two designs were used because of design changes over time by the manufacturer. From September 1988 to November 1992, nineteen total ankle arthroplasties were performed with the LCS mobile-bearing prosthesis (DePuy, Warsaw, Indiana) in one center, and, from March 1993 to December 1999, seventy-four total ankle arthroplasties were done with the Buechel-Pappas mobile-bearing prosthesis (Endotec, South Orange, New Jersey) in the two centers participating in this study. The institutional review boards of both centers approved the prospective character of this study. The first two arthroplasties were performed by the surgeon who designed the LCS total ankle prosthesis, seventy-nine arthroplasties were then performed by three experienced ankle-foot surgeons and the remaining twelve were performed by two orthopaedic surgeons experienced in the field of arthritis surgery. However, none of these five surgeons had any experience with TAA before the start of this study. Therefore, a learning curve effect has to be taken into account. Within a few years in both centers TAA became the procedure of choice for the severely affected ankle joint with inflammatory disease. Ankle arthrodesis was carried out mainly when there were contra-indications for prosthetic replacement, such as spontaneous ankylosis of the ankle joint, severe deformity, substantial bone loss at the ankle-hindfoot level, neurological or vascular disease, or infection either locally or at a distance and in heavy smokers. All patients in this study had IJD, and the majority had rheumatoid arthritis. During the study period, TAA was carried out for osteoarthritis in only six patients. Furthermore, in six more patients other implants had been used. Therefore, it was thought that performing a comparative study between diagnostic groups or the different implants would not be useful.

The demographic data on the patients at the time of surgery are summarized in Table 1. The radiographic degree of ankle joint destruction was assessed with us of the method described by Larsen et al.[22]. Fourteen ankles were classified as stage 3 (joint space narrowing); seventy-two, as stage 4 (complete loss of joint space); and seven, as stage 5 (periarticular bone loss). At the time of surgery, twenty-eight ankles (30%) had an ankylosed subtalar joint, which had developed spontaneously in seventeen ankles (18%) or was the result of prior surgical intervention in eleven ankles (12%). Twenty-four patients (32%) were taking corticosteroids. On the basis of radiographic findings and the gross appearance at the time of surgery, tibial bone quality demonstrated mild osteopenia in forty-nine ankles and severe osteopenia in eleven ankles. The remaining thirty-three ankles were considered to have a normal bone quality.

TABLE 1 Demographic Data on the Patients			
Diagnosis	No. of Ankles (No. of Patients)	Age* (yr)	Sex (F/M)
Rheumatoid arthritis†	70 (58)	59.4 (28.8-81.0)	59/11
RF positive			
RF negative	14 (12)	55.3 (33.5-78.8)	10/4
Juvenile idiopathic arthritis	4 (3)	38.4 (26.7-65.8)	4/0
Miscellaneous‡	5 (3)	55.7 (43.7-66.9)	3/2
All diagnoses	93 (76)	57.6 (26.7-81.0)	76/17

*The values are given as the average, with the range in parentheses. †RF= rheumatoid factor. ‡Two patients had ankylosing spondylitis, two had adult-onset Still disease, and one had psoriatic arthritis.

4.2.2 Implant Characteristics and Surgical Technique

The LCS prosthesis consists of a congruent polyethylene mobile bearing between a flat tibial component and a talar component with a shallow sulcus, both made of a cobalt-chromium alloy. For osseointegration, a porous coating is applied to the metal components, which are available in 2 sizes: standard and large. The design changes that were incorporated into the Buechel-Pappas prosthesis in 1989 were a deepening of the sulcus of the talar component in order to reduce the risk of bearing subluxation, the use of titanium for the metal components in order to improve osseointegration, the addition of two more sizes, and the creation of a somewhat thicker tibial stem.

Surgery is carried out through a straight anterior midline approach with the thigh in a leg holder in order to flex the knee to about 60o and to rotate the leg to a neutral position. After the incision of the skin and the extensor retinaculum, the interval between the anterior tibial and the extensor hallucis longus tendons is used to reach the anterior capsule, which is then opened medially and reflected laterally as a sleeve together with the dorsalis pedis artery and veins. In general, the tourniquet was inflated after the arthrotomy and before the preparation of the osseous surfaces is begun.

The preparation of the osseous surfaces is started with a flat resection of the lower surface of the distal aspect of the tibia, which is carried out with the aid of a resection guide. The aim is for a horizontal resection of the distal aspect of the tibia in the frontal plane and an anterior slope of 7° in the sagittal plane. For the stem of the tibial component, a window is created in the anterior part of the distal end of the tibia with use of a special box chisel. The talar dome is prepared by first making a groove for the central sulcus and then preparing the slots for the fixation fins. After the preparation of the talus is finished, an uncemented talar component and then an uncemented tibial component are implanted. The thickest possible polyethylene liner is then introduced in between these components with use of manual distraction of the joint. After bone graft (from locally resected bone) is placed around the tibial stem the removed tibial cortical window is then replaced and impacted. Routine wound closure with careful suturing of the extensor retinaculum completes the procedure. Lengthening of the Achilles tendon is not carried out. Postoperatively, the ankle is immobilized in a below-knee walking cast for six weeks, with weightbearing to tolerance allowed. The objective of the aftertreatment in a cast is mainly to protect the sutured extensor retinaculum and to promote soft-tissue healing.

4.2.3 Clinical Evaluation
All patients were evaluated preoperatively and were followed prospectively after the operation at six weeks, three months, one1 year and at one to two -year intervals thereafter. Clinical evaluation included the recording of any complications and assessment of three ankle scoring systems that are most frequently used in the literature: the LCS ankle score[16], the AOFAS (American Orthopaedic Foot and Ankle Society) ankle and hindfoot score[23], and the Kofoed ankle score[24]. All three scoring systems use somewhat similar items for pain, function, range of motion and deformity, and all have a maximum of 100 points. Although at the start of this study both the AOFAS and the Kofoed ankle score were not yet published, these scores could be calculated retrospectively as the items that differed from the LCS ankle score (e.g.: the use of orthopaedic footwear and walking on uneven surfaces) had been fully recorded since the beginning of the study. The range of motion of the ankle-hindfoot complex was measured manually with use of a goniometer; dorsiflexion, while the patient was standing; and plantar flexion as well as pronation and supination, while the patient was sitting.

4.2.4 Radiographic Evaluation
Radiographic evaluation was done with use of standardized AP and lateral radiographs of the ankle. Weight-bearing radiographs were made preoperatively, and non-weight-bearing radiographs were made at follow-up. The preoperative alignment of the ankle joint in the frontal plane was defined as the angle between the long axis of the tibia and the line perpendicular to the talar surface on the weight-bearing anteroposterior radiograph of the ankle (Fig. 1). Postoperatively, both radiographs were aimed so that they were parallel to the base plate of the tibial component, if necessary, fluoroscopy was used to obtain optimal radiographs. The serial radiographs were evaluated by the two clinical authors (H.C.D. and R.G.H.H.N.) and by an orthopaedic surgeon who was not involved in the care of these patients. The tibial component was considered to be undersized if there was > 2 mm of uncovered bone of the tibial plafond on the anteroposterior radiograph made shortly after surgery. The angular position of the tibial component was defined as the angle between the base plate of the tibial component

and the long axis of the tibia on both radiographs. The angular position of the talar component on the lateral radiograph was defined as the angle created by a line parallel to the fins of the talar component and a line drawn from the most dorsal part to the center of the anterior part of the talus (Figs. 2-A and 2-B). This talar reference line was chosen as it could be drawn reliably even with a fused subtalar joint.

Fig. 1 The preoperative alignment of the ankle in the frontal plane was defined as the angle ***'a'*** formed between the long axis of the tibia and a line drawn perpendicular to the talar dome on the preoperative weightbearing anteroposterior radiograph.

Fig. 1

Fig. 2-A

Fig. 2-B

Figs. 2-A and 2-B Postoperative alignment of the components. ***Fig. 2-A*** Anteroposterior radiograph showing the frontal angular position of the tibial component in relation to the long axis of the tibia (angle ***'b'***). ***Fig. 2-B*** Lateral radiograph showing the sagittal angular position of the tibial component in relation to the long axis of the tibia (angle ***'c'***) and the sagittal angular position of the talar component (angle ***'d'***), which was created by a line drawn parallel to the fins of the talar component and a line drawn from the most dorsal part to the center of the anterior part of the talus on the postoperative lateral radiograph.

The occurrence of radiolucent lines next to the prosthetic components was measured on the anteroposterior and lateral radiographs for the tibial component and on the lateral radiograph for the talar component. Our criterion to determine loosening of a component was an angular change in position of >3º, a subsidence of >3 mm in one of the radiographs, or a complete radiolucent line of >1 mm in both radiographs for the tibial component and on the lateral radiograph for the talar component. The polyethylene bearing (which had two small metallic markers) was assessed for subluxation in the sagittal and frontal planes and other abnormalities (gross wear or fracture).

4.2.5 Statistical Methods

All data were recorded in an especially developed ankle module of the Project Manager data management program (IMSOR, Leiden, The Netherlands). Statistical evaluation was done with SPSS software (version 11.5; SPSS, Chicago, Illinois). For the competing risk estimation NCSS software (version 2001; NCSS, Kaysville, Utah) was used.

Survival analysis techniques were used to estimate the probability of the occurrence of certain outcome events as a function of the time elapsed since operation. For noncomposite end points, curves were estimated with use of the Kaplan-Meier approach. When subgroups (the LCS and the Buechel-Pappas prosthesis) were compared, a log-rank test was used and the separate survival curves were estimated. Revision surgery with removal of the prosthesis was used as a noncomposite end point. Furthermore, revision of the ankle prosthesis for aseptic loosening, revision for other indications, and aseptic loosening without revision were used as the composite end points.

In the multivariate approach, we used the Cox proportional hazards model to analyze whether gender, age at the time of surgery, the year of surgery (as a factor indicating the learning curve), the type of prosthesis, ankylosis of the subtalar joint at the time of surgery, a varus position of the tibial component, or undersizing of the tibial component influenced the survival rate. The Cox proportional hazards model estimates the “probability of survival” of the prosthesis during follow-up time as a function of several risk factors[25].

When the event-of-interest was actually a composite outcome (for example “failure” as a noncomposite end point versus “failure due to infection, due to malalignment, due to …” as a composite outcome), we used a competing risk approach in order to obtain valid estimates for the cumulative incidences of the various subcategories of the overall outcome measure. By applying the competing risk estimation rather than the Kaplan-Meier estimation, the sum of the cumulative incidences of the various components added up to the usual Kaplan-Meier estimate on the noncomposite outcome.

One should note that the estimated hazard ratios within the Cox model framework are still valid estimates for the comparison of the LCS and the Buechel-Pappas prostheses even in the presence of competing risks. Only the survival curves themselves, are, in that case, not estimated correctly by the Kaplan-Meier method and should be calculated using the competing risk approach.

The results were considered significant if $p < 0.05$.

4.3 Results

Follow-up was completed as of January 2005, and no patients were lost to follow-up. Fifteen ankles were revised with either an arthrodesis (thirteen ankles) or an implant exchange (two ankles), which are described in detail below. Seventeen patients (twenty-one ankles) had died at an average interval of sixty-three months (range, five to 171 months) after the surgery. There was no apparent relationship between the cause of death and the TAA. The median duration of follow-up for the sixty-two patients (seventy-eight ankles) who had not reached the end point under study (that is, the patients who had died and those who had not had a revision) was 7.2 years (range, 0.4-16.3 years) and the average duration of follow-up was 7.6 years. These measures, therefore, reflect the distribution of follow-up times of the study itself.

4.3.1 Clinical and Radiographic Outcome

The clinical outcome at one year after surgery for eighty-seven ankles is listed in Table 2. The one-year result was not available for six ankles: two ankles were in two patients who had died, two ankles had been revised for early deep infection, one ankle had been revised for severe wound dehiscence and one ankle was in a patient who remained severely disabled following a cerebrovascular accident nine months after surgery.

For the fifty-seven total ankle arthroplasties seen at follow-up in 2005, the mean survival of the total ankle replacement was eight years (range, five to 16.3 years), the mean LCS ankle score was 83.3 (95% confidence interval 80.0 to 86.6), the mean AOFAS score was 77.0 (95% confidence interval 73.2 to 80.8) and the mean Kofoed score was 75.7 (95% confidence interval 72.1 to 97.3).

At the time of the final follow-up, nine of the fifty-seven ankles showed no radiolucency around the tibial component; thirty-nine ankles had partial radiolucent lines, especially around the tibial stem; six had a complete radiolucent line of ≤1 mm; and three tibial components showed subsidence. Complete osseointegration of the talar component was seen in fifty ankles, and partial radiolucent lines were seen around the component in seven, with subsidence in four of them. The seven ankles with subsidence of the implants had no important symptoms.

TABLE 2 Clinical Outcome for All Ankles at One Year					
	Ankle Scoring Systems[16,23,24]			Range of Motion	
	LCS*	AOFAS†	Kofoed‡	Dorsiflexion (deg) §	Plantarflexion (deg)
Preop.#	36.1	26.5	26.9	3.9	22.4
(n=93)	(33.6-38.7)	(24.1-28.8)	(24.1-29.7)	(2.9-5.0)	(20.2-24.6)
One year#	81.5	77.7	74.0	7.1	24.8
(n=87)	(78.4-83.9)	(74.7-80.6)	(70.9-77.2)	(5.8-8.4)	(22.6-27.2)
Gain	45.4**	51.2**	47.1**	3.2**	2.4

*At one year, fifteen ankles had a score of <70, forty had a score of 70 to 85, and thirty-two had a score of >85. LCS = low contact stress. †At one year, seventeen ankles had a score of <70, forty-four had a score of 70 to 85, and twenty-six had a score of >85. AOFAS = American Orthopaedic Foot and Ankle Society. ‡Thirty-two ankles had a score of <70, thirty-three had a score of 70 to 85, and twenty-two had a score of >85. §At one year, eighteen ankles had <5° of dorsiflexion, forty-eight had 5° to 9°, and twenty-one had ≥10°. #The values are given as the mean, with the 95% confidence interval in parentheses.
**Student paired t test; $p < 0.05$.

Fracture or substantial wear of the polyethylene bearing did not occur in this series, despite the fact that the thinnest bearing (3 mm) had been implanted in forty-eight ankles (52%).

Seventeen of the twenty-one ankles in the patients who had died had functioned well both clinically and radiographically. A sixty-one-year-old man who had a Buechel-Pappas ankle arthroplasty bilaterally for the treatment of severe ankle joint destruction due to longstanding rheumatoid arthritis developed early aseptic loosening of both tibial components. As the ankles were not symptomatic, no revision surgery was carried out. He died fifty months and thirty-eight months, respectively, after the two operations. A seventy-year-old woman who had a Buechel-Pappas ankle arthroplasty bilaterally for the treatment of severe ankle joint destruction due to longstanding rheumatoid arthritis also had early aseptic loosening of both tibial and both talar components. The ankles remained asymptomatic, and no revision surgery was carried out. She died 119 and seventy-one months, respectively, after the two operations.

4.3.2 Early Complications

At the time of surgery, osseous complications were seen in twenty-seven ankles: fifteen had a fracture of the medial malleolus, seven, an anterolateral distal tibial fracture (four were incomplete); and five, a fracture of the lateral malleolus (Table 3). Screw fixation was used to treat one ankle with an anterolateral distal tibial fracture, six ankles with a fracture of the medial malleolus, and two ankles with a fracture of the lateral malleolus. The other fractures were treated with routine immobilization of the ankle in a plaster cast and without osteosynthesis of the fracture. In only four ankles, these complications led to a change in postoperative management, which involved a non-weight-bearing period of four to six weeks. In two ankles with a pre-existing valgus deformity, the medial malleolar fracture went on to nonunion. Both ankles also had development of a stress fracture of the lateral malleolus, eventually leading to valgus instability and edge-loading, which required an arthrodesis (Figs. 3-A through 3-D). None of the other twenty-five fractures influenced the end result (Figs. 4-A through 4-D). Intraoperative fractures occurred both early and late in this series, but there were no injuries to nerves or vessels.

TABLE 3 Intraoperative and Early Postoperative Complications

	Medial Malleolar Fracture	Lateral Malleolar Fracture	Anterolateral Distal Tibial Fracture	Postoperative Distal Tibial Fracture	Wound Healing Disturbance	Infection
No. of ankles	15	5	7	4	8	3
No. of failures*	2	-	-	-	1	1

* See text and Table 4 for details

Fig. 3-A

Fig. 3-B

Figs. 3-A through 3-D Radiographs of a sixty-seven-year-old woman (Case 6, Table IV) who had rheumatoid arthritis for 36 years and a stiff hindfoot with a valgus deformity. ***Fig. 3-A*** The preoperative anteroposterior radiograph showing almost complete loss of joint space. ***Fig. 3-B*** Two months after implantation of a LCS prosthesis. The patient sustained a fracture of the medial malleolus, which was treated with osteosynthesis.

Fig. 3-C

Fig. 3-D

Fig. 3-C Despite repeat fixation with tension-band wiring and Kirscher wires, a nonunion of the medial malleolus developed, resulting in lateral edge loading and subluxation of the bearing. The radiograph shows the ankle after some of the hardware had been removed. ***Fig. 3-D*** Twenty-six months after the index operation, a successful arthrodesis was carried out.

Fig. 4-A

Fig. 4-B

Fig. 4-C

Fig. 4-D

Figs. 4-A through 4-D Radiographs of a forty-two-year-old woman who had rheumatoid arthritis for seventeen years and had undergone a prior hindfoot arthrodesis. The early post-operative anteroposterior (Fig. 4-A) and lateral (Fig. 4-B) radiographs showing a well-aligned Buechel-Pappas prosthesis and fixation of an intraoperative medial malleolar fracture with a screw and Kirschner wire. ***Figs. 4-C and 4-D*** Radiographs at ten years show excellent osseointegration of the tibial and talar components. The hardware in the medial malleolus was removed one year after implantation for the treatment of local symptoms. The ankle had 20º of motion, and the patient was satisfied with the result. Her walking distance was a few hundred meters and was limited by a symptomatic arthritis of the contralateral hip.

Postoperative complications included delayed wound healing in seven ankles, with additional surgery required in only two of them (one had a local flap and one had excision of a necrotic anterior tibial tendon) and one major wound dehiscence with an open ankle joint that required conversion to an arthrodesis after two months (Table 3). Three ankles had an early deep infection. Open lavage combined with culture-specific systemic intravenous antibiotics for two weeks, followed by oral antibiotics for four and ten weeks, respectively, resulted in resolution of the two infections. The third ankle had to be converted to an arthrodesis after four months.

Four ankles had an early spontaneous postoperative fracture of the distal aspect of the tibia, which occurred four to six months after surgery at the level of the tip of the stem of the tibial component. All four patients had severe osteopenia assessed intraoperatively. These fractures healed after treatment with cast immobilization, and they did not influence the end result. The distal tibial fractures (both intraoperative and early postoperative) were seen only with the Buechel-Pappas prosthesis.

4.3.3 Subsequent Surgery without Implant Exchange

Besides for the above mentioned complications, subsequent surgery included an arthroscopic debridement for talar-malleolar arthritis in three ankles, and valgus osteotomy of the calcaneus in one ankle in a patient who had a residual symptomatic varus deformity of the hindfoot. The results of these procedures were good in all four ankles.

4.3.4 Late Complications and Revisions

Two ankles had a secondary deep infection. In one of them, the infection developed after an arthroscopic debridement for talar-malleolar arthritis (described above). It was treated effectively with open lavage and culture-specific intravenous antibiotics for 4 weeks, followed by ten weeks of oral antibiotics. At eleven years, the ankle showed no signs of active infection. The second late infection developed after a subsequent hindfoot fusion that became infected. After a two-stage revision was performed, the infection resolved. The ankle eventually required conversion to an arthrodesis as a result of aseptic loosening of the cemented tibial component.

Six ankles were revised because of aseptic loosening. One ankle in a woman with juvenile idiopathic arthritis who was managed with an LCS prosthesis had subsidence of the talar component after 10 years. The ankle was successfully treated with an exchange of both the bearing and the talar component with use of Buechel-Pappas implants. Another five ankles (two with LCS components and three with Buechel-Pappas prostheses) had aseptic loosening of the tibial components. Four of them also had loosening of the talar component. They were converted to an arthrodesis 2.5 to thirteen years after implantation.

Another eight ankle replacements were converted to an arthrodesis because of an early deep infection (one ankle), major delay of wound healing with an open ankle joint (one ankle, which was described in the section on early complications), persistent varus deformity with edge loading (two ankles treated early in this series), valgus deformity with edge-loading due to nonunion of an intraoperative medial malleolar fracture (two ankles described in the section on early complications), secondary edge loading due to a late lateral malleolar fracture (one ankle), and secondary valgus instability with edge loading (one ankle).

In summary, a total of fifteen ankles (fourteen patients) were revised with an implant exchange (two ankles) or conversion to an arthrodesis (thirteen ankles). These data are listed in Table 4.

TABLE 4 Revisions

Case	Gender, Age (yr)	Year of Surgery	Diag-nosis ¶	Align-ment*	Type of Prosthesis	Months to Revision	Reason for Revision	Revision Surgery	Result
1	F, 27	1988	JIA	0	LCS	118	Loose talar compo-nent	Exchange talar component + bearing	Good
2	F, 27	1989	JIA	5	LCS	161	Loose tibial compo-nent	Arthrodesis with screws	Nonunion
3	F, 49	1989	RA	0	LCS	83	Secondary valgus + edge loading	Arthrodesis with blade plate	Union,, died 61 months later
4	M, 63	1990	RA	-20	LCS	91	Primary varus + edge loading	Arthrodesis with blade plate	Union
5	F, 59	1992	RA	0	LCS	115	Loose tibial + talar component	Arthrodesis with K-wires	Nonunion
6‡	F, 67	1992	RA	12	LCS	26	Valgus instability + nonunion medial malleolus	Arthrodesis with screws	Union
7	M, 51	1992	RA	-15	LCS	52	Primary varus + edge loading	Arthrodesis with blade plate	Union
8	F, 61	1995	RA	0	BP	44	Loose tibial compo-nent	Arthrodesis with retrograde nail	Nonunion
9‡	F, 44	1995	RA	10	BP	43	Valgus edge loading due to late lateral malleolar fracture	Arthrodesis with retrograde nail	Nonunion, Rearthro-desis
10	F, 72	1995	RA	10	BP	21	Valgus edge loading due to medial mal-leolar nonunion + late lateral malleolar fracture	Arthrodesis with screws	Union
11	M, 51	1995	RA	0	BP	11	Infection after secondary hindfoot arthrodesis	Two-stage revi-sion	Aseptic loosening of tibial compo-nent
12‡	F, 42	1996	RA	0	BP	5	Deep infection	Arthrodesis with screws	Union
13	M. 53	1996	RA	0	BP	39	Loose tibial + talar component	Arthrodesis with sliding tibial graft and 1 screw.	Nonunion
14	F, 45	1998	RA	4	BP	32	Loose tibial + talar component	Arthrodesis with retrograde nail	Union
15‡	F, 76	1999	RA	0	BP	2	Severe wound heal-ing disturbance	Arthrodesis with retrograde nail	Nonunion, Rearthro-desis

¶ JIA = Juvenile Idiopathic Arthritis, RA = Rheumatoid Arthritis. * Valgus = positive, varus = negative. ‡ Revision after intraoperative or early postoperative complication

4

4.3.5 Survival Analysis

Kaplan-Meier survival analysis and the cumulative incidence of survival in a competing risk scenario were used for the analysis of the risks for failure.

Separate endpoints for failure have been defined as (1) a revision (or conversion to an arthrodesis) for any reason, (2) a revision for aseptic loosening, and (3) radiographic evidence loosening without a revision.

The eight-year overall survival of both prostheses with revision or conversion to an arthrodesis for any reason as the endpoint was 83.6% (95% confidence interval, 73% to 93%).

Log-rank analysis revealed no significant difference, with the numbers available, in the mean overall survival rate between the LCS prosthesis and the Buechel-Pappas prosthesis during the observation period: p = 0.33 (Figs. 5-A and 5-B). However, a significant difference was demonstrated in the overall survival between ankles with or without a preoperative deformity in the frontal plane. Seventeen ankles had a preoperative varus or valgus deformity of >10º, and the overall survival rate of these ankles at eight years was 48% (95% confidence interval, 6% to 90%). Ankles with a neutral alignment preoperatively (<10º of varus or valgus) had an eight-year mean overall survival of 90% (95% confidence interval, 82% to 98%). Log-rank analysis demonstrated that the difference was significant (p = 0.03) (Figs. 6-A and 6-B).

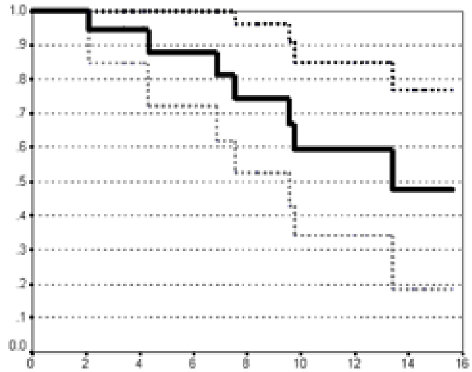

Fig. 5-A

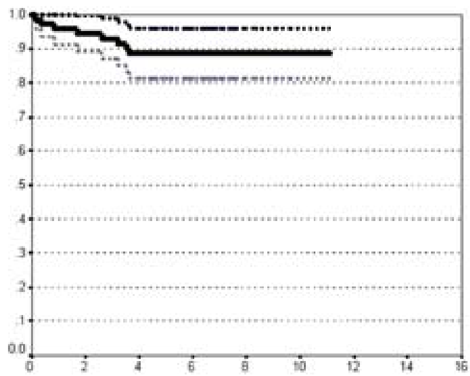

Fig. 5-B

Figs. 5-A and 5-B The Kaplan-Meier survival curves (solid black lines) with the 95% confidence intervals (dotted lines) of the LCS (Fig. 5-A) and the Buechel-Pappas (Fig. 5-B) total ankle replacements. With the numbers available, log-rank analysis showed no significant difference during the period of observation (p = 0.33).

The mean eight-year overall survival rate was 91% (95% confidence interval, 81% to 100%) for the sixty-six ankles that had a tibial component of correct size and 66% (95% confidence interval, 45% to 88%) for the twenty-seven ankles with an undersized tibial component. This difference was significant (p = 0.02).

The eight-year cumulative incidence of failure was calculated for the composite end points. At eight years, twenty-four ankles were in follow-up. The incidence of failure was 13% (95% confidence interval, 7% to 25%) with revision for other reasons (infection, edge loading, etc) as the end point, 3% (95% confidence interval, 1% to 10%) with revision because of aseptic loosening as the end point, and 16% (95% confidence interval, 9% to 30%) for aseptic loosening without revision as the end point (Fig. 7).

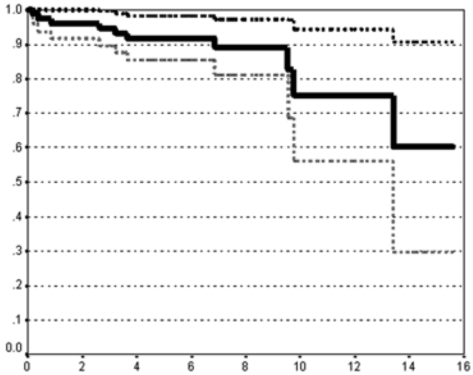

Fig. 6-A

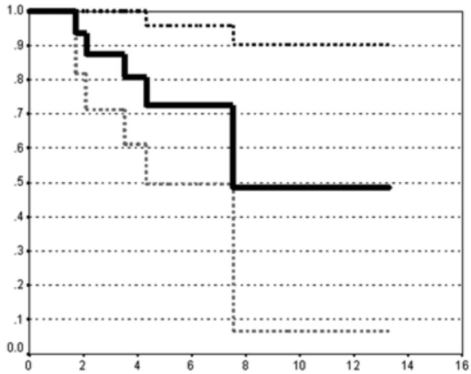

Fig. 6-B

Figs. 6-A and 6-B The Kaplan-Meier survival curves (solid black lines) with the 95% confidence intervals (dotted lines), for the ankle replacements in the neutrally aligned ankles (Fig. 6-A) and in the ankles with a preoperative deformity in the frontal plane of >10º (Fig. 6-B). Log-rank analysis showed a significant difference in favor of the ankles with a neutral alignment (p = 0.03).

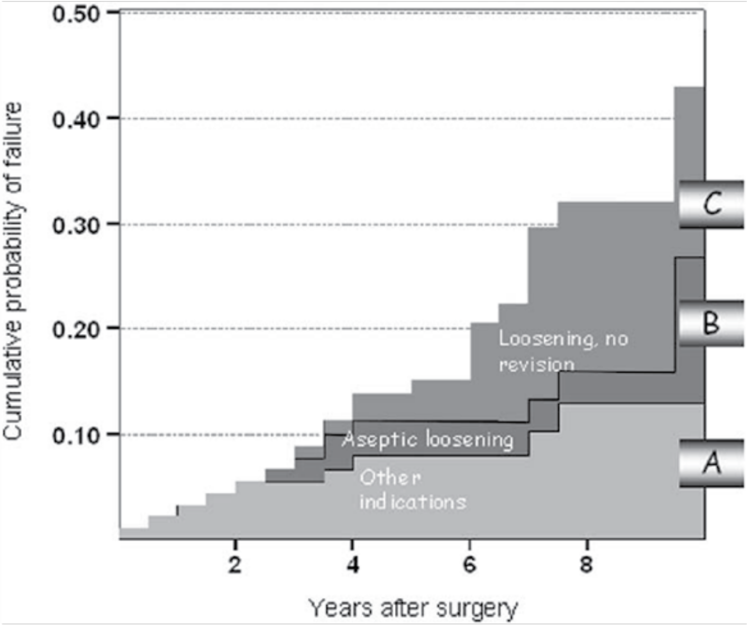

Fig.7 Cumulative incidence analysis of the risk of failure with revision for other indications (A), revision for aseptic loosening (B), and aseptic loosening, no revision (C) as the end points**.**

4.3.6 Cox Regression Analysis

With revision or conversion to an arthrodesis for any reason as the end point, a multivariate analysis was carried out to assess which factors influenced the survival of the replacement. The following items were evaluated: type of prosthesis (LCS or Buechel-Pappas), year of surgery, sex, age at the time of surgery (forty-eight patients were less than sixty years old, and forty-five were at least sixty years old), ankylosis of the hindfoot at the time of surgery, and the position of the tibial component in the frontal plane (sixty-six ankles had ≤4º of varus position of the implant and twenty-seven had ≥5º of varus position of the implant).

With the numbers available, no factor was identified as having a significant effect on survival. However, ankylosis of the hindfoot showed a trend toward significance ($p = 0.07$). As there was a high correlation between a varus position and an undersized tibial component, these factors could not be assessed together in one statistical model.

4.4 Discussion

There is considerable controversy with regard to TAA as a valid option in the treatment of inflammatory arthritis of the ankle joint. All studies that we know of on the various constrained two-component designs have shown unsatisfactory results[11,12,13,14], and these implants have been abandoned by most authors. Acceptable results have been reported only for the semiconstrained Agility TAA in two studies in mixed populations of posttraumatic arthritis, primary arthritis, and rheumatoid arthritis[15,26]. Both studies, however, used less strict radiographic criteria than we used for loosening without revision. With the same prosthesis, Spirt et al.[27] reported a relatively high rate of early failure, especially with respect to a high level of reoperations that occurred after the TAA. The inferior results in this study may have been the result of the younger age of the patients.

With a mean eight-year overall survival of 84% in patients with IJD, our study shows that mobile-bearing TAA is a valid treatment option. Above all, our study reflects the learning curve for this procedure, not only with regard to the surgical technique but also with regard to its application to the proper indications. Results were better if only well-aligned ankles (which had a mean survival rate of 90% at eight years) or tibial components of correct size (which had a mean survival of 91% at eight years) were considered. Younger age could not be identified as a risk factor for failure. A positive phenomenon was the absence of substantial wear or fracture of the polyethylene bearing in this series. The most important modes of failure were edge-loading in ankles with a preexisting deformity in the frontal plane, mainly occurring early in this series, and aseptic loosening of either the tibial or the talar component occurring with longer-term follow-up.

In 2003, Buechel et al.[18] in a prospective study reported a mean cumulative survival rate of 93.5% at eight years in a prospective study of the Buechel-Pappas prosthesis. Most of their patients had posttraumatic or primary osteoarthritis. In a prospective series of 200 ankles with both IJD and osteoarthritis that were managed with the mobile-bearing STAR (Scandinavian total ankle replacement) prosthesis, Wood and Deakin reported a survival rate of 87.9% at eight-year[20]. Aseptic loosening and edge loading as a result of persistent deformity were the most important modes of failure. Radiographic evidence of improved fixation was seen with a new version that had a dual-coating of hydroxyapatite on porous titanium. Using the same prosthesis, Anderson et al.[21] reported less favorable results, with a five-year survival rate of 70%. The mode of failure in that study was aseptic loosening in seven ankles and fracture of the polyethylene bearing in two out of the twelve ankles that had a revision. They commented that TAA was a challenging procedure and that the instrumentation used was unsatisfactory. Nevertheless, they still recommended this procedure in well-aligned ankles with sufficient bone stock and noted that the instrumentation in the mean time had been improved after their study had concluded.

In general, the results with mobile-bearing designs are better than of those describing for two-component designs. This is probably due to the unconstrained biomechanical characteristics of the mobile-bearing prosthesis: the vertical rotational forces occurring at the ankle during walking[28] should be better tolerated with an unconstrained implant. In vitro kinematical studies comparing the normal ankle, the fused ankle, the Agility two-component design and two mobile-bearing designs (STAR and HINTEGRA) demonstrated relatively normal kinematics at the ankle after implantation of the mobile-bearing prostheses; these findings were in strong contrast to the situation after ankle arthrodesis and also after arthroplasty with the Agility prosthesis[29,30,31]. The lower rate of early mechanical failure in the mobile-bearing designs compared with the constrained two-component designs could therefore be explained by their better kinematical characteristics.

The survival rate for mobile-bearing total ankle replacement is somewhat lower than that for replacements of the hip[32,33,34] and knee[35,36,37]. However, the long-term results are certainly acceptable, and the complication rate is similar to that for arthrodesis. Applying the threshold of ten-year survival as identified by SooHoo and Kominski[38], TAA with use of a mobile-bearing prosthesis appears to be a cost-effective procedure in the treatment of IJD.

Preoperative deformity in the frontal plane is quite difficult to correct during TAA and, if persistent after surgery, will frequently result in instability and a subluxation of the bearing, eventually leading to failure. This could be avoided by making varus or valgus deformity of >10° an absolute contraindication for TAA.

In our series, there was a relatively high rate of malleolar fractures. This can, at least to some extent, be explained by both the more osteopenic bone in these ankles in patients with rheumatoid arthritis and by the limited experience of the participating surgeons. No long-term adverse effects of malleolar fractures were seen in well-aligned ankles. Furthermore, there was an increased risk of distal tibial fractures both at the time of surgery and in the early postoperative period. This is most likely related to the window that is created in the distal part of the tibia for the stem of the tibial component and to the sometimes severely osteopenic bone in severely disabled patients with rheumatoid arthritis who take corticosteroids. Although no long-term adverse effects of such fractures were seen, they still carry a potential risk for failure. Finally, durable long-term fixation of the TAA might better be achieved by the use of implants with improved coating characteristics[20,39].

In conclusion, good clinical results can be achieved with the LCS or Buechel-Pappas total ankle prosthesis in the treatment of severe IJD if proper indications are applied. A varus or valgus deformity of the ankle and/or hindfoot of >10° should be considered a contraindication to this procedure. If such a deformity is present, we believe a procedure to correct it (e.g. triple arthrodesis) should to be performed prior to the ankle arthroplasty. Mobile-bearing TAA can certainly be considered an alternative to ankle arthrodesis in patients with rheumatoid arthritis. Because of the technical difficulties that can be encountered during surgery, and also the low frequency of this procedure in the typical orthopaedic practice, TAA probably should be restricted to the experienced ankle surgeon.

