

FACULDADE DE ENGENHARIA DA UNIVERSIDADE DO PORTO

Biomechanical study of surgical solutions in the treatment of ankle arthroplasty

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Master's in Bioengineering

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Resumo

A Artroplastia Total do Tornozelo tem-se afirmado como uma solução preservadora da mobilidade para o tratamento da osteoartrose terminal do tornozelo. Apesar dos avanços alcançados pelos implantes de terceira geração, a mobilização assética e a subsidência dos componentes continuam a constituir algumas das principais causas de falha, evidenciando a necessidade de uma melhor compreensão dos mecanismos de transferência de carga na interface osso-prótese.

O objetivo deste trabalho consistiu no desenvolvimento e análise de um modelo tridimensional de elementos finitos da prótese Zimmer Trabecular Metal™ Total Ankle System, com vista à investigação do seu comportamento biomecânico sob condições de carregamento representativas de diferentes posições funcionais da fase de apoio da marcha. Para tal, procedeu-se à reconstrução geométrica dos componentes protésicos e das estruturas ósseas envolventes, seguida da atribuição das propriedades dos materiais, definição das interações de contacto e implementação das respetivas condições de fronteira e cenários de carregamento.

Os componentes metálicos apresentaram um comportamento seguro relativamente aos limites mecânicos dos materiais em todas as posições analisadas, enquanto o componente de polietileno de ultra-alto peso molecular revelou um desempenho mecânico competitivo face a outras próteses contemporâneas. A análise das deformações no osso peri-implantar evidenciou uma assimetria médio-lateral pronunciada, caracterizada por uma subcarga crónica do compartimento medial e pela presença de concentrações de deformação superiores aos limiares de cedência na interface de fixação ântero-lateral, observadas mesmo sob condições mínimas de carregamento e em ambas as componentes compressiva e tensional. Comparativamente com implantes baseados em quilhas e pinos, a prótese Zimmer TM expôs um menor volume total de tecido ósseo a risco, embora associado a magnitudes máximas de deformação substancialmente mais elevadas.

Os resultados obtidos sugerem que este desenho proporciona uma fixação mecanicamente estável, em concordância com o desempenho clínico favorável reportado a curto e médio prazo, introduzindo simultaneamente vulnerabilidades espacialmente previsíveis que poderão adquirir relevância à medida que os implantes atingirem períodos de seguimento mais prolongados, e que não são evidentes apenas através dos dados de sobrevivência disponibilizados pelos registos ortopédicos.

Palavras-chave: Artroplastia Total do Tornozelo; Método dos Elementos Finitos; Biomecânica; Zimmer Trabecular Metal; Interface Osso-Prótese; Fixação Protésica.

Abstract

Total Ankle Arthroplasty has emerged as a motion-preserving treatment option for end-stage ankle osteoarthritis. Despite the advances achieved by third-generation implants, aseptic loosening and component subsidence remain among the leading causes of failure, highlighting the need for a better understanding of load transfer mechanisms at the bone–implant interface.

The aim of this work was to develop and analyse a three-dimensional finite element model of the Zimmer Trabecular Metal™ Total Ankle System in order to investigate its biomechanical behaviour under loading conditions representative of different functional positions of the stance phase of gait. Geometrical reconstruction of the prosthetic components and surrounding bone structures was performed, followed by material property assignment, contact definition, and implementation of appropriate boundary conditions and loading scenarios.

The metallic components operated safely within their material limits across all positions, and the ultra-high molecular weight polyethylene bearing demonstrated a mechanically competitive profile relative to other contemporary designs. Periprosthetic bone strain analysis revealed a pronounced medial-lateral asymmetry, with chronic underloading of the medial compartment and yield-exceeding strain concentrations at the anterolateral fixation interface, present even under minimum loading and across both compressive and tensile components. Compared to keel- and peg-based designs, the Zimmer TM placed a smaller total volume of bone at risk, but at substantially higher peak strain magnitudes.

These findings suggest that the design delivers mechanically stable fixation consistent with its favorable short- to mid-term clinical performance, while introducing spatially predictable vulnerabilities that may become relevant as implants reach longer follow-up horizons and that are not apparent from registry survivorship data alone.

Keywords: Total Ankle Arthroplasty; Finite Element Analysis; Biomechanics; Zimmer Trabecular Metal; Bone–Implant Interface; Prosthetic Fixation.

The United Nations Sustainable Development Goals

Beyond its clinical and mechanical contributions, this dissertation is aligned with the framework established by the United Nations Sustainable Development Goals (SDGs) [1]. In particular, the present work contributes to SDG 3 (Good Health and Well-being) and SDG 9 (Industry, Innovation and Infrastructure).

- **SDG 3: Good Health and Well-being** – Biomedical engineering has played a key role in improving healthcare and quality of life through the development of advanced medical devices [1]. By applying Finite Element Analysis (FEA) to investigate the biomechanical behaviour and failure mechanisms of total ankle replacements, this work contributes to the development of safer and more durable motion-preserving solutions.
- **SDG 9: Industry, Innovation and Infrastructure** – Engineering innovation is essential for scientific and technological progress [1]. The development of reverse-engineering procedures and computational simulation methodologies contributes to the advancement of numerical tools in biomedical engineering and promotes innovation in the evaluation of orthopaedic implants.

Table 1: Alignment of the dissertation with the United Nations Sustainable Development Goals.

SDG	Target	Contribution	Expected Impact
3	3.8	Finite Element Analysis of total ankle arthroplasty to investigate load transfer and implant failure mechanisms.	Improved understanding of bone response and factors influencing implant longevity.
9	9.5	Development of reverse-engineering and computational modelling methodologies for the assessment of contemporary ankle prostheses.	Advancement of numerical techniques and innovation in orthopaedic implant evaluation.

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Declaration of honour

I declare that this dissertation is my own work and has not previously been submitted or assessed at this or any other institution. References to other authors (statements, ideas, thoughts), as well as the use of published works, strictly comply with the rules of attribution and are duly identified in the text and in the bibliographic references, in accordance with the applicable referencing standards. I am aware that the practice of plagiarism or self-plagiarism constitutes an academic offence, as established in the U.Porto Code of Ethics.

I also declare that Artificial Intelligence tools were used to support the preparation of this dissertation, namely for language revision and writing assistance. Their outputs were critically assessed and verified, and the responsibility for the content, interpretations and conclusions presented in this work remains entirely my own.

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List of Acronyms

A/P	Anterior-Posterior
AJC	Ankle Joint Complex
AJRR	American Joint Replacement Registry
AOANJRR	Australian Orthopaedic Association National Joint Replacement Registry
ATFL	Anterior Talofibular Ligament
ATiTL	Anterior Tibiotalar Ligament
BW	Body Weight
CAD	Computer-Aided Design
CFL	Calcaneofibular Ligament
CJRR	Canadian Joint Replacement Registry
CoCrMo	Cobalt-Chromium-Molybdenum alloy
CPE	Conventional Polyethylene
CPRESS	Contact Pressure
CT	Computed Tomography
DAIR	Debridement, Antibiotics, and Implant Retention
FEA	Finite Element Analysis
FEM	Finite Element Method
FEUP	Faculdade de Engenharia da Universidade do Porto
GC	Gait Cycle
HXLPE	Highly Cross-Linked Polyethylene
LCL	Lateral Collateral Ligament
M/L	Medial-Lateral
MRI	Magnetic Resonance Imaging
NAR	Norwegian Arthroplasty Register
NJR	National Joint Registry
NZJR	New Zealand Joint Registry
OA	Osteoarthritis
PE	Polyethylene
PJI	Periprosthetic Joint Infection
PMMA	Polymethylmethacrylate
PROMs	Patient-Reported Outcome Measures
PSI	Patient-Specific Instrumentation
PTFL	Posterior Talofibular Ligament
PTiTL	Posterior Tibiotalar Ligament
SDG	Sustainable Development Goals
SEFAS	Self-reported Foot and Ankle Score
SPOT	Portuguese Society of Orthopaedics and Traumatology
STAR	Scandinavian Total Ankle Replacement

TAA	Total Ankle Arthroplasty
TAR	Total Ankle Replacement
Ti-6Al-4V	Titanium-6Aluminium-4Vanadium alloy
TM	Trabecular Metal
TTR	Total Talar Replacement
UHMWPE	Ultra-High-Molecular-Weight Polyethylene

List of Symbols and Units

E	Young's modulus	MPa
ν	Poisson's ratio	—
σ_{VM}	von Mises stress	MPa
P	Contact pressure	MPa
ε	Strain	$\mu\varepsilon$
ε_1	Maximum principal strain (tensile)	$\mu\varepsilon$
ε_3	Minimum principal strain (compressive)	$\mu\varepsilon$
μ	Coefficient of friction	—
BW	Body weight	N
θ	Joint angle	°

Chapter 1

Introduction

1.1 Motivation and Background

The human ankle joint complex is a biomechanically demanding structure that endures multi-axial cyclic loads reaching up to five to seven times the body weight during normal walking locomotion [2, 3]. When end-stage ankle osteoarthritis affects this complex, predominantly secondary to post-traumatic events such as fractures or severe ligamentous instability, it induces severe cartilage degradation, subchondral bone sclerosis, and marginal osteophyte formation [3, 4]. This pathological cascade transforms a smoothly gliding joint into a rigid and painful articulation, severely restricting weight-bearing activities and fracturing the patient’s quality of life [3, 5].

Historically, ankle arthrodesis (joint fusion) was deemed the absolute gold standard for treating end-stage degeneration due to its predictable structural stability [3, 6]. However, by permanently sacrificing the tibiotalar range of motion, arthrodesis inherently alters lower limb kinematics, which inevitably triggers secondary osteoarthritis in adjacent hindfoot and midfoot joints over long-term clinical horizons [3, 7]. Total Ankle Arthroplasty (TAA), also referred to in the literature as Total Ankle Replacement (TAR), has emerged as a major motion-preserving alternative, aimed at restoring a more physiological gait pattern and shielding adjacent joints from compensatory mechanical overload [4, 8].

Despite these clear kinematic advantages, the long-term survival rates of artificial ankles still lag behind those achieved in total hip and knee arthroplasties [7, 9]. Large-scale tracking data from national orthopedic registries indicate that modern total ankle implants remain susceptible to critical mechanical failure modes, dominated by aseptic loosening, component subsidence, and severe polymeric insert edge-loading or accelerated wear [9–11]. The primary driver behind these failures is a persistent biomechanical conflict: contemporary bone anchorage mechanisms and highly conforming articular geometries often trigger critical load imbalances and extreme, unphysiological localized stress concentrations along the bone-implant and bearing interfaces [9, 12, 13]. Furthermore, long-term empirical follow-ups are hindered by a substantial time lag, as years of clinical cyclic loading must accumulate before an interface defect manifests macroscopically, forcing a reactive approach that bears a severe patient morbidity burden [13].

Therefore, there is a need for advanced biomechanical research and high-fidelity computational optimization to overcome these limitations [14]. To address these highly specific research gaps, the primary motivation of this study is to implement a three-dimensional finite element analysis (FEA) of a contemporary, fourth-generation total ankle system: the Zimmer Trabecular Metal (TM) Total Ankle [12]. This specific prosthesis represents an innovative departure from traditional designs, combining a lateral transfibular approach with anatomically curved arcuate resections and highly porous tantalum structures engineered to minimize stress-shielding and promote rapid osseointegration [4, 15]. By deploying a detailed numerical framework across three functional positions of the stance phase of the gait cycle [14], this research seeks to preemptively map localized contact pressures, predict potential material yielding, and evaluate bone-implant interface load transmission. Pushing the boundaries of implant modeling provides orthopedic surgeons and biomedical engineers with predictive mechanical data, moving away from reactive empirical tracking toward proactive structural optimization, which ultimately aims to enhance the long-term survivorship and endurance of joint replacements.

1.2 Objectives

The main objective of this dissertation is to perform a biomechanical study of a surgical solution currently available for the treatment of end-stage ankle osteoarthritis, with particular emphasis on total ankle arthroplasty and the mechanisms governing load transfer, implant fixation, and long-term mechanical performance.

To accomplish this objective, the following specific goals were established:

- To review the current surgical treatment options for end-stage ankle osteoarthritis and investigate the evolution of total ankle arthroplasty, including implant design philosophies, fixation strategies, failure mechanisms, and clinical outcomes reported in national orthopedic registries;
- To reconstruct and virtually implement a representative contemporary ankle prosthesis, within an anatomical bone model;
- To develop a finite element framework capable of reproducing physiological loading conditions representative of the stance phase of gait;
- To evaluate stress distributions within the prosthetic components and strain patterns in the surrounding cortical and trabecular bone;
- To identify preferential load-transfer pathways and investigate biomechanical mechanisms associated with polyethylene wear, bone overloading, component subsidence, and aseptic loosening, while assessing the influence of implant geometry and fixation design on the long-term clinical performance of total ankle arthroplasty through comparison with published studies.

1.3 Document Organization

This dissertation is organized into eight chapters. Chapter 1 introduces the motivation, objectives, and structure of the work. Chapter 2 presents the anatomical and biomechanical foundations of the ankle joint and reviews the main surgical treatment options for ankle osteoarthritis. Chapter 3 provides an overview of the evolution and design principles of total ankle prostheses, while Chapter 4 summarizes findings from major national orthopedic registries.

Chapter 5 describes the geometric reconstruction and virtual implantation of the Zimmer Trabecular Metal Total Ankle System, and Chapter 6 details the finite element framework and numerical methodology. Chapter 7 presents and discusses the simulation results, and finally, Chapter 8 summarizes the main conclusions and outlines directions for future work.

1.4 Scientific Dissemination

Preliminary findings obtained during the development of this dissertation were disseminated through the submission of a scientific abstract to the 45th National Congress of Orthopaedics and Traumatology, organized by the Portuguese Society of Orthopaedics and Traumatology (SPOT). The abstract, entitled “*Estudo Computacional da Transferência de Carga no Tornozelo com a Prótese Trabecular Metal Total Ankle™*”, was submitted as a Free Paper under the Foot and Ankle category.

The contribution summarizes preliminary numerical results regarding load transfer mechanisms and stress distribution in the Zimmer Trabecular Metal Total Ankle prosthesis in neutral position. For completeness, the full text of the submitted abstract is reproduced in Appendix A.

Chapter 2

Biomechanical Background

To successfully implement a high-fidelity numerical simulation of a total ankle reconstruction, a comprehensive integration of biological and mechanical principles is fundamentally required.

Accordingly, this chapter establishes the clinical foundation necessary to contextualize the subsequent computational framework. The following sections delineate the standard anatomical reference systems and multi-planar motion characteristics of the foot and ankle complex. Furthermore, the structural composition of the joint is thoroughly analyzed, progressing from bone tissue mechanics and articular cartilage properties to the stabilizing ligamentous network and multi-phase gait dynamics. Finally, this chapter explores the clinical etiology of ankle osteoarthritis and provides a comparative evaluation of the primary surgical pathways: joint fusion and total replacement, establishing the clinical and mechanical trade-offs that justify the need for advanced 3D computational design.

2.1 Anatomical and Biomechanical Background

2.1.1 Anatomical Reference Systems

To accurately characterize the kinematics of the Ankle Joint Complex (AJC) and other articular movements, researchers rely on a standardized biomechanical vocabulary. This specific nomenclature is crucial for defining the exact spatial orientation and movement vectors of the system being analyzed.

The universally accepted baseline for investigating human biomechanics is the anatomical reference position [16, 17]. This standardized posture consists of an individual standing upright, with the gaze fixed forward, feet positioned parallel and slightly apart, and arms resting at the sides with the palms directed anteriorly. To accurately map musculoskeletal structures and determine spatial relationships within this coordinate framework, a specific set of directional terms is employed [17].

Spatial orientation relative to the head is defined by the terms superior and inferior, indicating positions closer to or farther away from it, respectively. Transverse alignment is described as

anterior, toward the front of the body, or posterior, toward the back. In the coronal dimension, structures are categorized as medial when located closer to the body's midline, or lateral when situated further from it. Furthermore, proximity to the trunk or point of structural attachment is designated as proximal or distal, while depth relative to the body's surface is distinguished using the terms superficial and deep [17].

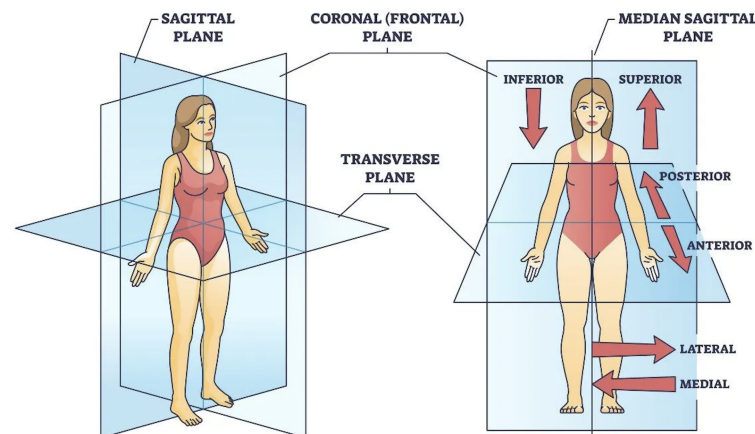


Figure 2.1: Representation of the primary anatomical reference system. The schema illustrates the three imaginary orthogonal planes intersecting the human body (sagittal, coronal/frontal, and transverse) along with key directional orientations, including superior/inferior, anterior/posterior, and medial/lateral relations. Adapted from [18].

To systematically quantify this spatial arrangement, the human body is geometrically divided by three imaginary orthogonal planes that intersect at the center of mass: the sagittal, frontal (or coronal), and transverse (or horizontal) planes [17]. The sagittal plane splits the body vertically into left and right portions, the frontal plane divides the structure into anterior and posterior halves, and the transverse plane separates the upper and lower sections. Perpendicular to each of these planes are the three primary anatomical axes around which geometric rotations and alignments are calculated. The mediolateral axis runs horizontally from side to side, perpendicular to the sagittal plane, the anteroposterior axis extends horizontally from front to back, perpendicular to the frontal plane, and the longitudinal axis passes vertically from superior to inferior, perpendicular to the transverse plane [17].

2.1.2 Local Planes and Joint Movements

Within the anatomical reference position, every joint angle is conventionally set to zero, providing a baseline to quantify any subsequent angular displacement [19]. For the foot and ankle complex, physiological movements are distributed across all three spatial dimensions, occurring as rotations around the previously defined orthogonal axes [3, 16].

Motion around the mediolateral axis within the sagittal plane generates dorsiflexion and plantarflexion, which describe the upward and downward movement of the foot, respectively [16, 19]. Meanwhile, rotation about the longitudinal (vertical) axis within the transverse plane results in abduction and adduction, dictating the outward and inward transversal deviation of the foot [2, 19]. Lastly, the foot undergoes inversion and eversion when rotating around the anteroposterior axis within the frontal plane, which tilts the plantar surface medially or laterally [2, 16, 19]. The spatial distribution of these anatomical planes and the corresponding isolated rotational movements of the foot are schematically represented in Figure 2.2 (a) and (b), respectively.

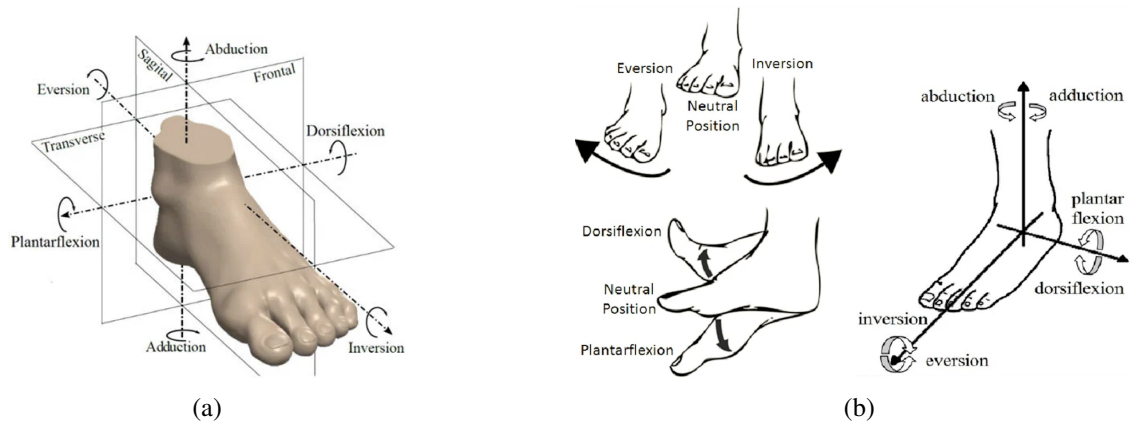


Figure 2.2: Kinematic reference of the foot and ankle complex: (a) spatial orientation of the anatomical planes (sagittal, frontal, and transverse) intersecting the foot, adapted from [20]; (b) isolated rotational movements (dorsiflexion/plantarflexion, inversion/eversion, and abduction/adduction) relative to the neutral position. Adapted from [21].

2.2 Morphological Structure of the Ankle Joint Complex

2.2.1 Bone Tissue Mechanics

To fully grasp the biomechanical complexity of the human ankle, it is essential to first understand its structural composition, progressing from the tissue level up to its multiaxial kinematics [22, 23]. The skeletal foundation relies on two distinct osseous tissues: cortical and trabecular (or cancellous) bone [23, 24]. Cortical bone forms the dense, hard outer shell of the skeletal structures, acting as the primary load-bearing framework that provides the structural rigidity required to resist massive axial, bending, and torsional forces [2, 23]. In computational biomechanics, this rigid nature is typically modeled with an elastic modulus ranging between 17 and 19 GPa [2, 22, 23, 25]. In contrast, trabecular bone, located within the inner regions of the distal tibia and constituting the bulk of the talus, is characterized by a porous, honeycomb-like lattice [22]. Mechanically, it is highly compliant, with elastic moduli often modeled between 400 and 1090 MPa, or mathematically defined as a density-dependent variable [2, 22, 23]. While cortical bone carries the majority of the structural load, this internal trabecular structure allows the bone to effectively absorb and

distribute localized articular stresses across the joint, with its strain energy density dynamically adapting to applied physiological loads [22, 23].

Furthermore, bone is not a homogeneous, isotropic structure. Rather, it is fundamentally characterized as a highly heterogeneous and anisotropic material [14, 25, 26]. In trabecular bone, this anisotropy is governed by its microstructural architecture, which is specifically organized to optimize load transfer by aligning its apparent principal material axes with the principal stress trajectories [26]. This optimized directional orientation enables the trabecular lattice to efficiently support and resist multi-directional physiological loads without significantly increasing the overall skeletal mass [26]. While cortical bone carries the majority of the structural load, this internal anisotropic architecture allows trabecular bone to effectively absorb and distribute localized articular stresses across the joint, with its strain energy density dynamically adapting to applied physiological loads [22, 23].

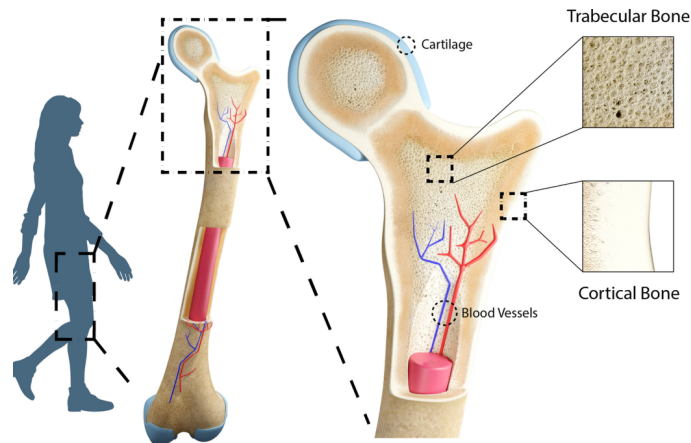


Figure 2.3: Hierarchical organization of bone tissue showing the cortical and trabecular compartments, vascularization, and articular cartilage. Adapted from [27].

2.2.2 Articular Architecture and Cartilage

These osseous tissues interact to form what is biomechanically known as the Ankle Joint Complex (AJC). Rather than a solitary articulation, the AJC is a highly coordinated system comprising three distinct joints, each involving specific bone interactions. The subtalar (talocalcaneal) joint is located inferiorly to the talus, connecting it to the calcaneus, the largest and strongest bone of the foot [16]. This joint forms a triplanar articulation that is paramount for frontal plane kinematics, dictating the inversion and eversion of the foot [2, 16]. The distal tibiofibular joint is formed by the inferior extremities of the tibia and the fibula, acting primarily as a stabilizing fibrous syndesmosis that securely holds the ankle mortise together [16].

The primary weight-bearing articulation is the tibiotalar (true ankle) joint, formed by the interaction of three distinct bones: the tibia, the fibula, and the talus [2, 16]. Functioning primarily

as a highly congruent hinge, it is responsible for the majority of sagittal plane motion, specifically dorsiflexion and plantarflexion [3, 16]. Anatomically, the distal ends of the tibia and fibula, along with their respective medial and lateral malleoli, create a deep rectangular socket known as the ankle mortise [16]. The trochlea of the talus fits firmly into this mortise. A defining anatomical feature of this joint is the wedge-shaped geometry of the talus, which is significantly wider anteriorly. This shape biomechanically locks the talus within the mortise during dorsiflexion, providing maximum stability [3, 16].

To ensure low-friction articulation and protect these specific bony structures, the surfaces of the tibiotalar joint are lined by articular cartilage, a viscoelastic connective tissue that facilitates load transmission [28]. Notably, the articular cartilage of the ankle is exceptionally thin, averaging approximately 1,6 mm compared to the 6 to 8 mm found in the knee [2]. Despite this reduced thickness, it possesses a higher compressive modulus to withstand immense localized forces [3], successfully supporting loads of 5 to 7 times the body weight during the stance phase of walking [2, 3].

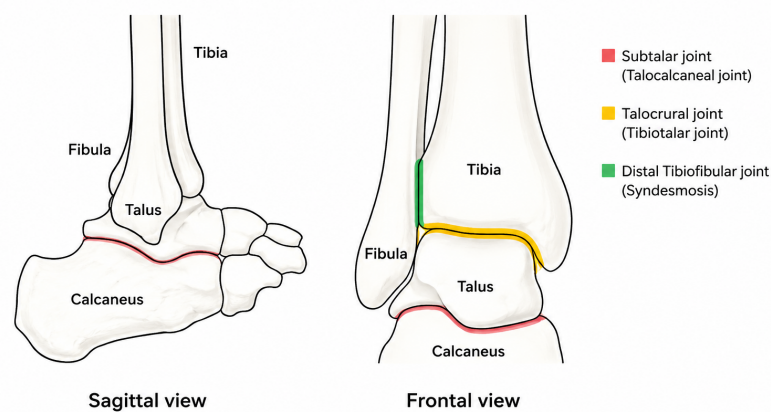


Figure 2.4: Sagittal and frontal views of the Ankle Joint Complex, highlighting the subtalar (red), tibiotalar (yellow), and distal tibiofibular (green) articulations. Adapted from [29].

2.2.3 Ligamentous Apparatus and Joint Stability

Because bony congruency alone is insufficient to maintain joint integrity, the Ankle Joint Complex is heavily reinforced by a sophisticated network of ligaments. These structures play a fundamental role in providing stability and guiding motion by withstanding high tensile forces and resisting movements beyond physiological limits [16].

On the lateral aspect, the joint is stabilized by the lateral collateral ligament (LCL) complex, comprising the anterior talofibular ligament (ATFL), the posterior talofibular ligament (PTFL), and the calcaneofibular ligament (CFL) [16]. Functionally, the ATFL acts as the primary restraint against anterior displacement of the talus and limits extreme extension [30]. The PTFL prevents

posterior displacement of the talus, while the CFL, the only ligament bridging both the tibiotalar and subtalar joints, provides critical stability during flexion [16, 30].

On the medial aspect is supported by the deltoid ligament complex. This robust, fan-shaped structure originates from the medial malleolus and extends to the talus, calcaneus, and navicular bones, primarily restricting excessive eversion and valgus stresses [16, 31]. Anatomically, it is a multifascicular structure divided into superficial and deep layers [31]. Highly relevant for tibiotalar kinematics is the deep layer, which includes the deep anterior and deep posterior tibiotalar ligaments. Notably, the deep posterior tibiotalar ligament is consistently the largest and thickest band of the entire deltoid complex, acting as the primary biomechanical restraint against external rotation of the talus [31]. Finally, the distal tibiofibular syndesmosis binds the tibia and fibula, preventing the bones from splaying apart during weight-bearing [16].

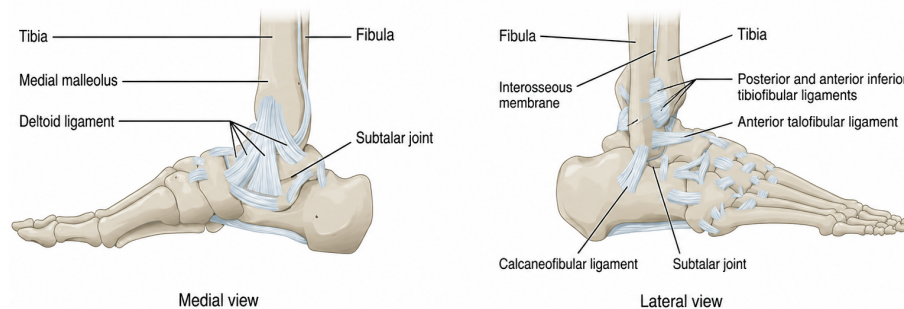


Figure 2.5: Medial and lateral views of the principal ligamentous structures stabilizing the ankle joint complex. Adapted from [32].

2.3 Gait Dynamics and Load Transmission

Human locomotion is a highly optimized biomechanical process. The gait cycle is defined as the continuous mechanical period between two successive heel strikes of the same foot, which is structurally divided into a stance phase and a swing phase [2].

During the stance phase, the tibiotalar joint undergoes a sequence of motions to absorb impact and provide forward propulsion. The cycle initiates with a heel strike, followed by a rapid plantarflexion [28]. As the body's center of mass advances, the ankle progressively moves into dorsiflexion. During the terminal push-off and toe-off stages, the ankle forcefully plantarflexes to generate the necessary propulsion before entering the swing phase [16, 28].

From a kinetic perspective, the ankle joint supports high-intensity forces during daily activities such as normal walking. The total articular contact force is primarily determined by the vector sum of the ground reaction force and the force exerted by the Achilles tendon [33]. Because the

moment arm of the Achilles tendon is significantly shorter than the moment arm of the ground reaction force, the tendon must generate a massive tensile force to maintain static equilibrium [33]. Consequently, the compressive force across the ankle joint is drastically amplified [33]. Literature highlights that the ankle joint complex bears an approximate force of five times the body weight during the stance phase of normal walking [16], with peak magnitudes potentially reaching five to seven times the body weight [2]. This peak of mechanical stress typically occurs at the end of the stance phase (push-off), precisely when the moment arm of the ground reaction force reaches its maximum amplitude [33].

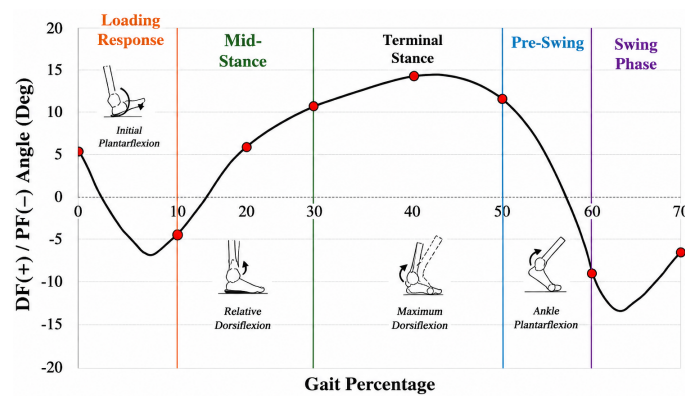


Figure 2.6: Angular profile of the tibiotalar joint during the gait cycle, showing the transition between plantarflexion and dorsiflexion throughout the loading response, mid-stance, terminal stance, pre-swing, and swing phases. Peak dorsiflexion is observed during terminal stance, followed by rapid plantarflexion during push-off. Adapted from [28].

2.4 Clinical Management of Ankle Osteoarthritis

2.4.1 Ankle Osteoarthritis

Ankle osteoarthritis (OA) is a progressive, debilitating degenerative joint disorder characterized by the gradual "wear and tear" and destruction of the articular cartilage that normally cushions the ends of the bones. In a healthy joint, articular cartilage provides a smooth, frictionless surface for motion. However, as OA progresses, this protective layer progressively thins and ultimately wears out, leading to direct bone-on-bone contact [14].

This severe mechanical friction triggers a detrimental pathological cascade that severely alters the joint's biomechanics [34]. First, the loss of cartilage leads to joint space narrowing and direct bone-on-bone friction, causing excruciating mechanical pain and progressive wearing away of the bone structure [14, 34]. Second, to withstand the extreme compressive loads without the cartilage's cushioning effect, the underlying bone undergoes subchondral bone sclerosis (a pathological thickening and hardening). By becoming excessively rigid, the bone loses its natural shock-absorbing capacity, which can lead to the formation of painful subchondral cysts and even

bone necrosis [34]. Finally, the altered mechanics and joint instability prompt the formation of osteophytes (bony spurs) at the joint margins [34, 35]. Rather than stabilizing the joint, these osteophytes act as mechanical blocks that collide during motion (impingement), leading to severe joint stiffness, restricted mobility, and acute pain [4, 34, 35]. Together, these changes transform a smoothly gliding joint into a rigid and painful articulation, significantly limiting normal weight-bearing activities and resulting in an overall physical and mental impact comparable to end-stage kidney disease or congestive heart failure [36, 37].

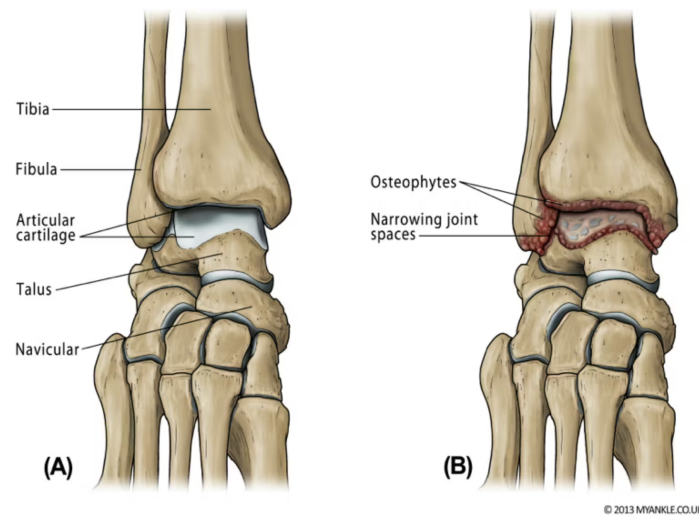


Figure 2.7: Schematic comparison between a healthy ankle joint and end-stage ankle osteoarthritis (OA): (A) Normal joint anatomy demonstrating healthy articular cartilage distribution and open joint spaces between the tibia, fibula, and talus; (B) Osteoarthritic joint displaying advanced cartilage degradation, prominent marginal osteophyte formation (bony spurs), and severe joint space narrowing resulting from structural erosion. Adapted from [38].

Affecting approximately 1% of the global adult population [6, 36], end-stage ankle arthritis differs significantly in its etiology compared to hip or knee arthritis. While hip and knee degeneration typically develops idiopathically over time (primary OA), up to 70% to 90% of ankle OA cases are post-traumatic, presenting as a sequela of severe fractures or chronic ligament instability [4, 37, 39]. Other significant causes leading to joint destruction include inflammatory conditions, most notably Rheumatoid Arthritis, as well as primary osteoarthritis and avascular necrosis (bone tissue death due to a lack of blood supply) [4, 37].

Because the predominant etiology is trauma-induced rather than age-related, the patient demographic for ankle replacement is uniquely challenging. The literature demonstrates that these patients are typically 7 to 11 years younger than those undergoing hip or knee arthroplasties at the time of surgery [40]. Consequently, they are often more physically active, have higher functional expectations, and possess longer life expectancies, placing extraordinary biomechanical demands on the implanted prosthesis [6, 39].

2.4.2 Surgical Solutions

2.4.2.1 Ankle Arthrodesis

Historically, ankle arthrodesis (joint fusion) has been widely considered the "gold standard" surgical intervention for the management of end-stage ankle osteoarthritis [6, 36]. This procedure effectively alleviates pain and provides a highly stable joint capable of withstanding high loads, making it a preferred treatment for younger, high-demand patients or those with severe uncorrectable deformities [41]. However, by joining the tibia and talus into a single bony unit, arthrodesis inherently sacrifices the joint's natural range of motion [6, 41]. This complete loss of mobility alters the biomechanics of the lower extremity, leading to a slower, asymmetrical gait and increased energy consumption during ambulation [37, 41]. Furthermore, these compensatory movements drastically increase mechanical stress on adjacent articulations [36]. Long-term clinical evidence has demonstrated that this altered load distribution frequently leads to the subsequent degeneration and development of secondary osteoarthritis in the adjacent hindfoot and midfoot joints (such as the subtalar joint) in up to 100% of cases over a 20-year period [36, 39]. Additional long-term complications include non-union, malunion, and persistent difficulty walking on uneven surfaces or climbing stairs [41].

2.4.2.2 Total Ankle Arthroplasty

To address the biomechanical shortcomings of ankle fusion, Total Ankle Arthroplasty has emerged as a significant advancement in orthopedic surgery, offering a viable motion-preserving alternative [6, 37]. By replacing the degenerated articular surfaces with a prosthetic implant, TAA aims to restore a more physiological gait pattern, including increased walking velocity and stride length, while maintaining the natural kinematics of the ankle [28, 41]. A key biomechanical advantage of TAA is its ability to absorb load and protect adjacent joints from excessive compensatory stress, thereby mitigating the risk of secondary arthritis and significantly decreasing the likelihood of requiring subsequent subtalar fusions [36].

Despite these clear biomechanical advantages, TAA presents distinct clinical challenges and disadvantages. The long-term survival rates of ankle implants remain inferior to those of hip and knee arthroplasties [7, 39]. This is primarily due to the unique mechanics of the ankle joint, which has a much smaller articular contact surface area yet must withstand immense axial forces, reaching up to five times the body weight during normal walking [37, 41]. Consequently, historically, TAA is associated with higher overall complication and revision rates compared to arthrodesis. The most prevalent cause of mechanical failure and revision is aseptic loosening, often preceded by implant subsidence, osteolysis, and wear of the ultra-high molecular weight polyethylene bearing. Additionally, depending on the surgical approach and implant design, TAA carries risks of delayed wound healing, periprosthetic fractures, and gutter impingement [7, 37, 42].

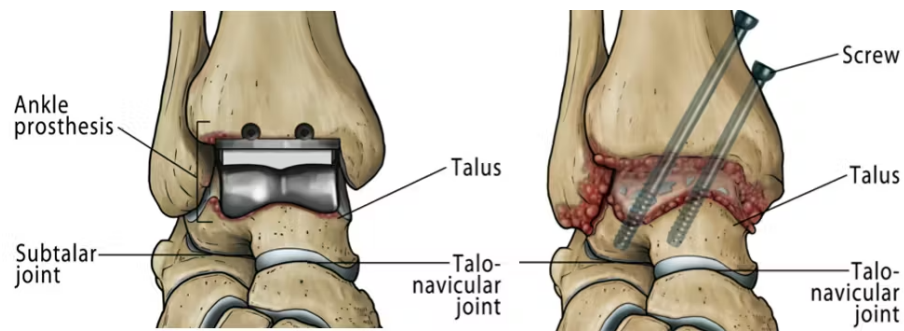


Figure 2.8: Surgical management of end-stage ankle osteoarthritis. (A) Total Ankle Replacement, demonstrating the replacement of articular surfaces with a prosthetic implant to preserve joint kinematics. (B) Ankle arthrodesis, illustrating the fusion of the tibiotalar joint using screws. Adapted from [43].

The literature explicitly states that both TAA and ankle arthrodesis are highly successful in achieving their primary clinical endpoint: substantial pain relief [44, 45]. However, from a strict biomechanical perspective, neither procedure offers a flawless restoration of a healthy joint. While arthrodesis sacrifices natural kinematics leading to adjacent joint degeneration, TAA remains hindered by the aforementioned higher long-term complication and revision rates [6, 36]. Therefore, rather than definitive cures, both interventions are more accurately characterized as clinical compromise solutions for end-stage disease. The ongoing debate in the literature underscores that it is currently difficult to definitively claim which procedure is superior [6]. Nevertheless, recent studies note a growing preference and exponential increase in popularity among both patients and orthopaedic surgeons for motion-preserving alternatives like TAA, driven by high patient satisfaction rates and the desire to avoid the severe restriction in range of motion inherent to joint fusion [41, 44, 45]. Ultimately, both procedures remain relevant, and the choice between TAA and arthrodesis must be highly individualized, carefully weighing the patient's specific functional demands against the inherent mechanical limitations and trade-offs of each surgical option [6]. Consequently, the optimal configuration for an artificial ankle joint has not yet been definitively established. There remains significant room for improvement in TAA, highlighting the critical need for continuous biomechanical research and computational modeling to optimize implant designs, better replicate natural joint kinematics, and ultimately enhance the long-term survivorship of these devices [46].

Chapter 3

State of the Art: Total Ankle Arthroplasty

While the preceding chapter established the fundamental biological and kinematic baselines of the native ankle joint, translating those organic mechanics into successful orthopedic implants represents one of the most demanding challenges in modern biomedical engineering. Total Ankle Arthroplasty has undergone a profound structural shift over the last few decades, transitioning from primitive, highly constrained configurations into advanced, patient-specific cementless systems. In prosthetic engineering, a meticulous historical perspective is paramount, mapping exactly where and why previous design paradigms failed is the only robust mechanism to drive continuous incremental improvements and optimize future implants. This way, a comprehensive technical and historical audit of implant engineering and its associated mechanical shortcomings is strictly required.

Accordingly, this chapter presents a detailed state-of-the-art review focused on the conceptual evolution of total ankle replacements. The following sections trace the mechanical lessons learned from early generations, explore the consolidation of modern fixed- and mobile-bearing design principles, and provide a segmented analysis of contemporary tibial, talar, and polymeric components. Furthermore, a comparative biomechanical evaluation of prominent clinical implementations is detailed.

3.1 Conceptual Evolution of TAA

3.1.1 Lessons from Early Generations

The development of total ankle arthroplasty occurred during an initial phase when the biomechanics of this joint were still poorly understood, and many designs were adapted from hip and knee prostheses, utilizing geometries and fixation concepts ill-suited for the ankle [37, 42]. These first-generation solutions frequently relied on cemented fixation and required extensive bone resection, sacrificing critical bone stock and compromising the bone-implant interface essential for

load transmission and potential future revisions [36, 42].

In an attempt to ensure stability, early designs varied significantly in their degree of constraint [42]. Constrained systems imposed rigid motion pathways but generated high shear stresses and torque at the bone-implant interface, contributing to premature aseptic loosening [36, 42]. However, unconstrained systems allowed for greater mobility but relied excessively on the surrounding ligaments, frequently leading to instability and component malalignment [42]. Furthermore, the lack of articular conformity in some implants resulted in high local contact stresses and accelerated wear [36]. These outcomes progressively highlighted the critical importance of anatomical congruency for joint stability and optimal load distribution [7, 36].

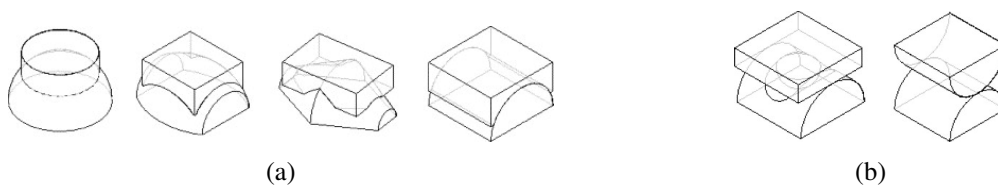


Figure 3.1: Geometric configurations of early total ankle replacement interfaces: (a) congruent surface designs (including spherical, spheroidal, conical, and cylindrical geometries) developed to distribute articular loads, adapted from [46]; (b) incongruent surface configurations, illustrating trochlear (left) and convex/convex (right) joint types that induce high localized contact stresses, adapted from [46].

Consequently, the clinical results of this phase were globally unsatisfactory, characterized by high rates of aseptic loosening, bearing component wear, prosthetic instability, and component subsidence [7, 42]. These failure modes reflected severe limitations in prosthetic design, material quality, and the contemporary understanding of tibiotalar biomechanics [37]. The high incidence of early revisions ultimately led to the complete abandonment of these implants, clearly demonstrating that concepts derived from hip and knee arthroplasty could not be directly transferred to the ankle [7, 42].

3.1.2 Consolidation of Modern Principles

It was in the context of the limitations observed in the first generation that the second generation of total ankle arthroplasty emerged, aiming to improve component fixation and achieve a more appropriate balance between joint stability and mobility. One of the most relevant evolutions was the progressive transition to cementless systems, utilizing porous surfaces and coatings intended to promote osseointegration and stabilize the bone-implant interface, thereby reducing the reliance on cementation and its associated loosening rates [7, 42].

In parallel, the preservation of bone stock became highly valued, defined as the reduction of bone resection necessary for prosthetic implantation through less invasive components and more

conservative bone preparations. This approach emerged as a response to the failures associated with the extensive resections of previous generations, seeking to improve the bone-implant interface and maintain viable options for eventual future revisions. This evolution reflected a growing understanding of the importance of respecting the anatomy and biomechanics of the tibiotalar joint, although many of the principles regarding prosthetic alignment and soft-tissue or ligamentous balancing would only be fully systematized in subsequent generations [4].

In terms of prosthetic design, many second-generation implants adopted semi-constrained configurations, seeking a compromise between joint stability and mobility, a context in which two main approaches were consolidated: fixed-bearing and mobile-bearing systems [4, 42]. In fixed-bearing systems, the polymeric insert is fixed to one of the metallic components, providing greater mechanical constraint and intrinsic stability. In mobile-bearing systems, the mobile insert functions as a freely mobile meniscus between the tibial and talar components, creating two separate articular surfaces. This design aims to redistribute stresses and reduce shear forces at the bone-implant interface through less mechanical constraint, while increasing articular congruency with the objective of minimizing the mobile insert wear [7]. However, this relative mobility is associated with specific risks, namely insert instability, wear, and undesirable translation or subluxation phenomena [42]. Furthermore, a geographic variation in the adoption of these systems is observed, with fixed-bearing implants being more commonly used in the United States, while mobile-bearing systems are historically more prevalent in Europe [4, 42].

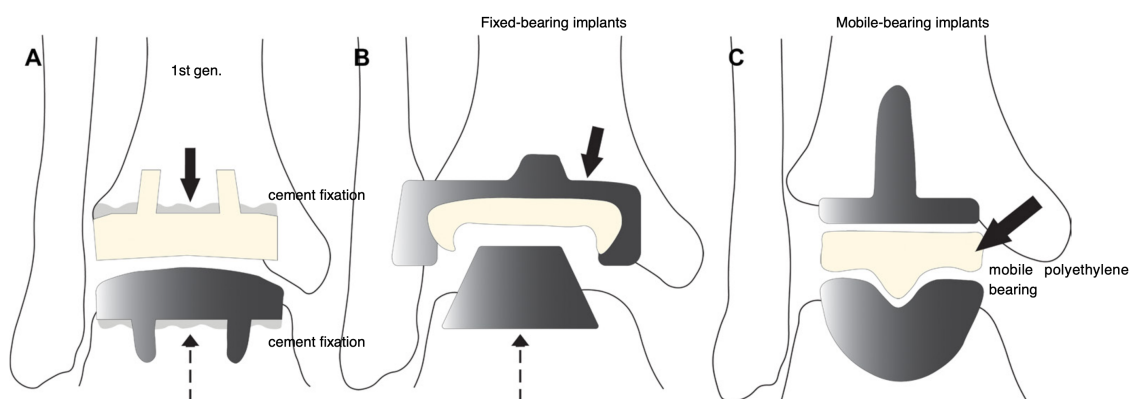


Figure 3.2: Schematic representation of total ankle replacement evolution and fixation strategies: (a) First-generation constrained or semi-constrained two-component designs utilizing bone cement fixation on both tibial and talar interfaces; (b) Modern cementless fixed-bearing designs featuring a polyethylene insert locked into a metallic component, allowing for more conservative bone resections; (c) Modern three-component mobile-bearing designs characterized by an unconstrained, freely mobile polyethylene meniscus that requires precise soft-tissue balancing. Adapted from [4].

Despite these conceptual and technical improvements, the clinical outcomes of the second generation remained heterogeneous. Complications such as aseptic loosening, polymeric insert wear, and subsidence of the talar component persisted, demonstrating that the optimization of

fixation alone was insufficient to resolve the complex load-transfer and stability challenges within the tibiotalar joint [42].

Implants such as the Agility, Buechel-Pappas, and STAR serve as prime examples of this technological transition. These systems introduced significant advances in biological fixation and articular congruency but remained highly dependent on factors such as bone quality, surgical technique, and ligamentous integrity. The accumulated clinical experience with this generation provided evidence that the long-term durability of TAA relies on a multifactorial interaction between implant design, bone conditions, and joint biomechanics, ultimately paving the way for the development of subsequent generations [4, 42].

The evolution towards the third and fourth generations of TAA represents primarily a phase of refinement and systematization of the strategies introduced during the second generation. Cementless solutions became the dominant approach, utilizing highly porous surfaces and bioactive coatings designed to promote more predictable osseointegration and consistent secondary stability [4, 42].

In parallel, there was a greater integration between prosthetic design and surgical strategy, recognizing that implant performance critically depends on lower limb alignment, ligamentous balancing, and the correction of associated deformities [4]. Within this context, the use of patient-specific instrumentation (PSI) and preoperative planning based on computed tomography (CT) have been increasingly explored to improve the accuracy of component positioning and the reproducibility of alignment. However, current literature remains insufficient to draw definitive conclusions regarding their long-term clinical impact [4, 42].

The two main design philosophies: fixed-bearing and mobile-bearing, were maintained, but are now embedded within a framework of enhanced biomechanical control and meticulous preoperative planning. Current evidence remains inconclusive regarding the absolute superiority of one philosophy over the other, suggesting that clinical outcomes are more heavily dependent on factors such as proper prosthetic alignment, soft-tissue balancing, and surgical technique precision [7, 42].

3.1.3 Contemporary Design and Fixation Strategies

3.1.3.1 Tibial Component

In contemporary total ankle arthroplasty prostheses, the design of the tibial component is guided by biomechanical principles aimed at maximizing the bone-implant contact area, reducing local stress concentrations, and ensuring sufficient primary stability to limit interfacial micromotion [9, 47]. These factors are crucial because excessive micromotion (typically $> 50\text{-}150\ \mu\text{m}$) impedes predictable osseointegration at the bone-implant interface, which can lead to aseptic loosening [47]. By promoting a more homogeneous load transfer into the tibial metaphysis, implants aim to withstand the high cyclic compressive loads of the ankle joint [48].

The distal tibia presents a favorable volume of trabecular bone and a geometry that allows for relatively broad contact interfaces, making it particularly suitable for biological fixation strategies. Modern philosophies favor preserving the subchondral bone and directing forces toward the highly vascularized, widening inner trabecular region of the bone, to mitigate stress shielding phenomena, a process where the rigid metallic implant shields the surrounding bone from natural physiological stimuli, leading to adaptive bone resorption [23]. Preserving this proximal bone stock is also critical to facilitate potential future revisions. Consequently, the majority of modern systems have transitioned from cemented designs to cementless fixation, utilizing porous surfaces, such as titanium plasma spray or trabecular tantalum metal, to promote robust osseointegration [4, 46].

From a geometric standpoint, tibial trays are designed to provide extensive coverage of the resected surface and adequate congruency with the polymeric insert, thereby contributing to a more uniform load distribution and reducing local stress concentrations [9]. Anchorage elements, such as central keels, pegs, or dual-cylinders, are incorporated to reinforce primary stability, control axial rotation, and increase resistance to shear forces [9, 47]. In contexts of significant deformity, poor bone stock, or revision surgery, intramedullary stems are frequently utilized to improve the control of the mechanical axis and enhance global stability [49].

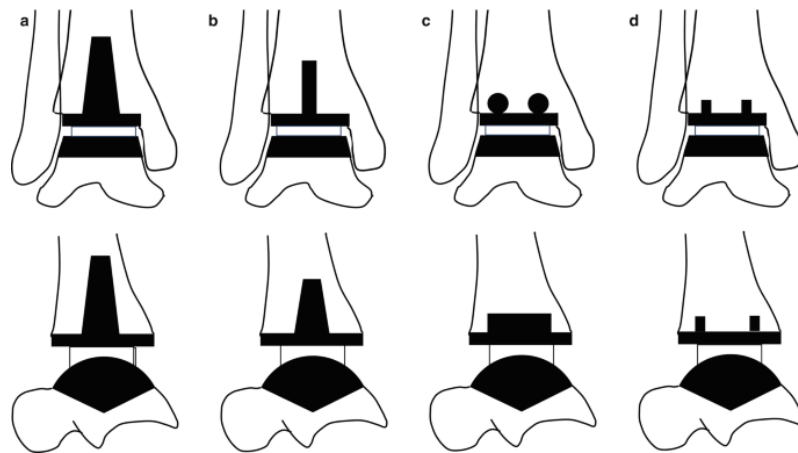


Figure 3.3: Schematic representation of different total ankle arthroplasty designs, illustrated in both coronal (top row) and sagittal (bottom) views: (a) high-anchorage central intramedullary stem combined; (b) thin, low-profile central anchoring tibial peg; (c) dual anterior-posterior cylindrical anchoring rails/keels; and (d) multiple vertical stabilizing tibial pegs paired with a multi-faceted. Adapted from [50].

Regarding materials, the structural base of the tibial tray is predominantly manufactured from titanium alloys (e.g., Ti6Al4V) [48, 51]. Titanium is favored for the tibial side due to its excellent biocompatibility and a modulus of elasticity that closely resembles that of native cancellous bone. This elastic matching helps minimize stress shielding, preventing subsequent adaptive bone resorption [51]. To ensure robust cementless fixation, the bone-interfacing surface incorporates

highly porous layers, such as titanium plasma spray or trabecular tantalum metal, and occasionally bioactive coatings like hydroxyapatite to stimulate initial bone ingrowth [51]. The opposite (inferior) surface is either highly polished or equipped with a locking mechanism to securely engage the bearing insert [34].

3.1.3.2 Talar Component

Talar fixation is widely recognized as biomechanically demanding, largely due to the anatomical characteristics of the talus. This bone features reduced dimensions, extensive cartilaginous coverage, a lower available volume of trabecular bone, and relatively vulnerable vascularization, factors that restrict its remodeling capacity and tolerance to prolonged cyclic loading. These particularities make achieving stable fixation significantly more challenging when compared to the tibial side [4, 46].

To address these limitations, the design of the talar component generally seeks to maximize talar dome coverage and preserve the subchondral bone, which is recognized as a central element in load transmission. The majority of current systems utilize cementless fixation with porous surfaces or bioactive coatings intended to promote osseointegration. However, the effective contact area remains inevitably constrained by the talar anatomy and the necessity to avoid excessive bone resection, a factor frequently implicated in talar subsidence phenomena [4, 42, 49].

In recent generations, there is an observable trend toward more anatomical geometries for the talar component. Many implants feature convex surfaces that aim to replicate the physiological talar dome and maintain articular congruency. Given that the natural tibiotalar joint is described as highly congruent throughout most of its range of motion, with biomechanical descriptions indicating congruency in approximately 96% of the functional arc, this approach aims to foster a more homogeneous load distribution and reduce local stress concentrations [46].

Some systems further incorporate kinematic control elements, such as sulcus-shaped designs or central talar ridges, which can help guide motion predominantly in the sagittal plane and limit unwanted rotations of the polymeric insert. These solutions seek to balance mobility and stability, contributing to a more predictable contact pattern [4].

In terms of anchorage, various strategies are employed, including pegs, fins, keels, or shields with screws, all designed to enhance primary stability and rotational control. The diversity of these solutions reflects the ongoing attempts to optimize the implant-talus interface within the strict constraints imposed by the available anatomy and bone quality [4, 42, 49].

Unlike the titanium-based tibial trays, the talar articular body must be predominantly manufactured from cobalt-chromium (CoCr) or cobalt-chromium-molybdenum (CoCrMo) alloys [48, 51]. These harder materials are strictly selected for their high mechanical strength and excellent wear

resistance when articulating against the polymeric insert [51]. While the articular surface is highly polished to minimize friction, the inferior bone-interfacing surface is similarly treated with porous coatings to promote biological fixation [51]. In cases of severe avascular necrosis or massive talar bone loss where standard implants are not viable, customized total talus replacements (TTR) utilizing 3D-printed cobalt-chromium or alumina ceramics have emerged as successful advanced alternatives [35].

3.1.3.3 Articular Philosophies: Fixed-Bearing vs. Mobile-Bearing

As previously noted in Section 3.1.2, two primary articular philosophies remain prevalent: fixed-bearing and mobile-bearing systems. Currently, there is no definitive consensus in the literature regarding the clinical superiority of one design over the other, although recent data highlight distinct biomechanical behaviors and failure profiles [7, 52].

In fixed-bearing systems, the bearing component is rigidly coupled to the tibial component. This design provides greater construct stability, simplifying load transfer and eliminating the risk of bearing dislocation [7, 12]. However, the reduced capacity to accommodate incongruencies, that is, the lack of relative mobility of the bearing component, can translate into a more direct load transmission, higher contact pressures, and greater shear forces at the bone-implant interfaces [7, 12].

Mobile-bearing systems feature an unconstrained meniscus that articulates freely between the tibial and talar components [53]. This mobility allows for a degree of self-alignment, greater articular congruency, and potential stress redistribution across the joint, theoretically reducing shear forces [53]. Despite these biomechanical advantages, the dual articulation introduces specific failure modes. Mobile-bearing designs are more prone to bearing subluxation or dislocation, malleolar impingement, and unfavorable wear patterns, particularly in scenarios of ligamentous imbalance [4, 7].

Regardless of the chosen articular philosophy, the choice of the bearing material is critical to mitigate wear and avoid debris-induced osteolysis. The bearing component is universally manufactured from ultra-high-molecular-weight polyethylene (UHMWPE), a polymer recognized for its chemical inertness, high ductility, and toughness [28]. Within this broad material class, it is crucial to distinguish between two main clinical formulations: conventional polyethylene (CPE) and highly cross-linked polyethylene (HXLPE) [54]. CPE refers to the standard, traditional UHMWPE used in earlier generations of implants [54]. Because CPE is susceptible to significant wear rates over time, modern TAA systems now frequently employ HXLPE, which is essentially UHMWPE that has been subjected to high-dose irradiation to increase the cross-linking of its polymer chains [10]. This structural modification provides HXLPE with superior resistance to wear, oxidative degeneration, and delamination compared to standard CPE [10, 51]. To further enhance durability and counteract the potential material embrittlement caused by the irradiation process, some

contemporary HXLPE inserts incorporate antioxidant technologies, such as Vitamin E infusion, which eliminates free radicals and promotes long-term implant survival [53]. Although alternative polymeric composites like polyether-ether-ketone (PEEK) and carbon-fiber-reinforced PEEK (CFR-PEEK) have been explored in recent computational studies as potential substitutes, the UHMWPE family remains the clinical gold standard due to its optimal performance in minimizing peak contact stresses at the articular interface [14, 48].

3.1.4 Mechanisms of Implant Failure

Despite advancements in implant design, contemporary total ankle arthroplasty systems remain susceptible to specific mechanical and biological failure modes. Current literature identifies four primary mechanisms driving component revision .

Aseptic Loosening and Subsidence: Aseptic loosening is the most frequently reported complication, occurring when the structural bond at the bone-implant interface fails. This is fundamentally driven by excessive micromotion, severe stress-shielding, or inadequate initial osseointegration. Concurrently, component subsidence - the vertical migration of the implant - frequently affects the talus. It is typically caused by localized avascular necrosis following surgical preparation, or by the mechanical insufficiency of the trabecular bone under repetitive loading [4].

UHMWPE Wear and Osteolysis: Abnormal stress concentrations and severe edge-loading accelerate the wear of the highly cross-linked or conventional polyethylene bearings [28]. The continuous generation of microscopic wear debris triggers a macrophage-mediated inflammatory response. This biological cascade up-regulates osteoclastic activity, resulting in periprosthetic expansile osteolysis (cyst formation) that catastrophically destroys the supporting bone stock [4]. In highly constrained designs, severe fatigue stresses can also lead to macroscopic fracture of the UHMWPE insert [10].

Impingement and Instability: Malalignment, ligamentous insufficiency, or overly conforming geometries can heavily alter the joint's center of rotation. This forced mechanical shift induces asymmetric load transmission, frequently promoting malleolar or gutter impingement [4, 55]. In these scenarios, the metallic components or the native bones repetitively collide during the gait cycle, causing severe pain, restricting the range of motion, and severely threatening overall implant survivorship [4].

Periprosthetic Joint Infection: While mechanical failures dominate late-stage revisions, periprosthetic joint infection (PJI) remains the most devastating biological complication, frequently precipitating early catastrophic failure [4, 42]. The clinical literature dictates that acute infections occurring in the immediate postoperative period can occasionally be salvaged through aggressive debridement, antibiotics, and implant retention (DAIR) alongside an isolated UHMWPE exchange. However, chronic or late-onset infections invariably mandate a highly morbid two-stage

revision protocol [42]. This severe intervention requires the complete extraction of all metallic components and the temporary insertion of an antibiotic-loaded cement spacer, culminating in either a complex revision arthroplasty or a definitive conversion to arthrodesis, which drastically limits patient mobility [4, 42].

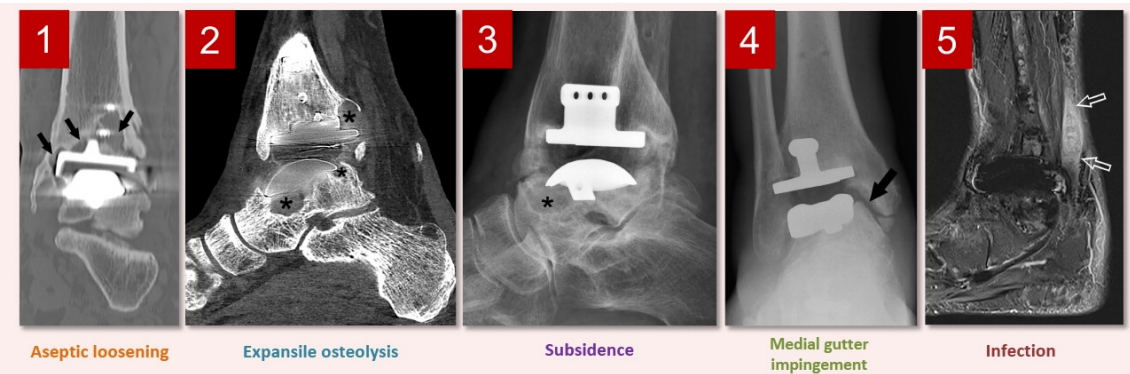


Figure 3.4: Overview of major imaging-depicted complications and failure mechanisms in contemporary total ankle replacement (TAR). Adapted from [4].