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Chapter 5

Early Migration of the Tibial Component of the Buechel-Pappas Total Ankle Prosthesis

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Abstract

Interest in mobile-bearing total ankle arthroplasty has increased in recent years. Clinical results show favorable but varying results, with survival rates between 70% and 90% at 10-year follow-up. Design-specific differences in early migration patterns might explain differences in the results and modes of failure. Using radiostereometric analysis (RSA) we prospectively followed 12 RA patients with a cementless mobile-bearing total ankle arthroplasty. The American Orthopaedic Foot and Ankle Society ankle score and radiostereometric radiographs were evaluated immediately postoperatively, 6 weeks postoperatively, 3 months, 6 months, and 12 months postoperatively and yearly thereafter. The postoperative clinical results improved. We observed increased migration of the tibial component during the first 3 months, but this stabilized by the 6-months follow-up. The mean lateral-medial migration was 0.8 mm, distal-proximal migration was 0.9 mm, and posteroanterior migration was -0.5 mm. The latter implies the total resultant migration was in anterior and valgus tilting of this tibial component. We believe the surgical technique (anterior cortical window for placement) and the method of tibial fixation likely explain this migration.

Level of Evidence: Therapeutic study, level IV. See Guidelines for Authors for a complete description of levels of evidence.

5.1 Introduction

The ankle and/or hindfoot is affected in up to 56% of patients with rheumatoid arthritis (RA)[1]. In the treatment of debilitated ankle joints, total ankle arthroplasty (TAA) potentially has certain functional advantages compared with ankle arthrodesis in monoarticular and polyarticular disease. Gait is compromised less and adverse effects on the other joints of the lower extremity are not expected with TAA because mobility of the ankle joint is preserved. Because of unsatisfactory results in the past with constrained two-component designs, which had 36 to 90% failure[2,3], TAA generally has been considered to be controversial in treating the severely affected ankle joint. Lesser-constrained TAAs using a three-component mobile-bearing prosthesis have distinct mechanical and kinematical advantages compared with two-component designs because there is full congruency of the articulating surfaces without restriction of rotational motion by the prosthesis, thereby reducing the rotational stresses at the bone-prosthesis interface. The latter is reflected in the improved survival rates of a mean of 91% at 5 years in a meta-analysis of this type of design[4]. However, outcomes with these prostheses still lack the excellent long-term results seen consistently with many types of total knee and total hip prostheses. In contrast, the mobile-bearing ankle prosthesis results show some variability (70–95% survival rates)[4,5,6,7,8,9,10,11,12,13].

Clinically symptomatic mechanical loosening is on of the most important modes of failure. Because initial progressive migration relates to long-term prosthesis survival with some implants[14,15], the mode of early migration of a prosthesis may predict whether the initial fixation is a factor of concern in a given type of prosthesis. Therefore we began a radiostereometric analysis (RSA) of early migration in mobile-bearing TAA. We focused on the tibial component since in a recently published study from our institution[8] mechanical failure occurred more often with the tibial component than with the talar component.

We sought to ascertain the amount and direction of initial migration of the tibial component after mobile-bearing TAA, and (2) if secondary stabilization occurred within the 1 year interval after surgery.

5.2 Materials and Methods

We prospectively followed fifteen patients (fifteen ankles) having Buechel-Pappas (BP) mobile-bearing TAAs (Endotec, South Orange, NJ) performed from October 2001 to November 2003. This device is our preferred TAA for severely destroyed ankle joints in RA (Larsen Stage 3 and higher[16]). Three ankles were classified as Stage 3 (joint space narrowing), nine were classified as Stage 4 (complete loss of joint space), and three were classified as Stage 5 (periarticular bone loss). All patients in this study had RA. There were 12 women and three men with a mean age of 61 years (SD 8.6 years). All ankle prostheses were implanted by an experienced TAA surgeon (RN). No patients were lost to follow-up. The mean follow-up period was 2 years (SD 0.4).

The BP[7,13] prosthesis consists of a congruent polyethylene (PE) mobile-bearing matching the curved talar component and the flat tibial and a deep-sulcus talar component, both made of titanium and available in four sizes. The PE insert is available in 3 mm, 5 mm, 7, 9 and 11 mm thicknesses. The tibial component has a central stem, the talar component has two rectangular fins, and the “bone surfaces” of these components are beaded to improve ingrowth (Fig. 1). Initially, one RSA marker was attached with a small dot of bone cement to the tip of the tibial component (7 prostheses). After this initial series the manufacturer provided RSA components with one marker attached to a metal tower at the tip of the central peg of the tibial component in 8 prostheses (Fig. 2).

Fig. 1 A mobile-bearing BP ankle prosthesis (rough beaded bone surface, polished articular surface).

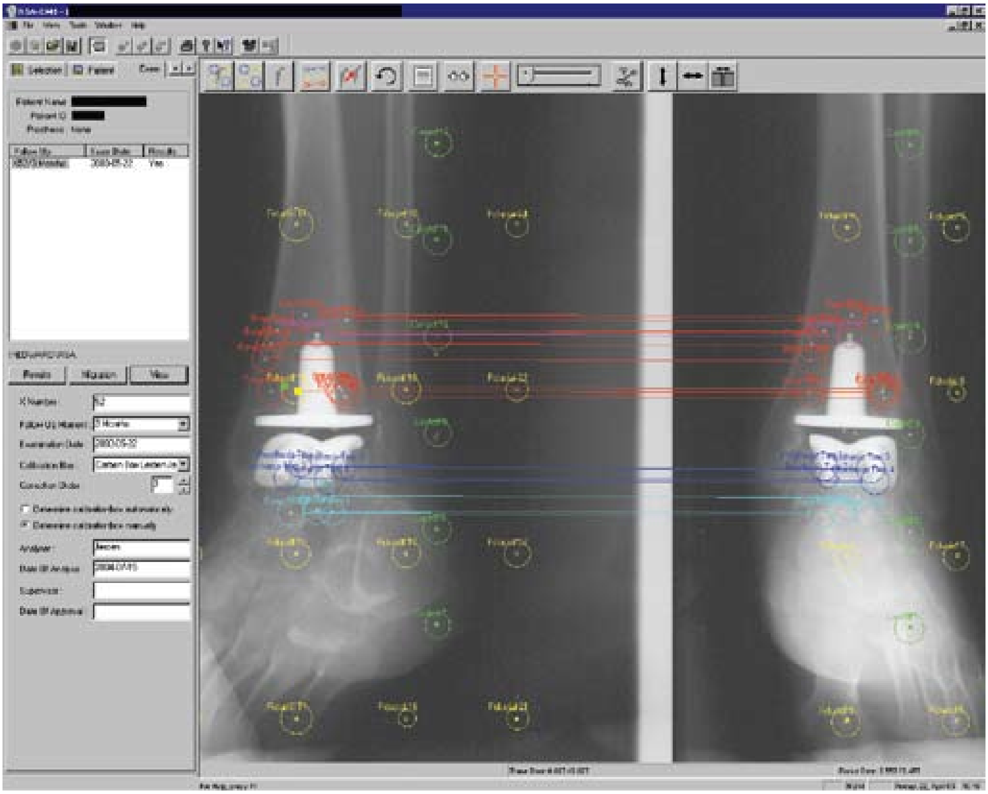

Fig. 2 An example of an analyzed RSA radiograph of the ankle prosthesis.

After administration of prophylactic antibiotics, a tourniquet was applied. This was used during the procedure until skin closure (all procedures were finished within 2 hours). Surgery was performed through a straight anterior midline approach with the leg slightly internally rotated on the surgical table. After incision through the skin and the extensor retinaculum, the interval between the ante¬rior tibial and the extensor hallucis tendons was used to reach the anterior capsule, which then was opened medially and reflected laterally as a sleeve together with the dorsal artery to the foot. The preparation of the osseous surfaces was started by a flat resection of the lower surface of the distal tibia with the aid of a resection guide. Our goal was to achieve a horizontal resection of the distal tibia in the frontal plane and an anterior slope of 7° in the sagittal plane, as recommended by the manufacturer. For the stem of the tibial component a window was created in the anterior part of the distal tibia using a special box chisel.

Preparation of the talar dome was performed by making a central groove antero-posteriorly using a round burr, and subsequently reaming of two slots for the fins of the talar component. After finishing the preparation of the talus, a noncemented talar and a noncemented tibial compo¬nent were implanted. The thickest possible PE liner was then introduced between these components by distracting the joint. After bone grafting around the tibial stem, the removed tibial cortical window was replaced and impacted. During the surgery six 1-mm tantalum markers (MEDIS Medical Imaging Systems, Leiden, The Netherlands) were inserted into the distal tibia. Routine wound closure with careful suturing of the extensor retinaculum was used. We did not lengthen the Achilles tendon in any of the cases. Postoperatively, the ankle was immobilized in a below-knee walking cast for 6 weeks, with weight-bearing allowed as tolerated. The cast was used mainly to protect the sutured extensor retinaculum and promote soft-tissue healing.

Due to logistical problems, replicate examinations for exact accuracy assessment were not made for this study. Therefore, the accuracy of the RSA measurements was extrapolated form an RSA study of a similar sized elbow prosthesis performed at our institution[17]. As the allowed volume for postioning markers in the distal tibia and distal humerus are comparable, we presumed this is a valid assumption. The accuracy for translations along the transverse, longitudinal, and sagittal axes in the earlier study was 0.13 mm, 0.14 mm and 0.34 mm respectively.

All patients were observed at regular intervals. The patients were evaluated preoperatively, then at 1 week, 6 weeks, 3 months, 6 months and 1 year postoperatively, and at yearly intervals thereafter. We recorded complications and evaluated patients with the American Orthopaedic Foot and Ankle Society (AOFAS) score[18]. This scoring system allows for evaluation of pain, function, ROM, and deformity based on a 100-point scale. ROM of the ankle-hindfoot complex was measured using a goniometer. Alignment and dorsiflexion were measured while the patient was standing, and plantar flexion, pronation and supination were measured while the patient was sitting.

Conventional anteroposterior (AP) and lateral radiographs were taken immediately after surgery. The angle between the long axis of the tibia and the line perpendicular to the talar surface on the weightbearing AP view of the ankle was defined as the preoperative alignment of the ankle joint. The angular position of the tibial component was defined as the angle between the base plate of the tibial component and the long axis of the tibia on both views. The angular position of the talar component on the lateral radiograph was defined as the angle parallel to the fins of the talar component and a line drawn from the most dorsal part to the center of the anterior part of the talus. A prosthesis aligned more than 3° off neutral was considered a malaligned prosthesis. The PE bearing was assessed for subluxation and other abnormalities.

The radiostereometric setup consisted of two synchronized roentgen tubes and a uniplanar calibration box (Carbon Box, MEDIS medical imaging systems) (Fig

2). The roentgen tubes were at an angle with the vertical of 20°, thus the angle between the tubes was 40º.

The RSA radiographs, with the patients in a supine position, were taken before weightbearing at the fifth postoperative day and thereafter at 6 weeks, 3 months, 12 months and 24 months. The first RSA examination served as the baseline reference. All subsequent evaluations of micromotion were related to the relative position of the prosthesis with respect to the bone markers at that time. Migration of the tibial component was expressed as translatory movements of the marker attached to the proximal tip of the tibial component along the three orthogonal axes: lateral-medial, posterior-anterior, and distal-proximal. Because only one marker was attached to the tibial component, only translatory migration along the three orthogonal axes could be measured. Rotation could not be measured. Because some of the markers in the bone were obscured, only 12 ankle prostheses could be studied using RSA.

Because this was a study on only one type of ankle prosthesis, a t test was used for calculation of 95% confidence limits. We compared the clinical scores from those obtained preoperatively and those obtained 2 years postoperatively. The SPSS software package (version 11.5; SPSS Inc., Chicago, IL) was used for analysis.

5.3 Results

Initial tibial component migration was seen mainly during the first 3 months. The mean mode of migration of the tibial component was along the distal-proximal axis after surgery, after which progression decreased between 3 and 6 months postoperatively along all three axes (Table 1 and Fig. 3). The common initial migration mode of this component was into anterior, proximal, and valgus tilting.

The clinical AOFAS score at the last follow-up improved ($p = 0.001$) from a preoperative mean score of 22 points (SD 9.7) to a mean 80 points at 2 years (SD 8.0). All patients reported they had little or no pain in their ankles at the last follow-up. If pain was present this was markedly reduced as evaluated by the AOFAS score. Nine patients had no pain in the ankle region and six patients had slight pain in the ankle region at follow-up. Comparing the preoperative with the last follow-up values, the ROM of the ankles improved ($p = 0.03$) from a mean of 2° (SD 5.1°) dorsiflexion to a mean of 7° (SD 4.8°). Plantar flexion did not improve significantly ($p = 0.18$). According to the subscore of the AOFAS score, the subtalar joint in three patients showed moderate function; in 12 patients this function was severely affected (< 20° pronation or supination).

At the last follow-up, we observed no radiolucencies around the tibial component and no signs of bone resorption around the tibial and talar components. All but two patients had their prostheses placed within 2° varus or valgus; the mean alignment was 0.8° (SD 3.1°). Fracture of the PE bearing did not occur in any patients, despite the fact that 10 patients (67%) had the thinnest PE bearing (3 mm).

We observed one superficial wound infection in a patient who had immunosuppressive therapy (anti- TNF alpha). The infection resolved with antibiotic therapy and cast immobilization. One year after this infection, no signs of deep (prosthetic) infection were present.

Two patients had spontaneous distal tibia fractures at the proximal part of the cortical window at 2 weeks while in the below-knee plaster. These fractures healed with prolonged immobilization (8 weeks in a cast postoperative instead of 6 weeks). The migration pattern of these two patients was similar to those of the other patients at 3 to 6 months and beyond.

TABLE 1 Migration of the Tibial Component along Three Orthogonal Axes*

	Migration (mm)**		
Follow-up	Lateral-Medial	Distal-Proximal	Posteroanterior
6 weeks	0.16 (0.43)	0.37 (0.24)	0.01 (0.24)
3 months	0.16 (0.52)	0.63 (0.43)	- 0.37 (0.70)
6 months	0.53 (0.64)	0.69 (0.46)	-0.56 (0.80)
1 year	0.45 (0.82)	0.80 (0.49)	-0.50 (0.77)
2 years	0.80 (0.58)	0.94 (0.26)	-0.51 (0.29)

* At the 1-year follow-up n=12; at the 2-year follow-up n=8; **Means and standard deviations are shown.

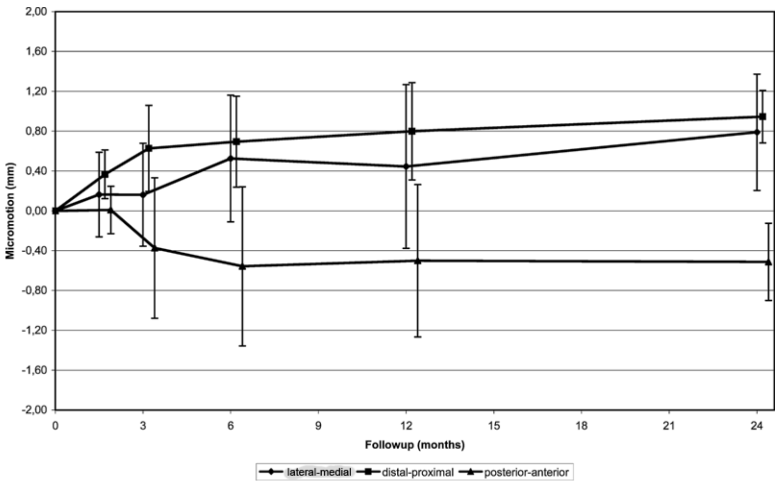

Fig. 3 Translation of the tibial component of the BP mobile-bearing ankle prosthesis (up to 1 year follow-up n=12; 2-year follow-up n=8). A positive lateral-medial migration indicates valgus tilt, a positive distal-proximal migration indicates upward migration, and a negative posteroanterior migration indicates an anterior tilt of the tibial component.

5.4 Discussion

Although this was a pilot study on a mobile-bearing ankle prosthesis with a central tibial peg fixation, some observations can be made on the migration pattern of this design. The migration of the BP ankle prosthesis showed an initial progressive migration that decreased at 3 months and stabilized at 6 months postoperatively. The main mode of migration of the tibial component was into anterior and valgus tilting during the initial postoperative months, after which the prosthesis stabilized. Failure can be related to prosthetic design factors, position of the prosthesis, and biologic factors. Several studies document a relation of ankle prosthesis design and failure[2,19], with revision rates of 30% to 50% for the more constrained and semi-constrained designs. Since the introduction of the nonconstrained mobile-bearing designs, survival rates of total ankle arthroplasty have improved considerably[4].

Limitations of the study include the use of only one marker (attached to the tip of the tibial component). However, translation measurements along the three orthogonal axes could be assessed accurately. Although the study population was small, and follow-up short, the accurate RSA measurements indicated some trends on prosthesis migration. Another limitation was that only RA patients were studied, who are generally considered a low demand population. On the other hand bone quality is in general more osteoporotic in a rheumatoid population and prosthesis migration might reflect a worst case scenario. Carlsson et al.[20] showed similar migration in TAAs between RA and osteoarthritis. In that study, tibial component fixation was different from that in our study. The design effect on migration is also accentuated by the different first 3–6 months migration pattern of the STAR[20] and the BP prosthesis.

Prosthesis alignment contributes to failure. The importance of alignment and stability of the hindfoot in patients with RA is reflected by a failure rate of 24% at 14-year follow-up examinations in patients with malaligned ankles[2,15]. The postoperative alignment of the ankle prosthesis in this series was moderate (87% within neutral alignment). The initial prosthesis migration in this series largely will be determined by biologic factors. Malaligned ankles are more likely to migrate at long-term follow-up[8]. Because an initial progressive migration existed for the studied ankles but stabilized after a few months, a dynamic biologic process is likely to be the cause. This has also been observed with knee prostheses[21]; if cementless prostheses stabilize, the bony-prosthesis interlock is likely to last longer than a slowly degrading bone-cement interface. In addition to these biological factors, mechanical factors will influence longevity of prosthesis survival as well.

We observed no osteolysis around the tibial component at this short follow-up. Importance of the initial biologic factor is not only related to bone ingrowth into the tibial component, but also to the healing of the tibial cortical window made during surgery to insert the pegged tibial component. Although no loosening occurred during this short follow-up, two patients experienced a fracture of the distal tibia at the level of the proximal part of the cortical tibia window during the plaster weight bearing period. Both patients were housebound preoperatively due to the incapacitating pain, suggesting osteopenia may have contributed to the fractures in addition to the stress riser of the cortical window.

Authors have shown varying mid-term and long-term results for different mobile-bearing ankle prosthesis designs[4,6,7,8,9,11,12]. The pegged tibial components of the BP prosthesis show tibial loosening in 12% of cases[8,13]. In contrast, the Scandinavian Total Ankle Replacement[11,12], Salto[6], and Hintegra[9] ankle prostheses have been shown to have lower loosening rates, although with shorter follow-up in the newer designs. The major differences between these designs are mainly focused on the fixation of the tibial side. The BP design needs a cortical window for insertion, whereas the other designs rely on the subchondral tibial bone without making a cortical window. On the tibial side, the cylindrical peg provides good compression of the metal base plate onto the distal tibial cut surface at impaction, eliminating the need for a tibial window. Being located away from the cut tibial surface, there is no risk of bone weakening during drilling. The high variability of migration of the BP design might be partly reflected in future differences in long-term survival rate. Ryd et al.[15] and Kärrholm et al.[14] showed that high initial migration is related to long-term failure, which implies the tibial fixation and insertion techniques used with the BP prosthesis in the current study might have to be changed in order to improve long-term survival.

Because all patients in the current study had RA and patients with RA generally have more osteoporotic bone, initial prosthesis migration might be higher compared with that seen in patients with osteoarthritis. However, in a small study the authors could not confirm this observation with that particular design[20]. Furthermore, survival rates for ankle prostheses implanted in patients with RA are similar to OA, with an 8-year survival of 82% (95% confidence interval 73% to 93%) in RA patients[4,8,13,22]. Other factors to consider when treating patients with RA are hindfoot problems, which are seen in 60% of patients with ankle joint arthritis[1,23]. This is either caused by tendon disease of the tibialis posterior, which contributes to the valgus shift of the hindfoot, or a valgus shift of the hindfoot resulting from subtalar and midfoot arthritis. These latter processes will cause more stress on the ankle prosthesis, which might be reflected in the lower implant survival rates in patients with RA. In our series all patients had deteriorated subtalar and midtarsal foot function preoperatively. In three patients the function was moderate and in 12 patients this function was affected severely (< 20° pronation or supination according to the AOFAS score).

We showed initial migration of this mobile-bearing ankle prosthesis into upward anterior and valgus tilting, which stabilized at 6 months. This pattern was observed in other cementless prostheses as well[21]. The surgical (anterior cortical window for placement) and tibial fixation techniques may explain this early migration pattern although such pattern might differ with a different design[20].

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Chapter 6

Medial Malleolar Osteotomy for the Correction of Varus Deformity during Total Ankle Arthroplasty: Results in 15 Ankles

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Abstract

Background Preoperative deformity in the frontal plane in the arthritic ankle is a risk factor for failure after total ankle arthroplasty. Medial malleolar lengthening osteotomy was developed to correct varus malalignment.
Materials and Methods From 1998 to 2005 total ankle arthroplasty combined with medial malleolar lengthening osteotomy was done in 15 ankles (13 patients) with a mean preoperative varus deformity of 14.9 (SD 7.8) degrees. Diagnosis was instability arthritis in 11 ankles (9 patients) and inflammatory joint disease in 4 ankles (4 patients). Two mobile-bearing designs were used. Osteosynthesis of the osteotomy was done in 2 ankles, for the remaining 13 osteotomies no fixation was carried out.
Results Follow-up was 5 (2-8) years. Neutral alignment was obtained in all ankles. In 3 patients residual hindfoot varus remained, for which a second-stage hindfoot correction was done. Two rheumatoid ankles developed a symptom-free nonunion of the medial malleolus, all other malleolar osteotomies united. One tibial component, implanted with too much anterior slope, developed early aseptic loosening and was revised. Debridement for talar-malleolar arthritis was done in two ankles. Of the 14 ankles in follow-up, 12 were rated as excellent or good, one as fair. One ankle with subsidence of the talar component was rated as unsatisfactory. AOFAS score increased from 30.8 preoperative to 81.0 at follow-up (p <0.01).
Conclusion Medial malleolar lengthening osteotomy is an easy technique for the realignment of the varus ankle at the time of total ankle arthroplasty, and serves as an alternative to medial ligament release or lateral ligament reconstruction.

6.1 Introduction

Posttraumatic arthritis of the ankle joint, either due to intra-articular fractures or to chronic lateral ligament insufficiency, and rheumatoid arthritis (RA) are the main indications for reconstructive surgery of the ankle joint[1]. In recent years total ankle arthroplasty (TAA) has gained popularity, although this procedure has not yet been widely accepted as a treatment option for the severely affected ankle joint[2,3].

Stability in neutral alignment is of paramount importance for a good long-term result after TAA. In posttraumatic arthritis, a varus deformity of the ankle joint is seen regularly, especially as a result of arthritis due to chronic lateral ligament insufficiency (instability arthritis). In the rheumatoid ankle, varus deformity also occurs. In ankles with a preoperative deformity in the coronal plane, edge loading of the prosthesis is a frequent complication[1]. In a series of TAA for inflammatory joint disease (IJD), ankles with a preoperative deformity in the coronal plane of more than 10 degrees had a significantly reduced survival rate compared to neutrally aligned ankles: 48% versus 90% after eight years[4]. In general, such a deformity is not easily correctible, and, therefore, has been considered a contraindication for total ankle arthroplasty[1,4].

Kofoed[5,6] described a talar sculpturing technique for correction of deformity in the coronal plane, if necessary combined with lateral ligament reconstruction in varus ankles with lateral ligament laxity. An alternative would be to perform a lengthening procedure at the medial side, just like for the correction of a varus deformity in the knee during total knee arthroplasty. Bonnin et al.[7] described good results after proximal and distal deltoid ligament release for varus deformity. Haskell and Mann[8] advised either lateral ligament reconstruction or deltoid ligament release or both to stabilize the replaced joint in neutral alignment. We used medial ligament release on a limited scale. However, after the occurrence of a severe skin necrosis in one ankle, we abandoned this technique. Then, lateral ligament reconstruction with use of the peroneus brevis tendon according to Kofoed was used. However, early recurrent instability complicated this procedure. Therefore, in 1998, as an alternative to deltoid ligament release, lengthening osteotomy of the medial malleolus was developed in order to balance the arthritic ankle with varus deformity during total ankle arthroplasty. As early results were encouraging (Fig. 1), a study of all varus ankles treated by this technique was started in order to investigate the correction obtained and the clinical and radiographic outcome. To our knowledge this technique and its results have not been described before.

Fig. 1-A

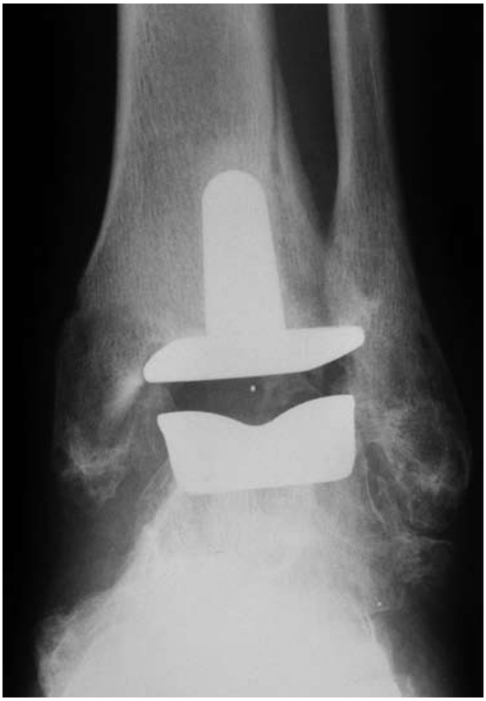

Fig. 1-B

Fig. 1-C

Fig. 1-D

Figs. 1-A through 1-D Radiographs of a 51-year-old-man with bilateral instability arthritis. This case was the first ankle where total ankle arthroplasty was combined with a medial malleolar osteotomy. ***Fig. 1-A*** Preoperative weightbearing view, showing a 15 degrees varus deformity of the left ankle joint. ***Fig. 1-B and 1-C*** Postoperative anteroposterior and lateral views, showing restoration to neutral alignment of the ankle by medial malleolar lengthening osteotomy and of the hindfoot by a calcaneal closing-wedge osteotomy. An incomplete fracture of the anterolateral aspect of the distal tibia was present, however, did not require fixation. ***Fig. 1-D*** At follow-up, the malleolar osteotomy has healed in a lengthened position, and the intraoperative distal tibial fracture has also united. The clinical result was excellent.

6.2 Materials and Methods

All ankle arthroplasty patients in our hospital enter a prospective follow-up protocol approved by the institutional review board. Patients participating in this study all gave informed consent. Our experience with mobile-bearing ankle arthroplasty started in 1988, when the Low Contact Stress (LCS®, DePuy, Warsaw, Indiana) was introduced in our hospital[9]. From 1993 till 2004 the Buechel-Pappas prosthesis™ (Endotec, South Orange, New Jersey)[10] has been used. The Buechel-Pappas prosthesis™ evolved from the LCS design. From 2004 onwards a new mobile-bearing design was used, the Ceramic Coated Implant (CCI) Evolution total ankle prosthesis, Van Straten Medical, Nieuwegein, The Netherlands (Fig. 2). In contrast to the Buechel-Pappas™ prosthesis, the tibial component of the CCI prosthesis has a fixation fin instead of a stem and the talar component requires a triple-V shaped resection of the talar dome instead of a curved resection.

Fig. 2-A

Fig. 2-B

Fig. 2-C

Figs 2-A through 2-C Radiographs of the Ceramic Coated Implant Evolution ankle prosthesis, as implanted in a 55-year-old-woman with instability arthritis. ***Fig.1-A*** Preoperative view, showing a 22 degree varus deformity. ***Figs. 2-B and 2-C*** Anteroposterior and lateral views, showing the two-year radiographic result with solid healing of the malleolar osteotomy, that was stabilized by two wires

6.2.1 Surgical Technique

The thigh was routinely placed in a leg holder, with the knee flexed about 60 degrees to relax the calf muscles and to rotate the lower leg to a neutral position. A straight anterior midline approach was used. After incision of the skin and the extensor retinaculum, the interval between the anterior tibial and the extensor hallucis tendons was used to reach the anterior capsule, which was then opened medially and reflected laterally as a sleeve together with the dorsal pedis artery. The tourniquet was inflated after the arthrotomy and before the beginning of the preparation of the osseous surfaces. The preparation of the osseous surfaces was started by a limited resection of the lower surface of the distal tibia, aiming for a horizontal resection of the distal tibia in the frontal plane and for a 7 degree anterior slope, as recommended by Buechel et al.[10]. After having finished the preparation of distal tibia and talus, a non-cemented talar and a non-cemented tibial component were implanted. Only at this time, if after insertion of the trial bearing a varus tilt remained, resulting in edge loading of the components, a lengthening osteotomy of the medial malleolus was performed. The osteotomy was directed in a slightly oblique direction. It was then mobilized with an osteotome and by downward sliding of the distal part of the medial malleolus, the ankle became balanced in a neutral position (Fig. 3). The downward sliding of the distal fragment was limited, rarely exceeding 4 to 5 mm, and in no case required traction at the distal fragment. The thickest possible polyethylene liner was then introduced in between these components. In general, no fixation of the medial malleolus was required. Routine wound closure with careful suturing of the extensor retinaculum finished the procedure. The ankle was immobilized in a below-knee cast, usually for a period of 6 weeks, with weightbearing to tolerance from 3 to 5 days after surgery.

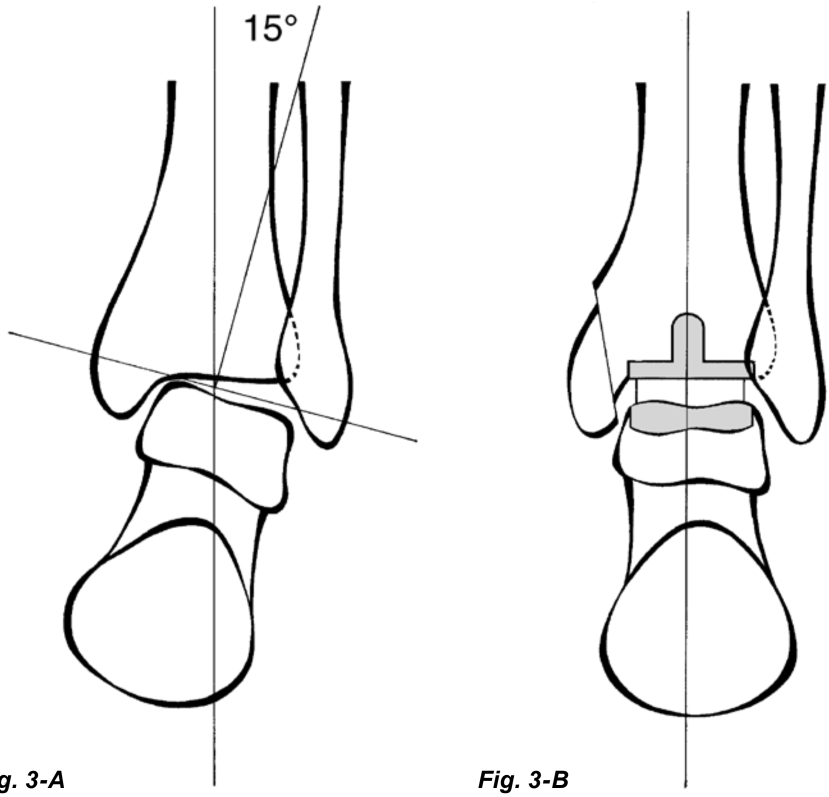

Fig. 3-A ***Fig. 3-B***

Fig. 3-A and 3-B Schematic drawings of the medial malleolar lengthening osteotomy. ***Fig. 3-A*** Ankle with incongruent varus deformity. ***Fig. 3-B*** Situation after implantation of a mobile-bearing prosthesis and correction of the deformity by medial malleolar osteotomy.

6.2.2 Clinical and Radiographic Evaluation

The results of all total ankle replacements for end-stage arthritis with a pre-existing varus deformity that were treated by this medial malleolar lengthening technique and with a minimum follow-up of 24 months were studied prospectively using the American Orthopaedic Foot and Ankle Society (AOFAS) ankle/hindfoot score[11] and a qualitative subjective scale for clinical evaluation. Evaluation moments were: pre-operatively and postoperatively at 6 weeks, 3 months, one year and then annually or bi-annually. If the medial malleolar osteotomy had not united after 3 months, in most cases a follow-up moment was arranged at 6 months. Postoperative ankle-hindfoot alignment was evaluated clinically with the patient standing, using a goniometer. Furthermore, at the most recent follow-up, a validated translation of the Foot Function Index (FFI) 23 items 5-point verbal rating scales questionnaire[12], and a Visual

Analogue Scale (VAS) for pain (scale from 0 to 100, where 0 is no pain and 100 the most severe pain) and for satisfaction (scale from 0 to 100, where 0 is fully unsatisfied and 100 fully satisfied) were also used for the evaluation of the clinical result. The FFI scores range from 0 to 100; the higher the score, the more limitation, pain and disability respectively.

Radiographic evaluation was performed in a standardized fashion as described by Doets et al.[4]. The angle between the long axis of the tibia and the line perpendicular to the talar dome on the preoperative weightbearing anteroposterior view of the ankle was defined as the preoperative alignment of the ankle joint. At follow-up, nonweightbearing anteroposterior and lateral radiographs aimed to be parallel to the base plate of the tibial component were made, if necessary using fluoroscopy. The serial radiographs were evaluated by a radiologist (JPK) who was not involved in the care of these patients. The angular position of the tibial component was defined as the angle between the base plate of the tibial component and the long axis of the tibia on both views. The angular position of the talar component on the lateral radiograph was defined as the angle parallel to the fins of the talar component and a line drawn from the most dorsal part to the center of the anterior part of the talus. This talar reference line was chosen as it could reliably be drawn also in the event of a fused subtalar joint. The position of the talar component in the frontal plane in relation to the hindfoot was difficult to assess accurately on plain radiographs, and was assessed semi-quantitatively. The occurrence of radiolucent lines next to the prosthetic components was measured on the anteroposterior and lateral radiographs for the tibial component and on the lateral radiograph for the talar component. Our criterion for migration of a component was an angular change in position of more than 3 degrees or a subsidence of more than 3 mm in one of the views or a complete radiolucent line of more than 1 mm in both views for the tibial component and on the lateral view for the talar component. The polyethylene bearing (which had two small metallic markers) was assessed for subluxation in the sagittal and frontal planes and for other abnormalities (gross wear or fracture). Any other adverse phenomenon occurring at follow-up was recorded.

6.2.3 Statistical Analysis

All data were recorded in an ankle module of the Project Manager data management program (IMSOR, Leiden, The Netherlands). Student's paired t test was used to evaluate the prospectively studied clinical outcome parameters. Differences were considered significant when $p < 0.05$. Statistical evaluation was done with SPSS software (version 14).

6.3 Results

6.3.1 Patient Demographics and Perioperative Data

From September 1998 to January 2005 fifteen ankles (thirteen patients, nine men (two bilateral) and four women) with a varus deformity or varus instability were treated by a primary TAA combined with a lengthening osteotomy of the medial malleolus. Diagnosis was instability arthritis in nine patients (eleven ankles) and IJD in four. In twelve ankles a Buechel-Pappas™ prosthesis was implanted, and in three ankles a CCI Evolution prosthesis. Mean preoperative varus deformity was 14.9 degrees (SD 7.8 degrees; range, 3 to 30 degrees). Four ankles had a preoperative varus deformity of less than 10 degrees (instability arthritis 2, IJD 2). Mean follow-up was 61 (range, 24 to 99) months. Patient demographics are listed in Table 1.

TABLE 1 Demographics of the Patient Population

Diagnosis	No. of Ankles (Patients)	Age* (yr)	Women/Men	Preoperative Varus* (degrees)
Instability Arthritis	11 (9)	55.8 (47-69)	2/9	14.4 (3-30)
RA	2 (2)	74.5 (69-79)	1/1	20 (10-30)
Miscellaneous Arthritis#	2 (2)	37.5 (31-44)	2/-	12.5 (5-20)
All diagnoses	15 (13)	55.5 (31-79)	5/11	14.9 (3-30)

* Mean values, with the range in parentheses. # Juvenile Chronic Arthritis 1, Oligoarticular Arthritis 1.

Internal fixation of the osteotomy was judged to be necessary in two ankles on the basis of a persisting anterior drawer sign after the prosthetic components had been inserted. All other medial malleolar osteotomies were not stabilized as there was sufficient stability by the soft tissue envelope. They were just immobilized by the routine postoperative plaster cast. Compared to ankle arthroplasty without a preoperative deformity in the frontal plane, relatively thick polyethylene bearings were used, having a height of between 7 and 11 mm. In the first ankle treated by this procedure, in addition to the ankle replacement, a valgus closing-wedge osteotomy of the calcaneus (Dwyer-type) was performed for correction of a varus hindfoot. Superficial wound healing disturbances were seen in two ankles (one male patient, suffering from bilateral instability arthritis). These wounds healed by prolonged cast immobilization.

6.3.2 Clinical Outcome

Follow-up was completed as of January 2007. No patients were lost to follow-up.

Neutral alignment was obtained in all ankles at the level of the ankle joint, but a slight residual varus at the hindfoot persisted in three ankles (in two as a result of an ankylosed subtalar joint with varus malalignment). These three ankles underwent a second-stage corrective procedure of the hindfoot. Another ankle underwent successful arthroscopic and open debridement for persistent pain due to talar-malleolar arthritis. Furthermore, in one ankle, where the tibial component had been implanted with excessive anterior slope (15 degrees), an anterior subluxation persisted, resulting in early anterior tilting with aseptic loosening of the tibial component. This tibial component was revised 16 months after the index surgery. Details of all reoperations are given in Table 2.

Mean follow-up of all ankles in follow-up (n=14) was 64 (range, 24-98) months. The AOFAS ankle score increased from 30.8 (SD 11.4; range, 14-53) preoperative to 81.0 (SD 14.3; range, 54-100) at latest follow-up (p <0.001). Dorsiflexion increased significantly (p <0.01) from a preoperative mean of 4.3 degrees (range, -12 to 15 degrees) to a mean of 9.5 degrees (range, 0 to 25 degrees) at latest follow-up. Plantarflexion remained the same, with a mean of 30.1 degrees preoperative and at follow-up.

Patients rated their ankles as very satisfactory in seven, satisfactory in five, and one ankle was rated as fair. Furthermore, one ankle was rated as unsatisfactory. This ankle developed mechanical loosening of the talar component in a rheumatoid ankle with pre-existing medially localized talar bone loss (discussed below).

At the most recent follow-up, patient-assessed results were as follows. The FFI total score was 19.7 (SD 12.4), the pain subscore was 24.1 (SD 16.4), the disability subscore 23.5 (SD 15.3), and the limitations subscore 11.6 (SD 8.3). The VAS for pain was 27.1 (SD 24.4) and the VAS for satisfaction was 76.1 (SD 23.5).

6.3.3 Radiographic Findings

In this study, the malleolar lengthening osteotomy was carried out at either the base of the medial malleolus (nine ankles) or halfway in the medial malleolus (six ankles). Two malleolar osteotomies, both in patients with IJD, and both located at the base of the medial malleolus, developed a stable and symptom-free nonunion. Both nonunions occurred in ankles with some residual varus deformity at the level of the hindfoot (Table 2). The remaining thirteen united after a mean interval of seven months (range, 3 to 12 months). Two tibial components showed initial migration, one into varus, and one an anterior tilt. The tibial component with varus tilt stabilized secondarily and showed no radiolucent lines with longer follow-up (77 months). Details of the tibial component that developed an anterior tilt were described above. Of the thirteen tibial components without signs of migration ten showed no radiolucent lines and three showed partial radiolucent lines. One talar component in a rheumatoid ankle with steroid-induced osteopenia developed aseptic loosening with subsidence of the talar component and edge loading 30 months after the index surgery. A revision procedure was offered to this patient but was refused as the symptoms remained at an acceptable level. All other talar components showed a complete osseointegration without migration. No edge-loading or bearing subluxation was seen in the thirteen ankles with stable implants.

TABLE 2 Overview of the Reoperations

Gender, Age (yr)	Year of Surgery	Diagnosis*	Preop varus (degrees)	Months to Reoperation	Reason for Reoperation	Subsequent surgery	Subjective result; AOFAS score
M, 54	1999	IA	15	11; 41	#1 Hindfoot varus; #2 Medial-sided pain	#1 Dwyer; #2 Arthroscopic debridement	Fair, 78
F, 44‡	1999	Oligoarticular Arthritis	20	15	Ankylosed varus hindfoot	Corrective triple	Very Satisfied; 78
M, 66‡	2001	RA	10	20;	Ankylosed varus hindfoot + malunion distal tibia	# 1Corrective triple; #2 Supramalleolar osteotomy	Satisfied; 74
M, 59	2003	IA	10	9; 18	Persistent anterior + medial pain	#1 Arthroscopic debridement; #2 Open debridement	Very Satisfied; 97
M, 54	2004	IA	14	16	Aseptic loosening tibial component + anterior subluxation	Revision of tibial component	Satisfied (after revision of tibial component); 77

* IA = Instability arthritis; RA = Rheumatoid arthritis. ‡ Patients with a nonunion of the medial malleolus

6.4 Discussion

Coronal plane deformity is a relatively frequent phenomenon in the arthritic ankle. In the posttraumatic ankle, chronic lateral ligament insufficiency can be considered the main underlying cause of varus deformity. Another factor leading to varus deformity can be a varus inclination of the tibial plafond[13]. Varus deformity can be also be seen in IJD, probably as a result of eccentric cartilage destruction and attenuation of the lateral ligaments by chronic synovitis. In a personal series, the incidence of varus deformity in patients with IJD requiring replacement of the ankle was twelve out of eighty (15 per cent). If surgical reconstruction is considered (either fusion or arthroplasty), coronal plane deformity should be corrected in order to obtain a good and lasting result[5,14]. Failure to correct such a deformity in TAA will lead to an increased rate of edge loading, bearing subluxation and early failure[1,4].

A simple way to improve the results of total ankle arthroplasty is to exclude patients with coronal plane deformity. Therefore, some authors considered a significant (>10 to 15 degrees) of talocrural or hindfoot malalignment a contraindication for total ankle arthroplasty[1,4,15]. The other, more difficult way would be to develop a reliable and reproducible surgical technique to produce a well-aligned and stable ankle.

Kofoed[5,6] advised a talar sculpturing technique for correction of coronal plane deformity, if necessary combined with lateral ligament reconstruction using the peroneus brevis tendon to improve stability in the varus ankle. Bonnin et al.[7] reported good results after extensive deltoid ligament release in a subgroup of fourteen varus ankles, as part of a study on ninety-three ankle replacements. Their follow-up was short: 35 months, and no specific details were given of this subgroup concerning diagnosis, preoperative alignment and amount of correction obtained at surgery. It has been suggested that arthrogenic muscle inhibition due to damaged mechanoreceptors in the lateral ligaments is present in patients with functional ankle instability[16]. As an extensive medial ligament release could result in additional damage to mechanoreceptors located in the deltoid ligament, such a procedure might worsen the muscular control of the replaced ankle. Haskell and Mann[8] advised either to stabilize the varus ankle by lateral ligament reconstruction, as proposed by Kofoed, or to realign it through a deltoid ligament release, or a combination of these procedures. They reported good results after reconstruction in congruent ankles, but had a high rate of early edge-loading after TAA in varus-incongruent ankles: four out of ten. Non-anatomic lateral ligament reconstruction is known to have mixed results for repair of chronic laxity in the non-arthritic ankle[17,18]. Therefore, it appears questionable that a non-anatomic peroneus brevis tendon reconstruction will durably restore stability after total ankle arthroplasty.

As we had inferior results with either medial ligament release (skin necrosis) and lateral ligament reconstruction (recurrent instability) after TAA in the arthritic varus ankle, it was decided to try to balance the ankle joint by a lengthening procedure of the medial malleolus without an additional reconstruction of the lateral ligament complex. Our results show that a varus deformity of up to 30 degrees at the level of the ankle joint can reliably be corrected by this procedure. Except for one ankle with talar component subsidence and one ankle with early failure due to tibial component malposition, all other ankles obtained a good result without a recurrence of varus deformity at the ankle joint after a mean follow-up of five years. However, residual deformity at the hindfoot, not corrected at the index operation, required subsequent surgery in three ankles. In retrospect, these hindfoot deformities should have been corrected either at the time of total ankle arthroplasty or as the first intervention of a two-stage procedure[7]. In view of the two implant failures and the four secondary procedures in this study, results with this procedure are until now still inferior compared to results of TAA in well-aligned ankles. To further improve our results, the following surgical modifications have been introduced: a) the lengthening osteotomy is done halfway down the medial malleolus; b) the medial gutter is routinely debrided; c) the tibial component is implanted with no or only minimal anterior slope; and d) any hindfoot deformity is corrected prior to or simultaneously with the arthroplasty.

In conclusion, lengthening osteotomy of the medial malleolus can be considered a simple and effective procedure for the correction of varus malalignment at the time of total ankle arthroplasty, and an alternative to medial ligament release or lateral ligament reconstruction. Fixation of the osteotomy by hardware is not required routinely. Time to union of the osteotomy can be relatively long, but a delay in radiographic union does not lead to a delay in rehabilitation. There is no increased risk of disturbed wound healing. Correct implant position and complete correction of a concurrent hindfoot varus remain crucial factors for a good long-term result.

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Chapter 7

Salvage Arthrodesis for Failed Total Ankle Arthroplasty Medium-term Results in Eighteen Ankles

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Abstract

Introduction. The objective of this study was to investigate the clinical, radiographic and subjective outcome after salvage arthrodesis for failed total ankle arthroplasty (TAA), with a focus on salvage in inflammatory joint disease (IJD).

Methods. Between 1994 and 2005, salvage arthrodesis for failed mobile-bearing TAA was performed in 18 ankles (18 patients). Primary diagnosis was IJD in 15 and osteoarthritis in 3. Tibiotalar fusion was performed in 7 and tibiotalocalcaneal fusion in 11 ankles (in 9 out of these, the subtalar joint was already ankylosed). Serial radiographs were studied retrospectively by an independent observer for time to union. Clinical outcome at latest follow-up was measured by the AOFAS score, by the Foot function Index (FFI) and by VAS scores for pain, function and satisfaction.

Results. Blade plates were used in 7 ankles, all united. Nonunion developed in 7 IJD ankles stabilized by either a nail or screws or multiple K-wires. Revision arthrodesis was done for 4 nonunions, 3 were successful. Eleven patients (8 fused ankles, 3 nonunions) were available for clinical evaluation. At follow-up, their mean AOFAS score was 62.4; mean overall FFI was 70.1; VAS for pain was 20.1, for function 64.3, for satisfaction 73.8.

Conclusions. Blade plate fixation is successful in salvage ankle arthrodesis. A high nonunion rate was found after salvage ankle arthrodesis in IJD with other methods of fixation. Clinical results were relatively good. The three nonunions in follow-up had subjective results similar to the fused ankles. Several publications on primary ankle arthrodesis also showed an elevated nonunion rate in IJD.

7.1 Introduction

Total ankle arthroplasty (TAA) with use of a mobile-bearing design can be seen as an alternative for arthrodesis for the treatment of the painful arthritic ankle. Several reports exist on the medium-term to long-term results with use of these third-generation designs, showing acceptable results with 83.7 to 93.5 per cent survival at eight years[1,2,3].

If TAA fails, mostly as a result of mechanical loosening, either salvage arthrodesis or implant exchange must be a reliable salvage procedure. In most studies dealing with results after third-generation TAA, some information is given on the failures after TAA. Some earlier studies have addressed the results of salvage after failed TAA[4,5,6,7]. These studies dealt with conversion to either tibiotalar or to tibiotalocalcaneal arthrodesis, and described the success rate of arthrodesis after failure of first-generation, constrained two-component designs, but gave limited information on the final clinical result. The number of ankles treated for rheumatoid arthritis (RA) in these studies is variable. Stauffer[4], and Groth and Fitch[5], were the first to report on the results of salvage arthrodesis after failed TAA. They found solid fusions in respectively all seventeen (underlying diagnosis not given) and eleven osteoarthritic (OA) cases. Kitaoka and Romness[6] included ten cases of inflammatory joint disease (IJD) in their series of thirty-eight ankles. Union was achieved in thirty-three out of thirty-seven ankles (one patient died early after the salvage surgery, before union could occur). Only the series reported by Carlsson et al.[7] included a relatively large number of patients with RA: sixteen out of twenty-one ankles. In their study, out of twenty-one arthrodeses seventeen united (four after a second attempt). In all of these studies patient satisfaction, pain and function were scored, but detailed subjective scores were not presented. Recently, some studies have been published on the salvage of mobile-bearing TAA. Hopgood et al.[8] published the results of fusion for failed TAA in twenty-three ankles (twelve OA, eleven RA). They found good results in osteoarthritic ankles and in RA ankles treated by a retrograde nail. However, all four RA fusions stabilized by screw fixation failed to heal. Culpan et al.[9] published successful results of fusion for failed TAA in sixteen ankles (mostly OA). They routinely used a tricortical strut graft from the iliac crest for defect filling and screws for fixation. Anderson et al.[10] reported on sixteen salvage fusions in RA ankles with use of a retrograde nail and either allograft or autograft bone. Eleven healed at the first attempt, another two after repeat arthrodesis.

The goal of this study was to determine whether salvage arthrodesis could be an adequate second line of defense after failed mobile-bearing TAA, with a special attention to salvage of the failed ankle prosthesis in patients suffering from IJD. Secondary research questions were: which salvage arthrodesis techniques were successful, and what was the subjective outcome of the salvage arthrodesis.

7.2 Material and Methods

7.2.1 Patient Demographics

This study was approved by the Institutional Review Board, and all patients seen at follow-up gave informed consent. Total ankle arthroplasty with use of a mobile-bearing design was introduced in our institution in 1988, and, until 2000, was mainly carried out in patients suffering from IJD (mostly RA)[2]. During the study period, conversion to a tibiotalar or tibiotalocalcaneal arthrodesis was the standard surgical treatment for failed TAA, and revision TAA was carried out only in a few selected cases.

Between 1994 and 2005, a total of eighteen patients (18 ankles) underwent a salvage arthrodesis for failed TAA at our institution (Table 1). Mean age at surgery was 55 years (range, 27 to 76 years). There were fifteen patients with a diagnosis of IJD (mostly RA) and three patients with a diagnosis of osteoarthritis. At the time of the salvage procedure, nine hindfeet in the arthritic population were ankylosed, either by a formal surgical fusion or having occurred spontaneously.

7.2.2 Surgical Technique

Sixteen out of eighteen salvage procedures were done by two experienced foot and ankle surgeons, two other surgeons each performed one salvage procedure.

The chosen stabilization technique depended on the following factors: condition of the subtalar joint at the time of surgery, quality of the local bone, and the extent of local bone loss. Blade plates (either an AO humeral plate or an AO child hip plate, Synthes GmbH, Solothurn, Switzerland) were used in seven tibiotalar arthrodeses, implanted either at the anterior or the lateral aspect of the ankle. Compression at the arthrodesis site was applied with the aid of an AO compression device. Blade plate fixation was our preferred technique in the presence of a normal subtalar joint, as rigid fixation could be obtained without interference with the subtalar joint in such cases. In six ankles a locking nail was used to stabilize the ankle, implanted in a retrograde fashion. In four ankles two to three compression screws were used, and one ankle in an elderly RA patient with severe osteopenia was stabilized by multiple K-wires.

In fourteen ankles a cancellous autologous bone graft was used to fill osseous defects, mostly harvested locally. In three ankles morselized allograft bone was used, in one of these combined with an autograft.

7.2.3 Clinical Evaluation

Medical charts of the failed ankle arthroplasties were reviewed in detail for patient characteristics, reason for failure of the prosthesis, operative technique of the arthrodesis, and for any postoperative complication or reoperation. At the time of the final evaluation, in 2008, the following clinical instruments were used to assess the clinical result of all ankles in follow-up:

1) The Ankle-Hindfoot Score developed by the American Orthopaedic Foot and Ankle Society (AOFAS). It is a 100-point score, constituted of both subjective and objective clinical parameters . The maximum attainable AOFAS score is 89 points after a tibiotalar arthrodesis, and after a tibiotalocalcalcaneal arthrodesis it is 86 points.
2) The Foot Function Index with verbal rating scales (FFI-5pt). This is a questionnaire of 23 items, which refers to complaints in the foot and ankle during daily life. The FFI-5pt measures pain and mobility limitation as the impact of foot problems on foot function. The scale consists of 23 items divided into 3 subscales: limitation (5 items), pain (9 items) and disability (9 items). The items of the FFI-5pt are rated on a 5 point-scale, ranging from 'never' (0) to 'always' (4) on the limitation subscale, from 'no pain' (0) to 'intense pain' (4) on the pain subscale and from 'no difficulty' (0) to 'impossible' (4) on the disability subscale. The respondent is asked, for each item, to tick the box that best corresponds with the impact of the foot complaints in terms of limitation, pain and disability. If function loss is not a result of foot complaints, the respondent is asked to tick the box "not applicable". That item is then not included in the calculations. To calculate the subscale scores the item scores are summed up, divided by the maximum possible sum of the item scores and then multiplied by 100 in order to calculate the definitive subscale scores. The total score is the mean of the subscale scores and ranges from 0 to 100. Higher scores indicate more limitation, pain and disability, respectively. A Dutch version of the FFI-5pt has been validated .
3) Visual Analogue Scales (VAS) with a scale from 0 to 100, to score pain (0 is no pain and 100 the most severe pain), limitation of function (0 is complete limitation of function, 100 is a normal function), and satisfaction of the treatment result (0 is fully unsatisfied, 100 fully satisfied).

The latter two instruments were used for a subjective evaluation of the outcome of the salvage procedure at latest follow-up.

7.2.4 Radiographic Evaluation

For the radiographic evaluation, the serial radiographs were evaluated for the time to fusion (at first or at second attempt) and for the alignment of the fused ankle in the coronal and the sagittal plane. This radiographic evaluation was done by a radiologist (JPK) not involved in the care of these patients.

7.2.5 Statistics

Data of all patients were recorded in an especially developed ankle module of the Project Manager data management program (Dpt. of Medical Statistics, Leiden University Medical Center, Leiden, The Netherlands). Two-sided Fisher's exact test was used to analyze the influence of fixation method (blade plate vs. nail or screws) in the IJD population. Confidence intervals and Fisher's exact test were computed with SPSS software (version 14; SPSS, Chicago, IL, USA).

7.3 Results

In 2008, at latest follow-up, the mean follow-up time of all salvaged ankles (patients deceased and in follow-up) was 7.3 years (range, 3.1 to 12.1 years).

7.3.1 Union Rate and Method of Fixation

The individual outcome of all ankles is given in Table 1. Eleven out of eighteen ankles healed after a first-attempt salvage arthrodesis. Mean time till solid fusion in this group was 6.3 months (range, 2 to 16 months; 95% confidence interval 3.5 to 9.1). All seven nonunions occurred in the rheumatoid ankles, accounting for 47 per cent nonunion rate in this group after first-attempt fusion. Four patients with a nonunion underwent a second-attempt salvage arthrodesis, resulting in union in two ankles. One more united after a third attempt, resulting in a final nonunion rate in the rheumatoid ankle group of 27 per cent. These cases will be described in the paragraph below.

In seven ankles a blade plate was used at the first-attempt salvage procedure, and all of these united (Fig. 1). Three out of six first-attempt procedures stabilized with a retrograde nail developed a nonunion (Fig. 2). Furthermore, two out of four screw fixations and the salvage arthrodesis stabilized by K-wires failed to heal. The difference in union rate between the rheumatoid ankles stabilized by a retrograde nail or screws and by a blade plate was not statistically significant ($p = 0.08$).

Seven out of eight rheumatoid ankles, in which autologous bone graft, derived from locally removed bone was used, healed after a first-attempt salvage. Three out of five rheumatoid ankles, in which an autologous bone graft from the iliac crest was used healed after first-attempt salvage arthrodesis. In two rheumatoid ankles only allograft was used, both resulting in nonunion. In three osteoarthritic ankles, treated by blade plate fixation and additional bone graft, an early fusion was seen.

Fig. 1-A

Fig. 1-B

Fig. 1-C

Figs. 1-A through 1-C Radiographs of a man with long-standing RA and a preoperative varus deformity of the ankle of 20 degrees (case 3). ***Fig. 1-A*** Anteroposterior view after implantation of an LCS prosthesis. There is a persistent varus deformity and edge-loading of the prosthesis. ***Figs. 1-B and 1-C*** Anteroposterior and lateral views after conversion to tibiotalar arthrodesis. The arthrodesis was stabilized by a humeral blade plate, implanted at the lateral side. Debris originating from the edge-loading of the metallic components is visible at the arthrodesis site.

7.3.2 Complications and Reoperations

One patient (# 13) required nail extraction at four months because of a deep infection. The ankle appeared to have fused, and no further surgical treatment was necessary. Another patient (# 17), stabilized by a blade plate, also required hardware extraction at four months because of low-grade infection. Although this ankle was solidly fused, further debridement and soft-tissue procedures became necessary at follow-up because of a small persisting fistula. One patient (# 7), stabilized by a locking nail, developed a delayed union. It eventually united 16 months postoperatively, 6 months after dynamisation of the nail. Seven patients with successful first-attempt salvage arthrodesis required hardware removal because of hinder from the material.

In seven patients, all suffering from IJD, a nonunion developed after the first-attempt salvage arthrodesis. Three of them refused further surgery, in two as they had a stiff and painfree fibrous nonunion. Four nonunions underwent further surgery. Their details are as follows:

One patient (# 9), underwent a salvage arthrodesis stabilized by screw fixation. Her hindfoot had already been fused prior to the TAA. Four months after the initial salvage, a re-arthrodesis with use of a blade plate had to be done for a nonunion, resulting in a solid fusion. However, some months after implant removal she developed a spontaneous talar neck fracture, for which a reoperation had to be carried out.

Case # 11 developed mechanical loosening four years after a 2-stage revision TAA for deep infection. The arthrodesis, stabilized by a retrograde nail, ended in a fibrous nonunion. For recurrent symptoms, a second attempt with a blade plate was done six years after the first salvage procedure. It was complicated by a wound dehiscence and an early deep infection, for which multiple debridements had to be carried out. Eventually, the ankle united after a third-attempt salvage with a compression locking nail.

Case # 12 developed mechanical loosening three years after her primary TAA. Her ipsilateral hip was ankylosed in a position of slight flexion and significant external rotation long before the primary TAA. An arthrodesis with use of a retrograde nail was done. After material extraction and debridement for an infected nonunion, a re-arthrodesis with use of an external fixator and autologous bone graft was done. Despite all efforts, a nonunion remained as end result. The ankylosed hip in this case probably has contributed to both the early mechanical loosening of the primary TAA and to nonunion of the salvage procedure. Total hip arthroplasty has been offered to this patient, which she repeatedly refused.

Case # 15 was a failed primary TAA due to a severe wound dehiscence with open joint. A 1-stage salvage arthrodesis, stabilized by a locking nail, was done two months after the index surgery. This resulted in a low-grade infected nonunion, for which a 2-stage rearthrodesis with use of an external fixator was carried out. The ankle united, but was complicated by a septic arthritis of the talonavicular joint, requiring subsequent surgery.

7.3.3 Clinical Outcome and Radiographic Alignment

The clinical outcome at latest follow-up could be collected in eleven patients (six patients had deceased and one elderly female RA patient with a solid ankle fusion, case # 15, was wheelchair bound due to generalized arthritic disease and judged herself not to be able to give a reliable subjective outcome). Mean interval since the salvage arthrodesis in this group was 7.8 years (range, 3.1 to 12.1 years). The mean AOFAS score was 62.4 (range, 38 to 89; 95% confidence interval, 53.6 to 71.1), and the mean overall FFI score was 70.1 (95% confidence interval, 61.9 to 78.3). The mean VAS for pain was 20.1 (95% confidence interval, 7.2 to 33.0), the mean VAS for function was

64.3 (95% confidence interval, 44.9 to 83.6), and the mean VAS for satisfaction was 73.8 (95% confidence interval, 61.1 to 86.5). The three rheumatoid ankles in follow-up with a persistent nonunion had subjective results similar to the fused ankles.

The radiographic alignment of the eleven ankles which healed after first-attempt salvage were as follows: the mean sagittal angle was 5.7 degrees of equinus (95% confidence interval, 0.6 to 10.8). Eight patients had a neutral alignment in the coronal plane (0 to 5 degrees valgus), two ankles had healed in slight varus and one ankle had healed in 15 degrees of valgus.

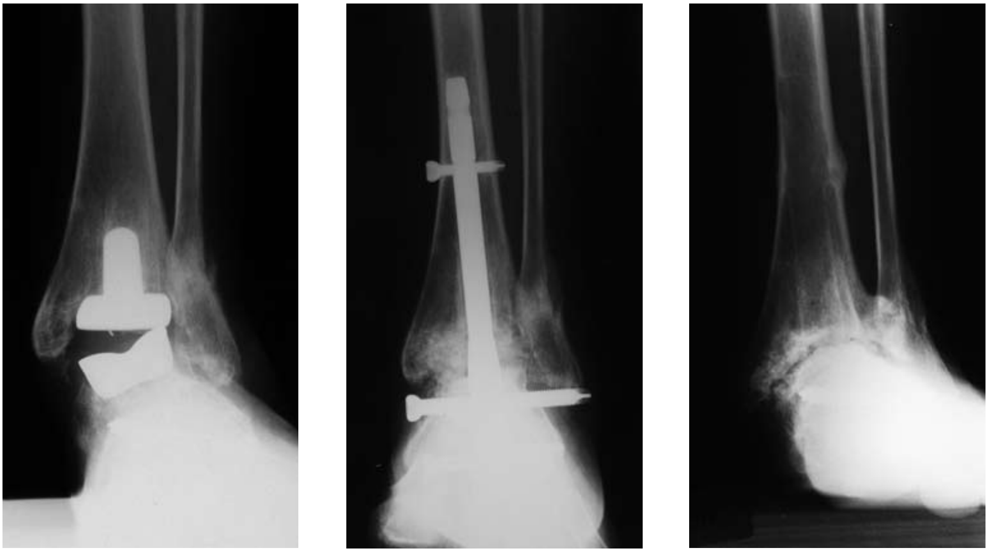

Fig. 2-A *Fig. 2-B* *Fig. 2-C*

Figs. 2-A through 2-C Radiographs of a woman with RA and a Buechel-Pappas prosthesis implanted at the age of 43 years (case 8). She had a preoperative valgus deformity of 10 degrees. ***Fig. 2-A*** Anteroposterior view three years postoperatively, when, due to a malleolar insufficiency fracture, a recurrent valgus deformity with edge-loading of the prosthesis had developed. ***Fig. 2-B*** A tibiotalocalcaneal arthrodesis was performed, stabilized by a retrograde locking nail. ***Fig. 2-C*** Two years after the salvage procedure the nail had been removed. Six years later, a fibrous nonunion exists. The ankle was fairly asymptomatic and did not require a second-attempt procedure.

TABLE 1 Demographic, perioperative and outcome data of all salvage procedures

Case	Gender, Age	Diagnosis‡	TAA¶	Failure Scenario	Interval since TAA (months)	Subtalar joint	Type of Fusion#	Fixation	Type of bone graft*	Time till Fusion (months)
1	F, 27	JIA	LCS	Loosening	161	Ankylosis	TTC	Screws	Allo	Nonunion
2	F, 56	RA	LCS	Instability	83	Arthritis	TT	Blade plate	AuP	3
3	M, 70	RA	LCS	Deformity	103	Normal	TT	Blade plate	AuL	4
4	F, 68	RA	LCS	Loosening	115	Fusion	TTC	K-wires	Allo	Nonunion
5	F, 68	RA	LCS	Deformity	25	Ankylosis	TTC	Screws	AuP	10
6	M, 55	RA	LCS	Deformity	51	Fusion	TTC	Blade plate	AuL	2
7	F, 64	RA	BP	Deformity	44	Fusion	TTC	Nail	AuL	16
8	F, 47	RA	BP	Postop. malleolar fracture	42	Fusion	TTC	Nail	AuFH (deep frozen)	Nonunion
9	F, 73	RA	BP	Instability	20	Fusion	TTC	Screws	AuP	Nonunion
10	F, 42	RA	BP	Infection	4	Arthritis	TT	Screws	AuL	3
11	M, 55	RA	BP	Loosening		Fusion	TTC	Nail	AuP	Nonunion
12	F, 76	RA	BP	Loosening	50	Arthritis	TTC	Nail	AuP	Nonunion
13	F, 47	RA	BP	Loosening	31	Ankylosis	TTC	Nail	AuL	9
14	M, 60	OA	BP	Pain	61	Normal	TT	Blade plate	AuL	2
15	F, 76	RA	BP	Wound dehiscence	2	Arthritis	TTC	Nail	None	Nonunion
16	F, 50	OA	BP	Pain	39	Normal	TT	Blade plate	AuL	4
17	F, 46	NsA	BP	Wound dehiscence	6	Normal	TT	Blade plate	AuP	4
18	F, 61	OA	CCI	Intraop. malleolar fracture	3	Normal	TT	Blade plate	Allo + AuL	10

‡ JIA = Juvenile Idiopathic Arthritis; RA = Rheumatoid Arthritis; NsO = Nonspecific Oligoarthritis; OA = Osteoarthritis.

¶ LCS = New Jersey Low Contact Stress; BP = Buechel-Pappas; CCI = Ceramic Coated Implant.

TT = Tibiotalar; TTC = Tibiotalocalcaneal.

*Allo = Allograft; AuL = autologous bone graft harvested locally; AuP = autologous bone graft from pelvis; AuFH = autologous bone graft from femoral head.

7.4 Discussion

Salvage arthrodesis should be a reliable treatment option if TAA fails and revision by implant exchange is impossible due to bone loss, deformity, or infection. Several reports show that salvage arthrodesis for failed ankle replacement has a mean fusion rate of 94 per cent (range, 74 to 100 per cent)[5,6,7,8,9,10,13]. These results are comparable to the success rate of primary ankle arthrodesis. Haddad et al.[14], in a meta-analysis, described a 90 per cent union rate after primary ankle arthrodesis. In general, the success rate of primary ankle arthrodesis in RA is less good. Dereymaker et al. had a 36 percent nonunion rate in a series of fourteen ankles[15]. Anderson et al. had a 26 per cent nonunion rate in ankles stabilized by screw fixation[16]. Better results were published from the same institution when a retrograde nail had been used: one nonunion out of 26 tibiotalocalcaneal fusions[17]. The largest series of ankle fusions in RA, performed through a transfibular approach, was published by Mäenpää et al. from the Rheumatism Foundation Hospital in Finland[18]. In their series of 130 ankles a 90 per cent union rate was found. They concluded that ankle arthrodesis in RA is a demanding procedure, and that the operation should be performed by an experienced surgeon, and furthermore that correction of malalignment and the use of bone grafts are of crucial importance for fusion.

In our series, primary nonunion occurred only in the rheumatoid ankles. This underlines the fact that it is apparently more difficult to obtain solid union of both primary and salvage ankle arthrodesis in these patients. Fortunately, three out of four second-attempt procedures were successful, and of the three non-reoperated patients two developed a stable fibrous nonunion. Subjective outcome after salvage arthrodesis was relatively good, with fair to good FFI scores, and mostly low VAS pain and good VAS satisfaction scores.

In our hands, blade plate fixation for tibiotalar arthrodesis was the most successful technique, although, with the numbers available, no statistically significant differences could be found in comparison with more widely used methods. The advantage of blade plate fixation is that a very stable fixation can be obtained and that no hardware is present inside the arthrodesis site. Good results with blade plate fixation for tibiocalcaneal and tibiotalocalcaneal arthrodesis have been published[19,20]. The advantage of blade plate fixation is furthermore supported by the biomechanical testing study by Chiodo et al.[21]. They found a greater stability of tibiotalocalcaneal arthrodesis stabilized by a blade-plate-and-screw construct in comparison with a retrograde intramedullary locking nail. As far as we know, no results have yet been published on blade plate fixation for tibiotalar arthrodesis.

In conclusion, in osteoarthritis the union rate of salvage ankle arthrodesis is good, and comparable to the outcome of primary ankle arthrodesis. In rheumatoid ankles, both primary arthrodesis and salvage arthrodesis are demanding procedures, and should probably best be done by experienced surgeons in specialized centers. Stabilization by a blade plate appears to be a promising technique for salvage ankle arthrodesis.

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Chapter 8

Gait Analysis after Successful Mobile-Bearing Total Ankle Replacement

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Abstract

Background The effect of total ankle replacement on gait is not fully known in terms of joint kinematics, ground reaction force, and activity of the muscles of the lower leg.

Methods A comparative gait study was done in 10 patients after uneventful unilateral mobile-bearing total ankle replacement and 10 healthy controls. A rigid body model was used to describe the motion of the knee and the three-dimensional motion of the ankle-hindfoot complex during barefoot walking. An opto-electronic motion analysis system was used to analyze bilateral movement patterns, synchronized with recordings of the ipsilateral vertical ground reaction forces and the electromyographic activity of four lower leg muscles.

Results Velocity was 6 per cent lower in the patient group. Dorsiflexion in the operated ankles was reduced ($p < 0.001$). No differences were found in the joint angular pattern of the knee and only minimal changes were found at the hindfoot-to-tibia and forefoot-to-hindfoot levels. The ground reaction force at midstance was somewhat increased ($p = 0.05$) while the magnitude of the vertical peak at terminal stance was decreased ($p < 0.001$). EMG activity patterns in the patient group were normal except for a higher activity of the gastrocnemius in early stance and the anterior tibial muscle in late stance.

Conclusions There is a near normal gait pattern in terms of joint kinematics of the knee, ankle, and foot after successful mobile-bearing total ankle replacement. The ground reaction forces and the EMG activity, however, do not fully normalize.

8.1 Introduction

The standard operative treatment for a severely painful and diseased ankle joint is arthrodesis. In general, at the expense of loss of motion, an arthrodesis produces a painless and stable ankle[1]. Clinical disadvantages of an arthrodesis are the risks for nonunion and malunion[13], and the long period of immobilization and non-weight bearing. With longer followup, there is an increased risk of developing osteoarthritis in the joints of the ipsilateral foot[2,3,4] and possible injury to the knee due to an increased extension movement at this joint[5,6]. For a comfortable gait with an ankylosed ankle an altered motion of the ipsilateral knee and midtarsal joints is required[5,6]. Beyaert et al.[7] described early heel-off after ankle arthrodesis during barefoot walking. This heel-off shift improved when walking with shoes. They also found a reduced walking speed. Wu et al.[8] described a three-dimensional ipsilateral gait pattern during barefoot walking in 10 patients with a unilateral arthrodesis and compared this to a control group. During stance they noted a shift to external rotation and eversion at the hindfoot-to-tibia level and a shift to plantarflexion, eversion and adduction at the forefoot-to-hindfoot level. They also noted an abnormal electromyographic (EMG) pattern of the soleus muscle. Furthermore, after ankle arthrodesis altered ground reaction force patterns have been demonstrated: a more posterior position of the ground reaction force with reference to the metatarsal heads[7] and a lower second vertical peak force[8]. These alterations of foot dynamics are thought to be associated with pathogenic stresses to the midtarsal joints[7].

Total ankle replacement using a mobile-bearing prosthesis is a valid treatment option for the severely affected joint and an alternative to arthrodesis[9,10,11,12,13,14]. Near-normal kinematics of the ankle joint after implantation of a mobile-bearing prosthesis were found in cadaver experiments[15,16,17]. Using fluoroscopy, Komistek et al.[18] reported normal motion in dorsiflexion and an increased translation and rotation at the ankle joint in plantarflexion using fluoroscopy under weight-bearing conditions in 10 patients after successful unilateral Buechel-Pappas total ankle replacement. In a pilot study, near-normal gait and EMG-patterns were found in patients after successful total ankle replacement[19]. A normal kinematic pattern at the tarsal joints after total ankle replacement would reduce the risk of secondary arthritic changes to these joints.

We postulated that gait after successful total ankle replacement would be either equal to or closely equivalent to normal during level walking and that there would be no disturbed kinematic pattern of the joints of the foot and knee and of the function of the lower leg muscles.