The principal drivers and consequences associated with these failure mechanisms are summarized in Table 3.1.

Table 3.1: Summary of the four primary mechanisms of implant failure in total ankle arthroplasty.

Mechanism	Primary Driver(s)	Key Consequence	Ref.
Aseptic loosening & subsidence	Excessive micromotion, stress-shielding, inadequate osseointegration (loosening); avascular necrosis or trabecular insufficiency under repetitive load (subsidence, mainly talar).	Failure of the bone-implant bond; vertical implant migration, most frequently affecting the talus.	[4]
Polyethylene wear & osteolysis	Abnormal stress concentration and severe edge-loading at the bearing surface.	Wear debris triggers macrophage-mediated inflammation and osteoclastic activity, driving expansile osteolysis (cyst formation); fatigue can fracture the insert in constrained designs.	[4, 10, 28]
Impingement & instability	Malalignment, ligamentous insufficiency, or overly conforming geometry shifting the center of rotation.	Asymmetric load transmission causing malleolar/gutter impingement; repetitive collision, pain, restricted ROM.	[4, 55]
Periprosthetic joint infection (PJI)	Biological/infectious, independent of mechanical loading.	Most devastating complication; acute cases salvaged via DAIR + PE exchange; chronic/late-onset requires two-stage revision (hardware removal, antibiotic spacer) preceding revision arthroplasty or arthrodesis.	[4, 42]

3.2 Contemporary TAA Implementations

Recent literature describes multiple total ankle arthroplasty systems featuring variations in design, materials, and fixation strategies. The following selection is not intended to be exhaustive, but rather to illustrate representative approaches currently in clinical use. These systems are frequently referenced in the literature as they present published clinical data and exemplify different prosthetic engineering philosophies within the latest generations of TAA [4, 42, 49].

3.2.1 STAR® Scandinavian Total Ankle Replacement



Figure 3.5: The Scandinavian Total Ankle Replacement (STAR) prosthesis system. Left: Isolated view of the STAR prosthesis components. Adapted from [56]. Right: Schematic representation of the implanted prosthesis within the ankle joint configuration, adapted from [57].

Design Characteristics The STAR prosthesis (originally Stryker, now DJO Global) is a widely utilized unconstrained, mobile-bearing, three-component system [4, 35, 42, 53]. The contemporary cementless version comprises a flat titanium or cobalt-chromium tibial tray anchored by two cylindrical barrels, a convex cobalt-chromium talar component featuring a central stabilizing fin, and a mobile UHMWPE meniscus [4, 34, 42]. This meniscus uniquely incorporates a radiopaque wire within its structure to facilitate postoperative radiographic monitoring of its position and potential wear [4]. A distinct geometric feature is the centrally placed talar ridge on the articulating surface, designed to resist rotational movement and guide motion predominantly in the anteroposterior direction [4, 35]. The bone-interfacing surfaces are treated with a titanium plasma spray to promote robust osseointegration [42, 53].

Biomechanical Discussion As a clinical benchmark for mobile-bearing designs, the STAR system demonstrates robust early survivorship, ranging from 89% to 96% at 5 years [4, 5, 42, 44]. Following this initial success, survivorship declines significantly to approximately 63% to 73% at 15 years [4, 5, 42]. Critical analysis links this late-term failure directly to specific geometric vulnerabilities of the STAR implant at both the osseous and articular levels [10, 58].

At the osseous level, the STAR's requirement for extensive flat bone resections severely compromises subchondral support [58]. Finite element analyses (FEA) of this specific implant reveal a drastic strain imbalance: lateral bone zones lose up to 96% of their physiological loading, triggering rapid stress shielding and resorption [59]. Simultaneously, medial and posterior zones experience compensatory strain increases of up to 990%, predisposing the bone to hypertrophy and fatigue fracture [59]. This precise mechanical mismatch drives the uniquely high 17,6% talar subsidence rate associated with this prosthesis [58].

At the articular interface, the unconstrained mobile bearing is highly sensitive to surgical positioning and soft-tissue balance [4, 60]. FEA demonstrates that minor coronal misalignments, as little as 5 degrees of varus or valgus, drastically shift the load vector, causing severe edge-loading [34, 60]. These concentrated forces frequently push the UHMWPE contact pressures beyond the 10 MPa yield threshold, accelerating delamination [59]. Consequently, retrieval studies report that 14% to 18% of STAR implants suffer macroscopic fracture requiring bearing exchange [4, 10]. Furthermore, poor intraoperative ligament balancing can lead to angular instability and excessive joint laxity, resulting in the subsequent migration or displacement of the mobile polyethylene meniscus [4].

The STAR system historically revolutionized the field by introducing cementless, mobile-bearing technology that offered significant early biomechanical advantages over older rigid designs [55, 59]. Ultimately, its long-term performance is overshadowed by these severe geometric and stability vulnerabilities [61]. For this reason, recent meta-analyses and international joint registries confirm a statistically significantly lower survival rate, explaining the progressive decline of its clinical utilization in contemporary practice in favor of newer, fourth-generation implants [55, 61].

3.2.2 Infinity® Total Ankle System (Stryker Corporation)

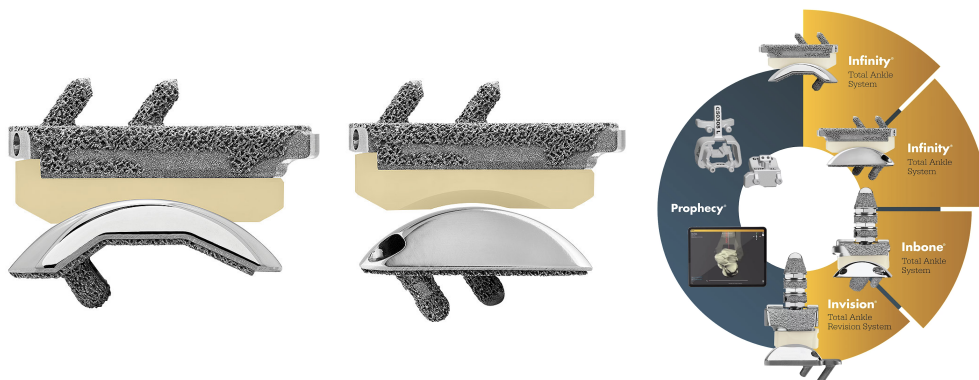


Figure 3.6: The Stryker total ankle replacement ecosystem. Left: Infinity Total Ankle System, adapted from [62]. Right: Commercial overview chart illustrating the clinical integration between the Prophecy preoperative navigation/patient-specific instrumentation (PSI) and the Infinity, IN-BONE, and Invision implant portfolios. Adapted from [63].

Design Characteristics The Infinity system is a contemporary fourth-generation, fixed-bearing TAR engineered for maximal bone preservation [4]. The tibial component features a low-profile titanium alloy tray anchored by three stabilizing pegs to resist rotational forces, coupled with a HXLPE insert that rigidly locks into the baseplate [9, 64]. Inferiorly, it utilizes a highly conforming cobalt-chromium (CoCrMo) talar dome equipped with two anterior pegs to enhance primary osseous fixation [4, 9]. Rather than relying strictly on a traditional straight flat resection, the standard talar component employs a bone-sparing chamfered cut design, which provides a more anatomical fit and preserves critical cortical bone stock [62]. A distinct geometric advantage of this system is its modular cross-compatibility, allowing its talar component to be interchangeable with modular stems from other systems to create hybrid constructs for complex revision surgeries [4, 49]. Precise initial component positioning is heavily facilitated through integration with computed tomography (CT)-derived patient-specific instrumentation (PSI) [4].

Biomechanical Discussion From a clinical perspective, recent large-cohort studies report exceptional early to mid-term survivorship for the Infinity system, ranging from 93% to 98,8% at two to five years, alongside significant improvements in functional outcomes [11]. Critical finite element analyses provide deeper insights into the mechanical mechanisms driving these clinical observations [9, 64].

At the osseous level, the system's reliance on three small tibial pegs instead of long intramedullary stems preserves a greater amount of dense cortical bone. This geometric strategy drastically reduces bone-implant interface micromotion, providing superior initial fixation and stability. Furthermore, the solid titanium tibial tray experiences high stress shielding compared to other modern implants, causing a reduction in physiological load transfer that can initiate long-term bone resorption [9].

At the articular interface, patient-specific wear prediction frameworks raise significant concerns regarding the long-term endurance of the fixed bearing. Computational studies demonstrate that the Infinity prosthesis exhibits a significantly higher volumetric wear rate compared to anatomical-design prostheses, with the latter showing a 49,5% lower wear rate. Progressive volumetric wear on the bearing surface causes slight increases in internal/external rotation and anterior/posterior translation of the ankle joint. This altered kinematics further varies the contact stresses, creating a negative feedback loop of progressive wear [64]. Driven by these biomechanical vulnerabilities, clinical registries frequently associate this implant with elevated early revision rates, primarily due to aseptic loosening of the tibial component [11, 49].

All this suggests that although the Infinity system demonstrates excellent early survivorship, classifying it as the definitive total ankle prosthesis remains premature [4, 65]. While current models confirm its superior initial stability [9], its fixed-bearing architecture could create a mechanical weakness: by rigidly locking the HXLPE insert, it transfers unabsorbed rotational shear stresses directly to the tibial bone-implant interface [9]. Over time, this dynamic overloading, coupled with accelerated HXLPE wear and stress shielding, directly drives its most frequent early clinical failures: aseptic loosening and tibial component subsidence [49, 64]. Consequently, since robust

long-term clinical data for this modern implant are inherently lacking [42], these well-documented early failure mechanisms prove that critical design improvements are still required before its ultimate longevity can be confirmed [4, 42].

3.2.3 Trabecular Metal™ Total Ankle (Zimmer Biomet, USA)

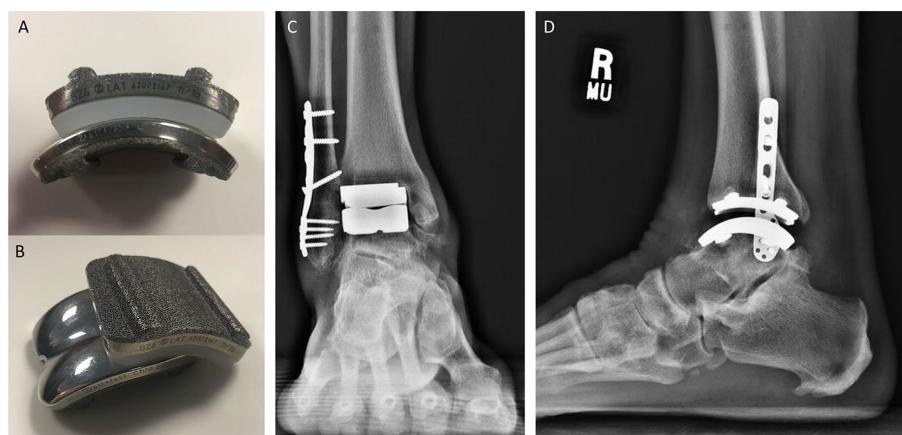


Figure 3.7: The Zimmer Biomet Trabecular Metal Total Ankle system overview. A: Coronal view of the assembled semi-constrained prosthesis components. B: Oblique view demonstrating the highly porous tantalum (Trabecular Metal) texture applied on the bone-implant interfacing surfaces to optimize secondary biological fixation. C: Postoperative weight-bearing anteroposterior (AP) radiograph showing the implant position and the fixation plate utilized to stabilize the mandatory transfibular osteotomy. D: Postoperative lateral radiograph displaying the dual-radius talar dome geometry and the curved sagittal profile of the components. Adapted from [8].

Design Characteristic The Zimmer Trabecular Metal (TM) Total Ankle is a contemporary semiconstrained, fixed-bearing system. Uniquely, it is the only current total ankle prosthesis to utilize a lateral transfibular approach, completely avoiding the traditional anterior incision [53]. This technique requires a fibular osteotomy, allowing direct visualization of the center of rotation and facilitating precise anatomical alignment [66]. Structurally, the implant utilizes a HXLPE bearing that rigidly snaps into a titanium tibial baseplate, coupled with a highly conforming cobalt-chromium-molybdenum (CoCrMo) talar component [10]. A defining hallmark is the integration of Trabecular Metal, a highly porous tantalum biomaterial applied to the bone-implant interfaces to promote robust osseointegration [51].

For primary stabilization, rather than relying on vertical pegs, both components are secured using a distinct rail fixation system [4]. The implants are inserted from lateral to medial along rails cut into the resected bone [67]. Geometrically, the talar articular surface replicates the frustum of a cone to safely accommodate natural joint kinematics [15]. Furthermore, guided by a specialized rigid external alignment frame, the system employs anatomically curved bone resections rather than traditional straight flat cuts, significantly preserving vital subchondral bone stock [51, 66].

Biomechanical Discussion From a biomechanical perspective, manufacturer-affiliated finite element models suggest distinct advantages over traditional fixed-bearing designs. Finite element analyses indicate that its highly conforming bicondylar geometry may create a significantly larger contact area at the articular interface, potentially reducing peak contact pressures [60]. Furthermore, computational models suggest that this specific geometry is highly resistant to edge-loading, remaining relatively insensitive to minor varus or valgus malalignments compared to other modern implants [60]. At the bone-implant interface, because the integration of highly porous tantalum mimics that of cancellous bone, it could theoretically minimize the severe stress shielding typically associated with solid titanium implants, while its high friction coefficient is proposed to promote exceptional primary stability and robust early osseointegration [51]. When examining independent systematic reviews and clinical registries, the literature reveals that the mandatory lateral transfibular approach creates a unique set of surgical vulnerabilities. The most frequent causes for clinical failure and reoperation are extra-articular: delayed union or nonunion of the fibular osteotomy, medial gutter impingement (frequently exacerbated by limited medial visualization during the lateral approach), and the mandatory removal of symptomatic lateral fixation hardware [45, 68]. Furthermore, independent registries also report persistent intra-articular complications, including the development of periprosthetic osteolytic cysts, deep infections, and component subsidence [15, 68].

Although the transverse lateral-to-medial insertion of the tibial rails could provide an initial advantage by effectively blocking sagittal shear forces [66], it could simultaneously create a concentration of maximum von Mises stresses along these specific fixation rails, while peak contact pressures could accumulate asymmetrically on the talar domes [34]. Furthermore, although the system employs anatomically curved resections purposefully designed to preserve vital subchondral bone stock, this press-fit design lacks deep intramedullary stems. Because it relies exclusively on shallow rails for primary stabilization, its mechanical success remains heavily dependent on the patient's local bone quality.

Ultimately, the current literature only permits us to infer that the Zimmer TM prosthesis provides excellent short- to mid-term biological fixation and stability, but at the direct cost of increased surgical morbidity related to the fibula. Because robust, independent, long-term clinical data are still inherently lacking, it remains impossible to definitively infer whether these theoretical biomaterial innovations will translate into superior long-term survivorship or successfully prevent late-stage mechanical failures [42, 45].

3.2.4 Exactech® Vantage® Total Ankle System



Figure 3.8: The Exactech® Vantage® Total Ankle System configuration, illustrating the complete two-component fixed-bearing assembly from an oblique perspective (left) and a coronal perspective (right). Adapted from [69].

Design Characteristics The Exactech Vantage is a contemporary fixed-bearing total ankle replacement system [28]. Structurally, it comprises a Cobalt-Chromium-Molybdenum (CoCrMo) talar component, a CoCrMo tibial tray, and an UHMWPE bearing insert infused with Vitamin E to theoretically decrease wear rates [4, 28]. Geometrically, the articulating interface features a highly conforming bicondylar design defined by specific talar and bearing condylar radii, as well as a distinct medial radius engineered to optimize joint congruency [28]. For primary stabilization, rather than employing transverse rails or deep stems, the tibial tray relies on a specific multiple-peg architecture, prominently featuring two anterior pegs coupled with a distinct posterior peg [4, 28].

Biomechanical Discussion While the talar component attempts to replicate the natural joint anatomy using a highly conforming bicondylar design, this geometric approach carries a severe structural consequence. The independent literature states that maximum von Mises stresses concentrate intensely on the condyles in the central region of the bearing, precisely where this anatomical geometry forces the UHMWPE to be the thinnest [28]. Furthermore, unaffiliated models suggest that its specific stabilization architecture creates a critical load imbalance: the posterior tibial peg sustains a distributed load twice as large as the anterior pegs. This uneven force transmission suggests the induction of severe edge-loading on the posteroinferior aspect of the bearing surface [28]. Over time, this intense localized overloading suggests a severe compromise of the bone-implant interface. Consequently, recent registries are beginning to report specific early clinical failures, most notably aseptic loosening and early subsidence of the tibial component [4, 49]. Ultimately, because robust independent long-term clinical data are still inherently lacking, it remains impossible to definitively infer whether the biological innovation of Vitamin E infusion will be sufficient to counteract these severe geometric stress concentrations and successfully prevent late-stage mechanical failures [28].

3.2.5 Cadence® Total Ankle System

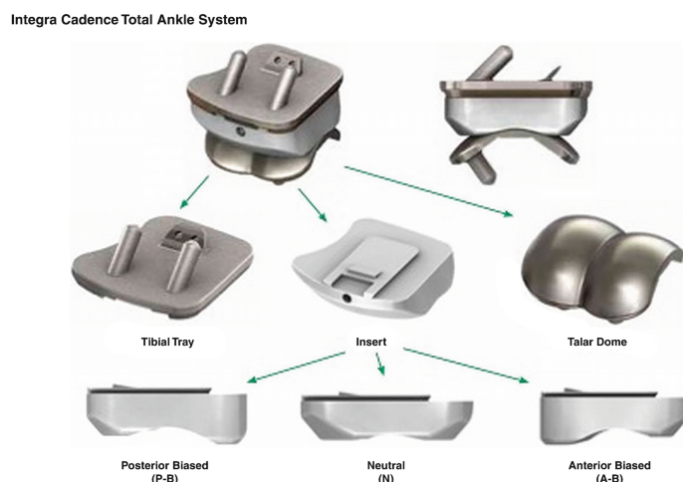


Figure 3.9: Component breakdown of the Cadence® Total Ankle System, showing the structural interplay between the modular tibial tray (featuring a targeted lateral fibular recess) and the distinct HXLPE insert varieties, including posterior-biased, neutral, and anterior-biased sagittal configuration options. Adapted from [70].

Design Characteristics The Cadence Total Ankle System is a contemporary fourth-generation, two-component, fixed-bearing implant [53]. Structurally, it comprises metallic tibial and talar components designed for cementless fixation, utilizing optimized surface technologies to promote robust osseointegration [7]. For primary stabilization, the unique tibial component is engineered with a specific lateral recess; this design aims to precisely accommodate the fibula, theoretically providing three cortices of bone support while minimizing the risk of extra-articular fibular impingement [4, 53]. Geometrically, the talar component is laterality-specific and highly anatomical, featuring a bicondylar design with a central trochlear groove specifically intended to minimize necessary bone resection [4]. The articulating bearing is manufactured from HXLPE to reduce wear-related complications, and uniquely offers anterior and posterior biased options to address the sagittal plane balance of the joint [4].

Biomechanical Discussion Early clinical reports note positive short-term outcomes, demonstrating significant improvements in patient activity and pain scores [4, 53]. However, the unique anatomical features of the Cadence system introduce specific biomechanical trade-offs. As previously discussed for similar highly anatomical designs, the bone-preserving bicondylar geometry of the talus inherently could force the central region of the HXLPE bearing to be its thinnest structural point, leading to concentration of maximum von Mises stresses directly on this vulnerable central area, potentially accelerating localized material wear. Furthermore, the unique implementation of anterior and posterior biased liners represents an elegant surgical strategy to restore sagittal balance [4]. However, it could carry some mechanical risk. Shifting the joint's center of rotation to correct malalignment theoretically induces asymmetric load transmission across the bearing.

This forced mechanical shift may promote severe edge-loading on the specific biased edges of the HXLPE insert, generating highly localized peak contact pressures.

This creates a fascinating biomechanical scenario within contemporary arthroplasty. The implant equips surgeons with unique modular tools for deformity correction, yet these exact geometrical innovations theoretically could predispose the system to severe asymmetric overloading. Clinically, these theoretical mechanical vulnerabilities strictly correlate with the established failure mechanisms documented in short- to mid-term literature for the Cadence system. As reported, the primary include delayed wound healing, periprosthetic and malleolar fractures, anterior impingement, and the critical occurrence of periprosthetic osteolysis [7]. The generation of microscopic wear debris (consequence of the aforementioned localized peak contact pressures and edge-loading), triggers an adverse biological response that promotes localized bone destruction [49].

3.2.6 INBONE® II Total Ankle System



Figure 3.10: The INBONE Total Ankle System: (a) 3D anatomical placement highlighting the intramedullary tibial stem fixation, and (b) Detailed lateral view of the modular prosthetic assembly. Adapted from [71].

Design Characteristics The INBONE Total Ankle System is a contemporary fixed-bearing implant, available in two progressive iterations (INBONE I and INBONE II). Structurally, both components consist of a titanium-coated cobalt-chromium (CoCr) tibial tray and talar dome, coupled with an ultra-high-molecular-weight polyethylene (UHMWPE) bearing insert [4, 53]. For primary stabilization, the system fundamentally diverges from shallow-peg or rail designs by employing highly invasive, modular intramedullary stems for both the tibia and the talus [4, 36]. Geometrically, while the INBONE I featured a flat talar articulation, the INBONE II was modified to include a highly conforming central sulcus on the talar surface to improve coronal stability, alongside two additional anterior pegs for enhanced fixation [4, 36].

Biomechanical Discussion While the robust modular stems provide exceptional initial stability and represent a powerful salvage option for patients with poor bone stock, this system carries

extreme biomechanical and surgical trade-offs. The mandatory surgical technique requires intramedullary reaming through a plantar approach, driving a drill directly through the calcaneus and the talus to access the tibia. The clinical literature states that this severe structural violation severely compromises the talar blood supply, frequently driving osteonecrosis, which directly explains the high incidence of catastrophic talar subsidence that plagued the INBONE I system [36, 42]. To counteract this instability, the INBONE II introduced an anatomical talar sulcus, a modification that inherently introduces the severe geometric downsides and stress concentrations discussed previously [28].

Despite these mechanical flaws, the system possesses a highly advantageous clinical feature: complete modularity. The INBONE II tibial component is specifically designed to be interchangeable with the low-profile Infinity talar component. This creates a popular hybrid configuration that allows surgeons to exploit the massive INBONE tibial stem for superior fixation in poor bone stock, while simultaneously utilizing the Infinity talus to minimize talar bone resection [4].

Fundamentally, the INBONE system embodies a severe biomechanical paradox. It provides unparalleled modular fixation capable of rescuing severely compromised bone, yet it remains plagued by specific mechanical and biological failure modes. When the system fails, current literature dictates that the most frequent causes are aseptic loosening and component subsidence [4, 36].

3.2.7 Other Notable Paradigms



(a) B.O.X.®



(b) Salto®



(c) HINTEGRA®



(d) 3D-printed TTR

Figure 3.11: Alternative design paradigms and customized solutions in TAR: (a) the three-component mobile-bearing B.O.X.® system, adapted from [72]; (b) the anatomical Salto® configuration, adapted from [8]; (c) the low-resection HINTEGRA® system incorporating unique anterior shield screw fixation, adapted from [73]; and (d) an exploded schematic of a personalized, 3D-printed customized Total Talar Replacement (TTR) coupled with a minimal-resection tibial baseplate, adapted from [74].

While the aforementioned systems represent the modern paradigm of highly modular, low-profile, and bone-preserving fourth-generation designs, a comprehensive biomechanical review must acknowledge other relevant design paradigms.

The B.O.X.® (Bologna-Oxford) Total Ankle is a three-component mobile-bearing system with a unique design: its tibial surface is curved in both anteroposterior and mediolateral planes to theoretically restore exact isometric ligament tensioning [55]. However, this complex geometry frequently fails in vivo. Independent long-term follow-ups report a concerning drop to a 66% survivorship rate at ten years, primarily driven by persistent malleolar impingement and inherent mobile-bearing complications, such as polyethylene wear and dislocation [55].

The Salto® Total Ankle prosthesis, initially introduced as a three-component mobile-bearing system and subsequently adapted into the fixed-bearing Salto Talaris®, incorporates an asymmetrical talar geometry with a larger lateral radius and a conical-shaped facet to precisely replicate the natural rotational axis of the ankle [4, 42]. To secure the implant against the bone surface, the titanium-coated cobalt-chromium tibial component utilizes a central keel or a tapered fixation plug [4, 42]. While computational models establish that this keel minimizes micromotion and confers superior primary stability [9, 13], its surgical insertion compromises the anterior tibial cortex. Coupled with an undersized tibial tray, this design forces joint loads onto the weaker cancellous bone, driving long-term stress-shielding and component subsidence [4]. Consequently, despite positive initial stability and excellent midterm survivorship, long-term clinical follow-ups strictly establish specific mechanical vulnerabilities. The primary documented complications driving surgical revision for this design include severe medial or lateral gutter impingement, medial malleolar stress fractures, and the formation of periprosthetic bone cysts. These cysts are primarily induced by mechanical stress and the subsequent biological response to the release of polyethylene wear particles [4].

Similarly, the HINTEGRA® system represents an unconstrained, three-component mobile-bearing paradigm uniquely characterized by a flat tibial surface and a conical talar profile [42]. Both components employ an anterior shield with holes for screw fixation, an anatomical strategy explicitly designed to minimize distal tibial and talar bone resection to a mere 2 to 3 mm [42]. Although mechanically preserving bone stock and demonstrating satisfactory short-term pain relief, comprehensive long-term registry data document a varying ten-year survivorship declining to between 68% and 84% [42]. The established causes for revision in this specific configuration are fundamentally driven by aseptic component loosening, mobile meniscus subluxation, and a remarkably high incidence of posterior tibial calcifications, which have been radiographically observed in up to 54,6% of cases [55].

Representing the forefront of personalized solutions, customized 3D-printed Total Talar Replacement (TTR) has emerged for patients with severe talar osteonecrosis [75]. By replacing the

entire talus, it entirely eliminates talar bone resection [75]. However, replacing the talus introduces extreme biomechanical complexity, as this bone acts as the structural keystone of the foot, simultaneously articulating with the tibia, fibula, calcaneus, and navicular. Furthermore, its complete implantation mandates the extensive resection of the surrounding collateral ligaments, severely challenging passive joint stability [75]. While computational models show TTR replicates healthy foot kinematics better than standard replacements, they warn that managing this multi-joint articulation is incredibly difficult: even a minor anatomical mismatch in the 3D implant heavily alters the load path, drastically increasing contact forces on the adjacent midtarsal joints and lateral column [75, 76]. Because robust long-term clinical data are lacking, TTR remains a fascinating but highly experimental biomechanical solution [42, 75].

To facilitate comparison between the currently available total ankle replacement systems, the principal design characteristics, fixation strategies, and their associated biomechanical advantages and vulnerabilities are summarized in Table 3.2.

Table 3.2: Comparative overview of contemporary total ankle replacement systems: design characteristics, fixation strategy, and biomechanical trade-offs.

Implant	Generation / Type	Approach	Fixation	Bearing	Biomechanical Advantages	Biomechanical Vulnerabilities
S.T.A.R.	3rd gen., mobile	Anterior	Tibial keel + talar pegs	UHMWPE	Unconstrained rotation reduces rotational shear; three-component design offloads ligamentous structures	Keel compromises anterior tibial cortex; high polyethylene wear and dislocation risk
Salto / Salto Talaris	3rd/4th gen., mobile/fixed	Anterior	Central tibial keel or fixation plug	UHMWPE	Asymmetric talar geometry replicates natural rotational axis; keel minimizes micromotion	Undersized tray forces load onto cancellous bone, promoting stress shielding and subsidence; periprosthetic cyst formation
B.O.X.	3rd gen., mobile	Anterior	Tibial and talar pegs	UHMWPE	Dual-curvature tibial surface theoretically restores isometric ligament tensioning; three-component kinematics reduce constraint	Complex geometry fails in vivo; persistent malleolar impingement; mobile-bearing dislocation risk
HINTEGRA	3rd gen., mobile	Anterior	Anterior shield screws	UHMWPE	Minimal bone resection; anterior screw fixation preserves bone stock; flat tibial surface reduces constraint	Mobile meniscus subluxation risk; high incidence of posterior tibial calcifications
INBONE II	4th gen., fixed	Anterior	Modular intramedullary stems	HXLPE	Exceptional primary stability; modular hybrid constructs possible; useful option for poor bone stock	Intramedullary reaming risks talar vascular compromise; conforming sulcus introduces condylar stress concentrations
Infinity	4th gen., fixed	Anterior	Multi-peg tibial + talar	HXLPE	Low-profile, bone-sparing design; compatible with INBONE talar for hybrid constructs	Peg tips concentrate stress locally; talar interface mechanically vulnerable under asymmetric loading
Cadence	4th gen., fixed	Anterior	Multi-peg tibial + talar	HXLPE	Lateral fibular recess provides three-cortex tibial support; biased liner options allow sagittal balance correction	Conforming geometry forces polyethylene to its thinnest centrally; biased liners introduce asymmetric edge-loading risk
Vantage	4th gen., fixed/mobile	Anterior	Multi-peg tibial + talar	UHMWPE (Vitamin E)	Anatomical bicondylar talar design; available in fixed and mobile configurations	Posterior peg sustains higher load, generating posterior bearing edge-loading; central polyethylene thinning risk
Zimmer TM	4th gen., fixed	Lateral transfibular	Transverse rails	HXLPE	Bone-sparing arcuate resections rest on dense subchondral bone; porous tantalum promotes osseointegration; rails resist coronal shear	Anterolateral rail terminus concentrates yield-exceeding strains under routine loading; persistent medial periprosthetic underloading; fibular osteotomy increases surgical complexity

3.3 Chapter Conclusion

The structural analysis compiled in this chapter highlights a persistent biomechanical trade-off: while highly conforming anatomical surfaces minimize local contact pressures, they frequently force the polymeric insert to its thinnest structural limit, and fixed-bearing architectures transfer unabsorbed rotational shear stresses directly to vulnerable bone-implant interfaces.

Ultimately, the wide array of contemporary designs proves that a definitive international consensus regarding the optimal implant geometry has not yet been achieved. This ongoing debate manifests as an extensive, multi-level design variability across modern portfolios, spanning from fundamental articular philosophies (fixed-bearing versus mobile-bearing unconstrained constructs) to radically different bone anchorage systems (including shallow multi-peg configurations, transverse geometry rails, central longitudinal keels, and highly invasive modular intramedullary stems). Furthermore, even the surgical access pathways remain unstandardized, as demonstrated by the divergence between traditional anterior instruments and specialized lateral transfibular approaches.

Nevertheless, amid this design polarization, the accumulation of long-term clinical registry data and retrieval frameworks has established definitive mechanical certainties that directly dictate contemporary design premises.

Chapter 4

National Orthopedic Registries

Orthopedic joint registries function as large-scale, systematic databases that aggregate clinical, demographic, and technical information regarding joint replacement procedures [77]. By providing longitudinal tracking of implant survivorship, complication rates, and patient outcomes in real-world settings, these databases act as crucial early-warning systems to identify underperforming implants and potential safety concerns [77]. While isolated clinical studies and developer-led reports provide valuable initial insights into implant performance, they often comprise small sample sizes and may be subject to publication bias [11]. In contrast, registries facilitate surgical benchmarking and the assessment of emerging technologies, offering a comprehensive and unbiased approach to evaluating arthroplasty outcomes at a population level [11, 44].

To accurately evaluate the true clinical efficacy, current market dynamics, and specific complication rates of Total Ankle Arthroplasty, this thesis draws upon the most recent and comprehensive data available from the world’s leading national joint registries. It is important to note that these documents represent the most recently published data available from each respective organization at the time of writing. Consequently, their publication years vary (spanning from 2024 to 2026) due to differences in national auditing and reporting cycles. Specifically, this analysis is grounded in the latest annual reports from:

- The **Australian Orthopaedic Association National Joint Replacement Registry (AOANJRR)** – 2025 Supplementary Report [78]. The AOANJRR is frequently highlighted in the literature for its rigorous methodology in actively identifying underperforming “outlier” prostheses [77, 78];
- The **National Joint Registry (NJR)** for England, Wales, Northern Ireland, the Isle of Man, and Guernsey – 22nd Annual Report 2025 [79]. Widely recognized as the largest orthopaedic registry in the world with millions of records, the NJR provides unparalleled statistical power for population-level trends and long-term implant survivorship [77, 79];

- The **New Zealand Joint Registry (NZJR)** – 26 Year Report 2025 [80]. The NZJR is highly valued in scientific literature for its longevity and for being a pioneer in systematically collecting Patient-Reported Outcome Measures (PROMs) directly alongside implant survivorship data [80];
- The **Swedish Ankle Registry (SwedAnkle)** – Annual Report 2024 [81]. This registry is unique for its long-standing and exclusive focus on the ankle joint (tracking replacements since 1993), its direct comparison between TAR and ankle arthrodesis, and its integration of the validated, region-specific SEFAS questionnaire [81].
- The **Norwegian Arthroplasty Register (NAR)** – Report 2024 [82]. Collecting ankle data since 1994, the NAR is renowned for its high data completeness, cross-validated with the Norwegian Patient Registry (NPR)[82].

Additionally, the **American Joint Replacement Registry (AJRR)** [83] and the **Canadian Joint Replacement Registry (CJRR)** [84] were consulted to establish robust survivorship benchmarks for hip and knee arthroplasties. Utilizing this large-scale, real-world data is particularly crucial for identifying universal trends and persistent biomechanical challenges in TAR that might not be evident in smaller cohorts [44].

4.1 Patient Demographics and Clinical Profile

The demographic profile of patients undergoing Total Ankle Arthroplasty presents a fascinating contrast to the broader joint replacement landscape. While total hip and knee arthroplasties are predominantly performed on female patients (constituting approximately 58% to 60% of cases globally) [83], national joint registries demonstrate that TAR exhibits a distinct gender distribution, often presenting as a male-dominated procedure in several major countries 4.1.

Table 4.1: Demographic profile (mean age and gender distribution) and primary clinical indications for primary TAA across major national registries.

Country (Registry)	Mean Age (M F)	Gender Distribution (Male)	Primary Diagnosis / Indication
Australia (AOANJRR)	67,4 (M: 68,0 F: 66,5)	62,1%	Osteoarthritis (94,4%) Rheumatoid Arthritis (3,6%)
United Kingdom (NJR)	~67,5 (M: 68,4 F: 66,7)	60,6%	Osteoarthritis (87,0%) Rheumatoid Arthritis (6,0%)
New Zealand (NZJR)	66,9 (M: 66,8 F: 66,9)	60,4%	Osteoarthritis (75,7%) Rheumatoid Arthritis (8,2%)
Sweden (SwedAnkle)	63,9 (M: 66,1 F: 60,8)	58,0%	Post-Traumatic OA (49,3%) Primary OA (27,2%)
Norway (NAR)	60,5	46,7%	Osteoarthritis (57,7%) Rheumatoid Arthritis (26,5%)

Unlike the hip and knee, where primary (idiopathic) osteoarthritis is the absolute leading cause, end-stage ankle arthritis presents a unique epidemiological profile. While the largest registries from the Anglosphere, such as the UK and Australia, broadly group the vast majority of primary indications under the umbrella term of "Osteoarthritis" (accounting for 94,4% of cases in Australia) [78, 85], global clinical literature establishes that ankle osteoarthritis frequently develops following severe trauma, with post-traumatic joint degeneration occurring in up to 70% of these ankle injuries[85]. Scandinavian databases provide a more granular clinical stratification. Both the Swedish Ankle Registry and the Norwegian Arthroplasty Register explicitly differentiate the types of arthritis, revealing that post-traumatic osteoarthritis (developing as a sequela of fractures or injuries sustained earlier in life) is actually the dominant indication for ankle replacement, actively surpassing primary idiopathic osteoarthritis [81, 82].

Furthermore, even in databases where women retain a slight majority, such as Norway (53,3% female), this demographic anomaly does not contradict the trauma paradigm. Instead, the Norwegian exception could be driven by an unusually high local incidence of Rheumatoid Arthritis (accounting for 26,5% of primary indications), which consists of an autoimmune disease predominantly affecting females, rather than a shift in the established post-traumatic osteoarthritis pattern [82, 85].

Global data also indicates that the average age at the time of primary ankle surgery generally falls within the 60s. In the largest registries, such as the UK and Australia, the 65 to 74-year-old cohort represents the most common age group. However, Scandinavian registries report slightly younger patient populations. The average age at primary surgery drops to approximately 60,5 years in Norway, while the latest data from the Swedish registry indicates a contemporary mean age of 63,9 years for patients undergoing total ankle replacement.

This younger demographic in Scandinavia directly reflects an epidemiological profile driven by early-onset trauma and aggressive autoimmunity [81, 82]. Unlike the gradual, age-related wear of idiopathic osteoarthritis, a joint fracture sustained in youth triggers a premature mechanical decline through localized chronic overload, while Rheumatoid Arthritis inflicts an active, systemic inflammatory destruction of the cartilage. Consequently, both etiologies accelerate joint degradation, leading to end-stage arthritis decades ahead of the normal biological timeline [81, 82].

4.2 Evolution of Implant Survivorship and Clinical Performance

Despite the fact that early evidence of first-generation implants introduced in the 1970s showed high rates of postoperative complications and revisions, which initially established ankle arthrodesis as the more common surgical treatment, the registries show that Total Ankle Arthroplasty has since undergone a significant clinical resurgence [11, 85].

To accurately assess the "life curve" of these devices, national registries globally employ the Kaplan-Meier survival analysis, a statistical method that estimates the probability of an implant

avoiding failure over time while accounting for patients lost to follow-up or death [79, 85]. While the underlying mathematics are identical, registries present these data using two inverse metrics. The Swedish (SwedAnkle), Norwegian (NAR), and New Zealand (NZJR) registries typically report *implant survivorship* (the percentage of prostheses remaining unrevised, descending from 100%). In contrast, the United Kingdom (NJR) and Australian (AOANJRR) registries report the *cumulative percent revision* (the probability of failure ascending from 0%).

Regardless of the reported metric, a recent consortium study analyzing survival curves across Australia, New Zealand, Norway, and Sweden provides a definitive historical baseline for modern arthroplasty up to 2019: aggregate data demonstrated that early survivorship was highly successful, with over 94% of implants surviving at 2 years, but gradually declined to an average survivorship of approximately 77% at 10 years [85].

However, while this global consortium establishes a historical baseline, contemporary joint registries provide strong evidence that this survivorship is actively improving. Data derived from the latest national databases, reflecting the most recent real-world performance across diverse patient populations, show that this positive shift is largely driven by advances in modern implant geometry, improved fixation strategies, and increased surgical experience [77].

The Australian Orthopaedic Association National Joint Replacement Registry (AOANJRR), for instance, reports a near 50% reduction in mid-term revision rates. The 7-year cumulative revision rate decreased significantly from 12,9% for procedures undertaken prior to 2015, down to just 6,4% for those performed from 2015 onwards. Similarly, data from the Swedish Ankle Registry indicates that modern prosthetic designs achieve substantially higher survivorship, exceeding 84% at 10 years, compared to approximately 69% for earlier legacy designs [81, 85].

Despite these generational improvements proven by the latest registry data, revision rates remain clinically significant. The persistence of complications indicates that challenges related to implant-bone interaction and biomechanical load transfer are not yet fully resolved. Furthermore, TAR survivorship still lags behind that of total hip and knee arthroplasties, which typically exceed 94% to 97% survival at 10 years [83, 84]. This discrepancy reflects the increased biomechanical complexity of the ankle joint and reinforces the absolute need for continued computational optimization of implant design.

4.3 Global Market Convergence and Regional Drivers

While individual countries exhibit unique evolutionary drivers, a transversal analysis of global registries reveals a clear convergence in market behavior across borders. The most striking universal trend is the definitive paradigm shift from mobile-bearing to fixed-bearing implants. Driven by clinical evidence demonstrating superior survivorship, fixed-bearing devices now dominate modern surgical practice, accounting for approximately 80% of all primary operations in the UK [79] and surging to nearly 90% of current annual procedures in Australia [78]. The justification for

this mass clinical transition is clearly quantified in the Australian registry, where mobile-bearing implants exhibit a severely higher 7-year cumulative revision rate compared to their fixed-bearing counterparts (13,3% versus 4,5%) [78]. The Norwegian registry further corroborates this transversal shift, reporting that recent long-term analyses suggest that fixed-bearing designs yield superior clinical outcomes and longevity.

Consequently, early-generation mobile-bearing prostheses have been universally abandoned, although the catalysts for their removal varied regionally. In Scandinavia, market correction was characterized by early clinical caution. The *S.T.A.R.* implant was completely abandoned in Sweden in 2007 and in Norway in 2013 following unacceptable failure rates and reports of high plastic breakage [81, 82]. On the other hand, in Australia and the UK, market shifts were heavily influenced by the active registry identification of underperforming systems as "outliers". The Australian registry explicitly shaped its domestic market by flagging the *S.T.A.R.*, *Mobility*, and the *Hintegra* (Hintermann Series H3) implants for having higher than anticipated revision rates. This rigorous monitoring has led to the quiet extinction of other once-prominent early designs. For instance, the *BOX* implant, despite accumulating over 800 historical procedures in the UK, has recorded exactly zero implantations since 2021, mirroring its complete disappearance from the Australian market [78, 79].

As legacy implants vanished, distinct regional profiles emerged to fill the void. In the Anglo-sphere (the UK and Australia), the market has been overwhelmingly captured by modern systems like the *Infinity*, *INBONE*, and *Zimmer TM* [78, 79]. However, not all market evolutions are driven by clinical evidence. The recent landscape in New Zealand exemplifies supplier-driven volatility. Until 2022, the *Salto*, *Salto Talaris*, and classic *Infinity* collectively accounted for over 75% of all new registrations. An abrupt corporate withdrawal of these models in late 2022 forced an overnight national transition, allowing the newly introduced *Infinity with Adaptis* and the *Vantage* systems to capture the vast majority of the market by 2024 [80]. This supplier-driven obsolescence of the *Salto* family was a global corporate maneuver, with both the UK and Australian registries also reporting a complete drop to zero procedures for the *Salto* and *Salto Talaris* by 2024 [78, 79].

Finally, regarding surgical volume, Scandinavia presents a fascinating dichotomy. Due to historically poor long-term outcomes (10-year survivorship of only 69%), Norway adopted a policy of extreme surgical centralization. The registry explicitly restricts the challenging decision between ankle arthroplasty and arthrodesis to highly specialized centers, resulting in a severe reduction in primary volume, with only 15 procedures performed nationally across four hospitals in 2023 [82]. In stark contrast, Sweden experienced a record high in surgical volume in 2024 [81]. However, despite this massive disparity in national volume, both nations have completely converged on their implant selection, crystallizing their markets into an identical strict duopoly. In both Norway and Sweden, current surgical practice is 100% saturated by the exact same two modern fixed-bearing systems: the *Infinity* and the *Zimmer TM* [81, 82].

Table 4.2: Top implanted primary Total Ankle Arthroplasty (TAR) systems by country in the most recent registry reporting year (2024). Data reflects the absolute number of primary procedures (shown in parentheses). Extracted from national registries [78–82].

United Kingdom	Australia	New Zealand	Sweden	Norway ^a
Infinity (891)	Infinity (458)	Infinity (178)	Zimmer TM (91)	Infinity (13)
Infinity/INBONE Hybrid (109)	INBONE II (122)	Vantage ^b (63)	Infinity (45)	Zimmer TM (2)
Hintegra (104)	Zimmer TM (79)	Zimmer TM (37)	-	-
Vantage (85)	Vantage (55)	INBONE II (13)	-	-
S.T.A.R. (77)	Hintermann H3 (15)	Salto Talaris (3)	-	-

^aThe most recent data available for the Norwegian Arthroplasty Register (NAR) reflects procedures performed in the 2023 calendar year. The overall volume remains severely restricted due to national centralization policies.

^bThe *Vantage* system in New Zealand includes both Mobile-Bearing (43) and Fixed-Bearing (20) variants.

Note: Empty cells (-) indicate that no further significant implant volumes were reported for that specific rank in the respective country's registry.

Beyond these regional differences, the sheer volume of contemporary applied implants reveals a modern surgical trend toward modularity and anatomical adaptability. Data from the UK National Joint Registry explicitly identifies the growing practice of "mix and match" componentry. Because the *Infinity* and *INBONE* systems were engineered by the same manufacturer to be interchangeable with a matched articulating surface, surgeons are actively combining them to suit patient-specific bone anatomy. This hybrid construct, typically an *Infinity* tibial component matched with an *INBONE* talar component, has rapidly emerged as the second most utilized system in the UK, accounting for over 100 primary procedures in 2024 [79]. This demonstrates a clinical shift toward modular hybrid constructs, allowing surgeons to customize the implant's biomechanics without diverging from a trusted manufacturer ecosystem.

While the *Infinity* ecosystem commands a massive global market share, registries also highlight the steady emergence of alternative designs exploring different biomechanical paradigms. The *Zimmer Trabecular Metal* total ankle system, for instance, introduces a radical departure from the traditional anterior surgical approach. As explicitly noted in the New Zealand registry, the *Zimmer TM* is implanted via a lateral transfibular approach, which requires a fibular osteotomy. This unique biomechanical philosophy justifies its absolute dominance in the Swedish market [81].

Furthermore, the *Vantage* system represents a recent anatomical design approach that has rapidly gained clinical traction. Within just a year of its introduction to the New Zealand market, it captured an impressive 22% of the primary volume, while simultaneously maintaining a robust and growing presence in both the UK and Australian registries [78–80]. The continued clinical adoption of these distinct design philosophies underscores that, despite current market consolidation, the optimal biomechanical solution for TAR remains a dynamic field of active engineering

and surgical debate.

The main regional trends and implant discontinuation patterns observed across the major national registries are summarized in Table 4.3.

Table 4.3: Synthesis of regional market drivers and implant discontinuations across major national TAR registries, complementing the implant volumes reported in Table 4.2.

Registry	Leading implant(s)	Primary regional driver	Notable shifts / discontinuations
UK (NJR) [79]	Infinity; Infinity/INBONE hybrid	Shift to fixed-bearing designs (~80% of primaries)	BOX: zero implantations since 2021; Salto family to zero by 2024; rise of “mix-and-match” Infinity/INBONE hybrid
Australia (AOANJRR) [78]	Infinity; INBONE II; Zimmer TM	Fixed-bearing surge (~90%); active flagging of underperforming systems	S.T.A.R., Mobility, Hintegra (H3) flagged as outliers; BOX and Salto family extinct by 2024; 7-year revision halved (12,9%–6,4%) post-2015
New Zealand (NZJR) [80]	Infinity (Adaptis); Vantage	Supplier-driven volatility, not clinical evidence	Salto, Salto Talaris, classic Infinity (>75% share) withdrawn by manufacturer in late 2022, forcing overnight transition
Sweden (SwedAnkle) [81]	Zimmer TM; Infinity (duopoly)	Early clinical caution; modern designs >84% survivorship at 10y vs. ~69% legacy	S.T.A.R. abandoned 2007 (failure rate, plastic breakage); record surgical volume in 2024
Norway (NAR) [82]	Infinity; Zimmer TM (duopoly)	Clinical caution plus extreme surgical centralization policy	S.T.A.R. abandoned 2013; only 15 procedures nationally in 2023 despite same implant selection as Sweden

4.4 Contemporary Survivorship Rates by Implant Design

The Swedish Ankle Registry provides an excellent conceptual framework for evaluating implant performance by explicitly grouping prostheses into distinct clinical generations. Based on this generational divide, global registry data demonstrates that new-generation implants achieve an 84% survivorship rate at 10 years, significantly outperforming the older generation of implants (such as the *AES* and *S.T.A.R.*), which drop to a 69% survivorship over the same period [81]. The statistical reality of these failure rates directly justifies the global market shift toward modern fixed-bearing devices.

While these legacy prostheses may exhibit seemingly acceptable early results, they degrade rapidly under long-term cyclic loading, leading to catastrophic failure rates. Implants such as the *S.T.A.R.*, *Mobility*, *Hintegra* (Hintermann Series H3), *Buechel-Pappas*, and *Agility* consistently present unacceptably high cumulative revision rates across multiple national databases [78–80]. Notably, the highly unconstrained *BOX* ankle exhibits some of the worst performance metrics

across both hemispheres [79, 80]. The statistical reality of these high failure rates directly justifies why these early-generation implants have been almost entirely abandoned in contemporary surgical practice.

In stark contrast, the newest generation of implants, which currently dominates the global market, exhibits exceptional survivorship. Modern systems like the *Infinity*, *Zimmer TM*, and *Vantage* report highly satisfactory and stable clinical profiles, drastically reducing the revision burden [79–81].

Beyond mere survivorship, the clinical data from these modern implants reveal highly interesting contemporary biomechanical trends. For instance, the *Vantage* system achieves remarkable early stability despite offering both fixed and mobile-bearing configurations, challenging the absolute dominance of rigid designs [80]. Simultaneously, the *Zimmer TM* proves that alternative surgical techniques, specifically its lateral transfibular approach combined with a porous trabecular metal structure, can yield exceptional osseointegration and longevity [80, 81]. Finally, the widespread success of the *Infinity* ecosystem has introduced the concept of modularity to the market. Registries like the UK NJR are actively tracking the successful clinical application of "mix and match" hybrid constructs (e.g., combining an *Infinity* tibial tray with an *INBONE* talar dome) allowing surgeons to perfectly adapt the device to patient-specific anatomy without compromising mechanical survival.

Table 4.4: Comprehensive comparison of implant-specific cumulative revision rates (%) at midterm (5 years) and long-term (10 years) milestones across the United Kingdom (NJR) and Australian (AOANJRR) registries [78, 79].

Implant Brand	United Kingdom (NJR)		Australia (AOANJRR)	
	5-Year Revision (%)	10-Year Revision (%)	5-Year Revision (%)	10-Year Revision (%)
Vantage	0,41%	-	-	-
INBONE	4,37%	-	2,80%	-
Trabecular Metal (Zimmer TM)	-	-	2,80%	-
Infinity (Standard)	2,76%	5,26%	3,70%	-
Infinity Tibial / INBONE Talar	2,32%	-	-	-
Salto Talaris	-	-	4,60%	-
Salto	4,92%	8,74%	9,00%	15,50%
S.T.A.R.	4,87%	10,22%	14,70%	29,90%
Mobility	8,48%	12,16%	10,20%	14,70%
Hintegra / Hintermann Series H3	5,03%	9,52%	11,40%	21,00%
Zenith	7,33%	11,30%	14,50%	16,30%
Buechel-Pappas	-	-	9,60%	16,60%
BOX	8,53%	16,81%	10,10%	14,70%

Note: Missing values (-) indicate that the specific implant has not yet accumulated sufficient clinical follow-up time in that specific registry to provide a statistically reliable estimate for that time milestone. Higher percentages indicate a greater rate of mechanical failure.

4.5 Primary Causes of Failure: The Biomechanical Problem

While survival curves quantify the overall lifespan of TAR, identifying the exact mechanisms of failure is the most critical step toward design optimization. A transversal analysis of global registries establishes that TAR failure is rarely driven by biological rejection or deep infection alone. Instead, the predominant cause of revision across all databases is a purely biomechanical phenomenon: aseptic loosening.

This mechanical failure is frequently preceded by "unexplained pain", which stands out in the registries of New Zealand and Norway as a leading clinical indication for revision, representing a critical early symptom of excessive micromotion and poor implant-bone fixation [80, 82, 85].

The direct relationship between aseptic loosening and poor initial osseointegration is clearly demonstrated by historical data from the Norwegian Arthroplasty Register (NAR). Their data proved that altering the implant's surface engineering has a massive impact on survivorship: when the *S.T.A.R.* implant transitioned from a single hydroxyapatite coating to a double titanium-calcium-phosphate coating to improve bone ingrowth, the specific failure rate due to aseptic loosening plummeted drastically from 14% to just 1% [82].

When analyzing which specific bone interface fails the most, data from both the United Kingdom and New Zealand unanimously point to the talar component as the weakest link in the prosthetic system. The UK National Joint Registry (NJR) quantifies this biomechanical vulnerability, reporting that isolated loosening of the talar component occurs at a significantly higher clinical rate (0,14 revisions per 100 prosthesis-years) than the tibial component (0,11). This disparity is also evident in osteolysis (bone resorption), where the UK reports a higher incidence rate in the talus (0,09) compared to the tibia (0,06). The New Zealand Joint Registry (NZJR) perfectly corroborates this pattern, recording 61 isolated talar loosening versus only 45 tibial loosening. These consistent registry findings highlight a distinct biomechanical vulnerability on the talar side of the joint, suggesting that its smaller bone volume struggles disproportionately to distribute multidirectional loads compared to the tibia.

Furthermore, looking at the overall volume of mechanical collapse, registries highlight the severe fragility of the bearing surface under cyclic loading. The Australian registry (AOANJRR) reports that while aseptic loosening accounts for 33,0% of all first revisions, material fatigue is a highly critical issue, specifically recording 38 cases of catastrophic polyethylene insert breakage (accounting for 9,0% of revisions) [78]. The Swedish Ankle Registry (SwedAnkle) provides the historical justification for these massive failure rates, explicitly noting that early flat-cut or unconstrained legacy designs failed to provide adequate long-term stress distribution, leading to isolated plastic wear and eventual mechanical failure [81].

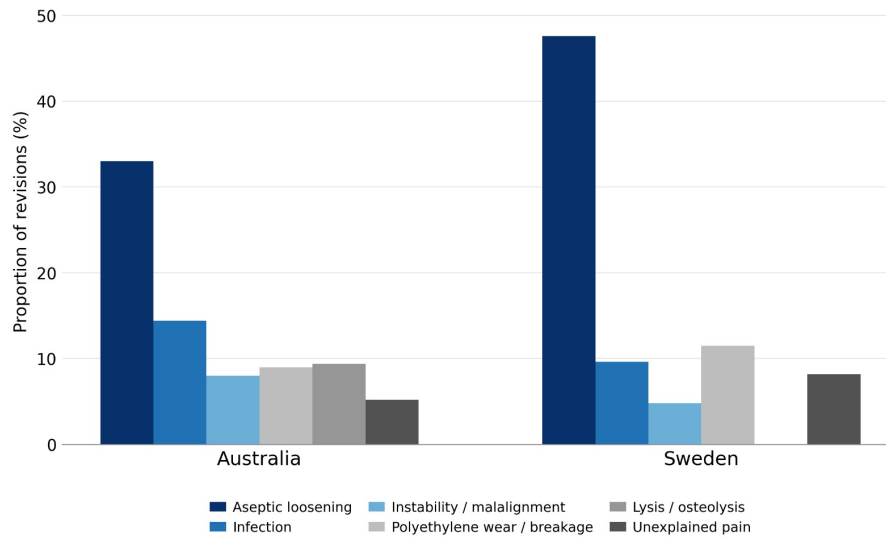


Figure 4.1: Comparative proportion (%) of primary indications for revision between the Australian (AOANJRR) [78] and Swedish (SwedAnkle) [81] joint registries, highlighting aseptic loosening as the leading failure mechanism. Percentages are calculated based on the total number of recorded revisions in each registry (Australia: $n = 424$; Sweden: $n = 355$).

It is evident that biological compatibility and surgical technique alone cannot resolve the inherent biomechanical conflict of the constrained ankle joint. To push the boundaries of implant longevity and overcome the limitations of the talar interface, future advancements must rely on rigorous mechanical optimization.

4.6 Concluding Remarks and Engineering Opportunities

This comprehensive review demonstrates that while Total Ankle Arthroplasty has undergone a significant generational evolution, the optimal biomechanical and geometrical configuration for an artificial ankle joint remains an open engineering challenge [53]. The clinical and commercial market has converged around a small cohort of fourth-generation fixed-bearing implants, such as the *Infinity*, *Zimmer TM*, *Vantage*, and *INBONE II*, due to their low-profile designs, advanced surface coatings, and cementless biological fixation [42, 55]. Nevertheless, registry data unequivocally establishes that the primary mode of failure is not biological rejection, but rather a purely mechanical phenomenon: aseptic loosening, which is heavily concentrated on the structurally constrained and poorly vascularized talar side of the joint [42, 49].

Crucially, a defining technological hallmark shared across modern designs like the *Zimmer TM*, and *Cadence* and *Exactech Vantage* is the implementation of highly conforming, laterality-specific bicondylar talar geometries that utilize asymmetry to replicate natural coupled motions. Simulating how these complex features behave under dynamic gait loads introduces severe computational friction, requiring the mathematical replication of highly non-linear tissue interfaces

and fundamentally heterogeneous, anisotropic bone structures [14]. Despite these modeling limitations, achieving high-fidelity numerical simulations is absolutely paramount to overcome the critical timeline lag inherent to national registries. Relying solely on empirical tracking requires decades of cyclic loading before defects manifest macroscopically, forcing a reactive clinical approach that accumulates severe patient morbidity and massive socio-economic healthcare burdens [14, 64].

Therefore, pushing the boundaries of implant longevity requires a definitive transition from slow clinical intuition to proactive, rigorous computational optimization. Deploying Finite Element Analysis (FEA) on contemporary configurations offers a vital opportunity to address highly specific research gaps in current literature [64, 86]. The globally dominant and most frequently implanted system, the *Infinity*, has been relatively well-studied [64]. In contrast, highly innovative alternative paradigms like the *Zimmer TM*, *Vantage*, and *Cadence* present unique mechanical features, such as arcuate resections, biomimetic condylar radii, or biased mobile liners, that remain profoundly under-represented in independent numerical literature. Furthermore, worldwide registry data indicate that the clinical adoption of these specific implants lacks broad consensus across different international markets. Evaluating these emerging design variables allows modern computational frameworks to act as active, predictive tools capable of optimizing the next generation of total ankle replacements [28, 64].

Chapter 5

Computational Modelling of the Prosthesis

For the core of this computational study, the *Zimmer Trabecular Metal™ (TM) Total Ankle System* was selected for investigation. This choice is justified by its unique position in contemporary orthopedic practice, representing an interesting departure from traditional paradigms.

As discussed in Chapter 4, the clinical adoption of this specific implant varies significantly across international markets. It has gained substantial integration in several countries, predominantly highlighted by its market presence in northern European databases, due to its specialized lateral transfibular surgical entry, yet it remains under-studied or less frequently utilized in others.

Despite its growing clinical adoption and its distinctive design philosophy, the Zimmer TM remains comparatively under-represented in independent computational literature. The only finite element study addressing this specific system to date focused exclusively on the isolated bearing component under dynamic loading, without evaluating the full bone-implant construct [12]. This gap is particularly relevant given the implant's growing adoption and its four defining features: anatomically curved arcuate resections, transverse fixation rails, biomimetic porous tantalum fixation surfaces, and a lateral transfibular surgical approach, each of which introduces geometric and material conditions that remain largely uncharacterised in the computational literature. Deploying a predictive finite element framework allows for an immediate assessment of these structural anchoring features, offering a vital methodology to investigate stress concentrations and potential mechanical vulnerabilities upfront [12, 13].

Consequently, this study aims to verify whether these radical geometric alterations successfully mitigate localized stress concentrations or if they introduce new biomechanical vulnerabilities at the bone-implant interface. Ultimately, this numerical evaluation seeks to determine if the Zimmer TM design principles represent a viable technological path toward eliminating the traditional failure mechanisms that historically lead to early revision surgeries, such as stress shielding and aseptic loosening.

Following this rationale, the present chapter details the methodology applied to generate the finite element model of the Zimmer TM, structured as follows:

- **Section 5.1** presents the design principles of the Zimmer TM system, covering its prosthetic characteristics and the lateral transfibular surgical technique that defines its geometric and mechanical boundary conditions.
- **Section 5.2** details the image-based reverse-engineering methodology adopted to reconstruct the three-dimensional CAD model of the prosthetic components, the structural simplifications implemented, and the virtual surgical implantation within the anatomical bone model.
- **Section 5.3** describes the kinematic configuration applied to simulate the three functional positions of the gait cycle: plantarflexion, neutral, and dorsiflexion.

5.1 Design Principles

5.1.1 Prosthesis Characteristics

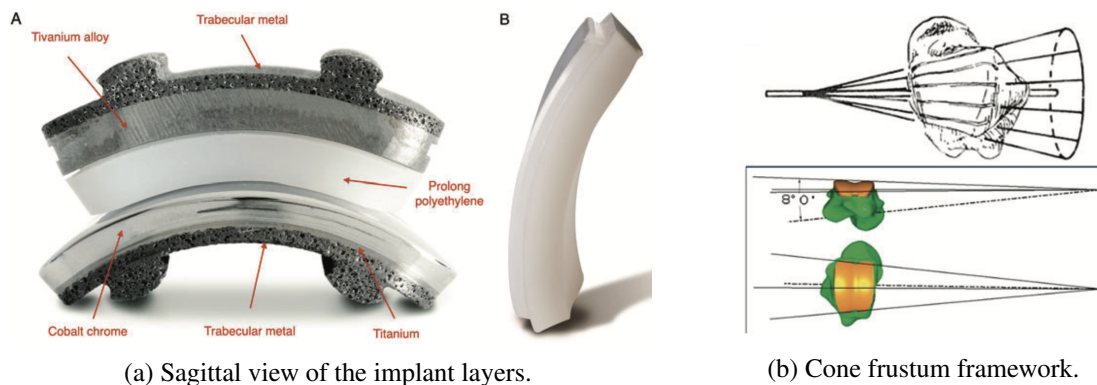


Figure 5.1: Design features of the Zimmer TM system: (a) sagittal profile showing the assembly (A) and the isolated polyethylene insert (B) with their respective materials, and (b) geometric model based on a cone frustum. Adapted from [51].