8

8.2 Materials and Methods

8.2.1 Patient Demographics and Implant Characteristics

A laboratory gait evaluation was done on a group of 10 patients, six men and four women, after unilateral total ankle replacement using the Buechel-Pappas™ prosthesis (Endotec, Orange, NJ) and a group of 10 healthy control subjects, matched for age and gender. The study was approved by the hospital review board and patients gave informed consent before data collection. The mean age of the patients was 59.8 (range 39 to 77) years, and of control subjects 59.0 (range 37 to 78) years. Demographic data are specified in Table 1. The patient group consisted of a rheumatoid arthritis subgroup (TAR-RA, one men and four women) and an osteoarthritis subgroup (TAR-OA, five men). None of these individuals had any major orthopaedic surgery, e.g. joint replacement or fusion besides the total ankle replacement, and non had major disturbance in the lower limb joints. The mean time from surgery was 41.5 (range of 11 to 126) months. Average duration of symptoms before surgery was 6.3 (range 1 to 22) years.

The Buechel-Pappas™ prosthesis is a fully congruent, unconstrained three-component prosthesis. The tibial and talar components are made of a titanium alloy and are titanium nitride-coated; the mobile meniscal bearings are made of ultrahigh molecular weight polyethylene (UMWPE). For bony ingrowth, a porous coating is applied to the metal components[20].

TABLE 1 Demographic data of the study group and subgroups

	Age (yr) [a]	Gender [b]	Operated side [c]	Time since surgery [d]	LCS / AOFAS score [e]
TAR (n=10)	59,8 (12,6)	6 M/ 4 F	5 R/ 5 L	41,5 (11-126)	93.5 / 92.2
TAR-RA (n=5)	58,8 (15,2)	1 M/ 4 F	2 R/ 3 L	46,6 (11-126)	91.6 / 91
TAR-OA (n=5)	60,8 (11,1)	5 M	3 R/ 2 L	36,4 (13-99)	95.4 / 93.4
Controls (n=10)	59,0 (12,1)	6 M/ 4 F	-	-	-

[a] mean, SD in parentheses; [b] M = male, F = female; [c] R = right, L = left; [d] mean, range in months in parentheses; [e] mean.

We used the following inclusion criteria for this gait analysis study: patients with a unilateral total ankle replacement, a mobile hindfoot with normal motion, a minimum interval since surgery of 10 months and a minimum ankle score of 80 points on both the Low Contact Stress (LCS)[21] and the American Orthopaedic Foot and Ankle Society (AOFAS) [15,22] scoring systems. These two scoring systems use somewhat similar items for pain, function, range of motion and deformity to a maximum of 100 points. Patients with a total ankle replacement were excluded when (1) the individual used walking aids, (2) was unable to walk more than 1 km, (3) had a functional impairment of any other lower extremity joint besides the operated ankle, or (4) a history of any joint replacement procedure other then the ankle because this might interfere with a normal gait pattern. These exclusion criteria were formulated because one of the main objectives of this study was to evaluate the motion patterns in the foot after total ankle replacement in otherwise well-functioning patients and so to exclude bias from comorbidity or residual symptoms of stiffness or pain in the ankle or hindfoot. Before 2000, our main indication for total ankle replacement had been in patients with a severely affected ankle joint due to rheumatoid arthritis. Only in recent years we have performed total ankle replacement more frequently in patients with primary or posttraumatic osteoarthritis. That only 10 patients could be included for this study can be explained by the fact that most patients with rheumatoid arthritis and total ankle replacement had one or more exclusion criteria.

8.2.2 Data Acquisition and Testing Procedures

Before testing, the subjects were informed about the aims and procedure of the experiment, after which they gave informed consent. First the LCS and AOFAS scores were assessed and ankle range of motion in the sagittal plane, subtalar joint motion in the frontal plane, and motion of the midtarsal joint in the transversal plane were measured by one of the authors (HCD) using manual goniometry. The gait analysis included the simultaneous recordings of body kinematics, muscle activation and ground reaction force in subjects walking at their self-selected speed.

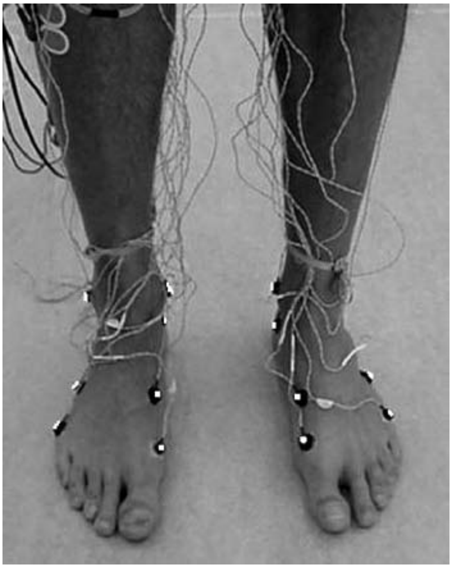

Fig. 1-A

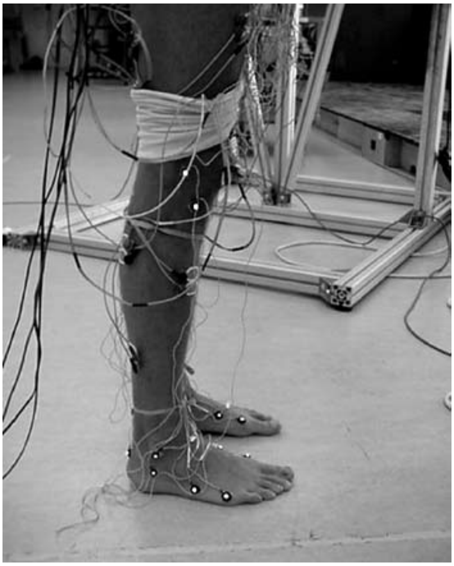

Fig. 1-B

Figs. 1-A and 1-B Photographs of a subject with LEDs and EMG surface electrodes on the lower leg and foot. ***Fig. 1-A*** Lateral view. ***Fig. 1-B*** Anterior view.

To analyze movement patterns a four-unit optoelectronic motion analysis system (Optotrak, Northern Digital Inc., Canada) was used. Each unit consisted of three cameras. Calibration of the camera position with use of a calibration grid was done before every test procedure. Twenty-eight small active light emitting diodes (LEDs) were placed on the subject's affected and unaffected hip, knee, ankle and foot (Figure 1 and Table 2). Marker placement was done according to the combined protocols of Kadaba et al.[23] and Wu et al[8]. In order to reduce variations in marker placement set up of the skin markers was carried out by one investigator (MVM.). Data were collected at a sampling rate of 100 Hz. A Kistler force plate (Type 9281A11, Kistler Instrument Corp., Winterthur, Switzerland) and an EMG system (Porti-17™, TMS Enschede, The Netherlands; input impedance 10^{12} Ω, Common-Mode-Rejection-Ratio > 90dB) were synchronized with the Optotrak system to measure ground reaction forces (GRF) and phasic activities of certain muscles of the lower extremities. The Kistler force plate used had dimensions of 0.60 x 0.40 m. Data from the force plate were sampled at a sampling rate of 100 Hz. The vertical ground reaction force was recorded for the operated side and on the right side for the control subjects. Surface EMG was sampled at 1000 Hz. Electrode placement (Medicotest, Blue Sensor, type N-00-S) for the different muscles in this study was done as described by Freriks and Hermens[24]. The EMG recordings were synchronized with the Optotrak computer by means of a pulse. Activity of the medial and lateral gastrocnemius, soleus and anterior tibial muscles was measured. Before placing the EMG surface electrodes, the skin was shaved and cleaned with alcohol to reduce skin resistance. Muscle activity of the ipsilateral side was measured. The operated side of patients and the right side of control subjects were considered the ipsilateral side. The unaffected side of patients and the left side of control subjects were considered the contralateral side.

After setup of markers for movement recording and electrode placement for surface EMG, subjects walked barefoot on a 8-m flat walkway, in which the force plate was incorporated. Before data collection started, subjects first walked on the walkway to become familiar with the surroundings. A minimum of 10 runs of gait data were collected for each subject walking at his/her self-selected speed, although patients were encouraged to walk at a speed of about 1.1 m/s to minimize velocity effects.

Table 2: Marker placement and anatomical frame definitions

Segment	Anatomical landmarks	Local coordinate system
Thigh	greater trochanter (GT)	$O = \frac{ME + LE}{2}$
	lateral epicondyle (LE)	$\vec{x} = \frac{ME - LE}{\|ME - LE\|}$
	medial epicondyle (ME)	$\vec{y} = \frac{GT - O}{\|GT - O\|} \times \vec{x}$
		$\vec{z} = \vec{x} \times \vec{y}$
Lower leg	Most lateral point on the ridge of the tibia plateau (LMT)	$O = \frac{LM + MyM}{2}$
	Most medial point on the ridge of the tibia plateau (MMT)	$\vec{x} = \frac{MM - LM}{\|MM - LM\|}$
	lateral maleolus (LM)	$\vec{y} = \frac{\frac{LMT + MMT}{2} - O}{\left\|\frac{LMT + MMT}{2} - O\right\|} \times \vec{x}$
	medial maleolus (MM)	$\vec{z} = \vec{x} \times \vec{y}$
Hindfoot	lateral side calcaneus (LC)	$O = \frac{LC + MC}{2}$
	medial side calcaneus (MC)	$\vec{z} = \frac{O - CP}{\|O - CP\|}$
	posterior side calcaneus (PC)	$\vec{y} = \vec{z} \times \frac{MC - LC}{\|MC - LC\|}$
		$\vec{x} = \vec{y} \times \vec{z}$
Forefoot	proximal head of 1th metatarsal (PM1)	$O = \frac{DM1 + DM5}{2}$
	proximal head of 5th metatarsal (PM5)	$\vec{z} = \frac{O - \frac{PM1 + PM5}{2}}{\left\|O - \frac{PM1 + PM5}{2}\right\|}$
	distal head of 1th metatarsal (DM1)	$\vec{y} = \vec{z} \times \frac{DM5 - DM1}{\|DM5 - DM1\|}$
	distal head of 5th metatarsal (DM5)	$\vec{x} = \vec{y} \times \vec{z}$

8.2.3 Data Analysis

A four-segment rigid body model was used to describe the motion of the knee, hindfoot-to-tibia and forefoot-to-hindfoot. The thigh included the femur and the lower leg the tibia and fibula. The hindfoot included the calcaneus and the forefoot the metatarsals. Motion of the four segments was expressed with respect to the neighboring proximal segment.

The joint angles were obtained from the relative orientations of the local coordinate systems (Table 2). Joint rotation matrices were expressed in joint angles following the standard decomposition order: dorsiflexion (extension)–(plantar)flexion, abduction–adduction, and pronation–supination. The anatomical zero for the hindfoot-to-tibia and forefoot-to-hindfoot joint angles were defined to occur at 15% of the gait cycle. This method was chosen as at 15% of the gait cycle full stance has been reached and the ankle and foot are known to be loaded in a vertical position[7]. The functional ranges of movement in degrees during walking of the knee, hindfoot-to-tibia and forefoot-to-hindfoot were calculated by taking the difference between the maximum and minimum joint angle during the gait cycle.

Vertical force plate data were presented by a percentage of the stance phase during the gait cycle and the amplitude was expressed as a percentage of the body-weight. Peak values of the vertical component of the ground reaction force (F1, F2 and F3) and the associated time of occurrences of the peaks (T1, T2 and T3) were represented according to Wu et al[8].

Joint kinematics, vertical ground reaction forces, and EMG-data were assembled and averaged for three trials around the median speed for each subject, leading to one average cycle for each subject. The EMG-data were processed according to the following steps: removal of artifacts (fourth order high pass Butterworth filter, 20 Hz), rectifying, low pass Butterworth filtering (2 Hz, time shift 80-100 ms) and normalization where timing was represented by percent of gait cycle and amplitude was a percentage of each subjects' maximum muscle activity during walking. EMG data were thus graphed as linear envelopes.

8.2.4 Statistical Analysis

Statistical analyses were done using a General Linear Model ANOVA for patients versus controls or for patient subgroups versus controls. When three groups were compared a Tukey post hoc test was used. In the analyses of the spatiotemporal parameters ipsilateral versus contralateral leg were considered to be a within-subject factor. Differences were considered significant when $p < 0.05$.

8.3 Results

8.3.1 Analysis of the Range of Motion of the Ankle and Foot

Data of the range of motion measured by clinical examination are specified in Table 3. Ipsilateral dorsiflexion in the ankle joint for patients was reduced in comparison with the contralateral side ($p < 0.001$) and to the control group ($p < 0.001$). Between the subgroups, a difference was found in dorsiflexion. Ipsilateral ankle dorsiflexion in the TAR-RA subgroup was less reduced than in the TAR-OA subgroup compared to control subjects ($p < 0.01$ and $p < 0.001$, respectively). There were no significant differences in range of motion between the groups at the subtalar and midtarsal joints.

TABLE 3 Range of motion in degrees (mean, SD in parentheses) measured by manual goniometry							
	TAR ipsi	**TAR contra**	**TAR-RA ipsi**	**TAR-RA contra**	**TAR-OA ipsi**	**TAR-OA contra**	**Controls ipsi**
Ankle							
Dorsiflexion	15.0 (2.9)*+	23.1 (3.8)	16.2 (3.6)‡#	22.8 (5.2)	13.8 (1.3)*#	23.4 (2.1)	25.4 (3.6)
Plantarflexion	36.8 (9.5)	45.3 (7.2)	40.6 (11.1)	46.6 (9.8)	33.0 (6.7)	44.0 (4.2)	44.9 (7.7)
Subtalar							
Supination	26.1 (4.6)	29.3 (6.9)	28.0 (5.7)	31.6 (7.8)	24.2 (2.4)	27.0 (5.7)	28.9 (5.9)
Pronation	14.0 (7.4)	19.5 (8.0)	16.0 (8.9)	21.0 (7.4)	12.0 (5.7)	18.0 (9.1)	16.3 (6.2)
Midtarsal							
Abduction	15.4 (3.0)	16.5 (6.0)	15.4 (3.8)	18.6 (7.4)	15.4 (2.3)	14.4 (3.8)	17.7 (3.4)
Adduction	13.3 (4.8)	14.0 (4.6)	11.2 (4.4)	13.0 (4.5)	15.4 (4.6)	15.0 (5.0)	14.6 (2.4)

* significant difference from the control group ($p < 0.001$)
\+ significant difference ipsilateral from contralateral ($p < 0.001$)
‡ significant difference from the control group ($p < 0.01$)
\# significant difference ipsilateral from contralateral ($p < 0.01$)

8.3.2 Spatiotemporal Parameters

As indicated in Table 4, the overall patient group walked 6% slower compared to control subjects ($p = 0.02$). Looking at the subgroups, however, only the patients with osteoarthritis walked significantly slower compared to the controls ($p = 0.001$). The slower walking speed was primarily caused by an increased stride time and to a lesser extend by a decreased stride length. Patients walked with a significantly larger ipsilateral step length. The effect showed up as a significant contrast within the patient population, but the effect was not significantly different between controls and patients.

TABLE 4 Spatiotemporal parameters (mean, SD in parentheses)				
	TAR	**TAR-RA**	**TAR-OA**	**Controls**
Velocity (m/sec)	1.02*(0.07)	1.07 (0.06)	0.97† (0.04)	1.09 (0.07)
Stride length (m)	1.22 (0.07)	1.21 (0.08)	1.23 (0.07)	1.27 (0.10)
Stride time (sec)	1.20 (0.10)	1.14 (0.08)	1.26* (0.07)	1.15 (0.07)
Ipsilat. step length (m)	0.62+ (0.04)	0.62+ (0.02)	0.62+ (0.07)	0.64 (0.06)
Contralat. step length (m)	0.59 (0.03)	0.60 (0.03)	0.58 (0.03)	0.63 (0.06)

* significant difference from control group ($p = 0.02$)

† significant difference from control group ($p = 0.001$)

\+ significant difference ipsilateral from contralateral ($p < 0.05$)

8.3.3 Analysis of the Gait Pattern

The knee flexion–extension and hindfoot-to-tibia plantar-dorsiflexion patterns of the patients agree with the data of the control group (Figure 2, A and B), but during the stance phase the standard deviations were larger in the patient group in both the knee and at the hindfoot-to-tibia level.

Joint angular motion data of forefoot-to-hindfoot during walking are shown in Figure 2, C through E. The plantar–dorsiflexion pattern of the patient group was similar to the pattern of the control group, except that the standard deviation in the patient group was larger from 50% to 90% of the gait cycle. Transverse plane motion demonstrated a slightly decreased adduction at early swing for the patient group in comparison with the control subjects in an otherwise similar abduction-adduction joint angle pattern. Coronal plane motion, supination–pronation, of the patients was similar to the supination–pronation pattern of the control subjects. The only difference was the standard deviation, which was larger in the patient group.

No differences were found between the overall patient group and the control group in functional angular range of the knee, hindfoot-to-tibia and forefoot-to-hindfoot (Table 5). Also, no differences were found between the subgroups and the control group.

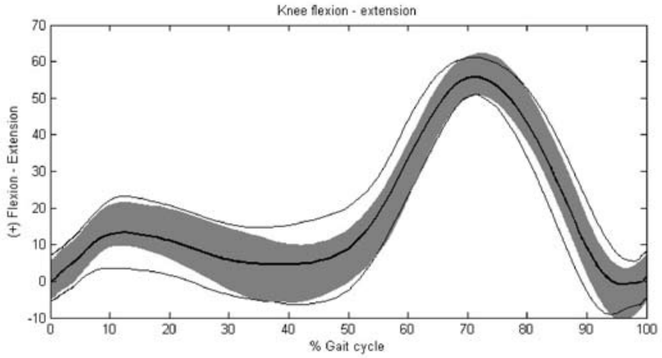

Fig. 2-A

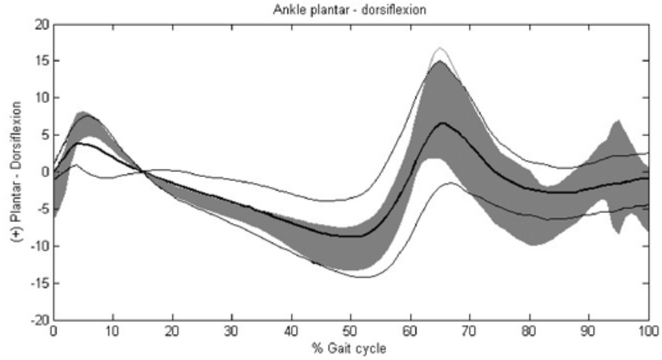

Fig. 2-B

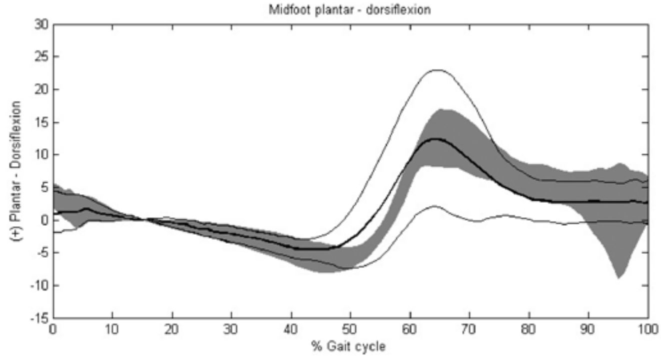

Fig. 2-C

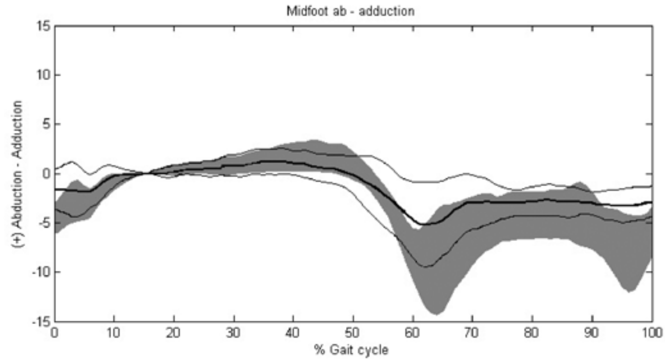

Fig. 2-D

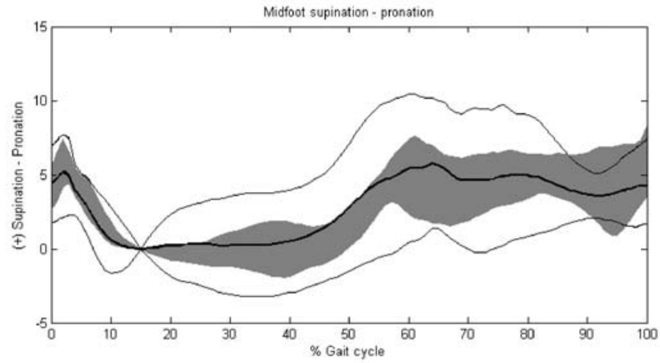

Fig. 2-E

Figs. 2-A through 2-E Joint angular motion patterns. For TAR patients, mean (thick black line) and standard deviation (thin black line). Mean plus and minus standard deviation for control subjects is represented by the gray shaded area. Normalization to 0 degrees was done at 15% of gait cycle for the hindfoot-to-tibia and forefoot-to-hindfoot.. ***Fig. 2-A*** Knee angular motion. ***Fig. 2-B*** Hindfoot-to-tibia angular motion. ***Fig. 2-C through E*** Forefoot-to-hindfoot angular motion.

TABLE 5 Ipsilateral functional angular range in degrees during walking (mean, SD in parentheses)						
	Knee	**Hindfoot-to-Tibia**	**Forefoot-to-Hindfoot**		**Forefoot-to-Tibia**	
	Flexion-Extension	Plantar-Dorsiflexion	Plantar-Dorsiflexion	Ab-Adduction	Plantar-Dorsiflexion	Pronation-Supination
TAR	62,2 (3,8)	21,0 (4,5)	20,7 (7,3)	11,1 (4,0)	36,4 (11,6)	11,0 (6,2)
TAR-RA	62,7 (3,7)	23,1 (5,9)	19,5 (9,9)	9,8 (4,1)	37,2 (16,6)	8,8 (1,8)
TAR-OA	61,8 (4,2)	19,3 (2,7)	21,7 (5,5)	12,2 (4,0)	35,5 (5,0)	12,7 (8,1)
Controls	61,0 (4,7)	23,5 (3,0)	20,3 (5,6)	14,7 (3,4)	42,3 (8,8)	10,3 (3,1)
TAR = Total ankle replacement; RA = rheumatoid arthritis; OA = osteoarthritis						

8.3.4 Analysis of the Vertical Ground Reaction Force

These data are presented in Table 6 and Figure 3. The midstance vertical valley force, F2, was higher and the second vertical peak force, F3, lower after total ankle replacement compared with values for normal subjects (p = 0.05 and < 0.001, respectively). Post hoc analysis revealed that in both the TAR-RA and TAR-OA patients the second vertical peak force, F3, was significantly lower than for the control group. Temporal force factors, T1, T2 and T3, were, both in the overall patient group and in the subgroups, not significantly different from normal subjects.

TABLE 6 Vertical ground reaction forces F1 through F3 as a percentage of body weight and temporal force factors T1through T3 as a percentage of stance phase (mean, SD in parentheses)						
	F1	**F2**	**F3**	**T1**	**T2**	**T3**
TAR-affected side	105 (7,4)	89 (4,2)*	108 (3,5)*	25 (2,8)	46 (2,2)	76 (2,5)
TAR-RA-affected side	105 (10,2)	89 (4,1)	107 (3,4)*	26 (3,3)	46 (2,3)	76 (2,9)
TAR-OA-affected side	104 (4,1)	89 (4,8)	109 (3,8)*	25 (2,4)	47 (2,2)	76 (2,5)
Controls	111 (8,8)	84 (6,4)	116 (5,4)	23 (3,2)	46 (2,9)	77 (3,0)
* significantly different from control (p < 0.05)						

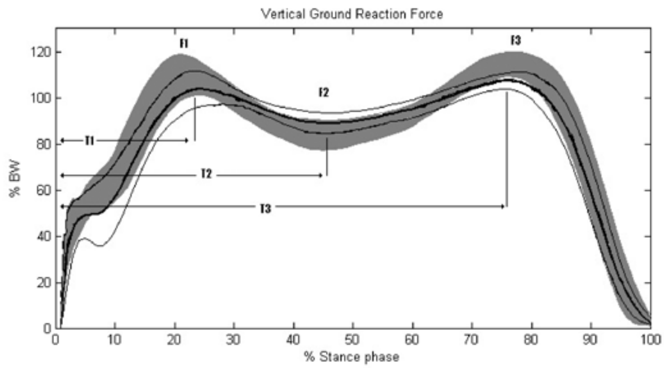

Fig. 3

Fig. 3 Vertical ground reaction force for TAR patients, mean (thick black line) and standard deviation (thin black line). Mean plus and minus standard deviation for control subjects is represented by the gray shaded area.

8.3.5 Analysis of the EMG Linear Envelopes

Figure 4 A through D show the EMG recordings on the ipsilateral side of the patient and the control group. The medial gastrocnemius was more active at early stance, and during terminal stance activity in the anterior tibial was greater in the patient group. As for the control group, there was a driving action of the medial and lateral gastrocnemius and soleus at about 40-60% of the gait cycle. The soleus and the lateral gastrocnemius were more active during the swing phase. The timing of the two peaks of the anterior tibial muscle, at 10% and 80% of the gait cycle, was similar to that of the control group. The magnitude of the second peak in the activity of the anterior tibial muscle was identical for both groups and indicated toe clearance during midswing.

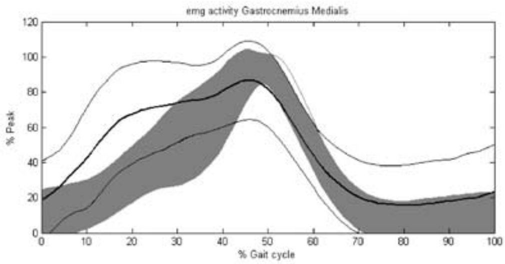

Fig. 4-A

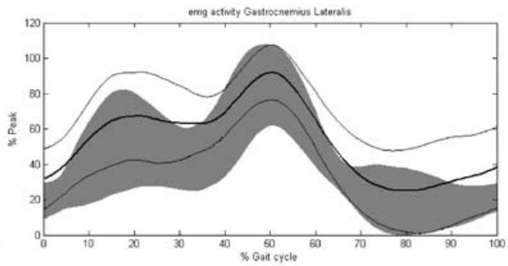

Fig. 4-B

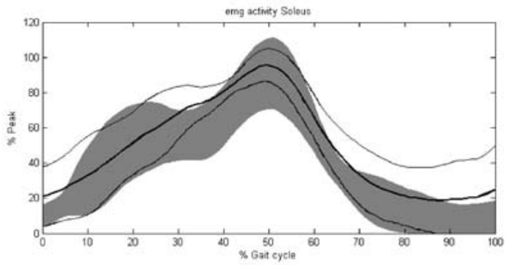

Fig. 4-C

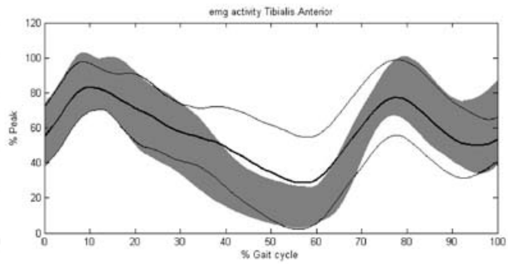

Fig. 4-D

Fig. 4-A through 4-D EMG envelopes for four lower leg muscles. ***Fig. 4-A*** Medial Gastrocnemius, ***Fig. 4-B*** Lateral Gastrocnemius, ***Fig. 4-C*** Soleus, and ***Fig. 4-D*** Anterior Tibial; mean (thick black line) and standard deviation (thin black line). Mean plus and minus standard deviation for control subjects is represented by the gray shaded area.

8.4 Discussion

The aim of this study was to compare the gait characteristics of a patient group with a good outcome after total ankle replacement with the gait characteristics in age- and sex-matched healthy controls and to discern whether the gait pattern in the patient group would show characteristics that might in the long-term lead to primary or secondary overuse injuries. Limitations in hindfoot-to-tibia, or forefoot-to-hindfoot functional ranges of motion might be a causative factor for loosening or ligament damage, while deviations in muscle forces might indicate higher joint contact forces and concomitant higher wear, or a higher energy expenditure.

In our study, differences were found in the range of motion of the ankle joint on physical examination. The patients had ipsilateral reduced dorsiflexion in the ankle compared to the control group. This was found for both the rheumatoid arthritis and osteoarthritis subgroups. Nevertheless, the reduced dorsiflexion had no effect on the functional angular range of the hindfoot-to-tibia motion found during walking. A similar angular motion pattern of hindfoot-to-tibia plantar-dorsiflexion during walking was found in patients and in controls. This means that, despite the reduced range of motion in the replaced ankle, during walking the functional angular range at this level was not affected. Furthermore, no differences between patients and controls were found in the range of motion of the subtalar and midtarsal joints and to a large extent in the average functional angular range during walking at the forefoot-to-hindfoot level in the sagittal, transversal and coronal planes. As most patients had symptoms for many years before surgery, it is likely that the increased standard deviation at the forefoot-to-hindfoot level existed already at the time of surgery. However, this question can only be evaluated in a prospective study comparing gait before and after total ankle replacement.

It has been shown that there is an increased risk of secondary osteoarthritic changes to the tarsal joints after ankle arthrodesis[2,3]. It is likely that the altered motion in the tarsal joints after ankle arthrodesis[8] is an important factor in the development of these osteoarthritic changes. We found that the function of the joints of the ipsilateral foot was minimally affected after total ankle replacement, which suggests that there might be a reduced risk of secondary overuse injuries to these joints after total ankle replacement in comparison with arthrodesis.

To obtain three-dimensional gait data we used an optoelectronic system with skin markers. It is well known that this procedure can introduce errors hampering the interpretation of results. A recent review[25] indicated the intra-examiner precision that can be obtained for the definition of the anatomical axis of the foot to be around 3 degrees (and inter-examiner precision between 5 and 10 degrees), depending on the defined axis. The standard deviation for the motion at the forefoot-to-hindfoot level was about 4 degrees, which will lead to a minimal detectable difference of 5 degrees at an 80% power level. It is difficult to state which difference is clinically relevant. A gait study after ankle arthrodesis showed that the difference in forefoot-to-hindfoot motion between patients and controls exceeded 5 degrees8. In this study the differences were less than 5 degrees and, thus, not significant. Whether this also implies a clinical insignificance remains to be seen. In this study, local anatomical axes were defined as having the same orientation at 15% of the gait cycle. This choice was based on the assumption that in this phase, the tibia is vertical. This assumption is fully supported by the study by Beyaert et al[7]. They showed that the tibia is vertical at 25% of the stance phase for healthy controls and that the stance phase is 61% of the gait cycle. The alternative of using the standing position at rest for normalization could also bring bias due to inter-individual differences.

The skin markers, surface electrodes and cables could theoretically cause some disturbance during the test runs. As ample time was given to become familiar with the study setup and several trial runs were done before the actual data collection started we feel that these factors are of little if any consequence.

Our patient group walked 6% slower than the controls. This might be related to the ankle replacement, although confounding effects can not be excluded. Despite the differences in velocity between groups, there was no difference in ipsilateral step length between the two groups. Since step length is an important factor related to joint kinematics, we conclude that the confounding effect of differences in gait velocity is negligible. Because of the small numbers in the subgroups no real comparison could be made between the osteoarthritic and rheumatoid ankles.

In our study, the midstance vertical valley force, F2, was higher and the subsequent vertical peak force, F3, lower after total ankle replacement compared to control subjects. Near heel-off, the vertical force peaks dropped resulting in a lower propelling force to help the foot leave the ground. Zerahn and Kofoed[26] showed that the vertical ground reaction force improved after successful total ankle replacement. However, they made no comparison with a normal control group. Therefore, it is likely that full restoration of the ground reaction forces cannot be expected after total ankle replacement. Differences in vertical ground reaction force were also found in patients after ankle arthrodesis[8]. These differences, both in patients with replaced ankles and with ankle arthrodesis, could be caused by the lower walking speed[27].

The EMG patterns of the four lower leg muscles of our patients were mostly the same as the control group and as published normative data[28], but there were some differences. In the patient group, the ipsilateral medial and lateral gastrocnemius muscles were somewhat more active during midstance (10%-30% of the gait cycle) and, together with the soleus muscle, were also more active during the swing phase. The anterior tibial muscle was slightly more active during terminal stance (40%-60% of the gait cycle). It appears as if there was more co-contraction in patients than in controls. In comparing our results to those after ankle arthrodesis from the study by Wu et al.[8] it should be kept in mind that due to differences in data processing there is a time shift of approximately 10% of the gait cycle between these and our results. In terms of activity however, they found a significantly abnormal soleus EMG amplitude after ankle arthrodesis, since activity of the soleus muscle was substantially reduced during stance and more active at terminal stance and early swing. Although the activity of the four most important muscles in the lower leg after total ankle replacement is not completely normal, the activity pattern after total ankle replacement is much better than after ankle arthrodesis. No differences were found between the TAR-RA and TAR-OA groups in terms of joint kinematics, vertical ground reaction force and muscle activity. This means that any possible comorbidity in the rheumatoid arthritis patients did not influence the gait pattern.

No full gait laboratory study with detailed three-dimensional kinematics has been published after successful total ankle replacement, but there are some studies describing certain aspects of gait after this procedure. Near-normal kinematics of the ankle joint after implantation of a mobile-bearing prosthesis were found in cadaver experiments[15,16,17]. Normal motion in dorsiflexion and an increased translation and rotation at the ankle joint in plantarflexion using fluoroscopy under weight-bearing conditions has been described in 10 patients after successful unilateral Buechel-Pappas total ankle replacement[18]. In a recent study, mainly focusing on kinetic aspects after total ankle replacement, walking velocity improved but did not normalize, and, furthermore, there was a significant kinetic improvement[29].

In summary, a near-normal gait pattern in terms of kinematics of the knee, ankle, and tarsal joints during level walking can be achieved after successful total ankle replacement despite a somewhat reduced dorsiflexion in the ankle joint. This might imply that there is a reduced risk of secondary overuse injuries to the tarsal joints after this procedure. As ground reaction forces and EMG activity of the lower leg muscles are somewhat different from normal, a complete normalization of gait could not be found.

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Chapter 9

Joint Stiffness of the Ankle during Walking after Successful Mobile-Bearing Total Ankle Replacement

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Abstract

Introduction It has been shown that walking kinematics remain near to normal after mobile bearing total ankle replacement (TAR). However, no information is available on mechanical joint loading. The purpose of this study was to determine whether mechanical load and 'quasi-stiffness' of the ankle joint after TAR differs from the normal load and stiffness of a healthy ankle joint during walking.

Methods Ten TAR patients and 10 age-matched healthy control subjects (CO) participated in this study. Participants walked barefooted on an indoor track of approximately 8 meters at self-selected walking speed. 3D ankle kinematics and ground reaction forces were measured and used for the calculation of 3D net joint moments, joint quasi-stiffness coefficients and internal net joint ankle work..

Results Between patients and control subjects no differences were observed in peak moments and stiffness coefficient at the ankle. Internal work at the ankle during the step differed however, significantly (-0.078 (0.088) vs. 0.005 (0.048) $J \cdot kg^{-1}$ for TAR vs. CO, p=0.02), although it could be argued that this difference was due to a minor difference in walking speed between both groups (1.02 (0.07) vs. 1.09 (0.07) for TAR vs. CO, p=0.02).

Conclusion Despite the small difference in internal work at the ankle, it could not be concluded that the mechanical loading of the ankle after TAR differed from normal.

Keywords: *Ankle; Arthroplasty; Gait; Joint Stiffness; Joint Moment*

9.1 Introduction

For painful arthritic ankle joints a mobile-bearing total ankle replacement (TAR) is a treatment option. In contrast to its alternative, an ankle arthrodesis, TAR preserves joint motion. This might prevent secondary overuse in the adjacent joints of the foot, which show compensatory motion in the case of an arthrodesis[1]. Literature on the success of mobile-bearing TAR describes varying, though generally good results[2-6]. In a recent study[7] we described the 3D kinematic pattern in ten patients after TAR and compared this to a control group. Although patients had a reduced range of motion in the ankle during clinical mobility testing, no effect on the functional range of motion during walking was found. The EMG patterns, however, showed more co-contraction of the anterior tibial muscle and gastrocnemius and soleus muscles during midstance in the patient group. Muscle co-contraction increases the internal joint reaction forces and the mechanical stress on the endo-prosthesis. This might indicate that, despite the kinematic similarities, the mechanical behavior of the replaced ankle is different compared to the healthy ankle.