The Zimmer Trabecular Metal (TM) Total Ankle is a semi-constrained, fixed-bearing ankle arthroplasty system [12, 51]. Unlike traditional devices that rely on flat osseous resections and an anterior surgical approach, this implant was specifically designed to be implanted through a lateral transfibular approach, preserving the natural arched contour of the tibiotalar joint [51, 87].

From a biomaterials perspective, the prosthesis comprises three main components. The tibial baseplate is manufactured from Tivanium (a Ti-6Al-4V titanium alloy) diffusion-bonded to Trabecular Metal. The talar component features an articular surface made of Zimaloy (a cobalt-chromium-molybdenum alloy), while its distal bone-contacting surface is also composed of titanium and Trabecular Metal. The intermediate bearing is a Prolong Highly Crosslinked Polyethylene (HXLPE) insert. To accommodate varying ligamentous tension, this insert is available in different modular thicknesses (standard +0 mm, +2 mm, and +4 mm) [12, 51].

Geometrically, the talar component is asymmetrical and replicates the frustum of a cone (Figure 5.1b), closely mirroring native ankle anatomy. The medial side of the prosthesis presents a smaller radius of curvature compared to the lateral side [12, 51].



Figure 5.2: Zimmer Trabecular Metal Total Ankle system: (a) assembled view illustrating the articulating motion between components, [87], (b) tibial baseplate (inferior view) showing the articular surface, and (c) talar component (superior view) highlighting the asymmetric bicondylar dome geometry. Adapted from [88].

To further enhance fixation and mechanical stability, both the tibial and talar components feature fixation rails on their bone-contacting surfaces [12, 51, 89]. These rails are oriented perpendicularly to the ankle joint's axis of motion, hoping to counteract coronal shear forces [51]. A notable geometric feature of this fixation mechanism is that the rails are shorter than the total medial-lateral width of the component. By design, the prepared rail holes are blind-ended and do not extend fully to the medial edge of the resected bone [89]. This intentional design prevents the drill from compromising the medial malleolus and medial neurovascular structures during bone preparation, while also acting as a physical stop to prevent medial implant overhang during insertion [51, 89].

The prosthesis is available in six proportional sizes, and the tibial and talar components are not interchangeable, requiring the selection of a matching pair to ensure articular congruency and avoid edge-loading [51, 89].

5.1.2 Surgical Technique: The Lateral Transfibular Approach

As mentioned before, the Zimmer TM employs a lateral transfibular approach, which requires specific surgical steps that directly influence the geometric and mechanical boundary conditions of the reconstructed joint [51]. Unlike traditional anterior approaches, this technique begins with an oblique osteotomy of the fibula, a controlled surgical bone cut performed to temporarily displace the distal fibula, to gain direct lateral access to the tibiotalar joint. After the definitive components are inserted, the fibula is anatomically reduced and rigidly fixed using a lateral neutralization plate and screws [51, 89].

To eliminate freehand errors and establish a strict global coordinate system for the resections, the patient's lower extremity is rigidly immobilized within an external Alignment Stand. The foot is secured to a footplate at a 90-degree angle with a mandatory 5° to 10° of internal rotation. This internal rotation is crucial because it vertically aligns the anterior half of the lateral talus, ensuring that the subsequent bone cuts follow the true anatomical center of rotation [51, 89]. The surgical procedure detailed throughout this section is strictly based on the manufacturer's official surgical guidelines [89] and corroborated by established clinical protocols [51].

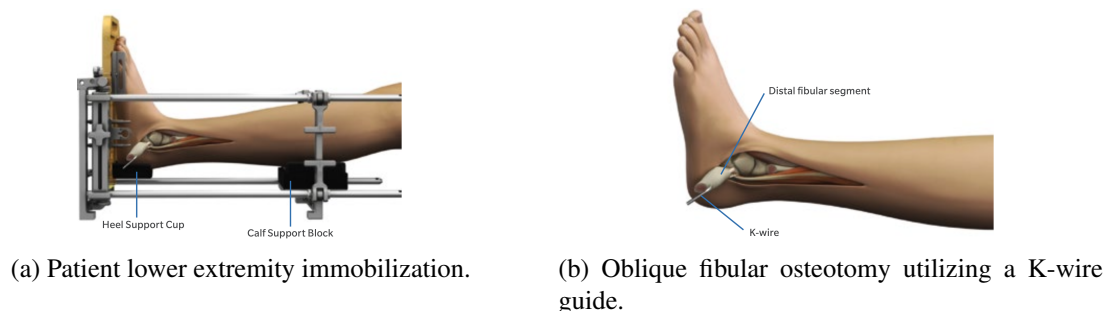


Figure 5.3: Initial surgical preparation for the lateral transfibular approach: (a) rigid immobilization within the external framework, and (b) execution of the oblique fibular osteotomy for temporary distal fibular displacement. Adapted from [89].

From a clinical and geometric perspective, the sizing of the prosthesis follows a strict protocol. The surgeon first evaluates the joint width using a Medial/Lateral (M/L) Sizer. A critical principle of this system is that the M/L measurement directly dictates the maximum allowable implant size, as it is imperative to avoid any medial or lateral metallic overhang that could impinge on the malleoli. Once the M/L size is established, the surgeon confirms the sagittal profile using an Anterior/Posterior (A/P) Sizer to ensure adequate cortical bone coverage without anterior or

posterior overhang. Furthermore, a strict design constraint of this system is that the talar and tibial component sizes are not interchangeable; the exact same size must be selected for both bones. If a patient's anatomy falls between two sizes, the smaller option is strictly chosen to reduce the likelihood of soft tissue impingement and component overhang [51, 89].

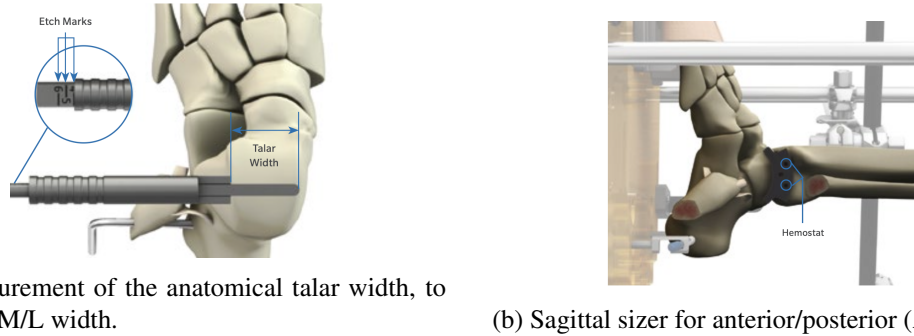


Figure 5.4: Intraoperative sizing protocol for the Zimmer TM system, explicitly comparing the: (a) medial/lateral (M/L) evaluation of the talar dome width using coronal etch marks, and (b) anterior/posterior (A/P) sagittal assessment to verify cortical bone coverage. Adapted from [89].

Once aligned, bone removal is performed using a guided burr that mills the bone in a sweeping motion rather than utilizing flat saw cuts. This milling process creates anatomically curved resections on both the tibia and the talus, matching the frustum-like geometry of the implant [51].

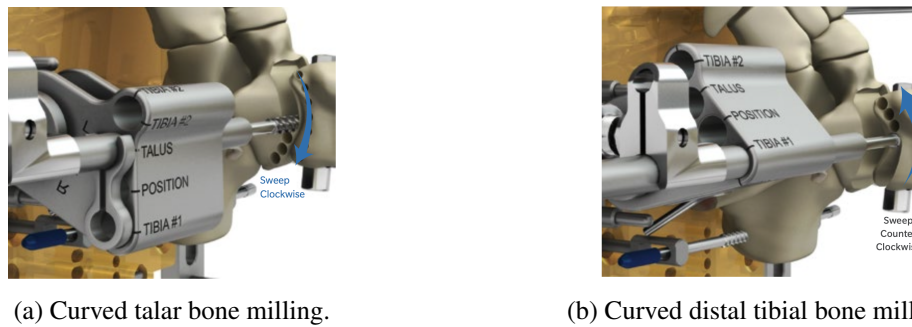
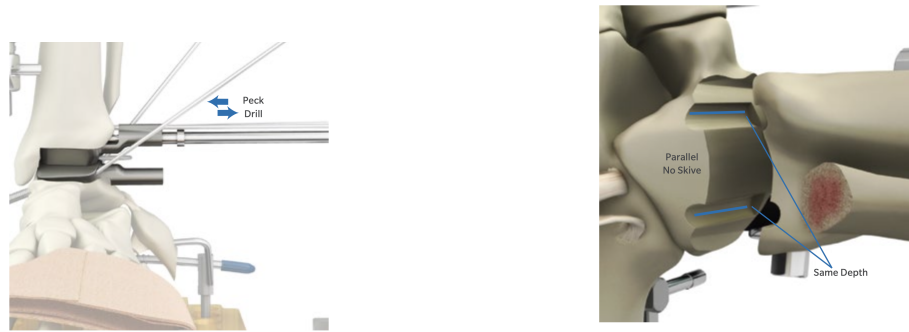


Figure 5.5: Sequential execution of the bone-sparing arcuate resections: (a) preparation of the curved talar dome, followed by (b) the matching distal tibial geometry configuration. Adapted from [89].

Following the curved bone milling, rail holes are drilled to accommodate the implant's fixation rails. The final tibial and talar components are then impacted into the joint space strictly following a lateral-to-medial vector. The blind-ended nature of the rail holes acts as a physical stop during insertion [89].



(a) Drilling process for the blind-ended cavities. (b) Final preparation of the bone anchoring tracks.

Figure 5.6: Preparation of the secondary stabilization mechanisms: (a) drilling guided blind-ended holes, and (b) final tracking configuration prior to prosthetic impaction. Adapted from [89].

If necessary, bone cement can be injected into the rail holes to augment early component fixation. Although the original surgical protocol and United States regulatory guidelines mandate the retrograde injection of polymethylmethacrylate (PMMA) cement into the four rail holes to secure the implant, contemporary clinical practice, particularly throughout Europe, frequently favors a completely cementless approach. Extensive multi-center studies dividing cohorts by continent demonstrate that relying exclusively on the press-fit biological osseointegration of the highly porous tantalum rails yields equivalent long-term survivorship, rendering the use of bone cement strictly optional and dependent on surgeon preference [67]. To conclude the procedure, the fibula is anatomically reduced and rigidly fixed using a neutralization plate and screws, and the lateral wound is closed.

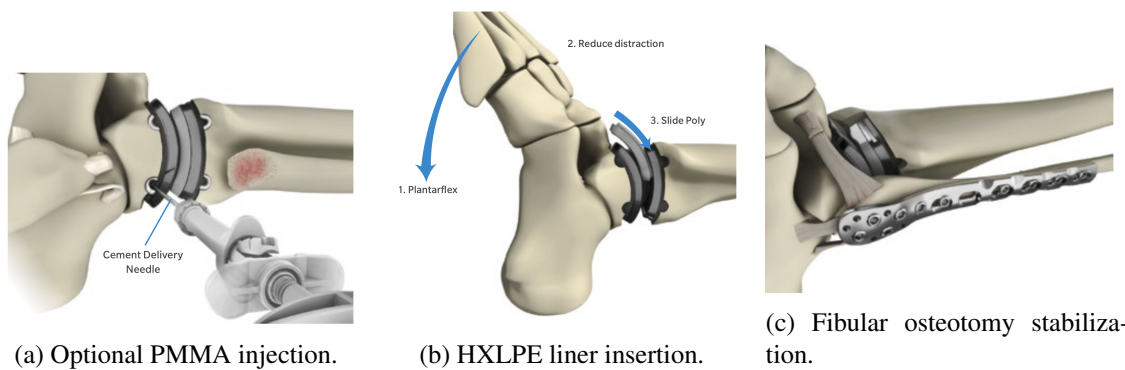
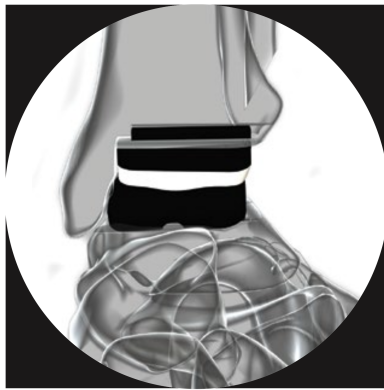
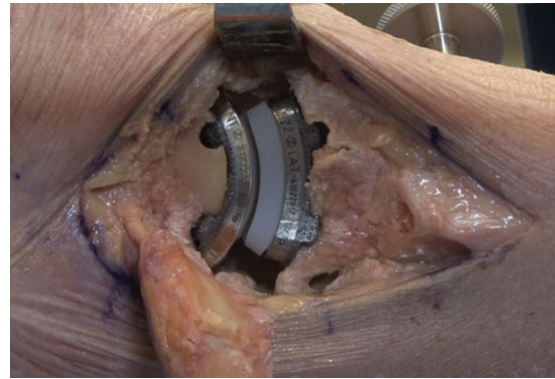


Figure 5.7: Final stages of the surgical workflow: (a) optional PMMA cement augmentation, (b) intermediate bearing placement, and (c) rigid anatomical reduction of the lateral corridor utilizing a neutralization plate. Adapted from [89].



(a) Fluoroscopic alignment verification layout.



(b) Intraoperative lateral view of the final assembly.

Figure 5.8: Final reconstructed total ankle arthroplasty assembly using the Zimmer TM system: (a) fluoroscopic verification layout utilized intraoperatively to validate proper component positioning and coronal plane alignment, adapted from [89] and (b) direct intraoperative lateral view demonstrating the surgery result, adapted from [51].

5.2 Drawing Methodology and Geometric Simplification

The development of accurate CAD models for commercial prostheses is a common and widely documented challenge in computational biomechanics. Because manufacturers rarely disclose original three-dimensional geometries due to strict intellectual property constraints, access to exact proprietary CAD files is frequently restricted to manufacturer-funded investigations. For instance, Casaroli et al. [19] explicitly report receiving the original 3D models and financial support directly from Zimmer [19]. In other specific scenarios, the manufacturer may solely provide the physical prosthesis without disclosing the virtual mathematical files, as documented by Martinelli et al. [12].

To overcome this critical restriction of physical and dimensional access to the implant, independent researchers systematically rely on reverse-engineering methodologies. As an example, Bingenheimer [34] acquired functional models utilizing 3D laser scanning on physical demonstration hardware, whereas Jain et al. [28] employed rigorous image-based geometric reconstruction. Similarly, the present study did not possess access to the proprietary mathematical files nor the physical prosthesis. Consequently, an image-based reverse-engineering approach was adopted, a method validated and frequently utilized in recent orthopaedic literature to generate functional 3D solid models [28]. Furthermore, this exact lack of direct physical access, combined with the strict requirement for numerical viability, dictated the necessary geometric simplifications implemented during the CAD reconstruction process. Nevertheless, the defining morphological hallmarks of the Zimmer TM were faithfully replicated, thereby guaranteeing the biomechanical validity of the computational assembly.

The three-dimensional modeling of the Zimmer Trabecular Metal (TM) prosthesis was developed entirely within the *SolidWorks* software environment. Orthogonal views, primarily high-resolution lateral/sagittal profiles extracted directly from the manufacturer's surgical technique manuals and radiographic templates, were imported as reference sketches into the workspace. To faithfully replicate the asymmetrical geometry of the implant, where the medial and lateral profiles differ, the outer shapes and key morphological hallmarks of the components were meticulously traced and generated via loft operations.

Furthermore, the fixation rails on the bone-contacting surfaces of both metallic components were explicitly drawn. The specific dimensions of these rails were also proportionally approximated based on the available surgical templates. Nevertheless, reflecting the real implant's safety constraints, their functional geometry was strictly preserved. This way, these rails were modeled to be shorter than the total medial-lateral width of the components. For completeness, technical drawings and the principal dimensions of the reconstructed components are provided in Appendix B.

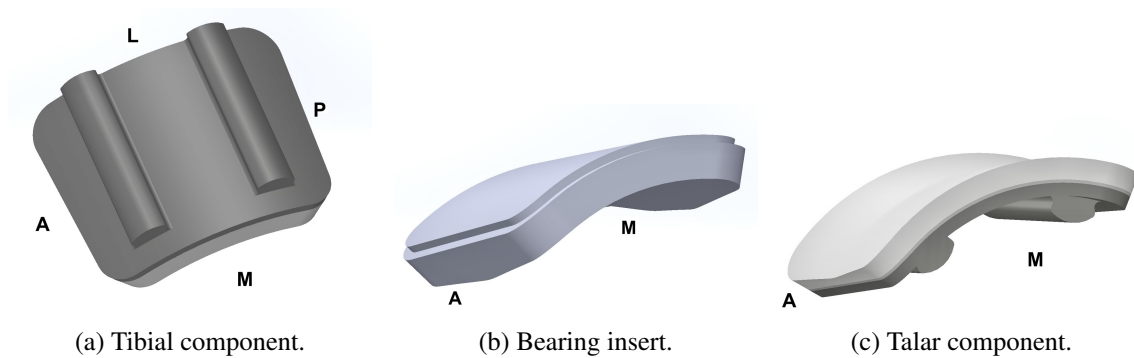


Figure 5.9: Developed CAD model of the total ankle replacement prosthesis components, showing: (a) the metallic tibial component, (b) the bearing insert, and (c) the metallic talar component.

Following the individual part reconstruction, the components were imported into the assembly environment to establish the complete prosthetic system. Concurrency and tangency mating conditions were carefully applied between the articulating curved surfaces of the polyethylene insert and the metallic talar components, replicating the mechanical behavior of the real implant.

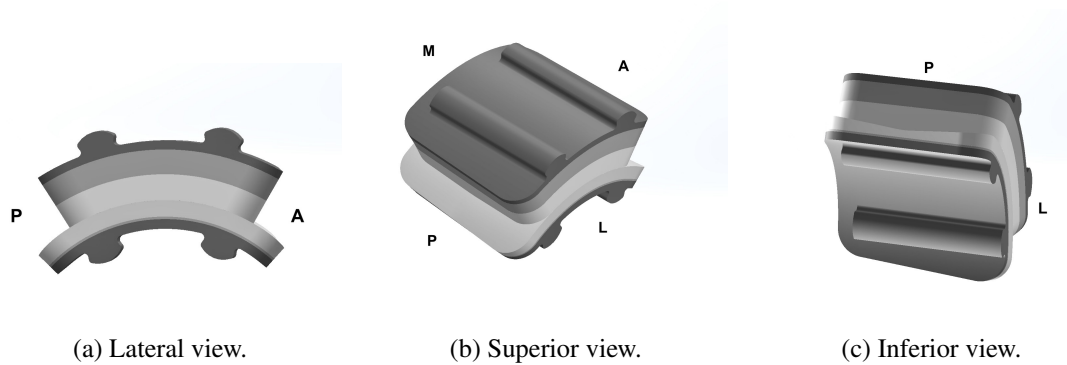


Figure 5.10: Developed CAD assembled model of the prosthesis: (a) lateral view, (b) superior view, and (c) inferior view.

A previously validated three-dimensional anatomical model of the human ankle joint, developed by Ramos et al. [24], was utilized as the anatomical bed. This intact model was rigorously reconstructed from a combination of computed tomography (CT) and magnetic resonance imaging (MRI) scans of a right-side ankle, accurately capturing the precise morphological boundaries of the cortical bone and the inner trabecular core [24].



Figure 5.11: Three-dimensional anatomical model of the human ankle joint: (a) anterior view and (b) lateral view, showing the structural relationship between the tibia, fibula, and talus.

Mirroring the actual surgical procedure, the medial-lateral (M/L) width of the native talar dome was measured within the anatomical model. This biological dimension served as the definitive scaling metric. By matching the prosthesis's talar width to the patient-specific anatomical boundary box, the initial proportional size of the Zimmer TM system was determined and successfully mapped.

Since the exact absolute dimensions of the manufacturer's tabulated implants were not publicly known, this alignment methodology and the final scaling of the prosthetic assembly were iteratively refined during the modeling phase, effectively simulating the Surgical Procedure. The prosthesis was uniformly resized to ensure an optimal fit relative to the specific anatomy of the intact 3D ankle bone model available for the study, strictly obeying the system's clinical design

constraints. Specifically, the scaling algorithm replicated the manufacturer's intraoperative protocol: the reconstructed M/L width directly dictated the maximum allowable implant scale to ensure no medial or lateral metallic overhang would impinge on the malleoli. Once this coronal constraint was satisfied, the sagittal profile was adjusted and verified against the anterior/posterior (A/P) dimensions of the bone model, guaranteeing adequate cortical bone coverage without anterior or posterior overhang. Furthermore, the strict design constraint mandating that the talar and tibial component sizes are not interchangeable was maintained throughout the virtual assembly, ensuring a perfectly congruent, non-interchangeable matching pair. If the virtual anatomy fell between two standard proportional scales, the smaller geometric option was strictly selected, ensuring a clinically realistic placement consistent with surgical reality.

Once the optimal implant size was established, the virtual bone resection (Boolean operations) was executed sequentially to accurately reflect the real operative workflow. According to the manufacturer's bone preparation protocol, the spatial position of the talar component was defined first relative to the anatomical center of rotation, and the corresponding curved talar bone resection was performed. Only after the talar geometry was fully prepared, the tibial component was aligned, and the distal tibial resection was subsequently executed.

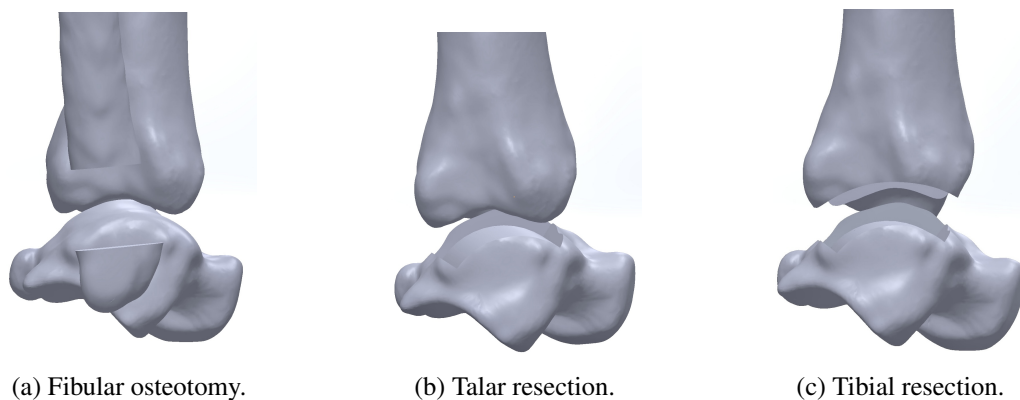


Figure 5.12: Sequential virtual surgical resections performed on the anatomical model to prepare the bone bed.

Following the completion of the curved bone resections, the preparation of the fixation rail holes was simulated. In the actual surgical protocol, specialized drills create cylindrical or semi-cylindrical channels to accommodate the implant's anchoring rails. However, to simulate a long-term clinical scenario with complete osseointegration, these cavities were modeled as perfectly congruent geometric slots matching the exact profile of the prosthetic rails. This geometric strategy ensures that, during the Finite Element Method preprocessing, the surrounding volume is discretized as continuous bone tissue. These virtual cavities were explicitly modeled as blind-ended holes, terminating just before the medial cortex, as dictated by the procedure.

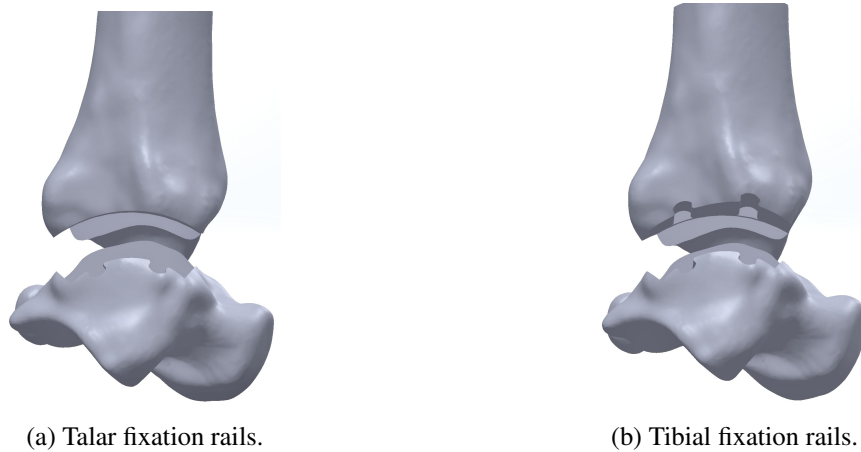


Figure 5.13: Final preparation of the bone bed showing the generated fixation rails: (a) proximal talar slots and (b) distal tibial slots.

Following the bone bed preparation, the engineered prosthetic components were systematically positioned onto the reconstructed anatomical surfaces. First, the metallic talar component was fitted into the proximal talar slots, ensuring structural alignment along the generated rails (Figure 5.14a). Subsequently, the remaining components, the metallic tibial component and the HXLPE bearing insert, were assembled to establish the complete total ankle replacement system within the joint geometry (Figure 5.14b).

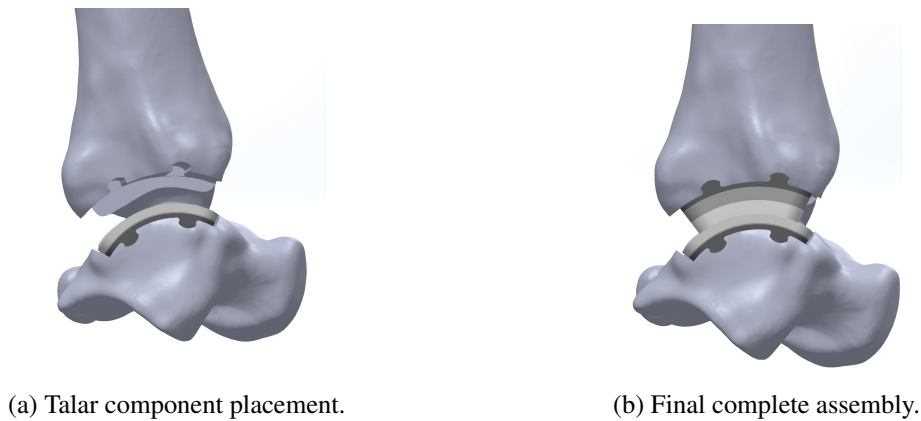


Figure 5.14: Final stages of the virtual implantation procedure: (a) insertion of the talar component into the prepared bone bed and (b) final assembled computational model of the total ankle arthroplasty.

5.3 Kinematic Configuration

To evaluate the biomechanical behavior of the reconstructed total ankle arthroplasty assembly under physiological boundaries, the range of motion was implemented according to typical gait extremes. To simulate both dorsiflexion and plantarflexion, the tibial bone component and its respective implant parts were angularly displaced relative to the talar component, establishing

a maximum dorsiflexion of 15° and a maximum plantarflexion of 10° . Because the assembly constraints (mates) were already active in the software, this movement allowed the components to naturally slide and glide against each other around the pre-defined talar center of rotation, perfectly replicating real anatomical motion.

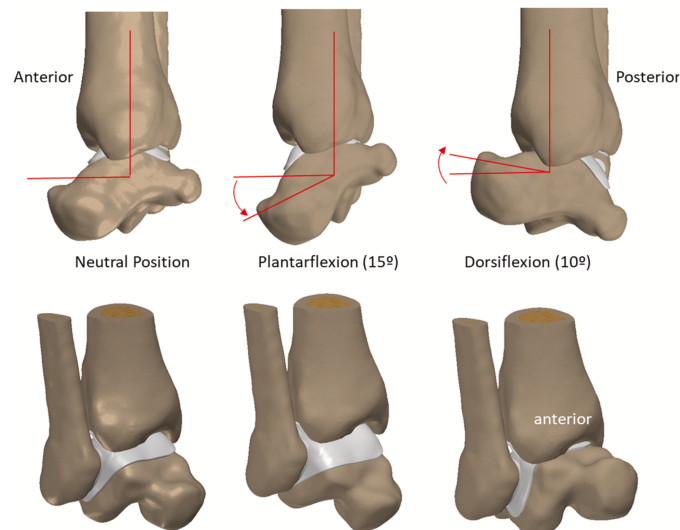


Figure 5.15: Kinematic configurations of the intact ankle joint model [24]: the top row shows the neutral position, maximum plantarflexion (15°), and maximum dorsiflexion (10°). The bottom row displays the corresponding anterior views. Adapted from [24].

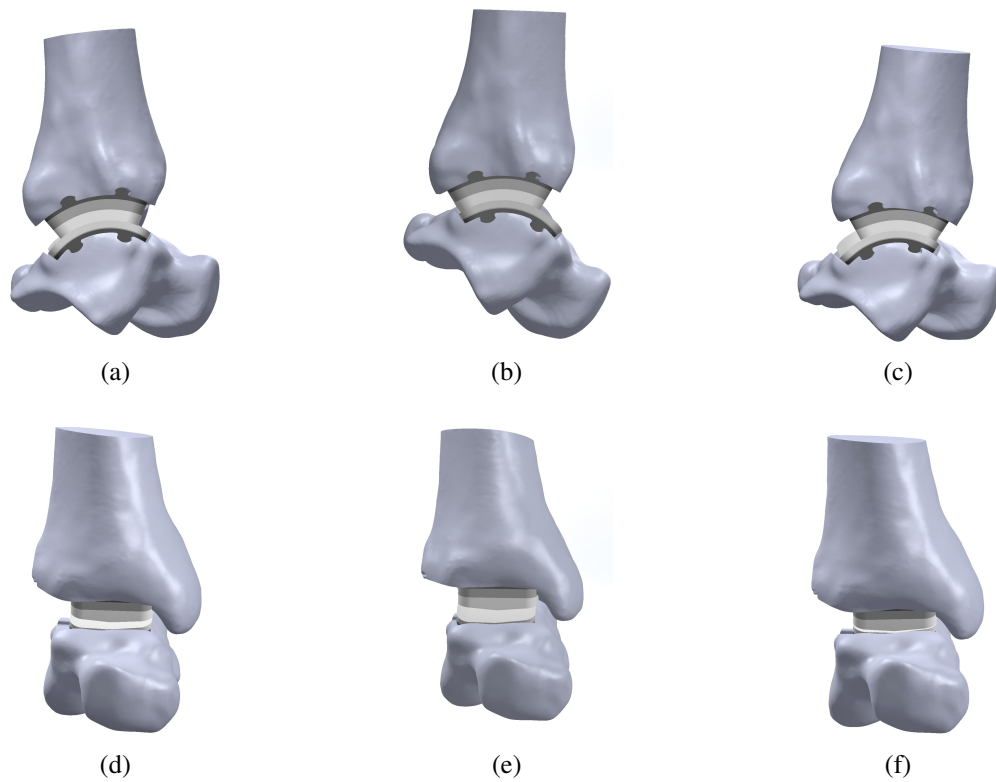


Figure 5.16: Kinematic configurations of this study. The top row illustrates the lateral perspective, while the bottom row displays their orientations from an anterior perspective. Where (a) and (d) correspond to neutral position, (b) and (e) to plantarflexion, and (c) and (f) to dorsiflexion.

Chapter 6

Finite Element Analysis

To evaluate the structural behavior and primary stability of the Zimmer Trabecular Metal Total Ankle prosthesis, a three-dimensional Finite Element (FE) model was developed. The computational framework is formulated as a non-linear static analysis within the *SolidWorks Simulation* environment.

This chapter details the specific methodological steps undertaken to construct, assemble, and analyze the three-dimensional Finite Element model of the implanted ankle joint. The numerical workflow consists of three interconnected stages summarized below:

- **Section 6.2** defines the material properties assigned to each component of the construct: cortical and trabecular bone, metallic alloys, and for the bearing insert.
- **Section 6.3** describes the boundary conditions and loading protocol, encompassing the axial compressive forces applied at each of the three simulated positions, the ligamentous constraint strategy implemented through tension-only spring connectors, and the contact interactions defined at the bone-implant and implant-implant interfaces.
- **Section 6.4** presents the mesh generation strategy adopted for the bone structures and prosthetic components.

To illustrate the development process, the pipeline from medical imaging data to structural simulation is schematically mapped in Figure 6.1. This diagram outlines the methodological sequence required to translate complex anatomical and prosthetic geometries into the finite element environment.

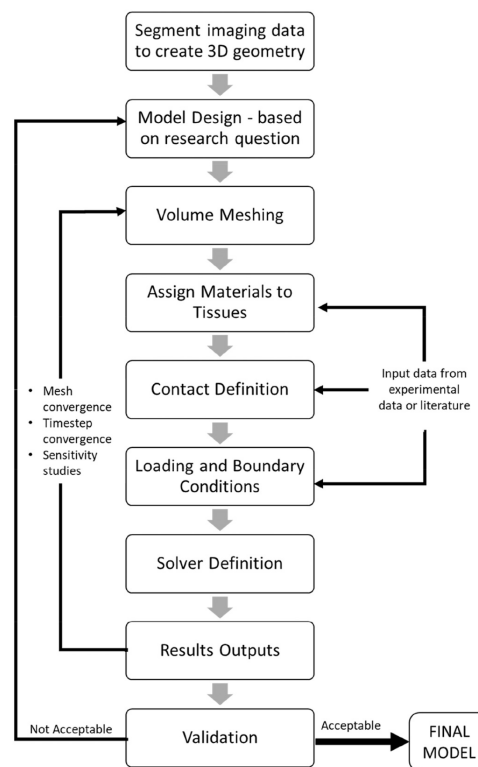


Figure 6.1: Schematic representation of a typical finite element analysis pipeline in orthopaedic biomechanics, illustrating the sequential progression from three-dimensional geometry definition through material assignment, contact and boundary condition setup, solver execution, and output validation. Adapted from [25].

6.1 The Finite Element Method in Biomechanics

The Finite Element Method (FEM) is a highly sophisticated numerical technique used to approximate the solutions of complex engineering problems governed by differential equations, which are otherwise mathematically intractable to solve analytically. In the field of orthopedic biomechanics, FEM provides a powerful computational framework to evaluate the internal mechanical behavior of biological structures and prosthetic implants under physiological loading environments [14].

The fundamental principle of the method relies on discretization, where a complex, continuous three-dimensional geometric domain (such as a reconstructed ankle joint and its corresponding implant) is subdivided into a finite number of smaller, simpler geometric subdomains called elements. These elements are interconnected at specific spatial coordinate points designated as nodes [59]. By discretizing the continuous structure, the governing partial differential equations that describe the overall physical behavior of the system (such as stress, strain, and displacement) are converted into a vast, simultaneous system of algebraic equations. Within the general framework of FEM, a comprehensive library of element typologies is available to researchers, ranging from one-dimensional (1D) trusses and springs, to two-dimensional (2D) shells, and three-dimensional

(3D) hexahedrons and tetrahedrons, each specifically formulated to address distinct geometric and physical requirements across diverse engineering fields [14].

However, in the specific context of this biomechanical model, the discretization strategy is strictly dictated by the complex anatomical morphology. To accurately capture the highly irregular boundaries of the 3D solid domains such as bones, cartilages, and metallic implants, tetrahedral elements (e.g., C3D4 or C3D10) are explicitly implemented, avoiding the severe mesh distortion frequently associated with regular hexahedral formulations [14].

For each individual element, the mechanical behavior is mathematically formulated based on the material's constitutive laws, defining properties such as elasticity, anisotropy, or viscoelasticity, and the element's geometric configuration. The individual equations are then assembled into a global stiffness matrix ($[K]$) that represents the structural integrity of the entire system, governed by the fundamental equilibrium equation:

$$\{F\} = [K]\{u\} \quad (6.1)$$

where $\{F\}$ represents the global vector of externally applied physiological forces and boundary conditions, and $\{u\}$ is the unknown vector of nodal displacements to be calculated. Once the boundary conditions, such as muscle forces, joint reaction loads, and physical fixations, are numerically applied, the system of equations is computationally solved to determine the precise displacements at every node. From these primary nodal displacements, secondary mechanical variables such as strains and internal stress distributions (e.g., von Mises stresses) are derived across the entire domain [59]. Consequently, FEM allows researchers to preemptively map localized contact pressures, predict potential material failure zones, and evaluate bone-implant interface micromotions with high spatial resolution, serving as an invaluable predictive tool in modern implant optimization [14].

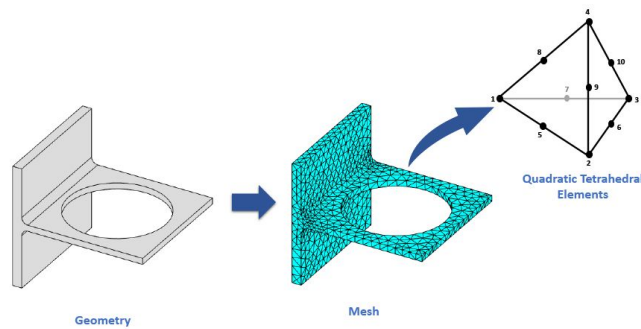


Figure 6.2: Conceptual workflow of the finite element discretization process, illustrating the conversion of a continuous CAD geometry into a discretized mesh composed of second-order quadratic tetrahedral elements. Adapted from [90].

6.2 Material Properties

6.2.1 Biological tissues

The biological tissues comprising the human ankle joint are intrinsically complex. In a true physiological state, bone tissue is heterogeneous, anisotropic (meaning its mechanical properties, such as stiffness, vary depending on the direction of the applied load, and viscoelastic (meaning its deformation is time-dependent and sensitive to the speed of the load). Attempting to digitally replicate this exact microarchitecture demands extreme and often prohibitive computational resources. When modeling bone, various methodological options exist in the literature to overcome this limitation. Some studies simplify the anatomy by assigning a single, averaged elastic modulus (e.g., 7,3 GPa) to the entire bone structure, treating it as a single solid block [14, 25]. By contrast, other authors utilize highly patient-specific models where the elastic modulus is calculated element-by-element based on computed tomography (CT) greyscale density using empirical formulas [9, 14, 25, 47]. While the latter captures the continuous heterogeneity of the tissue, it carries a significant computational burden [25].

To balance computational efficiency and anatomical realism without compromising clinical relevance, a widely accepted alternative in orthopaedic biomechanics is to explicitly model the cortical shell and the cancellous core as separate macroscopic regions. As demonstrated by extensive systematic reviews compiling dozens of previous computational models, idealizing these two distinct osseous zones as linearly elastic, homogeneous, and isotropic domains represents a highly validated and universally adopted simplification [14, 25]. This approach accounts for the drastic differences in mechanical strength between the dense cortical bone and the porous cancellous bone, allowing computational power to be efficiently allocated to evaluate the complex load transfer mechanisms and contact stresses at the resected bone-implant [12, 25].

When adopting this bi-material separation, there is less agreement in the literature on the exact properties to be used [25]. Historically, a Young's modulus of 19 GPa was established as the gold standard for cortical bone in classical biomechanical studies, and this value continues to be widely employed in numerous modern FE models [14, 25]. However, recent comprehensive reviews establish the accepted convention that the most common cortical bone elastic modulus utilized in current literature is 17 GPa, which is frequently combined with trabecular values ranging from 400 to 700 MPa [25]. Following this modern consensus, material properties were assigned to the bone structures to ensure the most appropriate load transfer simulation. The cortical bone was assigned a Young's modulus of 17 GPa, while the cancellous bone was assigned a much lower stiffness of 400 MPa. For both bone structures, a Poisson's ratio of 0,3 was maintained. Adopting this specific material combination ensures that the numerical stiffness perfectly matches the underlying morphology of the imported bone model [22, 24], while aligning precisely with the standard ranges validated in recent ankle FE literature [25].

6.2.2 Prosthesis materials

As previously established in Chapter 3 (Section 3.2.3), the Zimmer TM system features a multi-material architecture.

In the finite element analyses of total ankle replacements conducted by Martinelli et al. [12], Yu et al. [48], and Zhang et al. [9], the metallic alloys comprising the tibial and talar components were explicitly modeled as homogeneous, isotropic, and linear elastic materials. Following this established methodology, the metallic components were assumed under the exact same linear elastic conditions. This is a validated simplification, as these high-strength alloys do not undergo permanent plastic deformation under the typical physiological loads of the human gait [12, 48]. The proprietary prosthetic materials and the biological tissues were manually defined as custom materials within the software, utilizing the specific values extracted from the previously discussed biomechanical literature.

The highly porous Tantalum coating (Trabecular Metal) located at the bone-implant interface is characterized as a biomaterial with up to 80% porosity, structurally engineered to mimic cancellous bone [51, 91]. Its open-pore architecture provides an osteoconductive scaffold that promotes robust bony in-growth and long-term biological fixation [67, 91]. Furthermore, it is reported that its low elastic modulus can minimize stress-shielding effects on the underlying subchondral bone [51, 92]. In previous finite element studies evaluating this specific prosthesis, such as the model developed by Martinelli et al. [12], this porous coating was not explicitly discretized as a distinct material layer. Instead, the components were modeled solely based on their solid substrate alloys (Ti-6Al-4V and CoCrMo), assuming fully bonded bone-implant interfaces. To overcome this computational simplification and explicitly incorporate the coating's mechanical behavior, distinct properties were assigned to the Tantalum layer in the current model. The Young's modulus for this highly porous structure was strictly defined as 3 GPa. This metric was extracted from the manufacturer's official mechanical property charts, which explicitly report an elasticity range between 2,5 and 3,9 GPa for the universal Trabecular Metal biomaterial [91]. Since the manufacturer's technical literature does not report a specific Poisson's ratio, a value of 0,3 was adopted. This was considered a reliable and scientifically sound approximation, as it perfectly aligns with both the standard convention for isotropic metallic biomaterials and the intrinsic properties of the cancellous bone that the coating is designed to replicate [14, 22].

As previously noted regarding proprietary components, the exact stress-strain curves for these patented HXLPE formulations are not disclosed by manufacturers [28]. To overcome this limitation, the bearing's complex mechanical behavior under load is conventionally evaluated in finite element literature using the well-documented properties of its base polymer, virgin UHMWPE [9, 12, 28, 48]. As this polymer exhibits strict linear elastic behavior up to its yield point, the bearing was directly modeled exclusively within a linear elastic regime. To guarantee structural safety, the yield stress was fixed as detailed in the material properties table (Table 6.1), following the

validated methodology applied by Martinelli et al. [12] and corroborated by recent computational frameworks [12, 28, 48]. As demonstrated by Jain et al. [28], this specific threshold represents the lowest reported yield strength for standard UHMWPE, functioning as an absolute conservative boundary. By continually evaluating the calculated von Mises stresses against this strict limit, the linear model safely predicts initial yielding and potential material failure, entirely eliminating the need for computationally prohibitive post-yield plastic deformation protocols [28].

The specific mechanical properties inferred and assigned to both the bone structures and the distinct sub-materials are summarized in Table 6.1.

Table 6.1: Mechanical properties assigned to the bone structures and prosthetic materials in the finite element model.

Material	Young's Modulus (E), GPa	Poisson's Ratio (ν)	Reference
Cortical Bone	17	0,30	[22]
Cancellous Bone	0,4	0,30	[22]
Titanium (Ti-6Al-4V)	115	0,36	[12]
Zimaloy (CoCrMo)	210	0,30	[9, 48]
Trabecular Metal (Tantalum)	3	0,30	[14, 91]
Prolong Polyethylene (HXLPE)	0,6*	0,46	[12, 28, 48]

*Initial elastic modulus in the linear region, prior to plastic deformation.

The assignment of material properties was specifically tailored to the mathematical requirements of the finite element formulation. Since the computational model evaluates specific instances of peak loading during the stance phase of the gait cycle, the simulation was conducted as a static analysis. In this static condition, dynamic effects such as inertia and velocity are deemed negligible. Consequently, mass density and damping coefficients were excluded from the material definitions, as the structural response of the system is entirely dictated by the global stiffness matrix $[K]$.

This way, the macroscopic mechanical behavior is completely defined by two independent elastic constants: Young's modulus (E) and Poisson's ratio (ν). Other dependent variables, such as the shear modulus, are internally derived by the computational solver based on these parameters.

6.3 Boundary Conditions and Loading

6.3.1 Loading

To accurately simulate the biomechanical environment of the ankle joint, physiological boundary conditions and loading must be appropriately defined. During a normal walking gait cycle, the human ankle joint experiences severe compressive reaction forces. Biomechanical reviews and recent finite element studies have consistently demonstrated that the maximum compressive load on the tibiotalar joint can reach approximately 4,5 to 7 times the body weight (BW) [48, 86].

While some computational studies attempt to simulate these extreme peak loads by applying static forces exceeding 5 kN (e.g., 5,2 kN) [14, 25, 28], fully dynamic simulations of the gait cycle carry a massive computational burden and are often impractical for comparative implant contact mechanics.

To overcome this limitation without compromising clinical realism, a highly validated methodology was implemented. The simulation was discretized into three critical functional positions representing the stance phase of the gait cycle: dorsiflexion, neutral stance, and plantarflexion [14]. Following established and documented biomechanical protocols for finite element analysis of the ankle, specific static compressive loads were assigned to each kinematic position: 1 600 N during dorsiflexion, 600 N during neutral stance, and 400 N during plantarflexion [14].

Table 6.2: Summary of the static loading conditions and functional positions applied to the proximal tibia.

Gait Position	Angle	Axial Load (N)	Body Weight	References	Biomechanical Significance
Neutral Stance	0°	600	1,0 BW	[14, 59]	Single-leg standing baseline.
Dorsiflexion	10°	1 600	2,67 BW	[14, 59]	Peak force at terminal stance.
Plantarflexion	15°	400	0,67 BW	[14, 59]	Reduced force at toe-off phase.

These specific magnitudes directly correlate to the physiological weight-bearing fluctuations experienced during the gait cycle of a standard 60 kg patient. As systematically reviewed by Talbott et al. [25], compiling recent numerical frameworks, the 600 N load applied in the neutral stance represents exactly 100% of the average body weight (1,0 BW) utilized in computational orthopaedics, a standardized metric characteristic of single-leg standing. The 1 600 N load during dorsiflexion simulates the severe peak reaction force at terminal stance, accounting for approximately 2,67 times the body weight (267% BW). By contrast, the 400 N load applied during plantarflexion reflects a reduced weight-bearing state of approximately 67% BW, representing the toe-off phase [14]. This standardized distribution provides a rigorous baseline that allows for a direct comparison of the predicted contact stresses with previously published numerical models, completely avoiding the extreme complexities and potential instabilities of applying peak dynamic loads.

For each load case, the compressive force was applied axially to the superior cortical boundary of the proximal tibia [14, 25]. Restricting the load application exclusively to the cortical shell prevents artificial stress concentrations in the underlying trabecular core and ensures a physiological load transfer down the tibial shaft, guaranteeing the structural fidelity of the simulation [14].

Furthermore, while the fibula is anatomically present, it was deliberately excluded from direct axial loading. This simplification is widely adopted in ankle modeling because the literature demonstrates that the fibula transmits only a minimal fraction of the total axial load, reported to range from approximately 6,41% to 17% [16, 25]. Therefore, its direct structural contribution was

considered secondary to the primary objective of evaluating load transfer at the tibiotalar bone-implant interface, consistent with simplification protocols widely adopted in ankle finite element literature [14, 25].

6.3.2 Boundary conditions

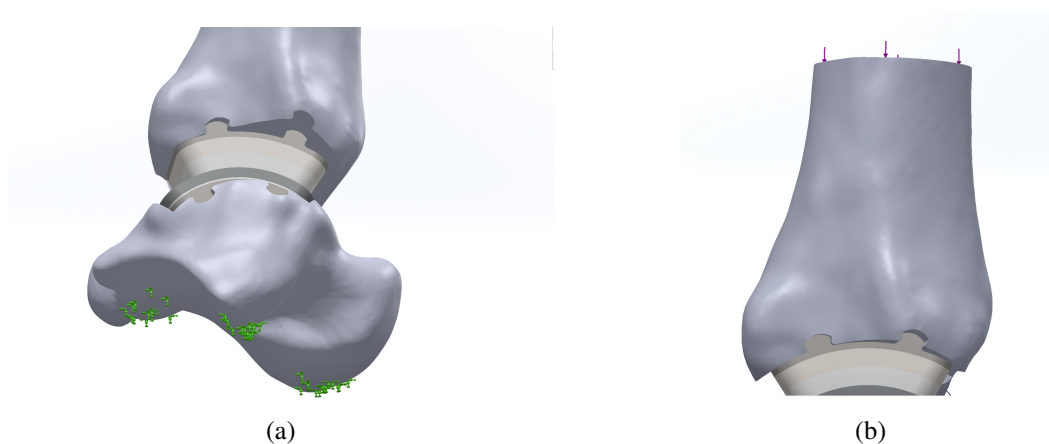


Figure 6.3: Boundary condition and loading configurations captured during the dorsiflexion gait phase, showing: (a) a detailed view of the fixed geometry constraints (green arrows) on the three planar distal surfaces of the talus, and (b) the setup including the axial compressive load (purple arrows).

Fixtures: Having established the magnitude and anatomical distribution of the compressive load, the next fundamental step is the application of appropriate boundary conditions. In a static finite element analysis, applying a load requires adequate structural constraints to prevent unphysiological rigid body motion and enable the resolution of the system's stiffness matrix. Following widely established protocols systematically documented in numerous previous numerical investigations of the ankle, the inferior aspect of the talus was rigidly fixed to provide a stable foundation [14, 47]. Rather than applying a blanket constraint, this fixation was strategically distributed across three distinct anatomical surfaces on the inferior talar base. This configuration guarantees a perfectly stable foundation for the joint while representing the missing subtalar support and isolating any potential boundary stress artifacts far from the critical articular evaluation zone [59, 93].

Connectors: To maximize clinical realism, the tibia was completely freed from any artificial rigid boundary constraints. Instead of unnaturally blocking the tibial degrees of freedom, frequently observed in conventional static frameworks, as documented in the comprehensive review by Mondal and Ghosh [14], physiological stabilization was exclusively achieved through the implementation of virtual one-dimensional (1D) spring elements. This methodology has also been successfully achieved in previous investigations through the implementation [12, 14]. Because this numerical model focuses strictly on the load transfer across the primary tibia-implant-talus

axis, secondary tarsal bones (such as the fibula and calcaneus) were omitted from the geometric domain. Consequently, native ligaments anchoring to these missing structures could not be explicitly modeled. To prevent unconstrained rigid-body motion or numerical dislocation under axial loading, these simplified elastic connections, strictly formulated to replicate the mechanical stiffness of the anterior tibiotalar (ATiTL) and posterior tibiotalar (PTiTL) ligaments, served as the anchoring mechanism for the tibia [12, 14]. This approach ensures that the tibia is guided naturally by the applied load and the prosthetic congruency, rather than being artificially "locked" in space, thereby generating any realistic contact stress evaluation.

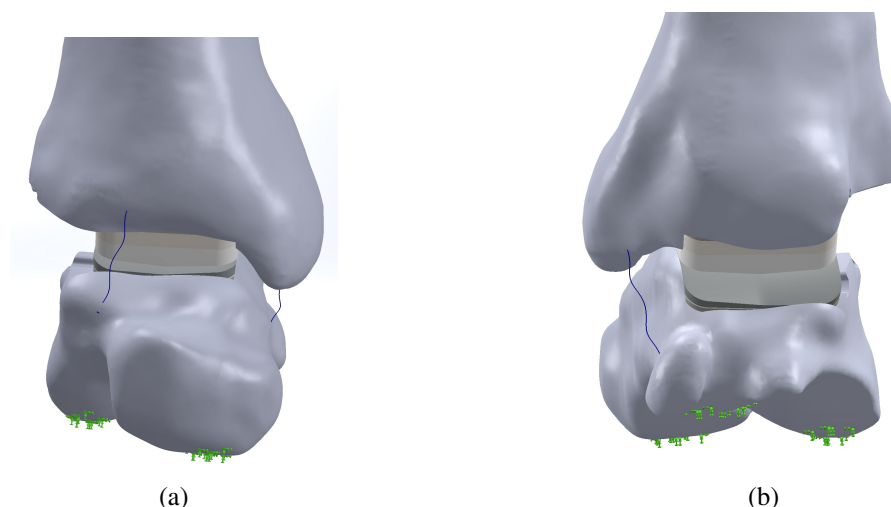


Figure 6.4: Detailed view of the ligamentous simulation during the dorsiflexion gait phase, displaying: (a) the anterior spring connector positioning, and (b) the posterior (blue line).

Structurally, an attempt was made to anchor both virtual ligaments at their anatomical origin and insertion footprints. Anatomically, both the ATiTL and the PTiTL belong to the deep deltoid complex on the medial aspect of the ankle [31]. However, implementing both elements strictly on the medial side within this simplified framework could induce mechanical imbalance under axial loading, causing the tibial component to pivot or tilt toward the unconstrained lateral side. To ensure three-dimensional structural equilibrium and solver convergence, a cross-lateral stabilization strategy was adopted: the anterior complex was maintained on its native medial aspect, while the posterior complex was strategically transposed to the lateral side. This bilateral distribution acts as an equivalent localized system that neutralizes artificial shear moments and stabilizes the assembly, substituting for the missing global anatomy.

As acknowledged in biomechanical literature, obtaining exact physiological parameters such as the subject-specific zero-load length and initial ligamentous strain remains challenging, frequently leading researchers to entirely exclude these stabilizing structures or artificially lock the joint degrees of freedom [25, 33]. This way, the specific pre-load conditions for these elements were calculated analytically based on the geometric displacement relative to the neutral position.

The neutral stance was established as the initial baseline. When the joint assembly was angularly displaced into gait extremes, the linear variation in length (Δx) from this neutral baseline was measured directly in the CAD environment.

The ligaments were modeled governed by tension-only, no-compression physiological mechanics. Operating under this rigorous physical regime, any directional rotation causing a ligament bundle to slacken drops its active pre-load entirely to zero. For instance, during dorsiflexion, the anterior bundle becomes completely slack (0 N), while the posterior bundle is subjected to tensile pre-load. This way, during plantarflexion, the posterior bundle relaxes to zero tension while the anterior bundle is actively stretched. For the bundle undergoing stretch, the structural effort was determined by multiplying its mechanical stiffness (k) by the calculated elongation (Δx) accumulated from the neutral state ($F = k \cdot \Delta x$), applying Hooke's Law of elasticity.

Table 6.3: Boundary conditions, structural fixtures, and virtual spring properties.

Entity	Element / Constraint Type	Stiffness (k)	Pre-load (F_{pre})
Inferior Talar Base	Rigid Fixture	—	—
Anterior Complex	Tension-only Spring	70 N/mm [14]	Neutral: 0,0 N Dorsifl.: 0,0 N Plantarfl.: 263,9 N
Posterior Complex	Tension-only Spring	80 N/mm [14]	Neutral: 0,0 N Dorsifl.: 239,2 N Plantarfl.: 0,0 N

Contacts: Following the definition of the boundary and loading conditions, the contact interactions between the distinct components of the model must be established. For the bone-implant interfaces, finite element methodologies generally distinguish between two scenarios: the immediate postoperative phase (non-osseointegrated), typically evaluated using frictional contact models [23, 47], and the long-term follow-up (fully osseointegrated), evaluated by assuming a completely tied interface [12, 24, 48].

In the immediate postoperative scenario, primary stability relies heavily on the high coefficient of friction between the porous tantalum layer (Trabecular Metal) and the cancellous bone. While general finite element methodologies often model this initial non-osseointegrated state using a baseline friction coefficient of 0,5 [23, 47], the specific Trabecular Metal utilized in the Zimmer prosthesis possesses an exceptionally high coefficient of friction, universally reported as 0,88 against cancellous bone [51, 87]. However, attempting to computationally replicate this extreme 0,88 coefficient using a nonlinear frictional contact algorithm frequently induces severe numerical convergence instabilities and artificial stress concentrations at the discrete nodal interface, rendering the short-term evaluation mathematically unrealistic.

Because this elevated friction inherently restricts initial micromotion to levels well below the critical threshold required for bone ingrowth, the implant achieves biological fixation remarkably fast. Therefore, to accurately represent a successful long-term postoperative scenario characterized by complete biological osseointegration, while simultaneously bypassing nonlinear contact artifacts, a fully bonded (tied) condition was explicitly assigned to all bone-implant interfaces. This rigorous configuration ensures a direct and realistic physiological load transfer without artificial micromotion [23, 48].

Regarding the prosthetic components, the modeled prosthesis employs a fixed-bearing design. Consequently, the UHMWPE insert was fully bonded to the tibial tray, preventing any relative displacement between these two parts [48].

At the dynamic articulating interface between the highly cross-linked polyethylene bearing and the CoCrMo talar component, a "No Penetration" contact condition was defined. Based on the established frictional properties of these specific biomaterials, an isotropic coefficient of friction of 0,04 was implemented within this contact definition [9, 12]. This specific formulation allows for relative sliding while strictly preventing non-physical volumetric penetration between the component meshes during load transfer [12]. To account for potential peripheral impingement and guarantee the absolute kinematic integrity of the overall assembly, a secondary contact interface was explicitly established between the surrounding native osseous boundaries and the polyethylene bearing. While static bone-to-implant fixation interfaces necessitate a high friction coefficient (e.g., 0,5) to ensure primary stability, applying such magnitude to the dynamic peripheral boundaries would induce unrealistic tangential shear upon incidental contact. Therefore, the same ultra-low isotropic coefficient of friction was assigned to this specific regional constraint. This advanced peripheral configuration acts strictly as a physical boundary to prevent unphysical mesh overclosure, avoiding the generation of artificial shear resistance and thereby preserving the unrestricted natural kinematics of the joint.

The complete set of kinematic constraints, applied loads, and contact definitions implemented within the *SolidWorks Simulation* environment is synthesized in Table 6.4.

Table 6.4: Summary of contact interactions, numerical formulations, and friction coefficients implemented in the FE model.

Interface Pair	Contact Type	Friction Coefficient (μ)
Bone-to-Implant	Bonded (Tied)	–
Polyethylene Bearing / Tibial Tray	Bonded (Tied)	–
Polyethylene Bearing / Talar Component	Contact	0,04 [12]
Bone Boundaries / Polyethylene Bearing	Contact	0,04

The position-specific parameters defining each simulated loading scenario, joint angle, applied load, and ligamentous pre-loads, are summarised in Table 6.5.

Table 6.5: Summary of the simulation-specific parameters applied across the loading positions.

Parameter	Plantarflexion	Neutral	Dorsiflexion
Joint angle	15°	0°	10°
Axial load	400 N	600 N	1 600 N
Body weight equivalent	0,67 BW	1,00 BW	2,67 BW
Anterior spring pre-load	263,9 N	0 N	0 N
Posterior spring pre-load	0 N	0 N	239,2 N

6.4 Mesh Generation

The subsequent phase of the computational framework involved the discretization of the three-dimensional solid models representing the bone structures and prosthetic components. The model was meshed utilizing tetrahedral elements. As highlighted by the comprehensive review of Mondal and Ghosh (2020) [14], tetrahedral formulations are universally preferred and highly recommended in foot and ankle biomechanics, as they are inherently more suitable for accommodating the complex, non-prismatic geometries of anatomical structures compared to hexahedral elements.

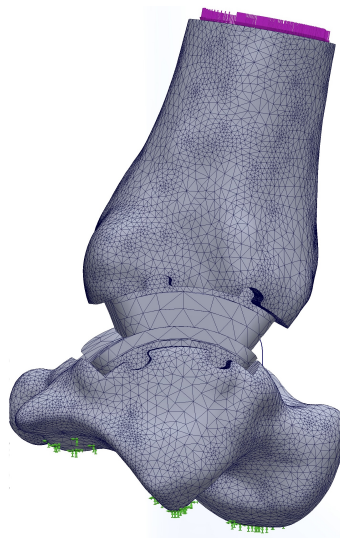


Figure 6.5: Finite element mesh generation of the ankle joint assembly configured for the dorsiflexion gait phase.

The mesh quality was ensured by maintaining highly stable element metrics throughout the model, successfully avoiding distorted elements that could lead to numerical singularities [24]. To validate the mesh density and ensure computational reliability, the discretization strategy was rigorously compared against converged mesh parameters established in seminal finite element studies on total ankle replacement. Biomechanical literature strongly emphasizes the importance of applying localized mesh refinement specifically at critical regions, such as the bone-implant interfaces and the articular contact surfaces, to accurately capture contact mechanics and stress gradients [12, 48]. Following this rationale, the element size at these specific critical regions in the present model was highly refined to sub-millimeter values, reaching a minimum of approximately 0,372 mm.

This degree of refinement strictly aligns with, and in fact exceeds, the optimal converged mesh parameters documented in the literature. For instance, comprehensive mesh sensitivity analyses by Martinelli et al. [12] and Zhang et al. [9, 64] demonstrated that peak contact stresses and contact areas in TAR models fully converge at an element size of 1,0 mm at the articulating surfaces. Similarly, Yu et al. [48] and Sopher et al. [47] established that element edge lengths of 1,0 to 1,5 mm at the articular and bone-implant interfaces are sufficient to guarantee accurate stress and micromotion predictions without discretization errors. Therefore, by adopting these pre-validated, highly refined element sizes, which provide double the resolution of the required literature threshold, in the regions of interest, the current computational framework inherently ensures the accuracy and stability of the contact mechanics outcomes.

Shape verification metrics confirmed good geometric stability across the entire assembly when compared to ideal finite element thresholds. Specifically, 96,4% of the total elements achieved an Aspect Ratio below 3,0, which represents an excellent quality metric, as an Aspect Ratio of 1,0 signifies a perfect equilateral tetrahedral shape and values below 3,0 are considered optimal for minimizing discretization errors. Furthermore, a negligible fraction of only 0,5% exhibited an Aspect Ratio greater than 10,0. The mesh validation routine also revealed exactly zero distorted elements, successfully meeting the ideal numerical threshold for geometric stability.

The details regarding the element types, prescribed element sizes, and the resulting number of nodes and elements for each component within the finalized assembly are summarized in Table 6.6.

Table 6.6: Mesh details for the bone structures and prosthetic components in the finite element model.

Component	No. of Elements	No. of Nodes
Tibia Bone	104 656	167 580
Talus Bone	144 091	219 470
Tibial Component	17 340	29 484
Talar Component	17 736	30 081
Bearing (UHMWPE)	472	982
Total Assembly	284 295	447 597

6.5 Chapter Conclusion

In conclusion, the development of this computational framework required a balance between engineering simplifications and anatomical realism to achieve solver convergence without losing clinical relevance. The study was formulated as a non-linear static structural analysis, a strategy specifically chosen to bypass the prohibitive computational burden of fully dynamic simulations while rigorously evaluate the joint behavior. The primary objective was to simulate a successful long-term osseointegrated scenario across three distinct functional phases of the gait cycle (Neutral, Dorsiflexion, and Plantarflexion), ensuring that the implant's biological fixation was accurately represented under true peak weight-bearing fluctuations.