The mechanical load of body joints can be analyzed in terms of net joint moments. However, net joint moments are unsuitable for the estimation of the effect of co-contraction on the internal joint loading. The net joint moment remains unchanged irrespective of the degree of co-contraction. An alternative procedure for analyzing the mechanical behavior of the joint, including the effect of co-contraction is to calculate the dynamic joint stiffness[8]. Dynamic joint stiffness can be defined as the resistance that a joint (i.e. the muscles and other soft tissue structures that cross the joint) offers during gait in response to an applied angular displacement[8]. It should be noted however, as Latash and Zatsiorsky[9] argue, that joint stiffness does not necessarily imply that the tissue around the joint behaves as a physical spring, storing and releasing energy. Latash and Zatsiorsky[9] therefore define this behavior of the joint as 'quasi-stiffness'. Joint quasi-stiffness can be an important determinant of stress generation in the joint, since joint quasi-stiffness depends heavily on and is therefore indicative of the degree of muscle co-contraction[9].

A relatively simple method for estimating joint quasi-stiffness is to examine the relation between the net moment at the ankle joint versus the angular displacement of the ankle[8]. This quantity (the quasi-stiffness coefficient) can reveal the overall mechanical system behavior of the joint (and hence joint stress) during the stance phase of walking, as was demonstrated previously[8, 10-12].

So far the ankle joint kinetics and stiffness in a 3D model in patients after TAR have not been examined. Dyrby et al.[13] investigated 2D net joint moments in nine patients using the STAR prosthesis but did not look at possible co-contraction and joint quasi-stiffness. The goal of the present study was to determine whether quasi-stiffness and mechanical load of the ankle joint after arthroplasty differs from the normal stiffness and load of a healthy ankle joint during walking. To investigate this, we analyzed 3D net joint moments and quasi-stiffness of the ankle joint of TAR patients and compared these to a group of age- and gender-matched control subjects.

9.2 Materials and Methods

9.2.1 Subjects

Ten post TAR patients and a group of ten healthy control subjects, matched for age and gender participated in this study. The ten patients, six males and four females, had received a unilateral mobile-bearing total ankle replacement using the Buechel-Pappas prosthesis[14]. Five replacements had been carried out for posttraumatic arthritis and five for rheumatoid arthritis. The mean time from surgery was 41.5 months (range 11-126 months). The study was approved by the local medical ethics committee. Prior to testing, subjects were informed about the aims and procedure of the experiment, after which they gave informed consent. After completing the measurement, an explanation of the expected results was given.

To reduce possible bias from co-morbidity the inclusion criteria for the patients were (1) that they had their joint replaced at least 10 months earlier and (2) that they should have a minimum ankle score of 80 points on both the LCS[15] and the AOFAS[16] scoring systems. In this way, patients with pain and a reduced function for whatever reason were excluded.

The sample size of ten subjects in each group was found to be sufficient to reveal a clinically relevant difference for peak moment and joint stiffness of 10per cent and 15per cent respectively, using a power calculation with alpha 0.05, beta 0.50, and the standard deviations as derived from the first five participants in each group.

9.2.2 Data Acquisition and Testing Procedures

The laboratory gait evaluation included the simultaneous recordings of body kinematics and ground reaction forces in subjects walking at their self-selected speed. To measure 3-D kinematics an opto-electronic motion analysis system (Optotrak: Northern Digital Inc., Canada), including 4 units with each 3 cameras, was used. Marker placement followed the protocol described earlier by Wu et al.[1]. This protocol was chosen instead of the ISB recommendations[17], since it was used in a previously reported part of this study[7] comparing kinematics after TAR with a group of patients after ankle arthrodesis reported by Wu et al.[1]. The accuracy and repeatability of this protocol was assessed by Wu et al.[1] and found to be satisfactory. In short, markers on the tibio-fibular segment were placed on the lateral and medial tibial condyles and lateral and medial malleoli. On the hindfoot segment, markers were placed on the lateral, posterior and medial calcaneus. One additional marker, placed at the pelvis was used to measure walking speed. Data were collected at a sampling rate of 100 Hz.

One Kistler force plate (Type 9281A11, Kistler Instrument Corp., Winterthur, Switzerland, dimensions 0.60 x 0.40 m) was synchronized with the Optotrak system to measure ground reaction forces (GRF) and the center of pressure (CoP). Data were sampled at a rate of 100 Hz. The ground reaction force was recorded on the operated side for the patients and on the right side for the control subjects. When the operated leg was the left one, data were mirrored to the right during the data analysis procedures.

After marker placement, subjects were asked to walk barefoot forward on a flat 8 m walkway, in which the force plate was incorporated. Before data collection started, subjects walked on the walkway several times until they indicated they were familiar with the walking conditions. Data from a minimum of three gait cycles were collected for each subject walking at his/her self-selected speed. When more than three successful runs were obtained, the three with the most closely matching walking speed were selected.

9.2.3 Data Analysis

The ankle angle was defined as the orientation of the hindfoot relative to the orientation of the local coordinate systems of the tibio-fibular segment. In the local tibio-fibular segment frame, first the x-axis (plantar-dorsal flexion) was fixed between both malleoli; subsequently, the y-axis (ab-adduction) was orientated perpendicular to the x-axis and a line from the midpoint between both malleoli and the midpoint between both tibial condyles; and finally the z-axis (internal-external rotation) was orientated perpendicular to the x and y-axis. Ankle angle was expressed following the standard decomposition order: dorsiflexion–plantarflexion, abduction–adduction and internal–external rotation[1, 7]. The anatomical zero of the ankle angle was defined to occur at 15 per cent of the gait cycle.

The 3D internal ankle net joint moment was calculated from the cross product of the ground reaction force and the distance between the CoP and the ankle center of rotation. The center of rotation of the ankle was chosen midway between the malleoli. The ankle moment was decomposed relative to the orientation of the local segment frame of the tibio-fibular segment. The local net ankle joint moments were presented as a percentage of the stance phase during the gait cycle and the amplitude was normalized to bodyweight. Peak values of the ankle moments around the 3 orthogonal axes of the tibia segment were quantified.

Ankle joint quasi-stiffness around the plantar-dorsal flexion axis (the primary axis of rotation) can be observed as the slope of the ankle plantar-dorsal flexion net joint moment plotted as a function of angular displacement around this axis. In accordance with Davis and DeLuca[8], we calculated the quasi-stiffness coefficient from the slope of the linear regression of joint moment versus angular displacement during the second rocker interval of the stance phase, i.e. the period from the first relative maximum plantar flexion in early stance to maximum dorsiflexion in midstance. The net internal work (Wnet) generated at the ankle during stance was estimated by determining the area under the ankle moment versus ankle angle curve, according to:

$$W_{net} = \int Mx \, d\text{Ø}_x, \qquad (1)$$

Where Mz is the ankle joint moment around the plantar-dorsiflexion axis and Øz is the ankle angular displacement around this axis. The integral in Eq. (1) was computed using trapezoidal approximation[10].

9.2.4 Statistics

The data were tested for normality using the Kolmogorov-Smirnov test, which indicated that the outcome parameters did not deviate from a normal distribution. Hence, statistical analyses between the patients with ankle arthroplasty and the control subjects were undertaken using the two-tailed independent t-test to compare walking speed, peak joint moments, ankle quasi-stiffnes s and net internal work.

9.3 Results

9.3.1 Analysis of the Gait Pattern

Subjects were asked to walk at a self-selected comfortable walking speed. As it can be observed in Table 1, the TAR patient group walked approximately 6 per cent slower compared to the control group (p=0.02).

Table 1 Subject characteristics (mean, SD)								
	Age (year)	Sex[a]	Height (m)	Mass (kg)	Operated side[b]	Time since surgery (months)	Stride length (m)	Gait speed (m·s-1)
TAR (n=10)	59.8 (12.6)	6 M / 4 F	174.7 (5.9)	74.8 (13.9)	5R / 5L	41.5 (41.2)	1.22 (0.7)	1.02 (0.07) *
Control (n=10)	59.0 (12.1)	6 M / 4 F	173.4 (8.0)	72.4 (s6.3)	-	-	1.27 (0.10)	1.09 (0.07)

[a] M = male, F = female, [b] R = right, L = left, * Significant difference (p=0.02)

No significant differences were observed between groups in peak net joint moments generated at the ankle (Table 2, Figure 1). However, small differences in joint moments at the onset of the stance phase during the period of initial contact and loading response are presented in Fig. 1, where the average joint moments of the TAR-group exceeded the bandwidth (mean ± 1 SD) of the control group.

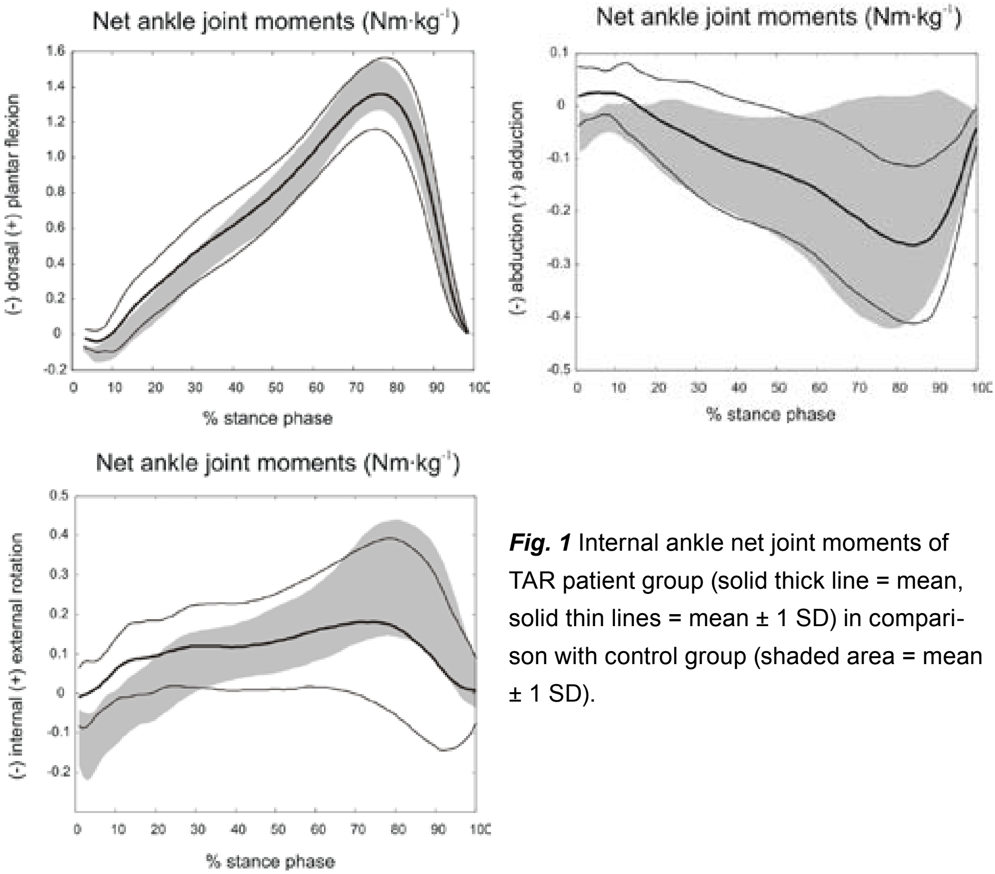

Fig. 1 Internal ankle net joint moments of TAR patient group (solid thick line = mean, solid thin lines = mean ± 1 SD) in comparison with control group (shaded area = mean ± 1 SD).

TABLE 2 Peak net ankle joint moments (mean, SD) in Nm·kg-1)			
	TAR	Control	*p*
Plantar-dorsiflexion	1.38 (0.18)	1.43 (0.13)	0.26
Ab- adduction	-0.28 (0.13)	-0.20 (0.23)	0.36
Internal-external rotation	0.22 (0.17)	0.30 (0.14)	0.31

The average ankle joint moment versus angular displacement around the plantar-dorsal flexion axis for both the TAR and control group is plotted in Fig. 2. There were no significant differences in the quasi-stiffness coefficient between the patient and the control group (Table 3). However, hysteresis of the angle-moment curve appeared to be different between the two groups. The angle-moment curve for the TAR patient group moved in a clockwise rotation, which implies that energy is dissipated around the ankle of the TAR patients during walking. In contrast, the angle-moment curve of the control group showed hardly any hysteresis. Indeed, the net work generated at the ankle during stance in the TAR group was significantly lower than in the control group (Table 3). Energy was dissipated at the ankle in the TAR group while the in the control group energy production at the ankle was, on average, close to zero.

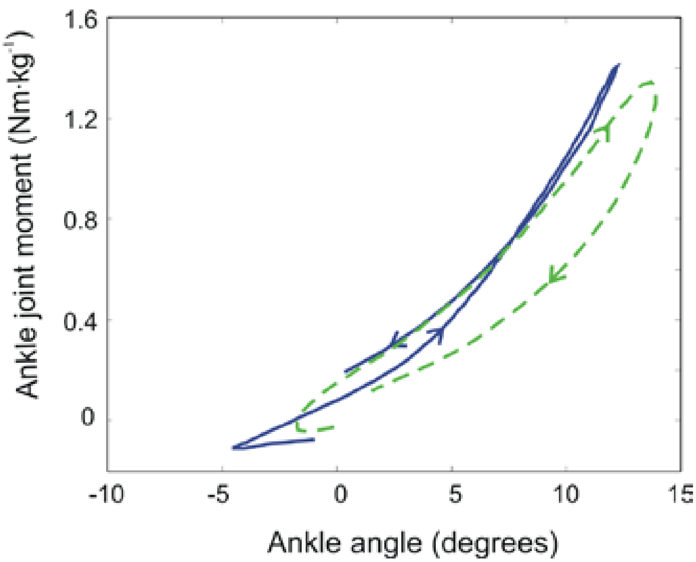

Fig. 2 Ankle joint moment versus ankle angular displacement graph for TAR patients (dashed) and control subjects (solid). Stiffness coefficient was calculated as the slope during the second rocker interval of the stance phase, i.e. the period from the first relative maximum plantar flexion (negative) in early stance to maximum dorsiflexion (positive) in midstance. Net joint work can be observed as the area enclosed by the curves.

TABLE 3 Stiffness coefficients (Nm·kg-1·deg-1) and net internal work (J·kg-1) for the ankle joint around the plantar-dorsal flexion axis (mean, SD)			
	TAR	Control	*p*
quasi-stiffness	0.089 (0.021)	0.092 (0.022)	0.41
net joint work	-0.078 (0.088)	0.005(0.048)	0.02*
* Statistically significant			

9.4 Discussion

In this study the quasi-stiffness and mechanical load of the ankle joint during walking was compared between a group of patients after a successful total ankle arthroplasty and a group of healthy gender and age-matched controls. Between these groups no significant differences were observed in the peak net ankle joint moment and quasi-stiffness of the ankle.

The equal peak plantar-dorsiflexion moment between groups in this study is in accordance with the results of Dyrby et al.[13]. However, these authors found a significant reduction in the abduction moment of the TAR group, which was not the case in our results. Ab-adduction moments are, however, difficult to compare between both studies since Dyrby et al.[13] used a frontal-plane 2D approach to analyze these moments. Despite equal peak moments between groups, the pattern of the net joint moments in our study differed a little between TAR patients and the control group during the first 20 per cent of the stance phase. This could suggest that the patient group walked with a less pronounced heel landing than the control group. This walking pattern, however, did not alter peak joint moments.

We analyzed joint quasi-stiffness in addition to the net joint moments, since net joint moments do not reflect the effect of co-contraction of muscles around a joint. In contrast, joint quasi-stiffness would increase with increasing co-contraction[9] and hence would be indicative of extra joint loading due to co-contraction. In our results we did not find a significant difference in ankle joint quasi-stiffness. This implies that, for this patient population and control group, differences in stiffness were absent or undetectable. Although the sample size in this study was low (and hence statistical power was somewhat low), the observed differences in peak plantar-dorsal flexion moment and stiffness between groups did not exceed 10 per cent and most patients fell within the range of values attained by the control group and within normal values reported in literature[8,11,12]. Hence, while increasing the sample size could result in significant differences, these differences do not seem clinically relevant.

Although there was no significant difference in the stiffness coefficient, the ankles in the TAR patient group were shown to dissipate energy while the ankles in the control group were not dissipating energy. Several explanations could be postulated to account for this difference. First, this energy dissipation in the replaced ankles could have been caused by an increased internal friction of the joint, which would be indicative for joint wear. However, the definition of joint quasi-stiffness[9] does not imply that the ankle would behave as a simple (damped) physical spring. Thus it is not possible to discern what structures, in and around the joint (i.e. joint surface, capsule/ligaments or muscles), would be responsible for energy generation and/or dissipation. The net negative joint work at the ankle could also indicate that the calf muscles of TAR patients may be affected by the ankle joint disease. They may be atrophic or patients may choose to protect their affected leg. In that case, the role of the calf muscles as power generators during walking would be replaced by other muscle groups, i.e. ipsilateral hip extensors or the contra-lateral leg. A third and perhaps most likely explanation for the difference in internal joint work may be related to the difference in walking speed between patients and controls. In the current study we have chosen to allow our subjects walk at their self-selected walking speed. However, the self-selected speed was different in the two groups. Hansen et al.[10] showed that the hysteresis loop of the ankle angle-moment curves changed from clockwise (net energy dissipation) to counter-clockwise (net energy generation) with increasing walking speed and intersected at zero net joint work at approximately 1.3 $m \cdot s^{-1}$. Palmer[18] found a similar crossover speed of 0.9 $m \cdot s^{-1}$. Our patient group had a mean walking speed of 1.02 $m \cdot s^{-1}$ while the control group had a mean walking speed of 1.09 $m \cdot s^{-1}$. The possible crossover speed from clockwise to counter-clockwise of our study (approximately 1.1 $m \cdot s^{-1}$), hence, lies in the range that was found in earlier studies. Thus, the difference in net energy generation could well be explained by a difference in walking speed between the two groups.

Finally, it should be acknowledged that we have chosen to include only successful TAR patients with AOFAS and LCS scores above 80. This was done to exclude patients with co-morbidities around the foot and ankle and with complications after TAR surgery. These co-morbidities and complications would affect the walking pattern and hence obscure the effects of TAR in itself. Hence, this study investigated the mechanical behavior of the ankle after successful TAR surgery and generalization of the results to other populations should be considered with care.

In conclusion, this study showed that the mechanical load of the ankle joint during walking after successful total ankle replacement does not differ from the mechanical load of the healthy human ankle joint. Peak net joint moments and joint quasi-stiffness do not differ, and the difference in net joint work could be attributed to difference in walking speed. These findings indicate that the function of the ankle joint in terms of mechanical load and stiffness appears not to be influenced by total ankle replacement.

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Chapter 10

Metabolic Cost and Mechanical Work for the Step-to-Step Transition in Walking after Successful Total Ankle Arthroplasty

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Abstract

Introduction The aim of this study was to investigate whether impaired ankle function after total ankle arthroplasty (TAA) affects the mechanical work during step-to-step transition and the metabolic cost of walking.

Methods Respiratory and force plate data were recorded in 11 patients and 11 healthy controls while they walked barefoot at a fixed walking speed (FWS, 1.25 m/s) and at their self-selected (SWS).

Results At FWS metabolic cost of transport was 28% higher for the TAA group, but at SWS there was no significant increase. During the step-to-step transition, positive mechanical work generated by the trailing TAA leg was lower and negative mechanical work in the leading intact leg was larger. Despite the increase in mechanical work dissipation during double support, no significant differences in total mechanical work were found over a complete stride. This might be a result of methodological limitations of calculating mechanical work. Nevertheless, mechanical work dissipated during the step-to-step transition at FWS correlated significantly with metabolic cost of transport: r=0.540.

Conclusion It was concluded that patients after successful TAA still experienced an impaired lower leg function, which contributed to an increased mechanical energy dissipation during the step-to-step transition, and to an increase in the metabolic demand of walking.

Keywords Energy Consumption; Mechanical Work; Barefoot Walking, Total Ankle Arthroplasty.

10.1 Introduction

The foot and ankle complex is a versatile complex of joints and muscles that forms a link between the floor and the body in bipedal movement tasks, such as walking. Due to its role as the interface between body and surroundings its behavior has large consequences for the behavior of the entire body on top. In case of pathology of the foot-ankle complex, this will have pronounced influence on walking ability.

People with end-stage arthritis of the ankle joint are often treated with a surgical fixation of this joint (i.e. ankle arthrodesis) as a final clinical intervention to relief the pain and restore functionality. This intervention has been shown to result in a reduced walking speed and step length[1,2], movement compensation in the adjacent tarsal joints[2] and increased energy demand during walking[3]. As an alternative to surgical fixation, in this group of patients ankle motion can be partially preserved with the use of an endoprosthesis, i.e. total ankle arthroplasty (TAA). In TAA the arthritic surfaces of the distal tibia and talar dome are replaced by either semiconstrained two-component fixed-bearing designs or unconstrained three-component mobile-bearing designs. Good medium-term clinical results of TAA with use of such designs have been described in recent years[4,5,6,7,8]. Although endoprostheses have been shown to preserve ankle motion during walking[9], walking speed and mechanical power output of the ankle joint remain reduced[10,11]. The effect of TAA on the metabolic cost of walking is yet unknown.

The metabolic cost of walking is regarded to be an important characteristic of gait, and reflects the ability of people to engage in prolonged walking activitiesi. Restriction of ankle function has been shown to affect metabolic cost of walking both in practice and theory. From experimental research it is known that restriction of ankle motion by an external immobilization[12,13] or through ankle arthrodesis[3] raises energy consumption during walking. However, a satisfying biomechanical explanation for this phenomenon has not been given until now.

A theoretical model on the role of ankle plantar flexion on metabolic cost of walking was recently presented by Kuo, Donelan and Ruina[14]. This so-called double inverted pendulum model describes and predicts the mechanical work necessary for the transition from one step to the next in walking. In its simplest form the human body can be modeled as two rigid legs with a point mass on top. During the step-to-step transition of this model, the center of mass (CoM) velocity has to be redirected from a circular trajectory around the trailing leg to a new circular trajectory around the leading leg. Redirection of the CoM can occur through an impulsive force generated by the leading leg during heel contact. However, this impact force dissipates mechanical energy. To walk at a steady velocity this negative work has to be restored. The double inverted pendulum model predicts that restoring mechanical work could best be done by the trailing leg at the instant of heelstrike through a powerful plantarflexion[14,15,16]. Alternatively, positive work could be generated substantially prior to heelstrike by torques generated around the hip. The model predicts however that this 'hip strategy' would result in an increase in negative work during collision, and consequently in an increase in the mechanical work necessary for the step-to-step transition, which would induce a higher metabolic energy cost for walking. This mechanism could explain the higher metabolic energy consumption found in people walking with a restricted ankle function, who are then forced to use a hip strategy.

In this study we set out to investigate the metabolic cost and the mechanical work performed on the CoM during walking in people after TAA, in order to find whether their impaired ankle function results in an increase in the mechanical work for the step-to-step transition during walking and whether this coincides with an increased metabolic cost. These results will contribute to the validation of the double pendulum model of walking and to our understanding of the crucial role of the ankle in locomotion, as well as to our understanding of the clinical implications of TAA on the metabolic cost of walking.

10.2 Materials and Methods

10.2.1 Subjects

The study was approved by the institutional review board and all subjects gave informed consent prior to data collection. Two groups were included: a control group and a patient group with a successful mobile-bearing TAA. The control group consisted of 11 healthy subjects without any impairment of the lower extremities. Inclusion criteria for the TAA group were: a primary diagnosis of osteoarthritis or rheumatoid arthritis of the ankle joint; a good clinical outcome as defined by ankle scores of more than 80 points on both the Low Contact Stress (LCS) ankle score[17] and the American Orthopaedic Foot and Ankle Society (AOFAS) ankle-hindfoot score[18]. Furthermore, alignment of the ankle-hindfoot complex should be neutral and range-of-motion, measured by manual goniometry, should be a minimum of 10 degrees of dorsiflexion and 20 degrees of plantarflexion, as measured by manual goniometry. The TAA group that was included in this experiment consisted of 11 subjects, 10 of whom had received a unilateral mobile-bearing ankle arthroplasty due to post-traumatic arthritis and 1 due to rheumatoid arthritis a mean 1.9 years (range 0.5 to 4 years) prior to the experiment. Mean duration of ankle symptoms prior to surgery was 11.7 years (range 2.5 to 28 years). Mean age at surgery was 51.5 years (range

40 to 61 years). In seven ankles operated between 2001 and 2004 a Buechel-Pappas (BP) prosthesis (Endotec, South Orange, NJ, USA) was implanted, and in four ankles operated in 2004 a Ceramic Coated Implant (CCI) prosthesis (Van Straten Medical, Nieuwegein, The Netherlands) was used. Both prostheses are mobile-bearing designs, and thereby have no intrinsic constraints. No ankle had a deformity in the frontal plane after surgery. Dorsiflexion averaged 13.4 degrees (range 10 to 32 degrees) and plantarflexion averaged 31.8 degrees (range 24 to 45 degrees). Hindfoot motion was considerably restricted in only one patient. At the time of the experiment the LCS ankle score of the patients was a mean 93 points (range 85 to 98), and the AOFAS ankle-hindfoot score was a mean 91 points (range 85 to 100). None of the patients used walking aids or had a functional impairment of any other lower extremity joint besides the operated ankle. All patients were satisfied by the result of the TAA, and all were able to walk more than one kilometer (Table 1).

10.2.2 Data Collection

Before the start of the experiment height and body mass of each subject were measured. Body mass was measured using the force plates in the experimental set up. In the patient group, the dorsiflexion and plantarflexion at the ankle, and pronation and supination at the hindfoot, were measured by manual goniometry.

The experiment consisted of two parts: stage one in which the mechanical work performed on the CoM (CoM work) was measured, and stage two in which the metabolic energy expenditure was determined. For each subject, the two parts of the experiment were carried out on a single day.

TABLE 1 Demographics of both study populations

	Mean, SD		*p*
	Control	**TAA**	
Gender	♂=7, ♀=4	♂=9, ♀=2	
Age (years)	45.4 (8.1)	54.7 (5.7)	0.005*
Mass (kg)	73.0 (8.0)	83.6 (10.3)	0.013*
Height (cm)	174.3 (7.5)	174.6 (8.4)	0.916
BMI	24.0 (2.2)	27.4 (3.1)	0.007*

TAA = Total Ankle Arthroplasty group; BMI = body mass index; * significant difference (p< 0.05)

CoM work was measured while subjects walked barefoot over a 10m walkway, with two built-in 1x1m custom made strain gauge force plates[i], for the measurement of the ground reaction forces. At the end of the walkway a camera unit of

[i] Resolution x,y direction 0.078 N/bit, z direction 0.159 N/bit, linearity <1%, hysteresis<1%, cross talk < 1%, resonance frequency 60 Hz.

an optical tracking system (Optotrak, Northern Digital Inc, Waterloo, Canada) was placed, which collected the positions of an infrared marker placed on the subject's belt buckle around the waist. Both Optotrak and force plate data were synchronized and collected at a sampling rate of 500Hz.

Before data collection, subjects were given ample time to get familiar with the settings. They were allowed to walk over the walkway without explicit instruction. In the meanwhile we recorded their self-selected walking speed and observed the starting point on the walkway from which they placed at least one foot on the first force plate and two on the second force plate at their natural cadence. Walking speed was determined automatically after each trial, using the collected position data. After each trial the subject was provided with feedback on his/ her walking speed in order to pace themselves to the required speed. Walking trials were carried out at two different speeds: a Self-selected Walking Speed (SWS, i.e the speed they self-selected on the 10 m walk-way) which allowed us to analyze walking cost at normal daily life walking speed, and a Fixed Walking Speed (FWS = 1.25m/s), which allowed us to control for the effect of speed on mechanical and metabolic work. In both conditions, subjects had to perform five successful trials in which their right foot was the trailing leg and five trials in which their left foot was the trailing leg. Trials were excluded when the feet did not align correctly with the force plates and when velocity of the FWS trial deviated more than 0.05m/s from the desired 1.25m/s or when subjects clearly accelerated or decelerated during the stride over the force plates (as could be assessed during data analysis).

Subsequently, subjects' metabolic energy expenditure was measured while walking on a treadmill. This was done by analyzing inspired and expired air with use of an Oxycon breath-by-breath gas analyzer (Jaeger GmbH, Hoechberg, Germany). Subjects walked barefoot at the same SWS and FWS as they had adopted during the walkway trials. Before the start of treadmill walking the resting metabolism during 3 minutes of quiet standing was measured. Subsequently, subjects were allowed to get accustomed to walking on a treadmill, which in general took less than 5 minutes. After this period subjects walked five minutes at SWS and after a period of rest at FWS while oxygen uptake was measured. Halfway during each treadmill trial step frequency was determined in order to test whether this was similar compared to the walkway test.

10.2.3 Data Analysis

From the walkway trials, data were analyzed for one complete step, starting with double support and ending after single support on the leading leg. Double support started with the placement of the leading leg on the second force plate and ended with the toe-off of the trailing leg on the first force plate. With this toe-off, the single support started, which ended with the placement of the former trailing leg on the second force plate. This latter instant could be found as a sudden large displacement of the center of pressure on the second force plate.

Using the force plate data, mechanical CoM work could be calculated according to the individual limbs methods outlined by Donelan, Kram and Kuo[19,20]. First, acceleration of the CoM was calculated from the summed ground reaction forces acting under each limb. Since velocity is the integral of acceleration, velocity of the CoM can be calculated using equation 1.

$$\vec{v}_{com} = \int \left(\frac{\vec{F}_{trail} + \vec{F}_{lead} + m \cdot \vec{g}}{m} \right) dt + \begin{bmatrix} c_x \\ c_y \\ c_z \end{bmatrix}$$

Where $\vec{v}_{com}$ is the velocity vector of the CoM, $\vec{F}_{trail}$ is the ground reaction force vector exerted by the trailing, push-off, $\vec{F}_{lead}$, is the force exerted by the leading, new stance, m is the subjects' body mass and $\vec{g}$ is the gravitational acceleration ([0, 0, -9.81] m•s-[2]). In accordance with Donelan, Kram and Kuo[19], the integration constant for the vertical direction (c_z) was obtained by assuming the average vertical CoM velocity over a step to be zero. The integration constant for the fore-after direction (c_y) was found by assuming the average velocity over a step to be equal to the average walking speed measured by the Optotrak system. For the medio-lateral direction (c_x), the integration constant was found by assuming the CoM velocity at the end of each step to be equal in magnitude but opposite in sign compared to the beginning.

Multiplying force under each separate limb by velocity of the body's CoM results in the calculation of the mechanical power generated by each limb on the CoM (equation 2 and 3).

$$P_{trail} = \vec{F}_{trail} \bullet \vec{v}_{com} \qquad (2)$$
$$P_{lead} = \vec{F}_{lead} \bullet \vec{v}_{com} \qquad (3)$$

The mechanical work performed on the CoM (CoM work) is equal to the cumulative time-integral of the mechanical power and was normalized to body mass. With these calculations of mechanical work, the net mechanical work during one step was calculated for both double and single support in both trailing and leading leg. Since during steady walking the net mechanical work over a complete stride will be zero, total mechanical work over a stride was calculated by summing the absolute (negative and positive) work over two subsequent steps. To compare with metabolic parameters the mechanical work per stride is expressed as average mechanical power ($W \bullet kg^{-1}$) by dividing the mechanical work per stride (J) by body mass (kg), divided by stride time (s). The mechanical cost of transport ($J \cdot kg^{-1} \cdot m^{-1}$) is calculated by dividing the mechanical work per stride (J) by body mass (kg), divided by stride length (m). Data of five successful trials with each leg for each subject were averaged after analysis of each separate trial.

Metabolic energy consumption ($\dot{E}_{met}$) was calculated from VO_2 ($ml \cdot s^{-1}$) and respiratory exchange ration (RER; Garby and Astrup)

$$\dot{E}_{met} = 4.94 \cdot RER + 16.04 \cdot VO_2 \qquad (4)$$

To derive metabolic power ($W \cdot kg^{-1}$), the metabolic energy consumption ($\dot{E}_{met}$) was divided by body mass (kg). Metabolic cost of transport ($J \cdot kg^{-1} \cdot m^{-1}$) was derived in a similar way by dividing metabolic energy consumption ($\dot{E}_{met}$) by body mass (kg) and divided by walking speed ($m \cdot s^{-1}$).

10.2.4 Statistics

The data were tested for normality using the Kolmogorov-Smirnov test, which indicated that the outcome parameters did not deviate from a normal distribution, and parametric statistics could be used. Differences between the healthy and affected leg of the TAA group and the control group were tested for significance using a Student t-test. For the comparison of step frequency between the walking conditions a paired sampled t-test was used. The correlation between mechanical work and metabolic energy cost was analyzed using Pearson's correlation coefficient. Differences were considered significant when $p < 0.05$.

10.3 Results

10.3.1 Spatiotemporal and Metabolic Parameters

Table 2 shows the spatiotemporal and metabolic parameters for the SWS and FWS condition. At SWS, control subjects walked faster than the patient group. The metabolic cost of transport was 6% higher for the patient group. This difference was not statistically significant. Both controls and patients were able to walk comfortably at FWS, which was for both groups slower than their SWS. At FWS, walking was significantly more demanding for the patient group. Metabolic power and cost of transport were respectively 29% and 28% higher for the patient group. Step length and step frequency did not differ significantly between groups at both SWS and FWS (Table 2), nor did step length and frequency differ between walking on the walkway and walking on the treadmill for both velocities.

TABLE 2 Gait parameters, metabolic power and cost of transport, and mechanical power and cost of transport for both groups during self-selected walking speed (SWS) and fixed walking speed (FWS)			
	Mean (SD)		*p*
	Control	TAA	
SWS			
walking speed (m•s $^{-1}$)	1.47 (0.17)	1.29 (0.14)	0.03*
step frequency (steps•s $^{-1}$)	1,98 (0.16)	1,83 (0.10)	0.07
step length (m)	0.74 (0.07)	0.71 (0.10)	0.34
metabolic power (W•kg^{-1})	3.42 (1.0)	3.62 (1.16)	0.74
metabolic cost of transport (J•kg^{-1}•m^{-1})	2.35 (0.54)	2.50 (0.68)	0.60
mechanical power (W•kg^{-1})	1.62 (0.28)	1.26 (0.24)	0.005*
mechanical cost of transport (J•kg^{-1}•m^{-1})	1.11 (0.11)	0.98 (0.11)	0.008*
FWS			
walking speed (m•s $^{-1}$)	1.25 (0.01)	1.25 (0.01)	
step frequency (steps•s $^{-1}$)	1.83 (0.12)	1,81 (0.10)	0.97
step length (m)	0.69 (0.04)	0.69 (0.04)	0.99
metabolic power (W•kg^{-1})	2.40 (0.55)	3.09 (0.54)	0.007*
metabolic cost of transport (J•kg^{-1}•m^{-1})	2.01 (0.46)	2.58 (0.45)	0.007*
mechanical power (W•kg^{-1})	1.30 (0.20)	1.19 (0.14)	0.14
mechanical cost of transport (J•kg^{-1}•m^{-1})	1.04 (0.16)	0.95 (0.11)	0.14
TAA = Total Ankle Arthroplasty group; * significant difference (p < 0.05)			

10.3.2 Mechanical Work

Table 3 shows the parameters dealing with the mechanical work performed on the CoM at both SWS and FWS during the separate phases of a step. For the patient group steps were analyzed separately: one step in which the healthy leg was the trailing leg (TAAh, healthy push-off) and one step in which the affected leg was the trailing leg (TAAa, affected push-off). In the two separate phases of a step, double support and single support, differences occurred, especially at the TAAa step. These differences can also be seen in the power curves of each step for SWS and FWS (Fig. 1). Note that the areas under the power curves visualize external mechanical work. Net work over a step did, on average, not exceed 0.0125 J•kg^{-1} for each condition.

TABLE 3 Center of mass work (J/kg) during the separate phases of a step					
	Mean (SD)			*p*	
	Control	TAAh	TAAa	Control -TAAh	Control -TAAa
SWS					
Wtrail DS	0.27 (0.05)	0.23 (0.05)	0.16 (0.04)	0.04*	< 0.001*
Wlead DS	-0.15 (0.07)	-0.18 (0.07)	-0.22 (0.09)	0.24	0.04*
W SS	-0.12 (0.09)	-0.04 (0.10)	0.06 (0.08)	0.09	< 0.001*
|Wstep|	0.82 (0.12)	0.67 (0.14)	0.72 (0.17)	0.013*	0.15
FWS					
Wtrail DS	0.26 (0.04)	0.22 (0.05)	0.16 (0.03)	0.061	< 0.001*
Wlead DS	-0.10 (0.06)	-0.17 (0.04)	-0.20 (0.07)	0.004*	< 0.001*
W SS	-0.15 (0.10)	-0.04 (0.05)	0.03 (0.07)	0.003*	< 0.001*
|Wstep|	0.72 (0.13)	0.62 (0.06)	0.69 (0.10)	0.044*	0.61
Parameters for a step in both test groups during self-selected walking speed (SWS) and fixed walking speed (FWS, 1.25 m/s): TAAh = a TAA step which started with the push-off by the healthy leg; TAAa= a TAA step which started with the push-off by the affected leg; W DS= net work in one leg (lead or trail) during double support; W SS = net work during single support; [Wstep| = absolute work over a full step; * significant difference (p< 0.05)					

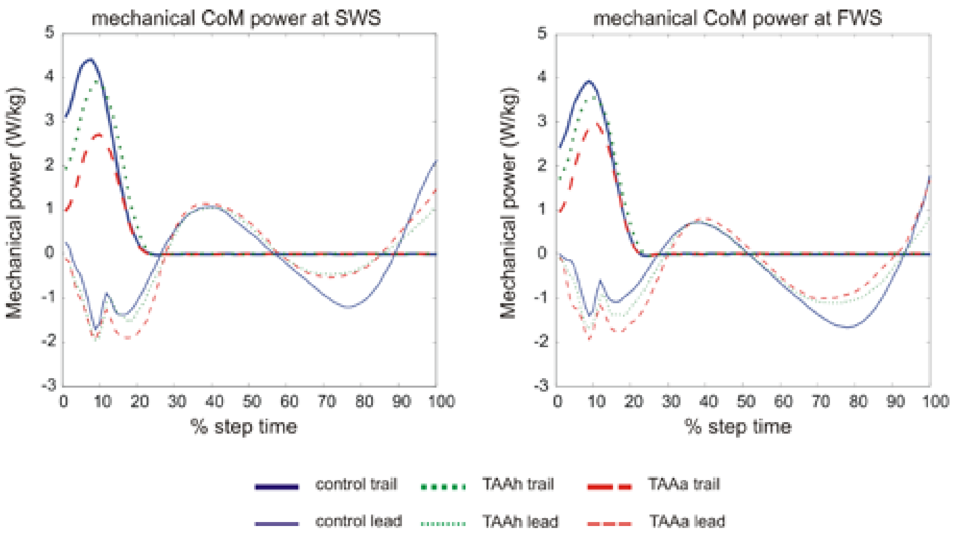

Fig. 1 Power curves from the trailing and leading leg during walking at their Self-selected Walking Speed (SWS) and at Fixed Walking Speed (FWS) of 1.25 m s-1. Areas under the power curves represent the positive and negative mechanical work. Mean curves for the control group in solid blue; mean curves for the step in which the healthy leg of patients was trailing (TAAh) in dotted green; mean curves for the step in which the affected leg of the patient was trailing (TAAa) in dashed red.

At SWS, as hypothesized, less positive mechanical work was generated with the trailing TAA leg during the double support phase, and simultaneously more negative work was generated by the healthy leading leg compared to the control group. During the single support phase in the control group net external mechanical work was negative. For the TAA group, negative mechanical work in the single support was significantly reduced on the affected leg and was even converted into positive work in the healthy leg. The total absolute mechanical work per full step did not differ between controls and patients for both a step with the affected and healthy leg being the trailing leg. Furthermore, the average mechanical cost of transport at SWS was significantly lower for the TAA group (see Table 2).

At FWS, less positive mechanical work was generated by the TAA trailing leg during the double support phase, and simultaneously more negative mechanical work was generated by the healthy leading leg compared to the control group. When the healthy leg was the trailing leg, during the double support phase a non-significant reduction in positive work of the healthy trailing leg was accompanied by a significant increase in the negative work in the TAA leading leg. During the single support phase, the net mechanical work was negative for controls but significantly less negative for the single support on the affected leg and even positive for the healthy leg. Differences in net mechanical work during single support were also significant between both the affected and healthy legs and the control group. As for the SWS condition, total absolute mechanical work did differ between controls and the healthy leg of patients being the trailing leg. Average mechanical cost of transport at FWS was not significantly reduced for the TAA group (see Table 2).

10.3.3 Relationship between Metabolic and Mechanical Energy

A correlation was found between metabolic and total absolute mechanical cost of transport at FWS (r=-0.519, p=0.013; see Fig. 2). Remarkably, this correlation appeared to be negative, meaning that the more absolute mechanical work over a stride was generated the less metabolic energy was consumed. In contrast, a positive correlation was found between the negative work performed during double support (expressed as $J \cdot kg^{-1} \cdot m^{-1}$) and the metabolic cost of transport at FWS (r=0.540, p<0.009) and at SWS (p=0.404, p=0.100), although the latter was not significant. This indicates that the larger the collision cost in the leading leg during heel strike the higher the metabolic cost of walking, i.e. the energy dissipated during the step-to-step transition explained 29% of the variance in metabolic energy cost of walking at FWS in the total group of subjects.

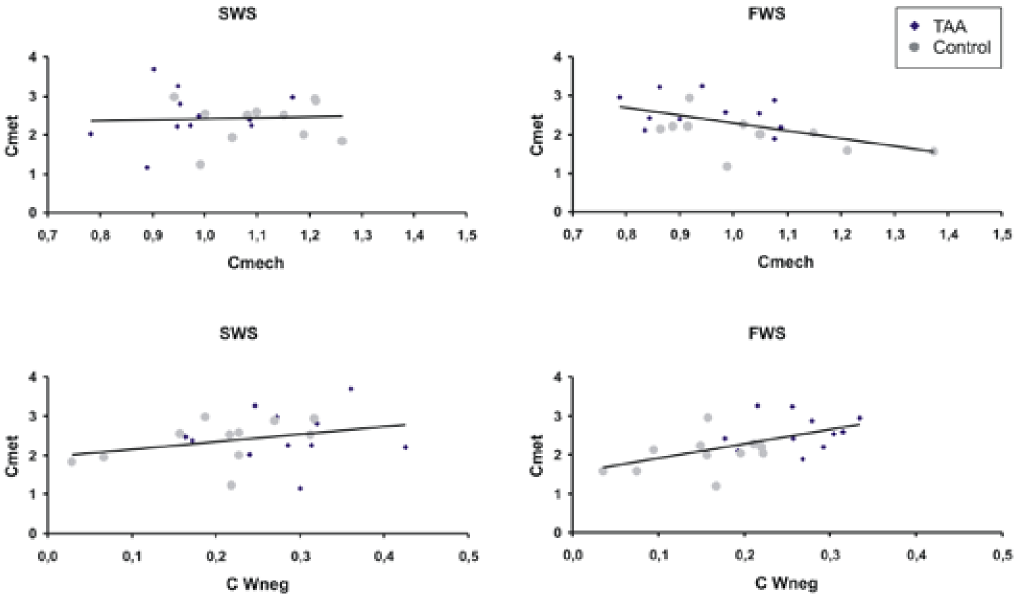

Fig. 2 Correlations between metabolic and total absolute mechanic cost of transport of TAA patients and controls. Upper panels: relation between metabolic cost and mechanical cost per stride (J kg m^{-1}) at SWS and at FWS. Lower panels: relation between metabolic cost per stride and negative mechanical work during the double support phase (J kg m^{-1}) at SWS and at FWS. Control subjects are displayed as gray circles, TAA subjects as blue diamonds.

10.4 Discussion

We investigated whether the impaired ankle function after TAA affected the metabolic cost of walking and the mechanical work during the step-to-step transition. At FWS, mechanical work performed by the trailing TAA leg was lower, and walking proved to be metabolically more demanding for the study group. Self-selected walking speed was 12% lower in the study group. At SWS, however, no significant difference in metabolic power or cost of transport was found.

The only study on energy expenditure after TAA we are aware of is the study by Detrembleur and Leemrijse[22]. In a prospective study, they found an increase in walking speed and a decrease in external mechanical work and energy expenditure at an average of 7 months after TAA surgery. However, in their patients walking speed remained well below (0.77 $m \cdot s^{-1}$), and metabolic cost of transport remained well above (3.18 $J \cdot kg^{-1} \cdot m^{-1}$) both the level of able bodied subjects and the level of the patient group included in this study, who were, on average, 1.9 years after surgery.

What makes TAA walking at FWS more energy consuming than normal walking? An explanation was expected to be found using the double inverted pendulum model of walking[14,19,20]. According to this model, the following hypotheses could be formulated regarding the result of a decrease in active plantar flexion power around the TAA ankle: 1) a decrease of positive work performed during push-off by the affected leg in the double support phase, 2) an increase of negative work performed during loading of the healthy leading leg in the double support phase, 3) an increase in work performed during single support, and subsequently 4) an increase of the total external mechanical work performed, and 5) an increase in the metabolic cost of walking at a constant velocity.

As expected, at FWS TAA patients showed a decrease in positive work during the push-off with their affected leg, an increase in negative external mechanical work performed by the healthy leading leg, and an increase in the net work during the single support phase. Hence, for these separate phases of the gait cycle our hypotheses could be confirmed and indeed we found that the mechanical work dissipated in the collision of the leading leg during double support was larger in the TAA group.

In contrast to our fourth hypothesis, however, total external mechanical work per step and per stride was lower in patients. Moreover, a weak though significant negative correlation was even found between total external mechanical and metabolic cost of walking, i.e. the higher the metabolic cost of walking, the lower the external mechanical energy proved to be. So, despite an increased mechanical energy dissipation during the step-to-step transition no increase in total mechanical work was found. It could be possible that TAA patients found a strategy to reduce total external mechanical work, despite their increased negative work at heelstrike. However, since this strategy apparently does not reduce metabolic cost, there seems no benefit in doing so. Alternatively, the relationship between total external mechanical work and metabolic energy consumption could be questioned. It should be realized that the total external mechanical work, as calculated in this study, represents the net work performed on the body's CoM. Simultaneous opposite work terms can cancel each other out. For the relatively static double support phase this cancellation is dealt with by the individual limbs method. However, especially during the single support phase, where energy is generated and exchanged between stance and swing leg, the relation between muscle work and external work is not warranted. For instance, it is not possible to derive whether the increase in external work in the single support phase found in this study, which actually was a decrease in net negative work, is the result of an increase in positive muscle work, a reduction in negative muscle work, or a combination of both. These limitations of external work calculation have been demonstrated and discussed previously[23,24,25].

Despite the above mentioned concerns with the validity of calculations of external work (which most importantly affect the single support phase), a significant increase in negative work in the leading intact leg during double support was found for the patient group. This means that extra energy is dissipated at heelstrike, which needs to be regenerated somewhere during the step. In addition, a positive correlation was found between the negative work done by the leading leg during heel strike and the metabolic cost of walking. The correlation between the negative work at heelstrike and metabolic energy cost indicates that the work dissipated at heelstrike during FWS accounts for 29% of the variance in energy cost of walking in our study group. Hence, with some prudence, it can still be concluded that the mechanical work required for the step-to-step transition explains at least part of the increased metabolic cost of walking after TAA.

The SWS condition was included in this study since it represents a more natural condition compared to FWS, although interpretation is more difficult due to the effect of walking speed on mechanical and metabolic cost. Self-selected walking speed appeared to be almost similar to the imposed fixed walking speed in the patient group, but controls walked significantly faster in the SWS condition. It was observed that work performed by the TAA leg during the separate phases of a step at SWS was similar to the work at FWS in patients, but due to the higher walking speed of the control group, the difference in collision cost with the control group became non-significant. Nevertheless, still a moderate, though not quite significant, correlation was maintained between the negative work at heelstrike and metabolic energy cost. This supports the conclusion that the energy cost for the step-to-step transition contributes to the increased metabolic cost of walking after TAA.

The explanation of the increased energy cost of pathological gait by the inverted pendulum model is encouraging, since past (biomechanical) studies aimed at explaining the increased energy cost of pathological gait have failed. For instance, the increased energy cost of amputees walking with a lower limb prosthesis has been shown to be unrelated to vertical CoM movement, external total CoM work (using combined limbs method), joint work or recovery index[24,26]. Looking in isolation at collision cost during the step-to-step transition might thus be useful to investigate the energy cost of other pathological gaits. This has recently also been demonstrated for the energy cost of walking in people after lower limb amputation[27].

However, the increased mechanical work for the step-to-step transition only seems able to account for part of the increased energy cost of walking after TAA. It therefore is reasonable to consider additional mechanisms as well. One of these mechanisms could be related to balance control. In an earlier study increased co-contraction of the lower leg muscles was found in walking after TAA, possibly in an attempt to stabilize gait[4]. This co-contraction is metabolically demanding but does not contribute to external mechanical work as measured with the use of ground reaction forces. An increased effort for balance control might be an intrinsic feature of walking after TAA, but might also be enhanced by the fact that subjects had to walk on a treadmill, which might be more challenging for the patients than for the able-bodied control subject. Besides the balance control issue, walking on a treadmill has been found to be mechanically similar to over ground walking[28], and hence will be less likely to account for differences between mechanical energy measured on the walkway and metabolic cost measured on the treadmill. Moreover, step length, which has an effect on step-to-step transition cost[16], was found not to differ between walkway and treadmill trials in this study. Another explanation could perhaps be found in the atrophy of the lower leg muscles that results from longstanding ankle disease. In atrophic muscle tissue fiber composition has changed as type 1 muscle fibers (slow, fatigue resistant) are lost predominantly. Thus, an atrophic muscle consists of a greater deal of fast, type 2, muscle fibers, which consume more energy than slow, type 1, muscle fibers[29]. Thus, persistent muscle atrophy could theoretically also account for the higher metabolic energy requirements found in this study.

With some limitations, our metabolic results can be compared with other studies in which ankle function was impaired or restricted. Waters et al.[6], found that patients with an ankle arthrodesis had an 11% increase in oxygen cost compared to healthy controls, both walking at their SWS. In healthy subjects walking with and without a below-knee plaster cast the oxygen cost at SWS was found to be 16% to 27% higher compared to unconstrained walking[12,13]. For these latter studies however, the weight of the cast and the fact that subjects did not walked barefooted should be taken into account. In addition, the increase of energy expenditure of walking with an externally immobilized ankle can be mitigated by the use of an appropriate rocker bottom sole[30,31]. With a 6% increase of metabolic cost at SWS, as found in this study, the metabolic demands for TAA subjects appear favorable in comparison with subjects with either an externally immobilized or a surgically fused ankle, the more so as our patient group walked at a higher SWS than the subjects in the referred studies.

In conclusion, in a patient group with a well-functioning unilateral total ankle arthroplasty we have found that metabolic power and cost of transport were significantly higher compared to an able-bodied control group. This coincided with, and is partially explained by a higher negative mechanical work during the collision of the leading leg in the step-to-step transition. This indicates that after a successful TAA an impaired ankle function remains, which contributes to an increased mechanical energy dissipation during the step-to-step transition and to a reduction in walking economy.

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Chapter 11

Overview of Currently Available Prosthetic Designs and A Review of the Clinical and Radiographic Outcome of Total Ankle Arthroplasty

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11.1 Current Designs of Total Ankle Prostheses

Most designs currently in use throughout the world for TAA are 3-component mobile-bearing designs. Exceptions are: the semi-constrained Agility (DePuy, Warsaw, Indiana, USA) total ankle prosthesis, mainly used in the USA[1]; the ESKA prosthesis, used on a limited scale in Germany[2]; and the TNK ankle prosthesis, used in Japan[3]. Of these 2-component designs, the Agility prosthesis is currently most widely used. Its upper component is designed to obtain support from both the distal tibia and fibula. Consequently, it requires an osseous fusion of the distal tibiofibular syndesmosis. This makes this design different from all other designs, and, from the surgical point of view, somewhat more difficult. In Europe, 2-component prostheses have almost disappeared from the market, and here currently mobile-bearing prostheses are used almost exclusively.

The mobile-bearing design is characterized by the use of a mobile polyethylene bearing in between the tibial and talar component. The major advantage of this concept is the reduction of shear and rotational forces at the bone-prosthesis interface. As a consequence, mechanical stresses on the implant are mostly compressive in nature, and therefore more physiological. Thus, the mobile-bearing concept appears to have biomechanical benefits for the endoprosthetic replacement of the arthritic ankle. This design rationale has proven to be successful in the endoprosthetic replacement of the knee as well[4,5,6]. However, in the knee joint, comparative clinical studies between fixed-bearing and mobile-bearing designs have failed to show superior clinical results with mobile-bearing over fixed-bearing designs[7,8]. In the ankle, no clinical studies comparing fixed-bearing and mobile-bearing have been carried out, although some cadaver and in-vivo studies investigating the kinematics of the replaced ankle exist. Michelson et al[9] in a cadaver experiment found no significant kinematic differences changes between the normal ankle and the ankle replaced by the fixed-bearing Irvine prosthesis (a 2-component spherical design). In contrast to this, Valderrabano et al[10,11,12], in cadaver experiments, showed a better kinematic behavior of two mobile-bearing prostheses: STAR (Waldemar Link, Hamburg, Germany) and Hintegra (NewDeal, Lyon, France), than with the semi-constrained fixed-bearing Agility prosthesis. Surgical advantages of 3-component mobile-bearing prostheses over 2-component designs are that: 1) adequate stability of the replaced ankle can be obtained by the use of a bearing of appropriate thickness; and that 2) in the event of wear or instability, a bearing exchange can easily be done. However, potential disadvantages of mobile-bearing prostheses are: 1) the risk of bearing subluxation or dislocation; and 2) the risk of impingement of the bearing with either flanges on the metallic components or with the malleoli. Impingement could give rise to symptomatic arthritis, accelerated wear and osteolysis.

All current ankle prostheses are designed for implantation without bone cement. A porous microstructure is applied to the osseous surface of most implants, often combined with a hydroxyapatite or similar calcium phosphate coating, in order to enhance the osseointegration process. Mobile-bearing prostheses can be classified in two main categories: symmetrical and asymmetrical. Symmetrical implants have no left-right version. Asymmetrical implants do have a left and right version, in most designs limited to the talar component. Concerning the fixation of the tibial component, there are several options: a stem or small fin(s), pegs or screws. In the past, stemmed tibial components were designed to be implanted with an anterior slope of the articulating surface of between five and seven degrees. Examples are: LCS (DePuy, Warsaw, Indiana, USA; its successor the Buechel-Pappas (Endotec, South Orange, New Jersey, USA); Salto (Tornier, Saint-Ismier, France); and the AES (Biomet, Valence, France) prosthesis. The philosophy behind this anterior slope is two-fold: a more anatomical bone resection and a better dorsiflexion. However, both claims remain to be proven. Newly designed stemmed tibial components have a stem that makes a 90 degree angle to the base plate. Examples are: Mobility (DePuy, Leeds, UK) and Zenith (Corin, Cirencester, UK). The articulating surface of the tibial component is mostly flat. The tibial component of the Salto prosthesis has a small flange to prevent contact of the bearing with the medial malleolus. A novel design, the BOX prosthesis (Finsbury, Leatherhead, UK), has a convex spherical articulating surface of the tibial component, in order to give a better mobility compared to flat-surface designs in both the sagittal and the frontal plane[13]. The talar component can be without (true tibiotalar replacement) or with articulating flanges for the malleoli (mostly hemiarthroplasty of the talarmalleolar joints) and can have a symmetrical curvature (examples: Buechel-Pappas, STAR, AES, Mobility, CCI (VanStratenMedical, Nieuwegein, Netherlands), Taric (Implantcast, Buxtehude, Germany), Zenith or an anatomical configuration with a smaller curvature medially compared to laterally (examples: Salto and Hintegra). Furthermore, the talar component can have either a cylindrical or a trochlear shape of the articulating surface. Most polyethylene bearings are fully congruent with both the tibial and talar component. Such a full-contact bearing will reduce the stresses on the polyethylene, thereby improving their wear characteristics, as has been shown by the endoprosthetic mobile-bearing replacement of the knee[14,15,16]. The bearing of the Salto prosthesis has purposely a slight incongruency, thereby allowing for some inversion-eversion motion, in order to compensate for a loss of motion of the hindfoot. However, it appears to be questionable if this concept is advantageous and, on the contrary, does not have other disadvantages such as accelerated wear.

11.2 Review of the Clinical and Radiographic Outcome of Total Ankle Arthroplasty

11.2.1 Clinical Outcome

A substantial number of reports on the clinical and radiographic outcome of TAA have been published in the last decade. Giannini et al[17] gave a historic overview of all designs that had been developed from the early 1970s until 2000. Feldman and Rockwood[18] gave an overview of 11 currently used fixed and mobile-bearing designs. Recently, Stengel et al[19] gave a literature review on the outcome of three mobile-bearing designs and of the fixed-bearing ESKA prosthesis. They found an increase in ankle score of 45.2 points. As recently new literature on the outcome of TAA has become available, and also to complete the above mentioned outcome reviews, an update of the outcome studies of currently used 2 and 3-component designs is presented. With the exception of the TNK ankle, only results with uncemented prostheses are reported.

Plantarflexion-dorsiflexion of normal ankles, as measured clinically (thus a combined motion of the ankle and hindfoot), varies from 50 to 70 degrees. Level walking, ascending and descending stairs require respectively 24 to 30, 37 and 56 degrees of range of motion. Range of motion during normal gait varies, depending on the method of measuring[20]. In the studies mentioned in Table 1, an increase in range of motion of between 3 to 13 degrees after TAA was found. Mean clinically measured range of motion at follow-up ranged from 23 to 41 degrees, clearly less than the motion of the healthy ankle joint. This range of motion at follow-up is sufficient for level walking, but not for descending stairs in a normal manner. Motion as measured clinically is larger than as measured radiographically, due to soft tissue deformation and motion in adjacent joints.[21] This was indeed observed in some of the studies reported.

Seven different scoring systems have been used in the studies referred to in Table 1: the Evanski[41], Mazur[42], LCS[43], AOFAS[44], Kofoed[45], the Ankle Osteoarthritis Scale (AOS)[46] and the almost similar Foot Function Index (FFI)[47], the latter two being the only self-reporting patient-assessment instruments. Except for the AOS and the FFI, the reliability and validity of the five scoring systems, quantifying subjective (pain) and objective (function, range of motion, etc) parameters has not been established. No gross differences have been observed between the five non-validated systems[48]. The clinical scores of these studies are summarized in Table 1.

TABLE 1 Overview of scores preoperative and at latest follow-up in individual studies

Fixed-bearing Designs*	n (n FU)	Mean FU (yrs)	Mean Clinical score			Scoring system
			Preoperative	At FU	*p*	
Agility[22]	132 (68)*,†	9	--	2.02 PS 3.36 DS	-	AOS
Agility[23]	43 (40)*,†	3.7	33.6	83.3	<.001	AOFAS
TNK[3]	70 (61)*,†	5.2	46.7	80.6	N.D.	Mazur
TNK[24]	21 (21)	2.8	52.0	74.1	<.05	Evanski
ESKA	159 102)*,†,‡	<15	34.3	94.5§	N.D.	Kofoed
Mobile-bearing Designs*						
BP[26]	15 (15)	2	22	80	<.001	AOFAS
BP[27]	31 (28)*,†	8.3	-	81	-	AOFAS
BP[28]	19 (19)	4.4	-	79	-	AOFAS
LCS/BP[29]	93 (57)†	8	36.1 26.5 26.9	83.3 77.0 75.7	<.05	LCS AOFAS Kofoed
STAR[30]	51 (39)†	4.3	39 -	70 74	<.0001	Kofoed AOFAS
STAR[31]	58 (48)†,||	3.1	-	81		AOFAS
STAR[32]	25 (25)	9.5	30	91.9	N.D.	Kofoed
STAR[33]	22 (20)†	2.2	-	75	-	Kofoed
STAR[34]	27 (26)*	1.3	35.4	74.8	N.D.	Kofoed
STAR[35]	22 (19)*	3.1	44.7	86.9	N.D.	Kofoed
STAR[36]	74 (68)*	3.7	24.7	84.3	<.05	AOFAS
STAR[37]	200 (182)†	3.8	28	70¶	N.D.	AOFAS
STAR[38]	49 (45)	2.3	- 59	68 35	<.001	Kofoed FFI
SALTO[39]	98 (93)*,†	2.9	32.3	83.1	<.005	AOFAS
HINTEGRA[40]	278 (271)*	2.8	40.3	85	N.D.	AOFAS

*: referred study; n: total number of ankles in study; n FU: number of ankles evaluated clinically at follow-up; Mean FU (yrs): mean follow-up duration in years; p: level of significance; PS: pain subscale; DS: disability subscale; N.D.: not described; LCS = Low Contact Stress; AOFAS = American Orthopaedic Foot and Ankle Society; AOS = Ankle Osteoarthritis Scale; FFI = Foot Function Index.

Details of some studies: * Some subjects lost-to-follow-up. † Revised prostheses excluded from analysis. ‡ Subjects with follow-up shorter than 10 months excluded. § Score at minimum 5-year follow-up. || Subjects with follow-up shorter than 1 year excluded, 1 subject with deep infection excluded. ¶ Score at 2-year follow-up.

All evaluated studies recording preoperative clinical scores report an increase at latest follow-up. Mean preoperative scores between the studies varied between 22 and 52 points. These differences might be attributable to differences in patient selection. Mean scores at follow-up varied between 70 and 94.5 points. The mean increase in score (weighted to sample size) of the 2-component prostheses is 47.6 points (121%, 39.2 points preoperative to 86.8 points at follow-up; n = 224). The mean increase in score (weighted to sample size) of the 3-component prostheses was 46.4 points (138% increase, from 33.4 points preoperative to 79.8 points at follow-up; n = 795). Clinical scores improved significantly after implantation of both 2 and 3-component total ankle prostheses. With regard to clinical score, 2 and 3-component prostheses showed similar results.

The following intraoperative and early postoperative complications have been reported to either influence the outcome of TAA or to necessitate a secondary intervention: malleolar and distal tibial fracture, persistent instability, persistent deformity, skin necrosis, delayed wound healing, deep infection, heterotopic ossification, persistent pain due to talarmalleolar arthritis, and, for the Agility prosthesis: nonunion of the distal tibiofibular synostosis.

11.2.2 Implant Survival

Adequate survival data with the end point defined as removal or replacement of components, conversion to ankle arthrodesis or below-the-knee amputation, and with a mean follow-up of five years or more are available from only a few of the studies referred to in Table 1. In a retrospective study, Knecht et al[22] reported an 8-year survival rate with the Agility prosthesis of 0.91 (95% CI: 0.84-0.96). The study population consisted of patients with posttraumatic arthritis, primary osteoarthritis and rheumatoid arthritis. All ankles had been operated by the designer of the prosthesis. San Giovanni et al[27] reported an 8-year survival rate with the Buechel-Pappas prosthesis of 0.93 (95% CI: 0.82-1) in a retrospective study of patients with the diagnosis of rheumatoid arthritis. In a prospective study with the LCS and Buechel-Pappas prosthesis combined, Doets et al[29] reported an 8-year survival rate of 0.84 (95% CI: 0.73-0.93) in a patient population with inflammatory joint disease (mainly rheumatoid arthritis). A better survival rate was found in ankles with a neutral alignment preoperatively and in ankles were a tibial component of correct size had been implanted. Kofoed[32] reported a 12-year survival rate with uncemented version of the STAR prosthesis of 0.94 (95% CI: 0.91-1) in a population mainly having osteoarthritis.

Stengel et al[19], in a meta-analysis of six studies, found a mean overall 5-year survival rate of 90.6 (95% CI: 84.1-97.1). Recently, reports from national arthroplasty registers have been published. Henricson et al[49] reported the results with six mobile-bearing designs from the Swedish Ankle Arthroplasty Register implanted between 1993 and 2005. They found a mean overall 5-year survival rate of 0.78 (95% CI: 0.74-0.82). The STAR prosthesis was used most frequently, but its number had decreased significantly in recent years. Aseptic loosening was the most frequent mode of failure, followed by technical errors (implant malposition) and infection. Better results were achieved by experienced surgeons who performed substantial numbers of arthroplasties. Fevang et al[50] reported the results with four designs implanted between 1994 and 2005 from the Norwegian Arthroplasty Register. They found a mean overall 5-year survival rate of 0.89 (95% CI: 0.84-0.93). Also in Norway the STAR prosthesis was the implant used most widely, followed by the 2-component TPR prosthesis. The TPR prosthesis was only used early in the study period, and almost exclusively in patients with inflammatory joint disease. Failure as a result of aseptic loosening of the STAR prosthesis occurred more frequently with the single-coated version (hydroxyapatite) than with the double-coated version (calcium phosphate coating on porous titanium). No other risk factor for failure could be identified. Hosman et al[51] reported the results with four designs from the New Zealand National Joint Registry implanted between 2000 and 2005. They found a mean overall 5-year survival rate of 0.86 (95% CI: 0.78-0.94). The 2-component Agility prosthesis was used most widely in New Zealand, followed by STAR prosthesis. Again, aseptic loosening was the most frequent mode of failure. In their series, patients with an inferior ankle score at six months postoperatively had an increased failure rate.

11.2.3 Radiographic Outcome

Comparison of radiographic loosening and/or migration of the prosthetic components between the studies of this review is difficult due to the fact that non-uniform criteria were used. Furthermore, almost half of the studies did not provide clear criteria for classification of components as being loose or having migrated. Table 2 summarizes the percentage of reported radiographic loosening or migration of the prosthetic components. With a relative short follow-up sometimes high percentages of loose prostheses were observed and there were differences in incidence of loosening with same designs. Partial radiolucent lines at the bone-prosthesis interface, which might indicate future failure, are not included in this overview. They were reported with an incidence ranging from 0 to 85 per cent in the studies referred to. It is surprising to note the difference of radiographically loose STAR-components (3.5% to 29.3%), reported in three studies with similar follow-up (3.6, 3.8 and 4.3 years), with use of the same criteria to assess subsidence and tilting[30,31,37]. Wood and Deakin[37] suggested that the difference in coating before and after 1999 could explain this difference in radiographic outcome, while Carlsson[31] suggested that the main factor probably was the surgical learning curve.

TABLE 2 Radiographic signs of loosening or migration of prosthetic components at follow-up in individual studies				
Fixed-bearing Designs*	n (X-FU)	Mean FU (yrs)	Aseptic loosening at FU % (n)	Failed component
Agility[22]	132 (117)*	7.2†	4.2% (5) 7.6% (9) 1.7% (2) 13.6% (16)	Tibial component Talar component Tibial + talar component TOTAL
Agility[23]	43 (40)‡,§	3.7	15% (6) 27.5% (11) 2.5% (1) 45% (18)	Tibial component Talar component Tibial + talar component TOTAL
Agility[53]	306 (85)||	2.7	34.1% (29)	Tibial and/or talar component
TNK[3]	70 (68)§	5.2	11.8% (8) 22.0% (15) 33.8% (23)	Tibial component Talar component TOTAL
Mobile-bearing Designs*				
BP[53]	75 (75)	5	4.0% (3)	Talar component
BP[27]	31 (28)‡,§	8.3	17.9% (5) 14.2% (4) 32.1% (9)	Tibial component Talar component TOTAL
BP[28]	19 (19)	4.4	10.5% (2)	Tibial component
LCS/BP[29]	93 (78)‡	7.2	9.0% (7) 7.6% (6) 16.6% (13)	Tibial component Talar component TOTAL
STAR[30]	51 (51)	4.3	7.8% (4) 13.7% (7) 7.8% (4) 29.3% (15)	Tibial component Talar component Tibial + talar component TOTAL
STAR[31]	58 (52)**	3.6	5.7% (3)	Tibial component
STAR[32]	25 (25)	9.5	4.0% (1)	Tibial component
STAR[33]	22 (22)	2.2	4.5% (1)	Tibial component
STAR[34]	27 (26)§	1.3	3.8% (1)	Talar component
STAR[35]	22 (19)§	3.1	0% (0)	Tibial and/or talar component
STAR[36]	74 (68)§,††	3.7	13.2% (9) 1.4% (1) 14.6% (10)	Tibial component Talar component TOTAL
STAR[37]	200 (200)	3.8	4.0% (8) 0.5% (1) 4.5% (9)	Tibial component Talar component TOTAL
STAR[38]	49 (45)	2.3	4% (2)	Tibial component
SALTO[39]	98 (95)§	2.9	1.0% (1)	Talar component
HINTEGRA[40]	278 (266)‡‡	2.8	0.8% (2) 6.0% (16) 6.8% (18)	Tibial component Talar component TOTAL

*: referred study; n: number of ankles in study; X-FU: number of ankles evaluated radiographically; Mean FU (yrs): mean follow-up duration in years; (n): number of components with signs of loosening.

Details of some studies: * Only ankles with a minimum of 2 years of follow-up. † Mean radiographic follow-up. ‡ Revised prostheses excluded from analysis. § Some subjects lost-to-follow-up. || Only information of reoperated ankles presented. ** Subjects with very short follow-up excluded, 1 subject with deep infection excluded. †† 2 revision arthroplasties (other prosthesis previously implanted) excluded. ‡‡ 5 prostheses revised to arthrodeses excluded.

11.2.4 Radiostereometric Analysis

Radiostereometric analysis (RSA) is a highly accurate technique for measuring micromotion, and an important tool for the assessment of implants[54]. So far, two studies have reported the results of RSA after TAA. Carlsson et al[55] carried out a RSA study with the tibial and talar components of the STAR prosthesis in 10 patients, 5 with a diagnosis of rheumatoid arthritis and 5 with osteoarthritis. Patients were followed for 4 (3-5) years. There was no difference in results between ankles operated on for rheumatoid arthritis or for osteoarthritis. A rapid initial migration was observed for the tibial component at 6 weeks, thereafter all but 1 implant seemed stable. The migration pattern for the talar component was similar. Rotation around the 3 axes was observed for the tibial component at 6 weeks, but not thereafter. The talar component became rapidly stable for rotation around the longitudinal and sagittal axes, but stable fixation around the transverse axis was not always observed. Nelissen et al[26] carried out a RSA study of the stability of the tibial component of the Buechel-Pappas prosthesis in patients with RA. Twelve completed a 1-year and eight a 2-year follow-up. An initial migration was observed during the first 3 months, but the component stabilized at 6-month. Mean lateral-medial migration was 0.8 mm, distal-proximal migration was 0.9 mm, and posteroanterior migration was −0.5 mm. This implies that total resultant migration was a proximal, anterior and valgus tilting of the tibial component. Both studies show that uncemented ankle prostheses can safely be used, and that implant fixation also occurs in the softer rheumatoid bone. No RSA studies have been carried out with 2-component ankle designs. By analogy with the study by Garling et al[56] on micromotion of fixed and mobile-bearing total knee replacements, it can be hypothesized that there is less variability in micromotion with mobile-bearing ankle designs.

11.3 Discussion

Due to their favorable biomechanical characteristics and their surgical advantages, 3-component mobile-bearing prostheses appear to be the most suitable concept for the endoprosthetic replacement of the arthritic ankle. A specific disadvantages of mobile-bearing designs is the risk of bearing subluxation. The incidence of this complication should be reduced by a good patient selection and by proper surgical technique. When comparing the characteristics of the specific designs, until now, no implant can claim that it will perform better than its competitors, and furthermore, randomized clinical studies comparing the outcome of different designs have not been done. However, from a clinical point of view, a deep-sulcus trochlear design of the talar component might have certain advantages, such as a reduced risk of bearing subluxation and also, as the result of the greater thickness of the polyethylene bearing, a reduced risk of fracture and wear of the bearing.

Range of motion of the replaced ankle does not fully normalize. In general, postoperative range of motion will be sufficient for level walking, but certain activities such as stair climbing and descending might still remain compromised. Ankle scores at follow-up show a significant gain compared to the preoperative level, indicating that functional results of TAA are relatively good. Sufficient long-term data of mobile-bearing TAA are still lacking, as only a few studies have a mean follow-up of more than five years. Survival data from the referred clinical studies and from the arthroplasty registers show that an eight-year survival rate of about 90 per cent can be expected with use of mobile-bearing designs and implanted by experienced surgeons. However, overall survival after TAA is somewhat inferior to the survival of total knee and total hip arthroplasty. For example, Robertson et al[57], in a study from the Swedish Knee Arthroplasty Register, found an 8-year survival rate of primary total knee arthroplasty of about 95 percent. A study from the Norwegian Arthroplasty Register showed a 9-year survival rate of around 95 per cent with the cemented Charnley and Exeter total hip prosthesis[58]. And Garrellick et al[59], in a prospective randomized study of two cemented hip prosthesis, found an 11-year survival rate of 96 respectively 93 percent. They also showed, that data from specialized centers tend to be better than data with the same implants from a national register.