To achieve this, general boundary conditions were calibrated to preserve structural integrity while respecting joint kinematics. Rigid-body motion was strictly controlled by fixing the talar base, yet physiological tibial guidance was carefully preserved through 1D tension-only springs that replicate native ligamentous constraints. Furthermore, the extreme non-linear friction of the porous coating was simplified into a fully bonded interface, while relative sliding with ultra-low friction was explicitly permitted at the dynamic articulating interfaces. Ultimately, this balanced numerical environment ensures that the calculated internal stress fields, contact pressures, and safety factors presented in the subsequent chapter are a reliable reflection of prosthetic performance rather than mathematical artifacts.

Chapter 7

Results and Discussion

This chapter presents and discusses the finite element analysis results of the Zimmer Trabecular Metal Total Ankle system across three static loading positions representative of the stance phase of gait. The analysis is structured as follows:

- **Section 7.1.1** evaluates the mechanical behavior of the HXLPE bearing insert, assessing contact pressure distribution and von Mises stress against established wear and yielding thresholds, and contextualizes the results against published data for other commercial TAR designs.
- **Section 7.1.2** evaluates the structural integrity of the metallic tibial and talar components, assessing peak von Mises stresses against the yield strength of their respective alloys across the three simulated loading positions.
- **Section 7.2** evaluates the periprosthetic bone response through minimum (ϵ_3) and maximum (ϵ_1) principal strain analysis within a 10-mm volumetric Region of Interest immediately adjacent to the bone-implant interface, interpreted against biomechanical thresholds for homeostatic remodelling and trabecular yield, and concludes with an overall assessment of the strain environment including a volumetric comparison of bone at risk against published data for other commercial TAR designs.

Where relevant, results are contextualized against existing computational studies, and the implications of model limitations are discussed throughout.

7.1 Mechanical Analysis of the Prosthetic Components

The von Mises stress is a metric for determining the equivalent stress for ductile materials, and comparing this equivalent stress to the material's yield strength provides a reliable failure criterion [28]. Because the safety factor quantifies how close the material is to yielding, it is critical to investigate the maximum von Mises stress generated on the implant to determine if the UHMWPE

is close to yielding and undergoing plastic deformation under the applied loads. From this relationship, the Safety Factor of the prosthesis can be quantified to predict how far the material is from yielding and potential rupture under peak physiological loads.

While the von Mises stress evaluates the risk of internal structural crushing (yielding), the contact pressure acts strictly on the articular surface to identify regions susceptible to mechanical overload. These localized peak pressures act as the primary mechanical precursors to tribological wear (the gradual surface degradation caused by friction) and delamination (the severe structural chipping and flaking of the polymer), which are the definitive mechanisms that restrict the long-term longevity of the prosthesis [28]. According to Archard's wear law, the volume of polyethylene debris generated over time is directly proportional to this localized contact pressure [64].

7.1.1 Bearing Component

In the biomechanical evaluation of total ankle arthroplasty, the polymeric bearing component is frequently the weakest link, being highly susceptible to wear and mechanical fatigue, which subsequently leads to aseptic loosening and implant failure [48, 64, 93]. To prevent accelerated wear, the biomechanical literature establishes a strict safety threshold. Above this limit, the surface undergoes severe degradation, generating plastic particulate debris that triggers macrophage-induced osteolysis and subsequent aseptic loosening [12, 28]. Therefore, alongside the von Mises yielding criterion, extracting the contact pressures at the tibio-talar interface is mandatory to evaluate the true wear performance of the Zimmer TM prosthesis.

To comprehensively evaluate the safety limit of the highly crosslinked polyethylene, this study adopted this dual-criterion approach. Regarding the internal material yielding (von Mises), the mentioned reference study [12] referenced standard yield stress values for medical-grade polyethylene ranging from 15 to 23 MPa. However, to ensure maximum safety predictability, the present study deliberately evaluated the yielding risk against a much more conservative and rigorous threshold of 11 MPa, as previously established in the methodology (Table 6.1), representing the lowest reported failure value in the literature [28]. Simultaneously, the contact pressure, which dictates the risk of surface tribological wear and delamination, was assessed with a critical safety limit established at 10 MPa [12].

The obtained results for the equivalent von Mises stress and contact pressure distributions across the three simulated positions are summarized in Table 7.1.

Table 7.1: Quantitative summary of contact pressures and von Mises stresses on the surface of the polyethylene component (UHMWPE) across the three evaluated joint positions.

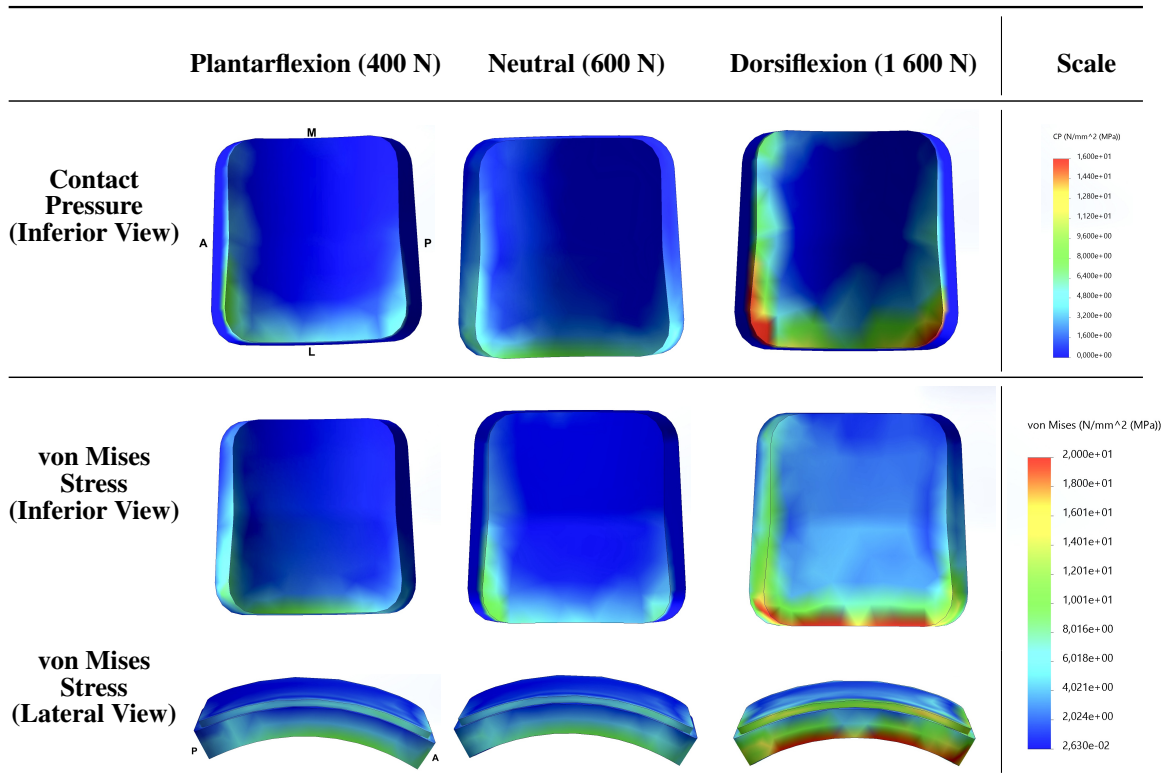
Simulated Position	Contact Pressure [MPa]		von Mises Stress [MPa]	
	Maximum	Average	Maximum	Average
Plantarflexion (400 N)	8,8	0,2	9,7	2,0
Neutral Position (600 N)	8,6	0,2	9,7	2,0
Dorsiflexion (1 600 N)	20,0	0,7	21,9	5,7

As mentioned, a review of the current biomechanical literature reveals a remarkable scarcity of computational studies specifically addressing the Zimmer Trabecular Metal (TM) Total Ankle system. To date, the finite element evaluation conducted by Martinelli *et al.* (2017) [12] remains the primary and almost exclusive reference for the mechanical behavior of this specific design. However, their investigation was strictly limited to the isolated analysis of the UHMWPE bearing under a continuous dynamic gait curve, without addressing the full bone-implant construct. Furthermore, their model adopted a highly simplified material approach by entirely excluding the surrounding bone structures and assigning a single, homogeneous material property to each component [12]. In contrast, the present study expands this scope by simulating a complete bone-implant assembly, incorporating the full spectrum of material layers, specifically differentiating the properties for cortical and heterogeneous cancellous bone, alongside the implant components, thus providing a much more realistic physiological boundary.

The computational results obtained in this study, are in agreement with the existing literature, successfully validating the numerical model. For the contact pressure, the model predicted a peak value of 20,0 MPa during maximum dorsiflexion. This result perfectly aligns with the 19,8 MPa peak contact pressure reported by the literature in the anterior section of the bearing at 40% of the gait cycle (terminal stance), an instant which corresponds to the maximum dorsiflexion position [12]. In the neutral and plantarflexion positions, the peak contact pressures recorded were 8,6 MPa and 8,8 MPa, respectively. These values remain safely below the 10 MPa critical threshold for delamination, suggesting that the prosthesis operates under safe surface loading conditions during the majority of the walking cycle, with critical wear risks being strictly experienced during the extreme push-off phase (dorsiflexion).

A similar mechanical behavior was observed regarding the internal structural yielding. The peak von Mises stresses during the neutral (9,7 MPa) and plantarflexion (9,7 MPa) positions were maintained below the conservative 11 MPa yield limit, translating to a Safety Factor greater than 1. However, during dorsiflexion, the peak von Mises stress spiked to 21,9 MPa, exceeding this threshold.

Table 7.2: Color maps of Contact Pressure (inferior view) and von Mises stress distributions (inferior and lateral views) on the polyethylene component (UHMWPE) across the three simulated positions.



Notably, as illustrated in the color maps presented in Table 7.2, across all three loading positions, the von Mises stress distribution was not uniform but consistently concentrated in the anterolateral region of the bearing, with dorsiflexion representing the most severe expression of this recurring pattern. Rather than a computational anomaly, this persistent localized concentration reflects the true kinematics of the ankle joint colliding with the geometrical constraints of the prosthesis. In a physiological scenario, as the foot rolls from plantarflexion through neutral and into dorsiflexion, the talar component progressively translates anteriorly [64]. Previous biomechanical studies have demonstrated that this motion is natively coupled with an eversion moment, causing the lateral side of the ankle to naturally sustain much higher asymmetric forces than the medial side [94]. In fact, musculoskeletal evaluations confirm that lateral joint forces are significantly higher than medial forces [12], and the physiological force per unit area is naturally greatest over the anterior and lateral portions of the talar dome [3]. To specifically address this complex and asymmetric native kinematics, the Zimmer TM prosthesis was engineered with a bicondylar articular surface featuring a smaller radius of curvature on the medial side compared to the lateral side, as mentioned in Chapter 5 [15, 51]. Because of this inherent geometric asymmetry, even under the simplified static loading applied in the present computational model, the center of contact is mechanically driven to shift toward the anterolateral edge during dorsiflexion [9, 51].

While the present model successfully captured this intended geometric shift, it simultaneously

exposed relatively high stress magnitudes at the implant's bearing edges. In a healthy joint, the compliant articular cartilage safely dissipates these asymmetric physiological forces, keeping peak contact stresses extremely low, in the range of 1,5 to 2,5 MPa, as demonstrated by Ramos *et al.* [24] in the intact model of the bone used in this study. However, when this biological dampener is replaced by the rigid constraints of a highly crosslinked polyethylene bearing, the intended anatomical shift of forces inevitably translates into the 20,0 MPa edge-loading observed in this study. It must be acknowledged that the finite element framework utilized in this study adopted a static approximation of the gait cycle, driven by angular variations, with the ligamentous capsule simplified into spring elements. By reducing the dynamic soft-tissue dampening effect, the model essentially simulates a "worst-case scenario" where load transfer occurs almost entirely through the rigid implant components [13].

When situating these results within the broader context of other commercial TAA designs, the Zimmer TM system demonstrates a highly competitive biomechanical profile. For instance, two-component designs like the Agility prosthesis have been computationally reported to generate contact pressures that widely exceed 18 MPa in standard configurations [12]. Three-component mobile-bearing designs, theoretically intended to distribute stresses more evenly, also present high peaks. The BOX prosthesis, for example, registers peak contact pressures ranging from 16,1 to 18,4 MPa during the terminal stance [12]. Furthermore, recent evaluations of contemporary fixed-bearing systems confirm that edge-loading is a ubiquitous challenge. Under physiological gait loading, the INBONE II and Vantage prostheses have demonstrated extreme edge contact pressures reaching 39,2 MPa and 45,4 MPa, respectively [28]. In more demanding dynamic multi-body simulations, peak contact stresses at the bearing edges for INBONE II and INFINITY systems have been reported to spike between 60,5 and 91,1 MPa [9], while their isolated internal von Mises stresses frequently surpass conservative yield limits, reaching approximately 14,9 MPa even under simple axial compression [12]. Ultimately, this comparison reveals that the 20-22 MPa edge-loading peaks observed in the Zimmer TM during extreme dorsiflexion are not only aligned with, but often substantially lower than, the critical stress concentrations reported across major contemporary prostheses.

In conclusion, the results indicate that the model successfully captures the natural anterolateral distribution of joint forces. However, by replacing the compliant native cartilage with a rigid prosthetic bearing, the system inherently loses the capacity to attenuate these loads as a healthy joint would. Instead, the rigid prosthetic constraints concentrate and amplify these forces at the bearing edges. As a consequence, the predicted peak stress exceeds the conservative yield limit adopted, although it remains within the absolute ultimate strength range reported for standard medical-grade UHMWPE. This theoretical vulnerability, however, must be carefully contextualized. First, the present study applied a simplified static load of 1 600 N for the dorsiflexion position, whereas *in vivo* dynamic loads during a full gait cycle can frequently exceed 3000 N [28]. Under such physiological loads, the absence of biological attenuation could plausibly drive the bearing to-

ward failure. Second, and more decisively, the real Zimmer TM system is manufactured from a proprietary HXLPE, not standard UHMWPE [87]. Experimental evaluations conducted by the manufacturer indicate that this advanced crosslinked material reliably resists fatigue and crack propagation even under severe edge-loading conditions and high dynamic loads [10]. Taken together, this suggests that although higher physiological loads could worsen the predicted stresses, the superior fatigue resistance of the real HXLPE bearing would likely offset this increase. On balance, rather than indicating an immediate risk of global failure, the results suggest that the bearing may simply operate near its upper mechanical limits in these isolated anterolateral regions during extreme push-off, remaining a mechanically safe and durable clinical solution overall while still translating, even if in an intensified form, the underlying loading reality of the native ankle joint.

7.1.2 Tibial and Talar Components

When evaluating the metallic hardware of the Zimmer TM prosthesis, the peak stress values for both the tibial and talar components are summarized in Table 7.3. For both components, the absolute peak stresses were unequivocally observed during dorsiflexion, a pattern that also perfectly corroborates recent computational studies confirming that internal stresses in metallic prosthetic components peak during this extreme push-off phase [95].

When evaluating the metallic hardware, the Zimmer TM system utilizes a Titanium alloy (Ti-6Al-4V) tibial baseplate and a Cobalt-Chromium-Molybdenum (CoCrMo) talar component [51]. The recorded maximum von Mises stresses remain well within the safe elastic regions of these high-strength medical-grade alloys. Therefore, the results do not suggest any risk of fatigue failure, macroscopic yielding, or fracture of the metallic components under the simulated physiological loading.

Notably, the tibial component registered substantially higher maximum stresses than the talar component (188,2 MPa vs. 71,8 MPa). This difference could result from both material and geometric factors: the lower elastic modulus of the Ti-6Al-4V tibial baseplate (115 GPa) versus the stiffer CoCrMo talus (210 GPa) [12] alters stress absorption, while its flatter profile and anchoring rails act as local stress concentrators. In contrast, the talar component's smooth bicondylar dome naturally dissipates contact forces [51]. Consequently, although both components remain structurally safe, the tibial tray acts as the primary receiver of intensified internal stresses.

Furthermore, the realistic mechanical behavior observed in the metallic components serves as an essential computational verification for the finite element model.

Table 7.3: Peak (Maximum) von Mises stress values [MPa] recorded on the metallic prosthetic components across the three simulated phases of gait.

Component	Plantarflexion (400 N) [MPa]	Neutral (600 N) [MPa]	Dorsiflexion (1 600 N) [MPa]
Tibial Component	85,7	84,9	188,2
Talar Component	16,9	24,2	71,8

7.2 Bone Remodeling and Structural Integrity: Tibia and Talus

Unlike the prosthetic components, the osseous structures of the ankle joint are living tissues that continuously adapt their mass and architecture in response to their mechanical environment, a phenomenon largely governed by Wolff's Law [59]. Following Total Ankle Replacement (TAR), the introduction of highly stiff metallic components, such as the Ti-6Al-4V tibial tray and the CoCrMo talar dome, drastically alters the physiological load transfer mechanism [13]. This bio-mechanical mismatch often leads to "stress shielding", a process where the rigid implant absorbs the majority of the physiological loads, thereby depriving the adjacent periprosthetic bone of its normal mechanical stimuli [14].

To evaluate this altered mechanical environment, finite element analysis software frequently defaults to metrics such as von Mises stress or Equivalent Strain (ESTRN) to provide a qualitative global map of the load transfer, allowing for the visual identification of large stress-shielded (underloaded) areas across the periprosthetic bone [23, 25]. However, because both von Mises stress and Equivalent Strain are scalar magnitudes that merge all directions into a single positive value, they fail to distinguish between tensile, compressive, and shear vectors [25]. Since human trabecular bone exhibits a highly asymmetric resistance profile, possessing high compressive strength but poor tensile and shear strengths, relying on these scalars is an inadequate predictor of fracture. Furthermore, in heterogeneous bone models, stress-based outputs are heavily dependent on the local elastic modulus assignment [25]. Therefore, accurately predicting structural failure requires isolating the mechanical response using strain-based metrics, which are independent of the elastic modulus, specifically Minimum Principal Strain (ϵ_3) for compression and Maximum Principal Strain (ϵ_1) for tension [47].

According to Frost's mechanostat theory, bone tissue requires a specific window of mechanical strain to maintain its homeostasis. When the load transfer is significantly reduced and local principal strains fall below physiological thresholds (e.g., dropping below $-1000\mu\epsilon$ in compression), an osteoclast-mediated adaptive remodeling is triggered, resulting in severe bone resorption and density loss [22]. On the other hand, abrupt geometric transitions in the implant design, such as anchoring pegs or keels, can act as severe stress concentrators. If the local strains transferred to the relatively soft trabecular bone exceed its ultimate yield limit, typically reported around 0,73%

($-7300\mu\epsilon$) in compression and 0,65% ($+6500\mu\epsilon$) in tension [47], the bone is highly susceptible to accumulating microdamage and subsequent trabecular collapse.

These extreme biomechanical environments are highly detrimental. Excessive strain concentrations lead to microfractures and severe subsidence of the components, whereas pronounced stress shielding leads to critical bone stock loss and aseptic loosening, which remains the leading cause of mid-to-long-term TAR failure [13, 49]. Therefore, to systematically interpret the numerical results and predict the long-term survivorship of the surrounding bone stock, the biomechanical thresholds and failure criteria adopted in this study are summarized in Table 7.4.

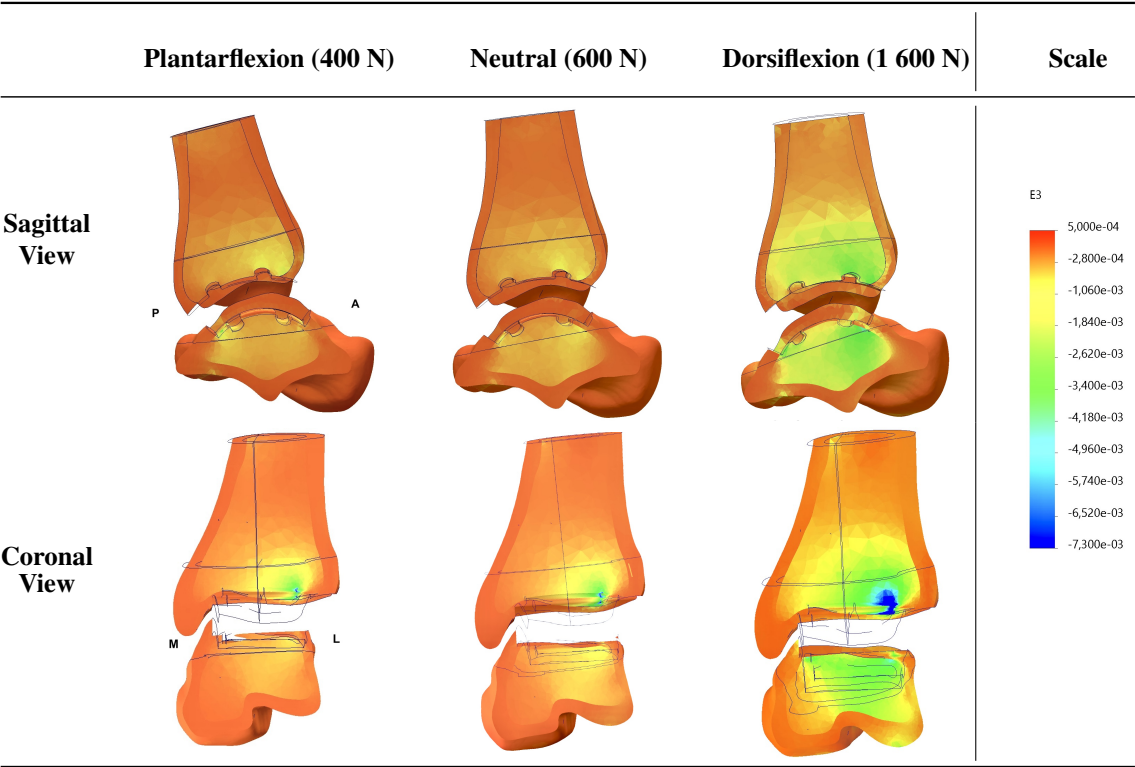
Table 7.4: Biomechanical Thresholds and Failure Criteria for Trabecular Bone

Parameter in SolidWorks	Loading Type	Critical Threshold	Clinical Implication	References
ϵ_3 (Minimum Principal Strain)	Compression (Trabecular Bone)	$> -1000\mu\epsilon$	Stress Shielding (Lazy Zone): Lack of mechanical stimulus leading to bone resorption and aseptic loosening.	[22]
ϵ_3 (Minimum Principal Strain)	Compression (Trabecular Bone)	$< -7300\mu\epsilon$	Compressive Fracture Risk: Trabecular collapse and microdamage leading to component subsidence.	[2, 47]
ϵ_1 (Maximum Principal Strain)	Tension (Trabecular Bone)	$> +6500\mu\epsilon$	Tensile Fracture Risk: Tearing of bone trabeculae due to shear and wedge effect forces.	[2, 47]

7.2.1 Minimum Principal Strain (ϵ_3)

The compressive strain response of the periprosthetic bone was evaluated through the Minimum Principal Strain (ϵ_3) across the three loading positions, spanning a fourfold increase in axial load from approximately 400 N to 1600 N [14]. Following the methodology proposed by Sopher et al. [47], a 10-mm volumetric Region of Interest (ROI) was defined immediately above the bone-implant interface to quantify load transfer in the most mechanically critical periprosthetic zone.

Table 7.5: Color maps of Minimum Principal Strain (ϵ_3) distributions (sagittal and coronal views) on the tibial and talar bone across the three simulated loading positions.



Tibial Bone For the tibial ROI, the volume was further subdivided into medial and lateral compartments to capture the medial-lateral load distribution imposed by the prosthetic design. The color map distributions are presented in Table 7.6, and the quantitative ROI results are summarized in Table 7.7.

Table 7.6: Color maps of Minimum Principal Strain (ϵ_3) distributions on the tibial bone (Inferior View) across the three simulated loading positions.

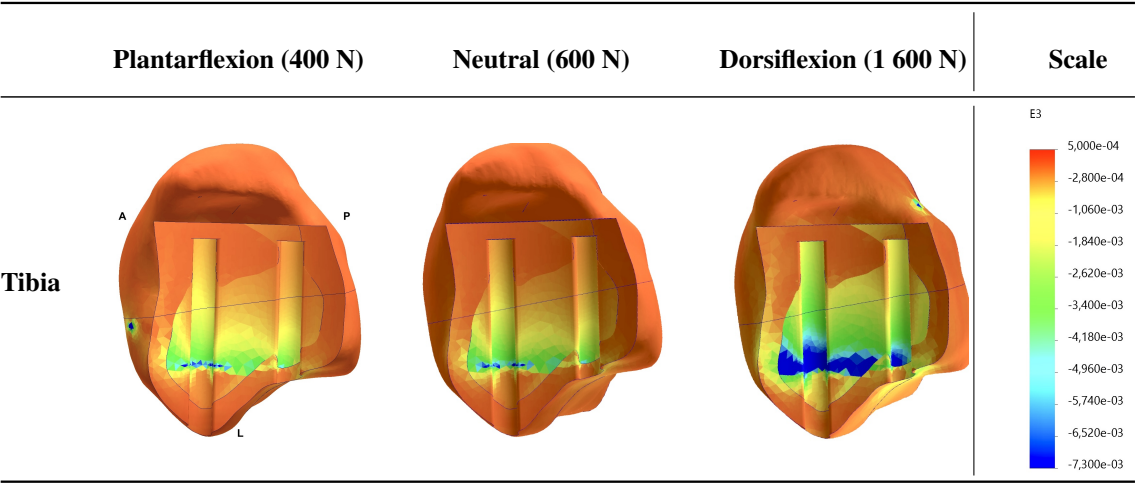


Table 7.7: Summary of Minimum Principal Strain (ϵ_3) in the periprosthetic tibial bone across the three loading positions.

Metric	Compartment	Plantarflexion (400 N)	Neutral (600 N)	Dorsiflexion (1 600 N)
ROI Mean [$\mu\epsilon$]	Medial	−371 [†]	−343 [†]	−1019
	Lateral	−1080	−1120	−3012
ROI Min/Peak [$\mu\epsilon$]	Medial	−1380	−1220	−4203
	Lateral	−9040*	−9621*	−26210*

* Exceeds ultimate trabecular compressive yield limit (−7300 $\mu\epsilon$).

[†] Below the homeostatic resorption threshold (−1000 $\mu\epsilon$).

The tibial ROI data (Table 7.7) reveal a consistent and pronounced medial-lateral asymmetry across all three loading positions. The medial compartment remains below the homeostatic resorption threshold at every load level, an observation that must be interpreted with caution at low loads. Reduced strains under 400 and 600 N are partially physiological and do not alone confirm chronic resorption risk [14]. At dorsiflexion, is particularly concerning the medial mean of −1019 $\mu\epsilon$, which is only marginally above the lower homeostatic boundary despite the maximum axial load, suggesting that the medial trabeculae are barely stimulated even when the joint is most heavily loaded. The lateral compartment means remain within the homeostatic window across all positions, with dorsiflexion reaching its upper boundary, indicating mechanically adequate stimulus on the lateral side, but with no safety margin.

This apparently healthy lateral average, however, conceals a critical local condition: the lateral peak already exceeds the trabecular yield limit under plantarflexion, the lowest load tested, and persists beyond it at every subsequent position, amplifying dramatically under dorsiflexion. The lateral peak therefore exceeds the yield limit not only under peak loading, but already under the lowest load tested, meaning that trabecular damage is an inherent feature of this design's load transfer, present even under routine weight-bearing demands. These yield-exceeding concentrations are spatially confined to the anterior and lateral tibial margins, specifically, the anterolateral edge of the tibial footprint, the same geometric extremities where the prosthesis transitions abruptly from supported to unsupported bone, consistent with the edge-loading pattern identified in the polyethylene bearing.

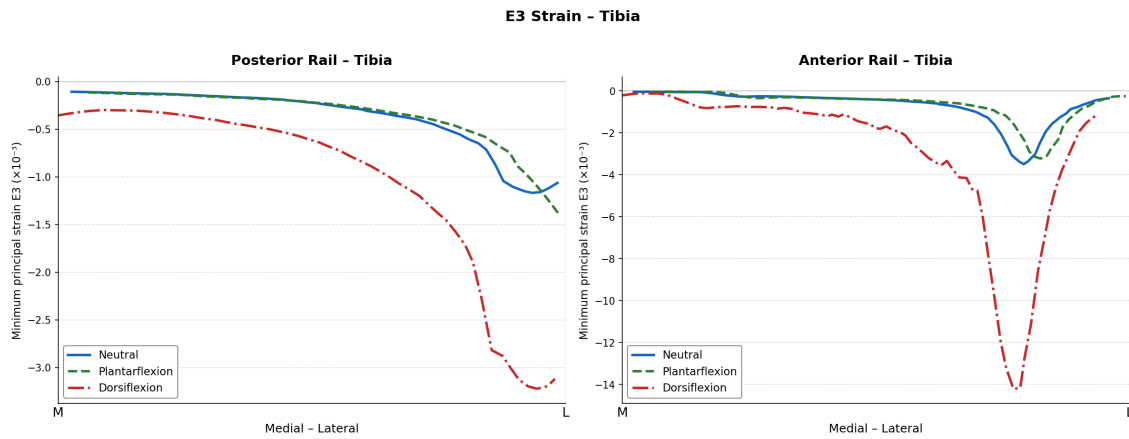
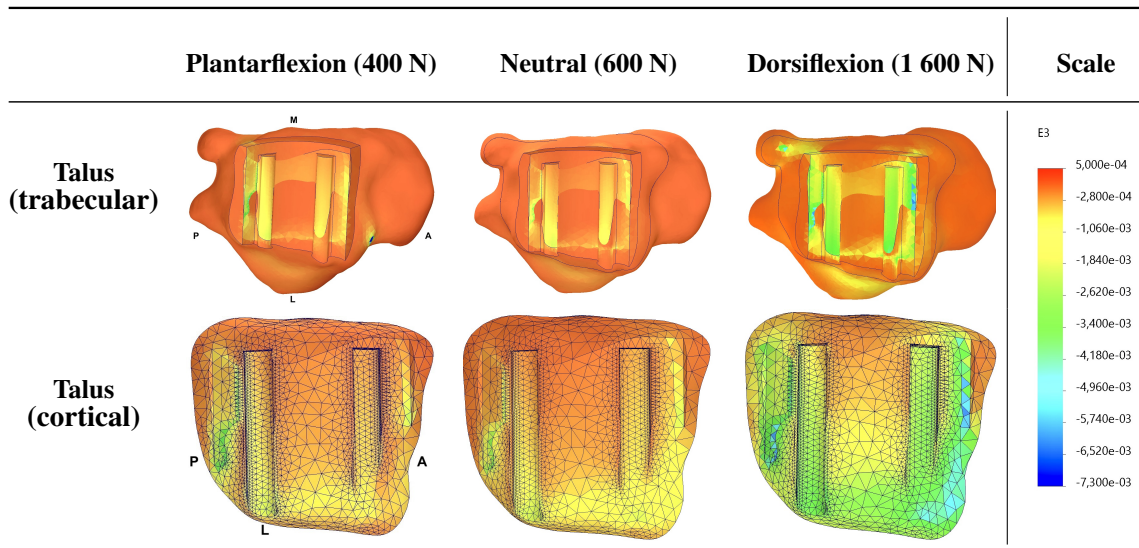


Figure 7.1: Minimum Principal Strain (ϵ_3) along the medial-lateral extent of the posterior (left) and anterior (right) tibial fixation rails, across the three simulated loading positions. The dashed horizontal line indicates the trabecular yield limit ($-7300 \mu\epsilon$).