Aseptic loosening remains the most important mode of failure of TAA. Radiographic loosening of the metallic components is a secondary end point of failure, and with time will usually lead to revision surgery. Current semiconstrained fixed-bearing designs show a relatively high incidence of component loosening at intermediate term follow-up, whereas most studies with mobile-bearing designs show a lower incidence of loosening. Furthermore, it is encouraging that with improved coating characteristics better osseointegration has been demonstrated[37]. Radiostereometric analysis has demonstrated an early migration of both the tibial[55,26] and talar component[55]. The migration pattern of the tibial component was a tilt in an anterior direction, the talar component sometimes lacked stable fixation around a transverse axis. In general, this migration stabilized at 3 to 6 months after surgery.

Stengel et al[19] concluded from their meta-analysis that TAA with use of a mobile-bearing implant provides an acceptable benefit-risk ratio. Haddad et al[60] carried out a comparative meta-analysis on the outcome of ankle arthrodesis and of currently used ankle prostheses. The main reason for failure after arthrodesis was nonunion, and after arthroplasty it was aseptic loosening. They concluded that the intermediate outcome of TAA appears to be similar to that of ankle arthrodesis, although they also noted that there was a higher risk of below-knee amputation after arthrodesis compared with arthroplasty: 5 versus 1 percent. SooHoo and Kominski[61], in a cost-effectiveness analysis, showed that TAA has the potential to become a cost-effective alternative to ankle fusion. Both Stengel et al[19] and Haddad et al[60] stated that randomized studies comparing ankle arthrodesis with arthroplasty are needed.

In summary, at intermediate term follow-up, TAA with use of a mobile-bearing design has a relatively good clinical outcome, equal to or better than ankle arthrodesis. Currently, the survival rate of this procedure is still lower than that of total hip or total knee arthroplasty. However, it has the potential to become a reliable treatment option for the arthritic ankle, and the more so if carried out by an experienced surgeon and after proper patient selection.

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Chapter 12

Preoperative Evaluation of Patients with End-Stage Ankle Arthritis and Recommended Surgical Technique of Total Ankle Arthroplasty

12.1 Preoperative Evaluation

A good result of TAA starts with proper patient selection and an adequate preoperative workup. Many inferior results of TAA could, at least in part, be attributed to improper patient selection and insufficient preoperative evaluation. Therefore, a complete medical history and physical examination, followed by an adequate preoperative imaging and planning are of paramount importance in order to obtain a good result. Any adverse phenomenon, such as a comorbidity influencing the neurovascular function of the lower leg (e.g. smoking, hypertension, diabetes mellitus), a history of wound healing disturbance, or a history of latent or overt infection should be recorded and thoroughly evaluated. A systematic assessment of the functional capacities should be done, preferably by filling out a commonly used ankle score like the American Orthopaedic Foot and Ankle Society score[1]. For research purposes, preferably a self-assessment instrument like the Foot Function Index[2] or the from this index derived Ankle Osteoarthritis Scale[3] should be used in addition to the physician assessment instruments.

Physical examination should start with an assessment of the alignment of both legs, with specific attention for the alignment of the ankle and hindfoot, followed by an evaluation of all the joints of lower leg. The active motion of the ankle and hindfoot should be recorded and the stability of the ankle joint should be tested. The skin condition should be evaluated, including scars from any previous surgery. Finally, the neurovascular status should be assessed, and in the event of abnormalities an evaluation by the neurologist and/or vascular surgeon is indicated.

After completion of the physical examination, adequate imaging is an essential next step. For preoperative imaging, standard anteroposterior and lateral weightbearing radiographs of the ankle and the whole foot are recommended. In the event of suspected or apparent bone loss a computer tomography (CT) scan with high-quality image reconstructions in the frontal and sagittal plane should be done, as only by this technique a trustworthy assessment of the osseous situation can be made. In the event of suspected or apparent tendon and/or muscular abnormalities a magnetic resonance imaging (MRI) scan should be done for a systematic assessment of these structures.

Contraindications for TAA are considered to be: neurovascular insufficiency (e.g. heavy smokers), a history of active or latent infection, gross deformity in the frontal or sagittal plane, substantial loss of bone stock, and patients who are noncompliant or are willing to function at a high level of physical activity.

12.2 Disease-Specific Characteristics and Recommendations

In view of the causes of ankle arthritis, the main indications for reconstructive surgery are posttraumatic arthritis (either post-fracture or instability arthritis) and inflammatory joint disease. Each of these indications has its disease-specific characteristics.

Patients with post-fracture arthritis frequently have undergone previous surgery, resulting in the presence of scars and often a reduced range of motion. If the fracture has healed in malunion, this could influence the alignment of the ankle-hindfoot complex. Some shortening of the lateral malleolus can be acceptable and should not be regarded as a contraindication for arthroplasty. Gross malalignment should be corrected as a first step by for example a supra-malleolar osteotomy before TAA is carried out. A widening of the ankle mortise due to syndesmotic injury should be corrected by performing a synostosis between fibula and tibia at the time of arthroplasty. In general, after an ankle fracture, the local bone stock has remained intact, so that implantation of the prosthetic components can be done without severe difficulty.

Instability arthritis is caused by chronic lateral ligament laxity. Due to this instability the talus has a tendency to tilt into varus. This persistent instability creates a problem when arthroplasty is considered, as for a good result of a mobile-bearing prosthesis stability in neutral alignment is a prerequisite. Restoration of alignment can be done by either a lateral ligament reconstruction or a medial ligament release. As an alternative to medial ligament release we developed a medial malleolar lengthening osteotomy for the restoration of neutral alignment (described in chapter 6). Mid-term results with this medial malleolar lengthening technique are encouraging[4]. Furthermore, with instability arthritis there is a high preoperative incidence of anterior subluxation at the ankle joint (Fig. 1).

12

Fig. 1 Preoperative lateral weight-bearing radiograph showing anterior subluxation of the talus with respect to the longitudinal axis of the tibia in a patient suffering from instability arthritis

Fig. 1

It is of paramount importance, that such an anterior subluxation is corrected. To achieve a proper alignment in the sagittal plane, we recommend implantation of the tibial component at 90 degrees to the vertical axis of the tibia in the sagittal plane, thus with no or only a minimal anterior slope (Fig. 2).

Fig. 2 Postoperative lateral radiograph of the same patient as in ***Fig. 1***, showing correct axial position of the talus with respect to the distal tibia.

Fig. 2

An increased anterior slope of the tibial component in an unstable ankle could lead to anterior subluxation of the replaced ankle, and thereby to edge-loading of the polyethylene bearing and/or to anterior tilting of the tibial component (Fig. 3).

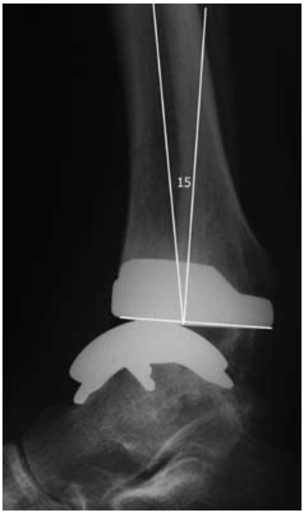

Fig. 3-A

Fig. 3-B

Figs. 3-A and 3-B Lateral radiographs of a patient with instability arthritis. ***Fig. 3-A*** An increased anterior slope of the tibial component is visible, leading to anterior subluxation and eccentric loading of the tibial component. ***Fig. 3-B*** He developed early tilting and aseptic loosening of the tibial component, requiring revision of the tibial component.

Some stemmed tibial components have an anterior slope incorporated, and extra attention should be paid to a correct positioning of such a tibial component. Also, a polyethylene bearing of sufficient thickness should be used as to adequately tension the stretched collateral ligaments.

Inflammatory joint disease (rheumatoid arthritis in particular) is characterized by polyarticular joint destruction, by osteopenia and by moderate to severe impairments of the patient. Localized bone loss and the formation of bone cysts occurs frequently and should be evaluated by preoperative CT scan (Fig. 4).

Fig. 4-A

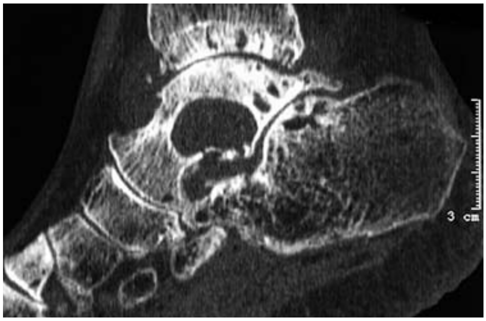

Fig. 4-B

Figs. 4-A and 4-B Preoperative radiographs of a 68-year old female patient with long-standing rheumatoid arthritis. ***Fig. 4-A*** Lateral weightbearing view of the foot and ankle, showing severe arthritic changes in the ankle and hindfoot but no gross osseous pathology. ***Fig. 4-B*** Computer tomography with reconstruction in the sagittal plane, showing a large cyst in the talar body, probably not visible on the standard radiographs due to overprojection.

Furthermore, the hindfoot frequently develops a valgus deformity with a concomitant lowering of the longitudinal arch of the foot due to midfoot arthritis. Correction of such foot deformities is mandatory for a good long-term result of TAA. An elevated failure rate has been demonstrated in the event of a persistent hindfoot deformity[5]. Due to the osteopenia an increased risk of an intra-operative fracture exists, necessitating a careful surgical technique.

12.3 Recommended Surgical Technique

After proper patient selection and informed consent, the patient can be scheduled for surgery. I prefer to position the patient supine with the thigh in a leg holder in order to flex the knee and to position the lower leg in neutral rotation (Fig. 5).

Fig. 5

Fig. 5 Recommended positioning of the lower leg with the thigh supported by a leg holder. The knee is flexed, the lower leg is in a neutral position and the calf muscles are relaxed.

Also, I prefer to start with the tourniquet deflated to reduce tourniquet time in an attempt to reduce wound healing complications and inflate the tourniquet usually at the start of or during the osseous preparation. A midline anterior approach is recommended. After skin incision, branches of the superficial peroneal nerve should be identified, and, if necessary, mobilized to the lateral side. After incision of the extensor retinaculum, the interval between the anterior tibial and the extensor hallucis tendons is used to reach the anterior capsule. Transverse vessels should be ligated. The anterior capsule is opened medially and mobilized to the lateral side. Bone preparation starts with removal of osteophytes at the anterior aspects of the distal tibia and the talus. For a correct distal tibial resection it is recommended to start with a short vertical osteotomy with use of a reciprocating saw at the base of the medial malleolus (Fig. 6).

Fig. 6

Fig. 6 This image shows the first step in the osseous preparation of the distal tibia. With aid of a reciprocating saw a small vertical osteotomy is made in the distal tibia at the base of the medial malleolus.

Then a limited resection of the articulating surface of the distal tibia is carried out. Sizing of the distal tibia is done by measuring the AP distance of the resection plane, for example with a depth gauge, and by the use of templates. One should aim for full coverage of the tibial resection plane by the tibial component. Tibial and talar preparation is done according to the specific recommendations of the manufacturer of the prosthesis. Special attention should be paid to a correct positioning of the prosthetic components, in order to prevent edge-loading of the polyethylene bearing and impingement with the malleoli. Furthermore, the talar component should not be oversized, as this also increases the risk of edge-loading and impingement. Optimal stability of the replaced joint should be aimed for by implanting a bearing of adequate thickness. Careful closure of the extensor retinaculum is an important final step to reduce the risk of wound healing problems. In the event of an equinus deformity, percutaneous achilles tendon lengthening should be done. Routine aftertreatment consists of 2-3 days of bed rest, followed by 4 weeks of immobilization in a below-knee plaster cast, with weightbearing to tolerance being allowed for. A major objective of cast immobilization is the promotion of proper wound healing. Cast immobilization is usually prolonged to 6 weeks in patients with inflammatory joint disease and instability arthritis. If the procedure is combined with a hindfoot fusion, cast immobilization usually is done for a period of 10 weeks. Delayed wound healing may also require prolonged immobilization in a cast.

A video demonstrating our surgical technique of TAA with use of the Buechel-Pappas prosthesis is available through Video Journal of Orthopaedics[6]. See their website for more information: www.vjortho.com.

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Chapter 13

General Discussion

13.1 Introduction

The ankle joint links the foot to the lower leg. It is one of the essential joints of the lower extremity and plays an important role in normal human bipedal locomotion. A disturbed ankle function will therefore, depending on the severity of the pathology, inevitably compromise gait to a certain extent. Traumatic injuries, either fracture or ligament rupture, are the most frequent causes of ankle arthritis[1,2,3]. Other common causes are repetitive sport injuries and inflammatory joint disease. The incidence of sports-related injuries of the ankle is relatively high. In contrast to the hip and the knee, idiopathic degenerative osteoarthritis of the ankle joint is fairly uncommon. Apparently, the normal ankle joint has a high resistance to degeneration[4]. Usually, symptoms caused by ankle arthritis remain at a tolerable level for a long time, until, after many years, severe cartilage damage might develop, leading to end-stage arthritis. As these pathologies frequently arise at a younger age, patients with end-stage arthritis of the ankle tend to be younger than patients with degenerative arthritis of the hip or the knee. Impairments related to end-stage ankle arthritis can have a severe impact on the ability of patients to fulfill their personal and social obligations and daily activities[5], as can furthermore be demonstrated by the low ankle scores of patients requiring surgical reconstruction.

For pain relief in patients with ankle arthritis, several treatment options are available. Local conservative treatment can be either corrective shoe wear (such as a rocker sole and/or a high shoe) or immobilization by an ankle-foot orthosis. Reconstructive surgery for the diseased ankle joint exists either of a surgical fusion (ankle arthrodesis) or of prosthetic replacement. Both treatments have their advantages and disadvantages. Due to inferior results with first-generation ankle prostheses[6], ankle arthrodesis has long been considered the “gold standard” for the surgical treatment of the diseased ankle[7,8,9]. Until now, despite promising clinical results with modern, so-called third-generation designs (Buechel, Buechel, Pappas[10]; Wood, Prem, Sutton[11]; chapter 4), throughout the world many members of the orthopaedic community consider prosthetic replacement of the ankle a second-best choice, only suitable for elderly people living a sedentary life or for patients suffering from inflammatory joint disease (IJD). However, patients with end-stage ankle arthritis are often quite reluctant to have their painful ankle fused, and frequently accept symptoms and disability instead of choosing for a painfree but stiff ankle. It is therefore of interest to critically evaluate the results of total ankle arthroplasty (TAA) with use of modern designs.

The aims of this thesis were to assess the medium to long-term clinical and radiographic outcome of mobile-bearing TAA, and to assess the functional outcome of TAA by investigating gait pattern and energy expenditure during level walking after successful unilateral TAA. Furthermore, assessment of the migration pattern of the tibial component by radiostereometry was considered to be of importance. Finally, the development of a reliable surgical technique for the endoprosthetic replacement of the arthritic ankle with varus deformity and the evaluation of the results of salvage arthrodesis after failed TAA were seen as important issues.

13.2 Design Characteristics and Survival of Total Ankle Arthroplasty

Total ankle arthroplasty can be seen as a real challenge for the foot and ankle surgeon. Both in the near past and nowadays, patients are expecting that artificial joint replacement should be made available to them, and that this should be a safe treatment option, not only for the arthritic hip and knee, but also for other essential joints in the human body like the shoulder, elbow and ankle. Thus, after the introduction on a larger scale of artificial hip and knee arthroplasty in the last decades of the 20th century, total ankle arthroplasty with use of first-generation designs was introduced, in an effort to alleviate pain and retain motion of the diseased ankle. However, in contrast to the success of total hip and total knee arthroplasty, total ankle arthroplasty did not at all fulfill the expectations of the pioneers in this field. Disappointing short-term results with all first-generation designs introduced on the market have been published[12,13,14,15,16]. Until today, this historic burden still rests on those who try to re-introduce this surgical treatment for the ankle with end-stage arthritis.

These first-generation designs have been developed without a thorough awareness of the complex biomechanics of the ankle joint, nor of the high loads exerted on a relatively small joint, nor of the specific pathological changes that occur in end-stage ankle arthritis. Thus, the inferior outcome of the first-generation ankle prostheses can easily be explained as inherent to the use of constrained designs in a joint that was regarded to behave like a simple hinge joint[17], but in fact does not. In the USA, the semiconstrained Agility prosthesis (DePuy, Warsaw, Indiana, USA) nowadays is the most widely used implant (mainly due to government restrictions on the use of mobile-bearing designs), and it seems to have an acceptable outcome[18], although a relatively high rate of complications and reoperations has been reported[19]. Outside the USA this prosthesis has not gained wide acceptance. Besides design aspects, these inferior results are also related to the necessity to fuse the tibiofibular syndesmosis.

The work on the clinical part of this thesis actually started in 1988, when the New Jersey Low Contact Stress (LCS®, DePuy, Warsaw, IN, USA) total ankle prosthesis was introduced at the Slotervaart Hospital. Its design rationale was immediately considered as being of great value, and providing a potential solution for the prosthetic replacement of the diseased ankle. The LCS® prosthesis had been designed by Fred Buechel, orthopaedic surgeon, and Michael Pappas, engineer, and early clinical experience had just been reported[20]. It was the first prosthesis applying the mobile-bearing concept to an ankle prosthesis. This mobile-bearing concept had been applied earlier to the endoprosthetic replacement of the knee joint by Goodfellow and O'Conner when they introduced a so-called meniscal-bearing unicompartmental prosthesis with use of a mobile intercalated polyethylene bearing[21]. Their objective was to create a prosthesis with low wear characteristics by congruency of the articulating surfaces, and low intrinsic constraints by making use of the crucial role the ligaments play in normal joint kinematics. Independently from Goodfellow and O'Connor, Buechel and Pappas had also developed a mobile-bearing total knee prosthesis, the LCS® knee prosthesis (DePuy, Warsaw, IN, USA). Early clinical results with both designs were promising, and both designs are still in clinical use more than 30 years after their introduction[22,23]. These mobile-bearing 3-component prostheses have distinct biomechanical and kinematic advantages over two-component designs, as there is full congruency of the articulating surfaces without restriction of rotational motion by the prosthesis. For the prosthetic replacement of the knee joint however, no distinct clinical advantages of mobile-bearing designs have been demonstrated[24,25,26,27], except that probably less patellar symptoms are present[28].

Almost all designs currently in use outside the USA are cementless 3-component, so-called third-generation designs. They have a mobile ultra-high-molecular-weight polyethylene bearing intercalated between metallic tibial and talar components. This mobile bearing reduces the rotational and translational constraints significantly, and can thus be seen as a more physiological solution for the endoprosthetic replacement of the ankle joint. Several medium to long-term studies with use of such designs are now available, showing that good results can be obtained with survival rates of between 80 and 92 per cent at 10 to 12 years of follow-up (Buechel, Buechel, Pappas[10]; Wood, Prem, Sutton[11]; chapter 4). It is of great interest to note that, in comparison to the situation in total knee arthroplasty, where no distinct advantages have been demonstrated of mobile-bearing over semiconstrained fixed-bearing designs, the arthritic ankle joint apparently favors a mobile-bearing design. Thus, it seems that rotational and translational stresses at the bone-prosthesis interface are less well tolerated in the ankle than in the knee.

13.3 Patient-related and Surgery-related Factors Influencing Implant Survival

The most important patient-related risk factor for failure of TAA is a preoperative deformity in the coronal plane of more than 10 to 15 degrees. This has been identified in chapter 4, and was also seen in other studies[29,30]. Haskell and Mann[29] categorized coronal plane deformity into congruent (< 10º difference between tibial and talar alignment) and incongruent (≥ 10º difference between tibial and talar alignment). Varus deformity in particular can be difficult to correct, as in this situation the lateral ligaments are often attenuated, leading to persistent instability and edge-loading of the replaced ankle. Several procedures have been developed for the correction of arthritic ankle with varus deformity. Kofoed[31] ad vised a talar-sculpturing technique for correction of coronal plane deformity, if necessary combined with lateral ligament reconstruction to improve stability in the varus ankle. No details on the outcome with this technique were given. Bonnin et al.[32] reported good results after extensive deltoid ligament release in a subgroup of 14 varus ankles, as part of a study on 93 ankle replacements. Their follow-up was short: 35 months, and no specific details were given of this subgroup concerning diagnosis, preoperative alignment and amount of correction obtained at surgery. Haskell and Mann[29] advised either to stabilize the varus ankle by lateral ligament reconstruction, as proposed by Kofoed, or to realign it through a deltoid ligament release, or a combination of these procedures. They reported good results after reconstruction in congruent ankles, but had a high rate (4 out of 10) of early edge-loading after TAA in varus-incongruent ankles. For the correction of varus deformity during total ankle arthroplasty we developed a medial malleolar lengthening technique in 1998. Good medium-term results have been described with this technique (chapter 6). In recent years, several other ankle surgeons in Europe and Japan have started to use this technique, and they report a similar outcome.

Henricson et al.[33], in their register-based study, reported that gender and diagnosis were no risk factors for failure, whereas lower age was found to have an increased risk for revision surgery. However, they did not give the cut-off point for lower age, neither did they specify which were the reasons for revision in the lower age group. Fevang et al.[34], in their register-based study, found no influence of lower age on revision rate.

Surgery-related risk factors for failure can be divided into surgeon-related factors and design-related factors. The most important surgeon-related factor influencing outcome is experience (or volume). Also for TAA, as with other procedures, experience of the surgeon with this procedure is probably an important factor influencing outcome. The difference between TAA and for example total hip arthroplasty is that, although less experienced surgeons usually have more complications[35], this higher complication rate will not necessarily lead to a higher revision rate in total hip arthroplasty. On the contrary, however, TAA performed by less experienced surgeons usually has a lower survival[33]. In chapter 4 we found that an undersized tibial component, which could be considered as a parameter of lower surgeon experience, was a risk factor for failure. Hosman et al.[36], in their register-based study, found that ankles with a longer operative time or an unfavorable clinical score at 6 months postoperatively had a higher failure rate. These parameters too might reflect lower surgeon experience.

Design-related risk factors for an unfavorable outcome, such as unsatisfactory instrumentation and bearing fracture, have been recognized with the STAR prosthesis (Waldemar Link, Hamburg, Germany)[33,37]. Hosman et al.[36] reported lower clinical scores with the Ramses total ankle arthroplasty (Laboratoire Fournitures Hospitalieres, Heimsbrunn, France).

13.4 Radiostereometry

Radiostereometry (RSA) can be considered the gold standard for assessment of implant stability, and a predictor for long-term implant survival[38]. Carlsson et al.[39] were the first to carry out a RSA study of a total ankle prosthesis. They investigated the stability of ten double-coated STAR prostheses in a marker-based RSA study, implanted with use of optimized instrumentation. The cementless prosthesis was the latest version of this design, and was provided with a modern coating on the anchoring surface in order to improve osseointegration. The double coating consisted of a 300-μm-thick ground layer of porous titanium, on top of which a 25-μm-thick hydroxyapatite layer had been applied. The tibial and talar components had been supplied with 3 tantalum markers, mounted on short metallic towers. They found a rapid initial migration along and rotation around all 3 axes for both the tibial and talar components, but stabilization thereafter, except for rotation of the talar component in the forward-backward direction in several ankles. No difference in migration pattern was found between the osteoarthritic and rheumatoid ankles in their cohort.

We investigated the stability of the tibial component of the cementless porocoated Buechel-Pappas prosthesis (Endotec, South Orange, NJ, USA) in a rheumatoid population (chapter 5). One single marker on the top of the tibial stem was used to measure migration. We observed migration of the tibial component during the first 3 months, but this stabilized by the 6 month follow-up. Mean lateral-medial migration was 0.8 mm, distal-proximal migration was 0.9 mm, and posteroanterior migration was −0.5 mm. This implies a total resultant migration in anterior and valgus tilting of this tibial component. We believe that the anterior cortical window, necessary for placement of the tibial component could explain this migration.

Both the study by Carlsson et al.[39] and our study show that cementless mobile-bearing ankle prostheses, in the patient groups most frequently treated by TAA, stabilize after a short initial period of migration. However, the migration pattern in the ankle appears to be somewhat less favorable compared to the migration pattern of hydroxyapatite-coated tibial components in the knee, where in most cases a stabilization phase occurs more rapidly[40].

13.5 Functional Aspects of the Normal and Reconstructed Ankle

The kinematics of the ankle joint are complex: it follows a motion pattern around a helical axis in the sagittal plane, combined with a rotation around a vertical axis[41,42]. Furthermore, there is a large inter-individual variance of this kinematic pattern of both the ankle and the tarsal joints. It is also important to keep in mind that the kinematics of the ankle and tarsal joints are closely linked in the loaded situation[43,44]. Therefore, loss of motion at the ankle joint is expected to lead to an abnormal kinematic pattern at the tarsal joints. That an abnormal tarsal kinematic pattern indeed exists after ankle arthrodesis was confirmed by Wu et al.[45]. They demonstrated significant shifts in the hindfoot-to-tibia and the forefoot-to-hindfoot motion patterns during barefoot walking, 1½ year (range, 0.5 to 4 years) after unilateral ankle arthrodesis. Suckel et al.[46] studied the forces in the midtarsal joint after ankle arthrodesis during simulated stance in a cadaver experiment. After ankle arthrodesis, a tendency of force redistribution from the lateral onto the medial column of the foot was found. The mean force increased upon arthrodesis in the talonavicular joint and decreased in the calcaneocuboid joint. Also the peak pressure in the talonavicular joint increased. These kinematic and kinetic changes after ankle arthrodesis can very well explain the high incidence of hindfoot arthritis that has been found by several authors. Compared to the situation after ankle arthrodesis, tarsal kinematics after TAA are close to normal, as was shown by Valderrabano et al.[47], and in the gait study described in chapter 8. Probably therefore, in contrast to the situation after ankle arthrodesis, TAA could protect the tarsal joints from becoming overloaded and from developing secondary hindfoot arthritis. Also, after TAA, there might be a diminished risk of co-existing tarsal arthritis to become clinically symptomatic (Hintermann, personal communication), although follow up of current studies has been to short to elucidate on this.

Besides the kinematic aspects mentioned above, load transmission across the ankle joint is another important aspect after prosthetic replacement. With use of data from the same study as described in chapter 8, we found that after successful unilateral TAA, compared to healthy controls, no differences could be observed in peak moments and in joint quasi-stiffness of the ankle (chapter 9). These findings indicate that the function of the ankle joint in terms of mechanical load and stiffness appears not to be influenced by TAA.

Energy cost of walking is another important characteristic of gait, and is known to be influenced by disturbances in ankle function. Energy cost is increased in immobilized and in surgically fused ankles[48,49,50]. As no information existed on energy cost of walking after TAA a study was done on energy cost and external mechanical work during barefoot walking after successful unilateral TAA (chapter 10). Self-selected walking speed was 12% lower in the study group, but step length and step frequency were similar. Compared to a control group of healthy subjects, metabolic cost of transport was significantly higher in the study group at a fixed walking speed (28%). It was also found that, during the step-to-step transition, positive external work generated by the trailing TAA leg was reduced, simultaneously with an increase in negative external mechanical work during heel landing of the intact leading leg. These findings are in accordance with the double inverted pendulum model of walking, as constructed by Donelan, Kram and Kuo[51,52]. However, despite the higher energy loss in the collision of the leading leg in the step-to-step transition in TAA subjects, the absolute external mechanical work over a complete step or stride did not differ between controls and patients. So, mechanical factors can not fully explain the increase in metabolic cost. Other factors that might explain, at least in part, the increased metabolic cost are co-contraction of the lower leg muscles, as was described in chapter 8. Furthermore, persistent muscle fiber atrophy after TAA could also play a role. In muscle atrophy as a consequence of chronic joint disease, type 1 muscle fibers, which are more energy efficient, are lost predominantly[53]. It is very well possible that the muscle quality remains impaired to a certain level after successful TAA, and this then might compromise energy efficiency of walking. From the study described in chapter 10 it was concluded that after successful TAA patients still experienced an impaired lower leg function, contributing to an increased mechanical energy dissipation during the step-to-step transition and an increase in the metabolic demand of walking.

13.6 Summary and Conclusions

Introduction of cementless unconstrained 3-component mobile-bearing prostheses has improved the outcome of TAA substantially. In comparison to constrained and semi-constrained prostheses, mobile-bearing prostheses allow for a more physiologic motion pattern of the replaced ankle joint and transmit lower torsional and translational stresses to the bone-prosthesis interface[54]. As results with 3-component mobile-bearing prostheses are promising (Buechel, Buechel, Pappas[10]; Wood, Prem, Sutton[11]; chapter 4), these designs are gaining popularity in many countries. An overview of currently available designs, with their published outcome is given in chapter 11. With increasing knowledge of the correct indications and contra-indications for TAA, the pitfalls during the surgical procedure (as described in chapter12), and with improvements in design and instrumentation, long-term results of mobile-bearing TAA are expected to improve in the years to come.

Several studies have shown that results of TAA are best when carried out by experienced ankle surgeons. However, the learning curve for new users might nowadays be shorter, as surgical techniques have been improved and more precisely described, and also because indications and contra-indications are better defined.

Finally, salvage arthrodesis of failed ankle prosthesis has been shown to be an amenable complication (chapter 7), although it remains difficult to achieve good results with ankle arthrodesis in the rheumatoid patient group.

Currently, for an optimal outcome of TAA, several challenges still exist, requiring additional research. Accurate patient selection, reproducible surgical technique, surgical training, and post-implantation monitoring are topics that have to be addressed by the orthopaedic community in order to improve the outcome of total ankle arthroplasty. Introduction of national registers in more countries would probably improve the quality of the outcome of TAA and would give a better and unbiased information of the performance of individual designs. Also, more RSA studies should be carried out, as this is the best method to give objective and accurate information of implant stability. Finally, the ultimate study for a thorough assessment of the best treatment for end-stage ankle arthritis would be to compare in a randomized setting the outcome of total ankle arthroplasty with that of ankle arthrodesis. However, to conduct such a study nowadays seems questionable in terms of ethics and of freedom of choice of the individual patient to choose his or hers preferred treatment: either a long-lasting painfree but stiff joint or a mobile joint that might require secondary surgery.

In summary, if proper indications are used and if a correct surgical technique is applied, total ankle arthroplasty with use of a mobile-bearing prosthesis is a reliable reconstructive procedure for end-stage ankle arthritis, and an alternative for ankle arthrodesis. Restoration to the level of a normal functioning ankle in terms of walking speed, muscle function and energy expenditure is not fully achievable, but a significant improvement of walking capacity can usually be obtained. Results in ankles with a neutral preoperative alignment are better than in ankles with a preoperative deformity in the frontal plane. Therefore, ankles with a preoperative deformity can best be treated by experienced ankle replacement surgeons if TAA is considered.

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Chapter 14

Summary

Samenvatting

Résumé

Summary

Ankle arthritis often leads to significant impairments for the patient. As total ankle arthroplasty (TAA) with use of fixed-bearing (2-component) total ankle prostheses has a high rate of early failures, fusion of the ankle joint is, until today, considered to be the standard surgical treatment for end-stage ankle arthritis. TAA with use of a mobile-bearing design became available in 1981 with the development of the New Jersey Low Contact Stress (LCS®, DePuy, Warsaw, Indiana) ankle prosthesis in the USA and in 1986 with the development of the Scandinavian Total Ankle Replacement (STAR, Waldemar Link, Hamburg, Germany) prosthesis in Denmark. The LCS ankle prosthesis was introduced in 1988 at the Slotervaartziekenhuis, Amsterdam, this hospital thus becoming the first non-designer hospital in Europe to use a mobile-bearing ankle prosthesis. With growing experience, this method became our preferred surgical treatment for the arthritic ankle with end-stage disease, at first in patients with inflammatory arthritis, later also in patients with osteoarthritis (mostly of posttraumatic origin). Since TAA as a reliable treatment option is, until today, still subject of debate, an in-depth evaluation of its clinical, radiographic and functional outcomes was performed in order to properly assess its current position in the armamentarium of the modern foot and ankle surgeon.

In ***chapter 1*** the aims and outline of this thesis are presented. Unconstrained 3-component prostheses that use a mobile polyethylene bearing have excellent biomechanical characteristics. Potentially therefore, the mobile-bearing concept offers a promising solution for the endoprosthetic replacement of the ankle joint. This thesis is divided into four sections: first an overview of the normal anatomy and kinematics, the pathology of the arthritic ankle, and the alternative treatment options of symptomatic ankle arthritis; second a clinical and radiographic outcome section of mobile-bearing TAA; third a functional outcome section; finally a review and discussion section.

Chapter 2 describes the anatomy and the kinematics of the normal ankle joint, followed by a description of the pathology of the arthritic ankle. In contrast to arthritis in other joints of the lower extremity, the most frequent causes of ankle arthritis are posttraumatic disorders and inflammatory joint disease. End-stage ankle arthritis mostly develops as a late sequel of either local or systemic joint disease. Furthermore, patients with ankle arthritis are generally younger than those suffering from end-stage arthritis of the hip and knee. These factors have consequences for the choice of both the optimal timing and the best modality of surgical treatment of end-stage ankle arthritis.

Chapter 3 describes the conservative treatment options for ankle arthritis

and the result of nonendoprosthetic surgical treatment. If conservative treatment is unsuccessful, for patients with moderate arthritis that have symptoms from osseous impingement, an arthroscopic debridement is probably the best surgical treatment. In the event of a deformity in the frontal plane, a corrective osteotomy should be considered, either at the level of the distal tibia or of the calcaneus. For end-stage disease, tibiotalar fusion is mostly considered to be the *"gold standard"*. If successful, it produces a painfree and stable ankle. Complications that have been reported, are non-union, malunion and infection. With longer follow-up, however, ankle arthrodesis is shown to carry a high incidence of hindfoot arthritis, and thus a significant risk of recurrent symptoms.

The next four chapters of this thesis are dedicated to the clinical and radiographic outcome of TAA and the salvage of failed TAA.

Chapter 4 describes the medium to long-term clinical and radiographic results of mobile-bearing TAA in a prospectively followed consecutive cohort of patients suffering from inflammatory joint disease. We conclude first that mobile-bearing TAA is a valid treatment option for the rheumatoid ankle if proper indications are applied, and second that aseptic loosening and persistent deformity are the most important modes of failure. An increased failure rate was encountered in ankles that had a preoperative deformity in the frontal plane of more than 10º, and in ankles where an undersized tibial component had been implanted.

In ***chapter 5*** the stability of the tibial component of the Buechel-Pappas prosthesis was assessed in a radiostereometric analysis (RSA) study. This study showed an initial tilting upwards, anterior and in valgus, which stabilized at 6 months. The creation of an anterior cortical window, required for placement of the tibial component, and the method of tibial fixation very likely explain this migration pattern.

Chapter 6 describes a new technique, developed in 1998, for the correction of varus deformity at the time of TAA. In a prospective investigation of fifteen ankles, in a mixed population of instability arthritis and inflammatory joint disease treated by this technique, medial malleolar lengthening osteotomy is shown to be an easy and effective technique for the realignment of the varus deformity at the time of TAA. Asymptomatic nonunion of the medial malleolar osteotomy was seen in two rheumatoid ankles that had the osteotomy done at the base of the medial malleolus (six in this series). Furthermore, early aseptic loosening of the tibial or the talar component developed in one ankle each. Residual deformity at the hindfoot, not corrected at the index operation, required subsequent surgery in three feet. In view of these results, TAA with a pre-existing varus deformity still has a somewhat inferior result compared to the outcome in well-aligned ankles. To further improve the outcome of TAA for the varus ankle, the following surgical modifications are proposed: a) the lengthening osteotomy is done halfway down the medial malleolus; b) the medial gutter is debrided routinely; c) the tibial component is implanted with no or only minimal anterior slope; and d) any hindfoot deformity is corrected prior to or simultaneously with the arthroplasty.

Chapter 7 describes the results of salvage arthrodesis of failed TAA in a patient population of eighteen ankles, mainly with inflammatory joint disease. High nonunion rates were seen with salvage of failed TAA in rheumatoid ankles that had been stabilized by screws or by a retrograde nail. By contrast, seven ankles (four rheumatoid ankles and three osteoarthritic ankles) stabilized by a blade plate all united. With the numbers available, however, no significant difference could be demonstrated between the three methods. Second-attempt fusion for a symptomatic nonunion, carried out in four ankles, was successful except in one patient that had an ipsilateral stiff hip. Three patients with a nonunion refused further surgery, as they had few symptoms. It was concluded that in osteoarthritis the union rate of salvage ankle arthrodesis is good, and comparable to the outcome of primary ankle arthrodesis. In rheumatoid ankles, however, both primary arthrodesis and salvage arthrodesis are demanding procedures. Such patients are probably best treated by experienced surgeons working in specialized centers. Stabilization by a blade plate appears to be a promising technique for salvage ankle arthrodesis.

The functional outcome part of this thesis consists of the following three chapters.

Chapter 8 describes a gait analysis study of patients after successful mobile-bearing TAA compared to a matched healthy control group. The patient group had a reduced dorsiflexion of the ankle joint. During barefoot walking, velocity was somewhat reduced in the patient group. This lower walking speed was primarily caused by an increased stride time and to a lesser extent by a decreased stride length. A near normal gait pattern was found in terms of joint kinematics of the knee, ankle, and foot. However, differences were found in the ground reaction forces of the leg with the replaced ankle. Also, the EMG activity pattern of the lower leg muscles with the replaced ankle showed some differences: the medial gastrocnemius was more active during early stance, and during terminal stance activity in the anterior tibia was higher. In contrast to studies on gait after ankle arthrodesis, our results show that, despite a somewhat reduced dorsiflexion, gait kinematics after successful TAA were comparable to normal. This also implies a low risk of secondary overuse injuries to the tarsal joints.

Chapter 9 is a kinetic analysis of the same study population as described in ***chapter 8***. No differences were observed in peak net joint moments and in joint quasi-stiffness of the replaced ankle, but internal work showed small differences, which is in line with the somewhat lower walking speed. This study showed that the mechanical load of the ankle joint during walking after successful ankle replacement does not differ from the mechanical load of the healthy ankle joint. This indicates that the function of the ankle joint in terms of mechanical load and joint quasi-stiffness appears not to be influenced by the ankle replacement.

Chapter 10 describes an energy consumption study during barefoot walking on a treadmill by patients after successful mobile-bearing TAA compared to a healthy control group. At a fixed walking speed the metabolic cost of transport was higher for the patient group. This increase correlated with dissipation of mechanical work during the step-to-step transition. At a self-selected walking speed, which was 12% lower in the study group, no increase in metabolic cost of transport could be found. It was concluded that the lower leg function in terms of energy consumption did not fully normalize after TAA.

The final part of this thesis consists of a meta-analysis of the literature on the prosthetic replacement of the arthritic ankle, the recommended preoperative workup and surgical technique, and a general discussion of the role of TAA in the treatment of the arthritic ankle.

Chapter 11 gives an overview of the currently available designs, followed by a meta-analysis of the literature of the most commonly used fixed and mobile-bearing designs. Range of motion of the replaced ankle does not fully normalize, but generally will be sufficient for proper level walking. Ankle scores at follow-up show a significant gain compared to the preoperative level. Survival data show that an eight-year survival rate of about 90 per cent can be expected with use of mobile-bearing designs when implanted by experienced surgeons. However, overall survival after TAA remains inferior to the survival of total knee and total hip arthroplasty. Compared to ankle fusion, a similar failure rate can be expected.

Chapter 12 discusses the preoperative evaluation of the patient with an arthritic ankle and presents recommendations for an optimal surgical technique of TAA, such as patient positioning, tibial preparation, implant position and correction of deformity.

Based on the work presented in this thesis, ***chapter 13*** gives a general discussion of the current and future role of TAA in the treatment of end-stage ankle arthritis. Full restoration of ankle function is not considered a realistic objective. Diagnosis (whether primary osteoarthritis, posttraumatic arthritis, or inflammatory joint disease), younger age and gender have not been identified as clear risk factors for failure. Risk factors for failure that have been identified are: a preoperative deformity in the frontal plane, limited experience of the surgeon (expressed as low volume surgeons, prolonged surgery time, and an undersized tibial component), and certain implant-specific characteristics. With proper patient selection, and implantation by surgeons dedicated to this topic, good long-term results of TAA with use of mobile-bearing designs can be expected with a ten-year survival of at least 90 per cent. Thus, TAA can be considered a valid treatment option, and, because salvage of the failed TAA is a manageable complication, it can be considered the preferred treatment for most patients suffering from end-stage ankle arthritis.

Recommendations for future research are: randomized studies comparing the outcome of ankle fusion and TAA, and RSA studies. Due to the high accuracy of RSA, only small numbers are needed in such studies. RSA studies of any new designs should be implemented. Finally, adequate monitoring of the result of TAA is of increasing importance owing to the growing number of TAAs performed in The Netherlands and elsewhere in Europe and in the world. Monitoring of the outcome of specific designs can best be performed by the implementation of national registers.

Samenvatting

Enkelarthrose leidt vaak tot duidelijke beperkingen voor de patiënt. Aangezien de totale enkel artrhoplastiek (TEA) met gebruik van vast-lager (2-component) totale enkelprotheses al na korte na-onderzoektijd een hoog faalcijfer kent, wordt het vastzetten van het enkelgewricht (ofwel enkelarthrodese) tot op heden gezien als de chirurgische standaardbehandeling voor ernstige enkelarthrose. De TEA gebruikmakend van een prothese met mobiel lager kwam in 1981 beschikbaar met de ontwikkeling van de New Jersey Low Contact Stress (LCS®, DePuy, Warsaw, Indiana) enkelprothese in de Verenigde Staten, en in 1986 met de ontwikkeling van de Scandinavian Total Ankle Replacement (STAR, Waldemar Link, Hamburg, Duitsland) prothese in Denemarken. De LCS enkelprothese werd in 1988 in het Slotervaartziekenhuis te Amsterdam geïntroduceerd. Het werd daarmee het eerste niet-ontwerper-ziekenhuis in Europa dat een enkelprothese met mobiel lager ging gebruiken. Met toenemende ervaring werd deze methode onze chirurgische behandeling van voorkeur voor ernstige enkelarthrose, eerst bij patiënten met reumatische gewrichtsziekten, later ook bij patiënten met degeneratieve arthrose (meestal van posttraumatische origine). Aangezien tot op heden de TEA als een betrouwbare behandelingsmogelijkheid nog steeds onderwerp van discussie is, werd een diepgaande evaluatie van het klinische, radiologische en functionele resultaat uitgevoerd teneinde de huidige positie ervan in het behandelingsarsenaal van de moderne voet- en enkelchirurg goed te kunnen beoordelen.

In ***hoofdstuk 1*** worden de doelstellingen en het kader van dit proefschrift gepresenteerd. Niet-restrictieve 3-component protheses, die gebruik maken van een mobiel polyethyleen lager, hebben uitstekende biomechanische eigenschappen. In potentie biedt dit concept dan ook een veelbelovende oplossing voor de behandeling met een gewrichtsvervangende prothese van het enkelgewricht. Dit proefschrift is onderverdeeld in vier secties: eerst een overzicht van de normale anatomie en kinematica, de pathologie van de arthrotische enkel, en de alternatieve behandelingsmogelijkheden van symptomatische enkelarthrose; gevolgd door een sectie over het klinische en radiologische resultaat van de TEA met mobiel lager; daarna een sectie over het functionele resultaat; en eindigend met een overzicht- en discussiesectie.

Hoofdstuk 2 beschrijft de anatomie en de kinematica van het normale enkelgewricht, gevolgd door een beschrijving van de pathologie van de arthrotische enkel. In tegenstelling tot arthrose in andere gewrichten van de onderste extremiteit zijn de meest voorkomende oorzaken van enkelarthrose posttraumatische afwijkingen en reumatische gewrichtsziekten. Ernstige enkelarthrose ontwikkelt zich meestal als een laat gevolg van ofwel een lokale aandoening dan wel een systemische gewrichtsziekte. Voorts zijn patiënten met enkelarthrose over het algemeen jonger dan patiënten lijdend aan ernstige arthrose van de heup en de knie. Deze factoren hebben consequenties voor de keuze van zowel het optimale moment als ook de beste wijze van chirurgische behandeling van ernstige enkelarthrose.

Hoofdstuk 3 beschrijft de conservatieve behandelingsmogelijkheden van enkelarthrose en het resultaat van de niet-gewrichtsvervangende chirurgie. Als de conservatieve behandeling niet succesvol is, is voor patiënten met matige arthrose, die ossale impingement klachten hebben, een arthroscopische nettoyage waarschijnlijk de beste chirurgische behandeling. In geval van een standafwijking in het frontale vlak zou een corrigerende osteotomie overwogen moeten worden, hetzij van de distale tibia dan wel van de calcaneus. Voor ernstige enkelarthrose wordt een tibiotalaire arthrodese meestal beschouwd als de *"gouden standaard"*. Wanneer deze behandeling succesvol is, ontstaat een pijnvrij en stabiel gewricht. Bekende complicaties zijn pseudoarthrose, malunion en infectie. Na langere tijd heeft de enkelarthrodese een verhoogd risico op arthrose van de achtervoetgewrichten, en dus op een aanzienlijk risico van het ontstaan van recidiefklachten.

De volgende vier hoofdstukken van dit proefschrift zijn gewijd aan het klinisch en radiologisch resultaat van de TEA en van de hersteloperatie van de gefaalde TEA.

Hoofdstuk 4 beschrijft het klinisch en radiologisch resultaat op middellange tot lange termijn van de TEA met mobiel lager in een prospectief gevolgd cohort van patiënten met reumatische gewrichtsziekte. Wij concluderen dat de mobiel-lager TEA een valide behandelingsmogelijkheid is voor de reumatische enkel op voorwaarde dat de juiste indicaties worden gehanteerd, en verder dat aseptische loslating en blijvende standafwijking de belangrijkste faalmechanismen zijn. Een verhoogd faalrisico werd gevonden bij enkels met een preoperatieve standafwijking in het frontale vlak van meer dan 10º, en verder in enkels waar een ondermaatse tibiacomponent was geïmplanteerd.

In ***hoofdstuk 5*** is de stabiliteit van de tibiacomponent van de Buechel-Pappas prothese beoordeeld met behulp van een RSA–studie (radiostereometrische analyse). Deze studie toonde aan dat er sprake was van een initiële kanteling naar boven, naar voren en in valgus, die stabiliseerde na 6 maanden. Het maken van het corticale luik, noodzakelijk voor het plaatsen van de tibiacomponent, en de methode van fixatie van de tibiacomponent kunnen goed dit migratiepatroon verklaren.

Hoofdstuk 6 beschrijft een nieuwe techniek, ontwikkeld in 1998, voor de correctie van varusdeformiteit tijdens de TEA. In een prospectief onderzoek van 15 enkels, in een gemengde populatie van instabiliteitsarthrose en reumatische gewrichtsziekte die met deze methode waren behandeld, werd aangetoond dat de verlengingsosteotomie van de mediale malleolus een eenvoudige en effectieve techniek is voor de correctie van varusdeformiteit tijdens de TEA. Een symptoomvrije pseudoarthrose van de mediale malleolus werd gezien in twee reumatische enkels waar de osteotomie aan de basis van de mediale malleolus was verricht (zes in deze serie). Verder werd een aseptische loslating van zowel de talus- als van de tibiacomponent eenmaal gezien. Een resterende deformiteit van de achtervoet, niet gecorrigeerd bij de eerste ingreep, noodzaakte tot aanvullende chirurgie in drie voeten. Gezien deze resultaten kan gesteld worden dat de TEA bij een pre-existente varusdeformiteit nog een minder goed resultaat heeft in vergelijking met TEA zonder deformiteit. Om het resultaat van de TEA bij de varusenkel verder te verbeteren worden de volgende chirurgisch-technische wijzigingen aanbevolen: a) de verlengingsosteotomie wordt gedaan halverwege de mediale malleolus; b) het mediale talo-malleolaire gewricht wordt routinematig zorgvuldig genettoyeerd; c) de tibiacomponent wordt geïmplanteerd in een neutrale of hooguit in een minimaal naar voren gerichte stand; en d) een deformiteit van de achtervoet wordt vóór of gelijktijdig met de TEA gecorrigeerd.

Hoofdstuk 7 beschrijft het resultaat van de herstelarthrodese na gefaalde prothese in een populatie van achttien enkels, voornamelijk lijdend aan een reumatische gewrichtsziekte. Er werd een hoog aantal pseudoarthroses gevonden bij herstelarthrodese na gefaalde TEA in reumatische enkels die waren gefixeerd met schroeven of met een retrograde pen. Daarentegen werd een benige fusie bereikt in alle zeven enkels (vier reumatisch, drie met degeneratieve arthrose) die waren gefixeerd met een hoekplaat. Vanwege de relatief kleine subgroepen kon er geen significant verschil worden aangetoond tussen het resultaat van de drie fixatiemethoden. Voor symptomatische pseudoarthrose werd in vier enkels een re-arthrodese uitgevoerd. Deze was succesvol in drie enkels, terwijl in een patiënt, met een ipsilateraal ankylotische heup, de pseudoarthrose bleef bestaan. Drie patiënten met een pseudoarthrose hadden geen behoefte aan verdere behandeling vanwege beperkte klachten. We concludeerden dat bij degeneratieve arthrose een herstelarthrodese goede resultaten geeft, vergelijkbaar met het resultaat na primaire enkelarthrodese. In reumatische enkels daarentegen zijn zowel de primaire arthrodese als de herstelarthrodese technisch veeleisende procedures. Dergelijke enkels dienen bij voorkeur te worden behandeld door gespecialiseerde orthopeden die in gespecialiseerde centra werken. Stabilisatie met een hoekplaat lijkt een veelbelovende techniek te zijn voor de herstelarthrodese.

Het deel van dit proefschrift dat het functionele resultaat behandelt, bestaat uit de volgende drie hoofdstukken.

Hoofdstuk 8 beschrijft een gangbeeldstudie bij patiënten na geslaagde mobiel-lager TEA in vergelijking met een bijpassende gezonde controlegroep. De studiegroep had een verminderde dorsaalflexie van de enkel. Tijdens blootsvoets lopen was de loopsnelheid in de studiegroep wat lager. Deze lagere loopsnelheid werd vooral veroorzaakt door een verlengde schredetijd en in mindere mate door een verkorte schredelengte. Een vrijwel normaal gangbeeld werd gevonden in termen van de kinematica van de knie, enkel en voet. Daarentegen werden verschillen gevonden in de grondreactiekrachten in het been met de enkelarthroplastiek. Ook het EMG-activiteitenpatroon van de onderbeenspieren in het been met de enkelarthroplastiek liet enige verschillen zien: de mediale gastrocnemius was actiever in de vroege standfase, en tijdens de late standfase was de activiteit van de tibialis anterior hoger. In tegenstelling tot studies die het gangbeeld na enkelarthrodese beschrijven, laten onze resultaten zien dat, ondanks de verminderde dorsaalflexie, het gangbeeld na geslaagde TEA goeddeels vergelijkbaar is met het normale gangbeeld. Dit impliceert tevens dat er een laag risico is van secundaire overbelastingsletsels van de achtervoetgewrichten.

Hoofdstuk 9 is een kinetische analyse van dezelfde populatie als beschreven in ***hoofdstuk 8***. Er werden geen verschillen gevonden in netto-piek-gewrichtsmomenten en in gewricht-quasi-stijfheid van de geopereerde enkel, maar de interne arbeid liet kleine verschillen zien, mogelijk veroorzaakt door de wat lagere loopsnelheid. Deze studie laat zien dat de mechanische belasting van het enkelgewricht tijdens het lopen na een geslaagde enkelarthroplastiek niet verschilt van de mechanische belasting van het gezonde enkelgewricht. Dit maakt aannemelijk dat de functie van het enkelgewricht in termen van mechanische belasting en gewricht-quasi-stijfheid niet wordt beïnvloed door de enkelvervanging.

Hoofdstuk 10 beschrijft een energieverbruikstudie tijdens blootsvoets lopen op een lopende band bij patiënten na een geslaagde mobiel-lager TEA in vergelijking met een gezonde controlegroep. Bij een vaste loopsnelheid waren de metabole kosten hoger in de studiegroep. Deze toename correleerde met een wegvloeien van mechanische arbeid tijdens de overgang van de ene naar de volgende stap. Bij een zelfgekozen loopsnelheid, die in de studiegroep 12% lager was, vonden wij geen significant verschil in metabole kosten. We concludeerden dat de onderbeenfunctie in termen van energieverbruik niet volledig normaliseerde na de TEA.

Het laatste deel van dit proefschrift omvat een meta-analyse van de literatuur van de vervanging met een kunstgewricht van de arthrotische enkel. Het bevat verder de aanbevolen preoperatieve voorbereiding en operatieve techniek, gevolgd door een algemene discussie over de rol van TEA bij de behandeling van de arthrotische enkel.

Hoofdstuk 11 geeft een overzicht van de op dit moment beschikbare protheses, gevolgd door een meta-analyse van de literatuur van de meest gebruikte protheses met zowel een vast als mobiel lager. De beweeglijkheid van de door een kunstgewricht vervangen enkel herstelt niet volledig, maar is in het algemeen voldoende voor het goed lopen op een vlakke ondergrond. Enkelscores bij na-onderzoek laten een significante toename zien t.o.v. het preoperatieve niveau. Overlevingsdata laten zien dat een 8-jaars overleving van rond 90 procent kan worden verwacht met mobiel-lager protheses als deze zijn geïmplanteerd door ervaren orthopeden. Echter, de overleving van de TEA blijft achter bij de overleving van de totale knie en de totale heup arthroplastiek. In vergelijk met de enkelarthrodese kan een vergelijkbaar faalcijfer worden verwacht.

Hoofdstuk 12 bespreekt de preoperatieve evaluatie van de patiënt met enkelarthrose en geeft aanbevelingen voor een optimale chirurgische techniek van de TEA, zoals positionering van de patiënt, implantaatpositie en correctie van deformiteiten.

Gebaseerd op de onderzoeken die in dit proefschrift zijn beschreven geeft ***hoofdstuk 13*** een algemene discussie van de huidige en toekomstige rol van de TEA in de behandeling van ernstige enkelarthrose. Een volledig herstel naar een normale enkelfunctie wordt niet als een reëel doel beschouwd. Diagnose (hetzij primaire arthrose, ofwel posttraumatische arthrose, of reumatische gewrichtsontsteking), jongere leeftijd en geslacht zijn niet geïdentificeerd als duidelijke risicofactoren voor falen. Risicofactoren voor het falen die wel zijn geïdentificeerd zijn: een preoperatieve deformiteit in het frontale vlak, een beperkte ervaring van de orthopeed (uitgedrukt in laag-volume chirurgen, toegenomen operatietijd en een ondermaats geplaatste tibiacomponent), en sommige implantaat-specifieke kenmerken. Met een goede patiëntenselectie, en geïmplanteerd door orthopeden met toewijding voor dit onderwerp, kunnen goede lange termijn resultaten van mobiel-lager TEA worden verwacht, met een 10-jaars overleving van tenminste 90 procent. Daarmee kan de TEA worden beschouwd als een valide behandelingsmogelijkheid, en, aangezien de gefaalde TEA, behandeld met een herstelarthrodese, een behandelbare complicatie is, als de behandeling van voorkeur voor de meeste patiënten lijdend aan ernstige enkelarthrose.

Aanbevelingen voor toekomstig onderzoek zijn: gerandomiseerde studies die het resultaat vergelijken van enkelarthrodese en TEA, en RSA studies. Vanwege de hoge nauwkeurigheid van RSA zijn maar beperkte aantallen patiënten nodig in dergelijke studies. RSA studies zouden moeten worden verricht van elk nieuw protheseontwerp. Tot slot is een adequate beoordeling van het resultaat van TEA van toenemend belang gezien de toename van het aantal geïmplanteerde enkelprotheses, zowel in Nederland als in Europa en elders in de wereld. Beoordeling van het resultaat van afzonderlijke protheses kan het beste worden uitgevoerd door middel van het instellen van nationale implantaatregisters.

Résumé

L'arthrite de la cheville provoque souvent des affaiblissements importants pour le patient. Du fait que l'arthroplastie totale de la cheville (ATC) par utilisation de prothèses totales de la cheville avec patin fixe connait un taux élevé de défaillances précoces, l'arthrodèse de la cheville est, jusqu'à aujourd'hui, considérée comme le traitement chirurgical standard pour l'arthrite de la cheville en phase terminale. L'ATC avec utilisation d'une prothèse avec patin mobile est devenue possible en 1981 avec le développement aux Etats-Unis de la prothèse de cheville New Jersey LCS®.(DePuy, Warsaw, Indiana) et avec le développement en 1986 au Danemark de la prothèse STAR (Scandinavian Total Ankle Replacement, Waldemar Link, Hambourg, Allemagne). La prothèse de cheville LCS a été introduite en 1988 à l'hôpital Slotervaart-ziekenhuis à Amsterdam, devenant ainsi le premier hôpital non-auteur en Europe à utiliser une prothèse de cheville avec patin mobile. Avec l'expérience, cette méthode est devenue notre traitement chirurgical préféré pour la cheville arthritique en phase terminale, d'abord pour les patients atteints d'une polyarthrite inflammatoire, ensuite également pour les patients souffrant d'une arthrose (la plupart du temps d'origine post-traumatique). Du fait que l'ATC en tant que traitement fiable est jusqu'à ce jour sujet à discussion, une étude détaillée de ses résultats cliniques, radiographiques et fonctionnels a été exécutée afin d'évaluer de façon adéquate sa position actuelle dans l'outillage moderne à disposition du chirurgien spécialiste du pied et de la cheville.

Au ***chapitre 1*** sont présentés les objectifs et les données générales de cette thèse. Les prothèses à 3 composants sans contrainte ont d'excellentes caractéristiques biomécaniques. Le concept de patin mobile offre donc potentiellement une solution prometteuse pour le remplacement endoprothétique de l'articulation de la cheville. Cette thèse est divisée en quatre sections: d'abord une vue d'ensemble de l'anatomie et la cinématique normale, de la pathologie de la cheville arthritique et des possibilités de traitement de l'arthrite symptomatique de cheville; ensuite les résultats cliniques et radiographiques de l'ATC avec patin mobile; puis une section consacrée aux résultats fonctionnels; enfin une section de critiques et discussion.

Le ***chapitre 2*** décrit l'anatomie et la cinématique de l'articulation normale de la cheville, suivi d'une description de la pathologie de la cheville arthritique. Contrairement à l'arthrite des autres articulations de l'extrémité inférieure, les causes les plus fréquentes de l'arthrite de la cheville sont des affections post-traumatiques et la polyarthrite. L'arthrite de la cheville en phase terminale est la plupart du temps une séquelle d'une polyarthrite ou d'une arthrite locale. En outre, les patients souffrant d'une arthrite de la cheville sont généralement plus jeunes que ceux qui souffrent d'une arthrite de la hanche ou du genou. Ces facteurs ont des conséquences pour le choix du moment et du traitement chirurgical de l'arthrite de la cheville en phase terminale les mieux adaptés.

Chapitre 3 décrit les options conservatrices de traitement de l'arthrite de la cheville et le résultat du traitement chirurgical non-endoprothétique. Si le traitement conservateur échoue, pour les patients présentant une arthrite modérée avec symptôme d'empiétement osseux, un débridement arthoscopique est probablement le meilleur traitement. En cas de difformité dans le plan frontal, une ostéotomie corrective doit être envisagée, ou bien au niveau du tibia distal ou bien du calcanéum. En cas de maladie en phase terminale, l'arthrodèse tibiotalaire est la plupart du temps considérée comme étant la voie royale. Si l'opération réussit, on obtient une cheville stable et sans douleur. Les complications constatées sont non union, mal union et infection. Avec un suivi à plus long terme, cependant, il apparait que l'arthrodèse de la cheville comporte une incidence élevée d'arthrite de l'arrière- pied et de ce fait un risque significatif de symptômes récurrents.

Les quatre chapitres suivants de cette thèse sont consacrés aux résultats cliniques et radiographiques de l'ATC et aux solutions en cas d'échec de l'ATC.

Chapitre 4 décrit les résultats cliniques et radiographiques à moyen et long terme de l'ATC avec patin mobile résultant d'une étude de patients souffrant de polyarthrite inflammatoire. Nous pouvons conclure que l'ATC avec patin mobile est une option valide de traitement de la cheville rhumatoïde en cas d'indications appropriées et que le desserrement aseptique et une difformité persistante sont les causes d'échec les plus importants. Un taux d'échec accru a été constaté en ce qui concerne des chevilles ayant une difformité préopératoire dans le plan frontal de plus que 10º et des chevilles où un composant tibial trop petit avait été implanté.
Dans le chapitre 5 la stabilité du composant tibial de la prothèse de Buechel-Pappas a été évaluée au moyen d'une étude radio-stéréométrique. Cette étude a montré une inclinaison initiale vers le haut, antérieure et dans le valgus, qui se stabilise au bout de 6 mois. La création d'une fenêtre corticale antérieure, nécessaire au placement du composant tibial et la méthode de fixation tibiale expliquent probablement cette tendance à la migration.

Chapitre 6 décrit une nouvelle technique, développée en 1998, pour la correction de la difformité du varus par l'ATC. L'étude de quinze chevilles, dans une population mélangée d'instabilité arthritique et de polyarthrite traitée par cette technique, montre que l'ostéotomie rallongeant la malléole médiane s'avère une technique facile et efficace pour le réalignement de la difformité du varus par l'ATC. La non-union asymptomatique de l'ostéotomie malléolaire médiane a été constatée dans deux des chevilles rhumatoïdes concernées (six de cette série). En outre, le décèlement aseptique du composant tibial ou de l'astragale s'est développé dans une cheville pour chaque cas. Une difformité résiduelle de l'arrière-pied, non corrigée à l'opération, a nécessité une intervention chirurgicale additionnelle dans trois cas. A la vue de ces résultats, l'ATC en cas de difformité du varus préexistante a toujours un résultat quelque peu inférieur aux résultats dans des chevilles bien-alignées. Pour améliorer les résultats de l'ATC de la cheville avec varus, les modifications chirurgicales suivantes sont possibles: a) l'ostéotomie de rallongement est faite à la moitié de la malléole médiale; b) la gouttière médiale est débridée. de manière habituelle; c) le composant tibial est implanté sans ou avec seulement une inclinaison antérieure minimale; d) toute difformité de l'arrière-pied est corrigée avant ou simultanément à l'arthroplastie.

Chapitre 7 décrit les résultats de l'arthrodèse lorsque l'ATC échoue, sur un groupe de patients comportant dix-huit chevilles, la plupart souffrant de polyarthrite inflammatoire. Des taux élevés de non-union ont été constatés dans des chevilles rhumatoïdes qui avaient été stabilisées par des vis ou par clou rétrograde. Au contraire, les sept chevilles (quatre chevilles rhumatoïdes et trois chevilles ostéoarthritiques) stabilisées par une plaque en lame étaient toutes jointes. Sur le nombre de cas disponibles, cependant, aucune différence significative n'a pu être démontrée entre les trois méthodes. Un deuxième essai de fusion sur une non-union symptomatique, effectuée dans quatre chevilles, a été réussi à l'exception d'un patient souffrant d'une hanche ipsilatérale raide. Trois patients présentant une non-union refusèrent l'intervention en raison de symptômes légers. On peut conclure qu'en cas d'arthrose le taux de fusion d'arthrodèse de récupération est bon, et comparable aux résultats de l'arthrodèse primaire de cheville. Pour les cas de chevilles rhumatoïdes, cependant, aussi bien l'arthrodèse primaire que l'arthrodèse de récupération sont des procédures exigeantes. De tels patients peuvent le mieux être traités par des chirurgiens expérimentés travaillant dans des centres spécialisés. La stabilisation par une plaque en lame semble être une technique prometteuse pour l'arthrodèse de cheville de récupération.

La partie de cette thèse consacrée aux résultats fonctionnels comprend les trois chapitres suivants :

Le chapitre 8 décrit une étude sur l'analyse de la marche de patients après ATC avec patin réussie, comparée à un groupe de contrôle similaire en bonne santé. Le groupe de patients souffre d'une dorsiflexion réduite de la cheville. Pendant la marche nu-pieds, la vitesse du groupe patient est légèrement inférieure. Cette vitesse de marche inférieure est surtout provoquée par un pas plus lent et, à un moindre degré, par un pas plus court. La marche était presque normale en termes de cinématique du genou, de la cheville et du pied. Cependant, des différences ont été trouvées dans les forces de réaction au sol de la jambe à la cheville opérée. En outre, le mode d'activité d'EMG des muscles de la jambe à la cheville remplacée présente quelques différences : le gastrocnème médial est plus sollicité au début de l'appui et le tibia antérieur à la fin de l'appui. Comparé aux études sur la marche après arthrodèse de la cheville, nos résultats prouvent que, en dépit d'une dorsiflexion légèrement réduite, la marche après ATC réussie est comparable à la normale. Ceci implique également un faible risque de problèmes secondaires de surcharge des articulations de l'arrière-pied .

Le ***chapitre 9*** est une analyse cinétique de la même population qu'au ***chapitre 8***. Aucune différence n'a été observée en ce qui concerne les moments maximaux et la quasi-rigidité de la cheville opérée, mais le travail interne présente de petites différences, peut être dues à la vitesse de marche légèrement réduite. Cette étude a démontré que la charge mécanique de la cheville pendant la marche après ATC ne diffère pas de la charge mécanique de la cheville saine. Ceci indique que la fonction de la cheville en termes de charge mécanique et de quasi-rigidité ne semble ne pas être influencée par le remplacement de la cheville.

Le chapitre 10 décrit une étude de consommation d'énergie pendant la marche nu-pieds sur un tapis roulant par des patients après ATC réussie comparé à un groupe similaire en bonne santé. À une vitesse de marche fixée la dépense métabolique était plus grande pour le groupe patient. Cette augmentation est corrélée à la dissipation du travail mécanique pendant la transition de pas à pas. À une vitesse de marche autodéterminée, qui était 12% moins vite pour le group patient, la dépense métabolique n'était pas different. On peut conclure que la fonction de la jambe n'est pas entièrement normalisée après ATC.

La partie finale de cette thèse se compose d'une méta-analyse de la littérature sur le remplacement prosthétique de la cheville arthritique, les préparatifs pré opératifs recommandés et la technique chirurgicale, et une discussion générale du rôle de l'ATC dans le traitement de la cheville arthritique.

Chapitre 11 donne une vue d'ensemble des conceptions actuellement disponibles, suivie d'une méta-analyse de la littérature sur les types les plus utilisés, à patins fixe et mobile. L'amplitude du mouvement de la cheville opérée ne se normalise pas entièrement, mais sera en général suffisante pour une marche de bon niveau. Les chevilles suivies ont montré un gain significatif comparé au niveau pré opératif. Les données de survie après huit ans ont prouvé qu'un taux de survie d'environ 90 % peut être prévu avec l'utilisation de prothèses à patin mobile, implantées par des chirurgiens expérimentés. Cependant, la survie globale après ATC reste inférieure à la survie après arthroplastie totale du genou ou de la hanche. Comparé à l'arthrodèse, le taux d'échec est similaire.

Le chapitre 12 discute l'évaluation pré opérative du patient avec une cheville arthritique et présente des recommandations pour une technique chirurgicale optimale de l'ATC, tel que le positionnement du patient, la préparation tibiale, la position de l'implant et la correction du défaut de forme.

Fondé sur le travail présenté dans cette thèse, ***le chapitre 13*** présente une discussion générale sur le rôle actuel et futur de l'ATC. La pleine restauration de la fonction de la cheville n'est pas considérée comme un objectif réaliste. Le diagnostic (ostéoarthrite primaire, arthrite post-traumatique ou polyarthrite inflammatoire), l'âge et le sexe n'ont pas été identifiés en tant que facteurs de risque certains d'échec. Les facteurs de risque d'échec qui ont été identifiés sont la difformité préopératoire dans la zone frontale, l'expérience du chirurgien (chirurgiens faisant peu d'interventions, durée prolongée de l'intervention et composant tibial trop petit), et certaines caractéristiques spécifiques aux implants. Avec un choix approprié des patients et implanté par des chirurgiens spécialisés, de bons résultats à long terme de l'ATC avec patin mobile peuvent être attendus, avec une survie à dix ans d'au moins de 90%. De ce fait, l'ATC peut être considéré comme une possibilité de traitement valide et, vue les possibilités de secours en cas d'échec, le traitement préféré pour la plupart des patients souffrant d'arthrite de la cheville en phase terminale.
Recommandations pour futurs recherches:

- études comparant les résultats de l'arthrodèse et de l'ATC,
- des études d'analyses radiostéréométriques (ARS). En raison de la grande précision de l'ARS, seulement de petits nombres de cas sont nécessaires. Des études d'ARS devraient être appliquées à toutes les nouvelles prothèses.
- Enfin, le suivi des résultats de l'ATC est d'importance croissante due au nombre de plus en plus important d'ATC effectuées aux Pays Bas et ailleurs en Europe et dans le monde. Le suivi des résultats des différentes prothèses peut être le mieux réalisé par l'introduction de registres nationaux.

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About the Author

Hendrik Cornelis (Kees) Doets was born on July 16, 1947 in Zaandam, The Netherlands. He graduated from secondary school in 1964 (HBS-B, Zaanlands Lyceum, Zaandam). With the aim to become a surgeon he started his study of medicine at the University of Amsterdam (formerly: Gemeente Universiteit). During his internship in surgery at the Onze Lieve Vrouwe Gasthuis, Amsterdam, he became interested in orthopaedic surgery while working under dr. Th.J.G. van Rens. In 1972 he obtained his medical degree at the University of Amsterdam. Thereafter, he worked for a short period as a physician in a nursery home and as a resident at the Department of Pathology, Wilhelmina Gasthuis, Amsterdam.

In 1974 he started his two years of training in general surgery at the Department of General Surgery, Julianaziekenhuis, Zaandam (head: dr. S.A. Bouma), and in 1975 continued this training at the Department of General Surgery, Sint Lucas Ziekenhuis, Amsterdam (head: dr. J.N. Keeman). From 1976 to 1980 followed his orthopaedic training, first at the Department of Orthopaedic Surgery, Andreasziekenhuis, Amsterdam (head: dr. S.A. Cohen), then from 1977 to 1979 at the Department of Orthopaedic Surgery, Sint Lucas Ziekenhuis, Amsterdam (head: dr. R.J.B. Gastkemper, later: M.C. van Joost), and from 1979 to 1980 at the Department of Orthopaedic Surgery, Onze Lieve Vrouwe Gasthuis, Amsterdam (head: dr. B.E.E.M.J. Veraart). He qualified as orthopaedic surgeon in 1980.

From 1980 to 1981 he worked as a fellow at the Department of Orthopaedic Surgery, Radboudziekenhuis, Nijmegen (head: prof.dr. Th.J.G. van Rens). Since 1981 he has been working as an orthopaedic surgeon at the Department of Orthopaedic Surgery, Slotervaartziekenhuis, Amsterdam (head: prof.dr. R.G. Pöll), and as a consultant in arthritis surgery at the Jan van Breemen Instituut, Amsterdam. He developed a special interest in arthritis surgery and in reconstructive surgery of the hip, and from 1988 onwards, also in ankle reconstruction. Growing experience with total ankle arthroplasty stimulated him to carry out a fundamental assessment of the clinical, radiographic and functional outcomes of this procedure, as, until recently, little was known about these aspects of ankle reconstruction. His other fields of interest are foot, hip and wrist surgery.

He became an active member of the Dutch Orthopaedic Society, of the Netherlands Rheumatoid Arthritis Surgical Society, of the European Rheumatoid Arthritis Surgical Society, and of the Dutch Orthopaedic Foot and Ankle Association. Furthermore, from 2002 to 2005, he worked part time as an orthopaedic surgeon at Kliniek de Lairesse, Amsterdam. He discontinued this position in order to have more time for scientific work.

Since 1989 he is happily married to Nadia Harize. They have one daughter, Irena ('91), who this year will graduate from secondary school (gymnasium, N&T). From his first marriage he has two fine sons, Jurriaan ('77) and Wouter ('79). Besides for orthopaedics, he has a great passion for sailing: regatta sailing both inshore and offshore; as a member of race committees; as a supporter (together with Nadia) of Irena's dinghy sailing; and as a leisure sailor with family and friends.

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