The strain profiles along the tibial fixation rails (Figure 7.1) reveal a clear contrast between the two rails. Along the posterior rail, all three loading positions produce a gradual medial-to-lateral gradient, with neutral and plantarflexion tracking each other closely and remaining well within the homeostatic window throughout. Dorsiflexion diverges progressively from the midpoint onward, reaching approximately $-3200 \mu\epsilon$ at the lateral extreme, without exceeding the yield limit. Load transfer along the posterior rail is therefore relatively distributed, with no localized stress concentration.

The anterior rail presents a fundamentally different pattern. Under neutral and plantarflexion, strains remain near-zero across the entire medial-to-central extent before dropping sharply into a narrow compressive trough at the lateral edge, exceeding the yield limit at this single location. Under dorsiflexion, the same spatial pattern is preserved but at substantially higher magnitudes. The lateral concentration intensifies markedly, far exceeding the yield limit, while the medial and central portions remain near-zero. The confinement of this concentration to the anterolateral region, absent along the posterior rail, confirms a geometry-driven edge-loading mechanism independent of the applied load level, and identifies the anterolateral fixation rail as a site of structural risk in the tibial periprosthetic bone.

Talar Bone The talar ROI was evaluated as a single volume, as the compact geometry of the resected talar dome does not lend itself to a reliable medial-lateral subdivision for the purposes of this analysis. The color map distributions are presented in Table 7.8, and the quantitative ROI results are summarized in Table 7.9.

Table 7.8: Color maps of Minimum Principal Strain (ϵ_3) distributions on the talar bone (superior and cortical views) across the three simulated loading positions.Table 7.9: Summary of Minimum Principal Strain (ϵ_3) in the periprosthetic talar bone across the three loading positions.

Metric	Plantarflexion (400 N)	Neutral (600 N)	Dorsiflexion (1 600 N)
ROI Mean [$\mu\epsilon$]	−558 [†]	−488 [†]	−1475
ROI Max [$\mu\epsilon$]	−46	−41	−211
ROI Min/Peak [$\mu\epsilon$]	−5470	−3400	−10590*

* Exceeds ultimate trabecular compressive yield limit (−7300 $\mu\epsilon$).

[†] Below the homeostatic resorption threshold (−1000 $\mu\epsilon$).

Under plantarflexion and neutral loading, the talar ROI mean remains below the homeostatic threshold, applying the same interpretive caution as for the tibial medial compartment at these low loads. Peak strains remain within or close to the physiological window, indicating mechanically safe conditions under low-demand loading. However, the plantarflexion peak exhibits a compressive magnitude of −5470 $\mu\epsilon$, suggesting a higher local mechanical overload. Although this magnitude has not yet reached the structural compressive yield threshold, this elevated strain state could place the trabecular bone at a significant risk of microdamage even under minimum load conditions.

Under dorsiflexion, the overall mean enters the homeostatic window, suggesting adequate compressive stimulus is achieved only under peak loading. However, this average masks a critically non-uniform distribution: the peak strain substantially exceeds the yield limit, concentrated at the anterior and posterior resection edges, as observed in Table 7.8, in the cortical perspective, and at the antero-lateral border of the talar dome. Qualitative inspection of the strain maps indicates that the medial and central zones remain consistently below the homeostatic threshold, in the range

of -240 to $-1060 \mu\epsilon$. This spatial distribution mirrors the anterolateral edge-loading identified in both the polyethylene bearing and the tibial bone, and introduces a new region: the posterior resection edge.

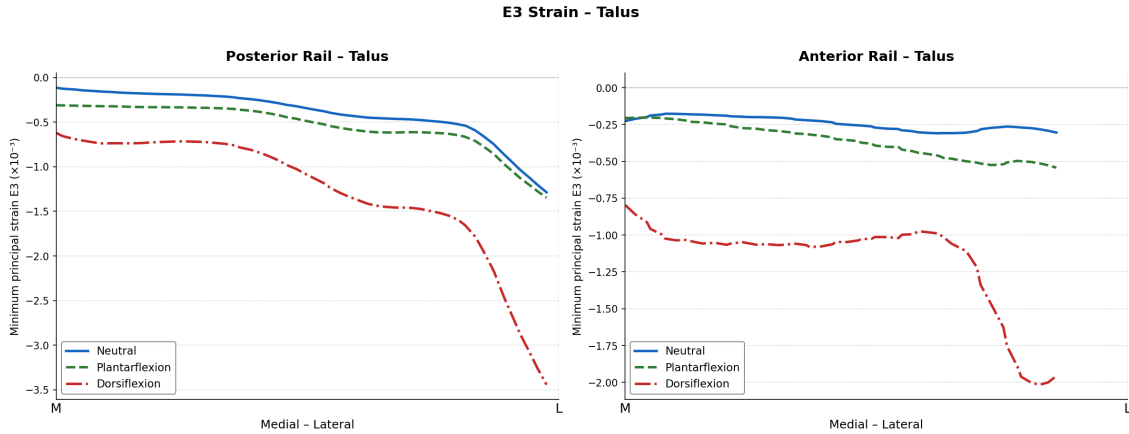


Figure 7.2: Minimum Principal Strain (ϵ_3) along the medial-lateral extent of the posterior (left) and anterior (right) talar fixation rails, across the three simulated loading positions.

The talar rail profiles (Figure 7.2) further characterize this spatial pattern, and again reveal a clear contrast between the two rails. Along the posterior rail, neutral and plantarflexion track each other closely, remaining well within the homeostatic window across the medial two-thirds before declining gradually toward the lateral end. Dorsiflexion follows the same gradient but diverges progressively from the midpoint onward, reaching approximately $-3500 \mu\epsilon$ at the lateral extreme, not exceeding the homeostatic window. Load transfer along the posterior rail is therefore progressively distributed, with lateral amplification under peak load but no localized concentration.

The anterior rail tells a different story. Neutral and plantarflexion remain low and nearly flat across the entire rail, consistently below $-550 \mu\epsilon$ and well under the lower homeostatic boundary, confirming the relative medial and central underloading inferred from the strain maps. Under dorsiflexion, strains are substantially higher from the medial end, sustaining a plateau near $-1000 \mu\epsilon$ (resorption threshold) through the central portion before rising sharply at the lateral extreme to approximately $-2000 \mu\epsilon$. Unlike the tibial anterior rail, no yield-exceeding trough is observed; however, the plateau-then-rise profile confirms that compressive load is preferentially channelled toward the lateral border of the talar dome under peak loading, while the medial and central segments remain at or below the homeostatic threshold across all positions.

Comparison with the Literature When contextualized against existing finite element studies of other commercial TAR designs, the ϵ_3 results reveal both qualitative similarities and quantitatively important distinctions. Sopher et al. [47] reported mean ϵ_3 values within the 10-mm tibial ROI ranging from -2500 to $-3200 \mu\epsilon$ for the BOX, Mobility, and Salto prostheses, broadly comparable to the lateral compartment means observed here. Furthermore, while Sopher et al. reported localized maxima ranging from -6800 to $-8900 \mu\epsilon$, these values are substantially lower than the

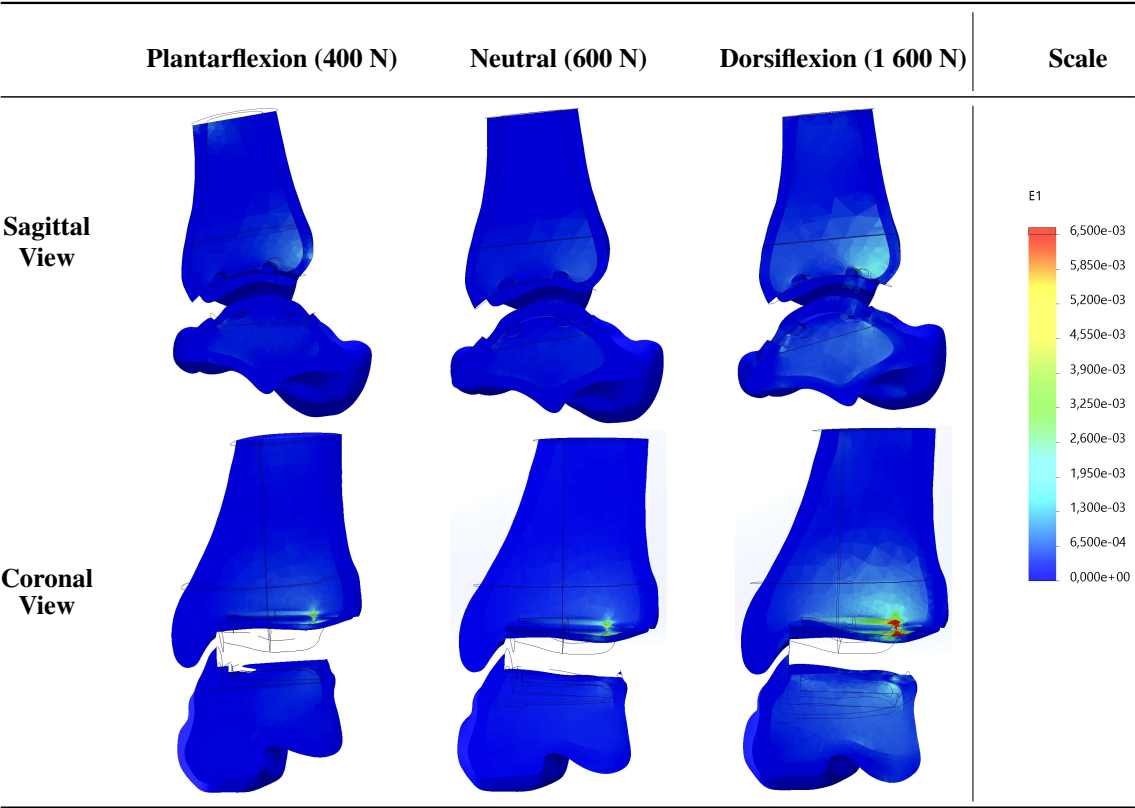
lateral tibial peaks recorded in the present model, which exceed the structural yield limit already under plantarflexion. The persistent medial underloading is also more pronounced than reported for those keel- and peg-based designs, likely reflecting the asymmetric bicondylar load transfer of the Zimmer TM rather than a more symmetric fixation geometry.

For the talar component, Sopher et al. [47] reported mean compressive strains of only -1600 to $-1800 \mu\epsilon$ beneath the implants for traditional designs, well below the yield limit and drastically lower than the massive $-10590 \mu\epsilon$ peak recorded here under dorsiflexion. Ramos et al. [22] documented subthreshold periprosthetic talar strains (below $-1000 \mu\epsilon$) following implantation relative to the intact bone, a pattern qualitatively consistent with the relative medial and central underloading identified here. Taken together, these comparisons suggest that the saddle-type talar component of the Zimmer TM concentrates compressive load more severely at the resection periphery than comparable designs, while leaving medial and central talar bone more consistently understimulated.

7.2.2 Maximum Principal Strain (ϵ_1)

The maximum principal strain (ϵ_1) distribution captures the tensile strain field within the periprosthetic bone, complementing the compressive picture established by ϵ_3 , using the same 10-mm volumetric ROI and medial-lateral subdivision [47]. The trabecular yield limit under tension is established at $+6500 \mu\epsilon$ [2, 47], notably lower in magnitude than the compressive yield limit of $-7300 \mu\epsilon$, reflecting the inherently greater vulnerability of trabecular bone to tensile failure.

Table 7.10: Color maps of Maximum Principal Strain (ϵ_1) distributions (sagittal and coronal views) on the tibial and talus bone across the three simulated loading positions.



Tibial Bone The tibial E1 data, mirror the medial-lateral asymmetry established under ϵ_3 , as seen in Table 7.12.

Table 7.11: Color maps of Maximum Principal Strain (ϵ_1) distributions on the tibial bone (inferior view) across the three simulated loading positions.

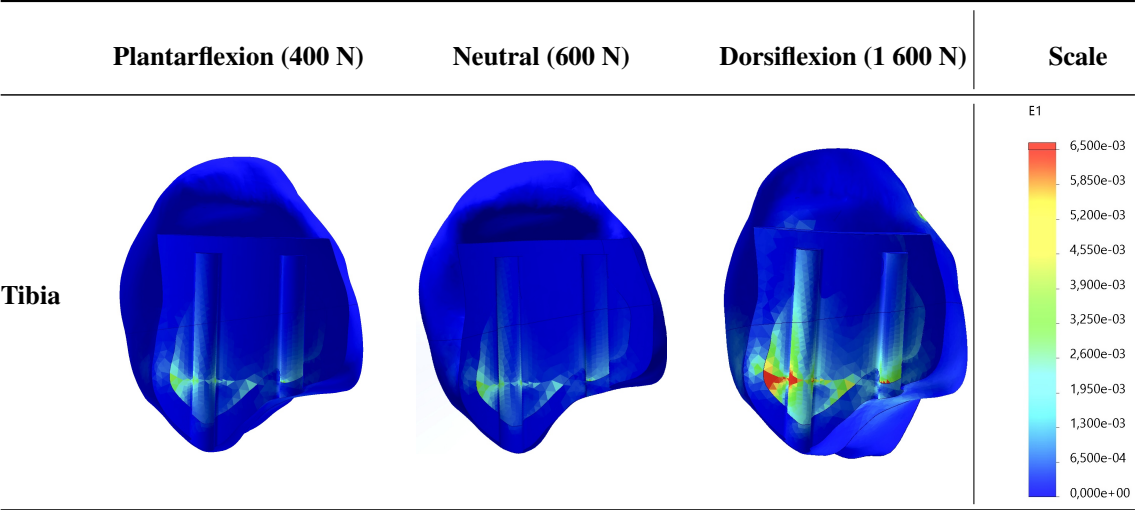


Table 7.12: Summary of Maximum Principal Strain (ϵ_1) in the periprosthetic tibial bone across the three loading positions.

Metric	Compartment	Plantarflexion (400 N)	Neutral (600 N)	Dorsiflexion (1 600 N)
ROI Mean [$\mu\epsilon$]	Medial	+205	+190	+564
	Lateral	+596	+632	+1699
ROI Max/Peak [$\mu\epsilon$]	Medial	+1380	+845	+2390
	Lateral	+5990 [†]	+6170 [†]	+18990 [*]

* Exceeds ultimate trabecular tensile yield limit (+6500 $\mu\epsilon$).
[†] Within 600 $\mu\epsilon$ of the tensile yield limit.

Mean tensile strains remain low across both compartments at all three loading positions, well within the elastic range. The lateral compartment peak, however, tells a different story: it approaches the tensile yield limit already under plantarflexion and neutral loading, values of +5990 and +6170 $\mu\epsilon$ respectively, both within 600 $\mu\epsilon$ of the yield threshold. This near-yield condition at low loads establishes the same early mechanical precursor observed under ϵ_3 : critical tensile straining at the anterolateral corner is not a peak-load phenomenon but a persistent feature of routine loading. The medial compartment, by contrast, peaks well below yield at every position, reinforcing the pattern of lateral concentration and medial underloading.

Under dorsiflexion, the lateral compartment peak far exceeds the tensile yield limit, indicating risk of tensile failure within the lateral trabecular volume. Spatially, the critical tensile zones closely track the compressive hotspot pattern: localized regions exceeding +6500 $\mu\epsilon$ are evident at the lateral edge of the posterior rail and, more prominently, at the anterior rail extending into the adjacent anterolateral tibial bone. The surrounding metaphyseal zone sustains a diffuse tensile field that, while below yield, is mechanically significant across a broader volume.

The co-localization of yield-exceeding compressive (ϵ_3) and tensile (ϵ_1) strains within the same anterolateral volume is the most clinically relevant finding of this analysis. This multiaxial state, simultaneously driving compressive crushing and tensile microcracking in the same trabecular bone, is mechanically more damaging than either component in isolation, creating conditions highly conducive to progressive fatigue failure and implant loosening at the anterolateral interface [96].

Talar Bone The color map distributions for the talar bone under ϵ_1 are presented in Table 7.13, and the talar ROI results in Table 7.14.

Table 7.13: Color maps of Maximum Principal Strain (ϵ_1) distributions on the talar bone (superior and trabecular views) across the three simulated loading positions.

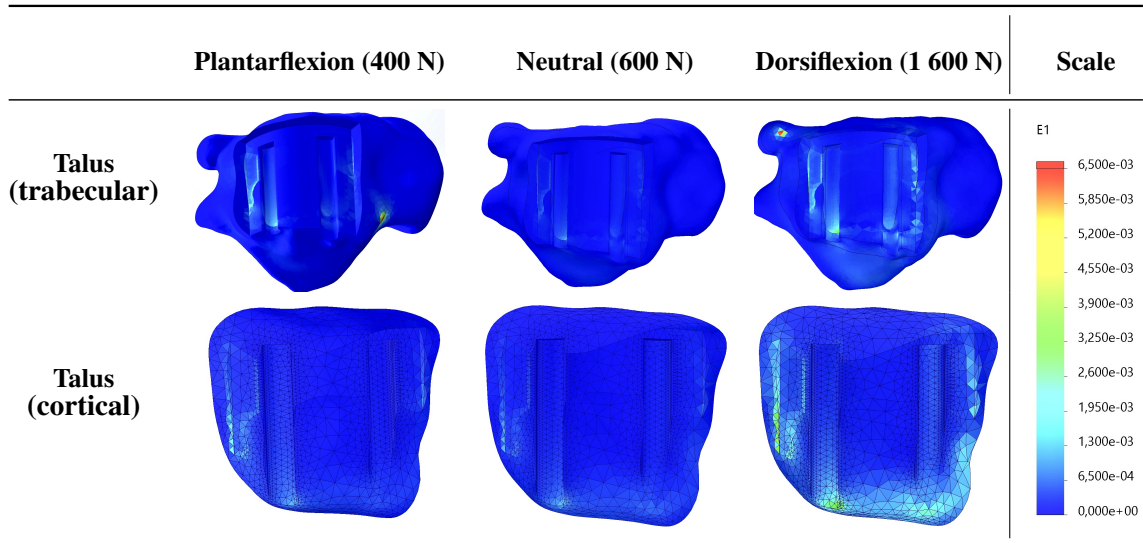


Table 7.14: Summary of Maximum Principal Strain (ϵ_1) in the periprosthetic talar bone across the three loading positions.

Metric	Plantarflexion (400 N)	Neutral (600 N)	Dorsiflexion (1 600 N)
ROI Mean [$\mu\epsilon$]	+191	+156	+483
ROI Max/Peak [$\mu\epsilon$]	+4620	+2000	+8531*
* Exceeds ultimate trabecular tensile yield limit (+6500 $\mu\epsilon$).			

Under plantarflexion and neutral loading, talar tensile strains remain well within the elastic range, with both means and peaks safely below yield. Under dorsiflexion, the volumetric mean confirms overall elastic accommodation, but a highly localized peak of +8531 $\mu\epsilon$ exceeds the tensile yield limit, strictly at the anterolateral border and at the anterior and posterior resection edges of the talar dome. This is consistent with the edge-loading pattern identified under ϵ_3 , reinforcing that these peripheral margins represent the dominant mechanical risk zone regardless of strain component.

Comparison with the Literature The ϵ_1 results reveal a markedly more critical tensile environment than reported for comparable TAR designs. Sopher et al. [47] reported mean (ϵ_1) values between +1300 and +1500 $\mu\epsilon$ and peak values from +3700 to +4500 $\mu\epsilon$ for the BOX, Mobility, and Salto prostheses, all below the tensile yield limit. By contrast, the lateral tibial compartment of the Zimmer TM records a peak under plantarflexion already approaching the yield limit (+5990 $\mu\epsilon$), a condition not reported for any of those designs, and a dorsiflexion peak (+18990 $\mu\epsilon$) more than fourfold their reported maxima.

Furthermore, while Sopher et al. [47] identified the anterior and posterior extremities of keels and pegs as the primary tensile hotspots, driven by symmetric axial rocking of the tibial tray, the critical tensile zone here is concentrated, as observed with the compressive strains, at the antero-lateral rail edge. This reinforces the distinct spatial pattern potentially driven by the asymmetric bicondylar load transfer of the Zimmer TM.

Sopher et al. [47] and Mondal et al. [23] demonstrated that even modest malpositioning dramatically amplifies localized bone strains and interfacial micromotion, increasing the risk of aseptic loosening. Given that the present model assumes optimal implant seating, the massive tensile peaks reported here likely represent a lower bound. This way, any clinical deviation from ideal positioning would be expected to further exacerbate an already critical anterolateral tensile environment.

7.2.3 Overall Assessment of Periprosthetic Strain Environment

The combined ϵ_3 and ϵ_1 analysis reveals three consistent features of the periprosthetic strain environment of the Zimmer TM.

Medial-lateral asymmetry: Across all loading positions and both strain components, the medial tibial and talus compartment remain at or below the lower homeostatic boundary ($-1000 \mu\epsilon$). This pattern of relative mechanical under-stimulation is present at every load level tested. Using the same intact-bone model as the baseline, Ramos et al. [22] documented periprosthetic talar strains falling below $-1000 \mu\epsilon$ in the proximal region following implantation — consistent with the medial and central underloading observed here. While the present study does not include an intact-bone comparison, these observations strongly suggest that, under sustained loading conditions, the medial periprosthetic bone receives insufficient mechanical stimulus to maintain homeostasis. This chronic underloading initiates stress-shielding, a phenomenon where the lack of physiological strain triggers adaptive bone remodelling characterized by progressive bone resorption. As confirmed by extensive computational remodelling studies in both the tibia and talus by Mondal and Ghosh [23, 97], this localized loss of bone density severely compromises the mechanical support of the implant over time, ultimately culminating in component subsidence and aseptic loosening, the primary mechanisms of late-term TAA failure.

Anterolateral stress concentration: All yield-exceeding values, compressive (ϵ_3) and tensile (ϵ_1), localize consistently to the anterolateral corner of the tibial footprint, where the anterior fixation rail terminates at the transition between supported and unsupported bone. The co-occurrence of compressive and tensile peaks in the same trabecular volume creates a multiaxial mechanical state that is more damaging than either component in isolation, and which, if sustained cyclically, creates conditions conducive to progressive damage at this interface. Exceeding the ultimate yield limits in these highly localized regions could initiate a progressive osseous failure cascade. The immediate structural consequence is hypothesized to be trabecular microcracking and localized bone collapse [98]. As the supporting trabecular architecture deteriorates under cyclic loading,

this mechanical insufficiency may possibly lead to asymmetric component subsidence, specifically, the anterolateral tilting of the tibial tray [98]. This progressive malalignment directly compromises the bone-implant interface, severely increasing interfacial micromotion and preventing proper long-term osseointegration [47, 53]. Ultimately, the combination of structural subsidence and the loss of stable fixation is highly likely to culminate in aseptic loosening, which remains one of the most definitive and devastating endpoints for TAR failure [28, 53].

Rail-specific load transfer divergence: The posterior rail distributes load progressively along its medial-to-lateral extent, with strains remaining within physiological bounds across most of its length. The anterior rail behaves differently: the medial and central portions are largely unloaded, while a narrow concentration at the lateral terminus exceeds the yield limit under all loading conditions. This divergence is consistent across both bones and both strain components, and identifies the anterolateral termination of the anterior rail as a structural liability of this design.

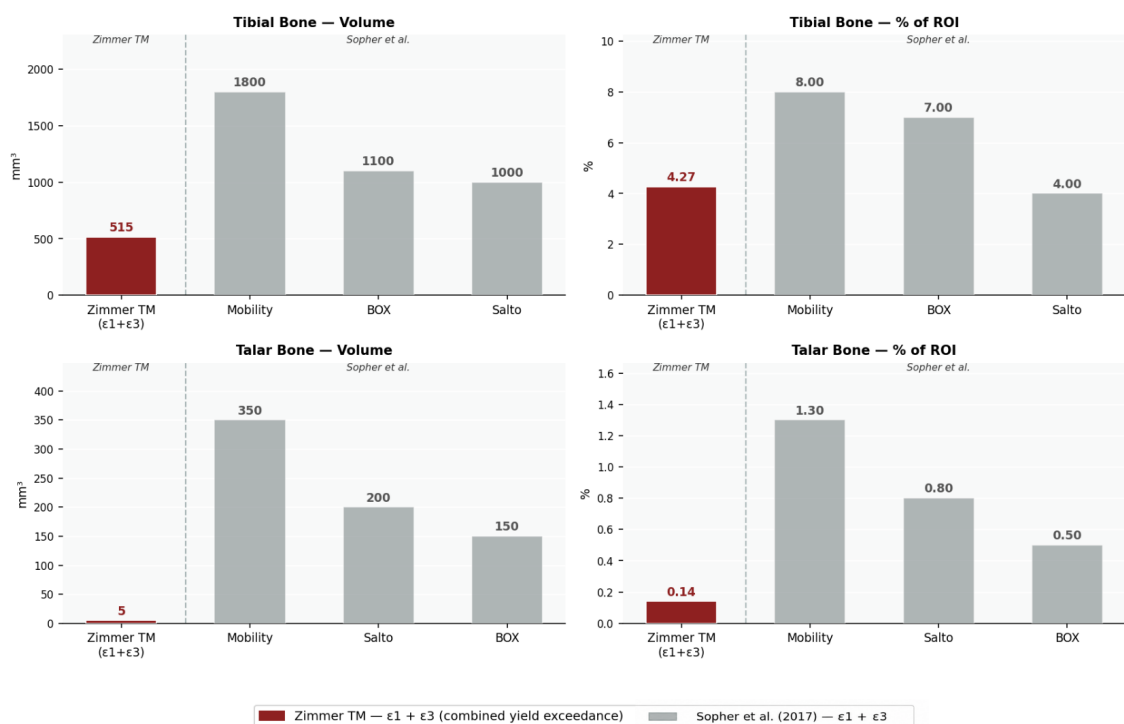


Figure 7.3: Bone volume exceeding the trabecular compressive (ϵ_3) and tensile (ϵ_1) yield limits for the Zimmer TM (present study) and the Mobility, BOX, and Salto prostheses [47], at dorsiflexion (1 600 N).

In order to contextualize the volumetric damage predicted for the Zimmer TM, the results were compared against the failure metric established by Sopher et al. [47]. As previously mentioned, in this reference biomechanical study, three commercial ankle prostheses (BOX, Mobility, and Salto) were evaluated using an equivalent 10-mm trabecular ROI, and a combined damage metric was defined as the total percentage of bone volume failing simultaneously in compression (ϵ_3

$< -7300 \mu\epsilon$) and tension ($\epsilon_1 > +6500 \mu\epsilon$), establishing the standard benchmark for mechanical risk assessment in total ankle arthroplasty (Figure 7.3). It should be noted that Sopher et al. simulated a single static loading condition: dorsiflexion at $5.2\times$ body weight, whereas the present study evaluated three loading positions spanning a fourfold load range, with dorsiflexion at 2.67 BW. The comparison presented here is therefore restricted to the Zimmer TM dorsiflexion results, the loading condition most comparable in magnitude and joint position to Sopher et al. .

Against this benchmark, the Zimmer TM presents a consistently smaller volume of bone at risk than all three designs reported by Sopher et al., at both the tibial and talar levels. This result, however, should not be interpreted as evidence of a mechanically superior design. As established in the preceding sections, the yield-exceeding zones in the Zimmer TM are spatially confined to the anterolateral fixation rail, which is a narrow region subject to substantially higher peak strains than any value reported for the keel- and peg-based designs. The lower volume at risk thus reflects geometric concentration rather than mechanical safety: fewer elements are critical, but those that are carry a disproportionate mechanical burden.

Whether focal, severe overloading of a small bone volume is clinically preferable to more distributed overloading of a larger volume remains an open question, as the two failure modes — focal trabecular crushing versus distributed fatigue damage — carry different implications for fixation stability and long-term bone remodelling [26, 47, 96].

Talar strains in the present model are consistently lower in magnitude than their tibial counterparts across both ϵ_3 and ϵ_1 , mirroring the pattern reported by Sopher et al. [47], who similarly found talar trabecular bone strains to be lower than tibial strains across all three implant designs studied.

Chapter 8

Conclusion and Future Work

8.1 Conclusion

Total Ankle Arthroplasty continues to lag behind hip and knee replacement in long-term survivorship, with aseptic loosening and component subsidence remaining the dominant failure mechanisms across international registries [49]. These failures are ultimately rooted in a persistent biomechanical conflict: the difficulty of transferring physiological joint loads into the bone without either overloading localized fixation features or under-stimulating the surrounding trabecular structure. Addressing this conflict requires predictive engineering tools capable of identifying mechanical vulnerabilities before they manifest clinically, rather than relying on the reactive, decades-long timeline of registry follow-up. This dissertation contributed to that effort through an independent finite element evaluation of the Zimmer Trabecular Metal (TM) Total Ankle, a fourth-generation system whose lateral transfibular approach, arcuate bone resections, and porous tantalum rail fixation remain comparatively under-represented in the computational literature.

The results show that the Zimmer TM delivers what its design intends: a mechanically stable, well-anchored interface under the loading conditions simulated. The metallic tibial and talar components operate safely within their material limits across all three positions of the gait cycle, and the UHMWPE bearing presents a mechanically competitive profile relative to other contemporary designs, with edge-loading peaks substantially lower than those reported for the Agility, BOX, INBONE II, and Vantage systems [28]. The high friction coefficient reported for porous Trabecular Metal [13] is expected to restrict destructive shear forces and limit micromotion at the bone-implant interface, favoring early osseointegration. In this sense, the implant represents a genuine advancement over legacy designs, broadly consistent with the favorable short- to mid-term survivorship reported in national joint registries.

This rigid fixation mechanism, however, comes at a specific and spatially predictable mechanical cost. Across the polyethylene bearing, the tibial bone, and the talar bone, compressive (ϵ_3) and tensile (ϵ_1) strains consistently concentrate at the anterolateral edges of the prosthesis – the

region where the implant's geometry is most exposed and where load transfer abruptly transitions from supported to unsupported bone – and simultaneously exceed the trabecular yield limit at this location [47]. This co-localization is mechanically significant: compressive overload crushes trabeculae while tensile overload tears them apart, and when both occur in the same bone volume the result is a multiaxial damage state that drives microdamage accumulation far faster than either mode alone, creating the conditions for progressive fatigue failure, fixation loosening, and implant subsidence at exactly this location.

The inverse problem occurs on the medial side. Across all three loading positions, including peak dorsiflexion, the medial tibial and talar trabecular bone remains at or below the homeostatic strain threshold. Per Frost's mechanostat theory, strains this low fail to provide the mechanical stimulus bone needs to maintain its density, which over time triggers osteoclast-mediated resorption. While confirming true stress-shielding requires an intact-bone comparison, these strain levels align with the periprosthetic patterns reported by Ramos et al. [22] using the same bone model. This chronic medial under-loading offers a plausible biomechanical explanation for two complications reported in mid-term clinical follow-up of this implant: medial malleolar stress fracture, from a progressively weakened medial cortex [4], and periprosthetic cyst formation, from resorbed bone being replaced by fibrous or cystic tissue. Combined with the lateral transfibular approach's limited direct visualization of the medial joint space during surgery [4], this could also help explain the high reoperation rates for medial gutter debridement reported in registries, reaching up to 23,7% at five years [68], a complication mechanism not visible in standard survivorship statistics, since it does not register as implant loosening.

Compared to the keel- and peg-based BOX, Mobility, and Salto prostheses [47], the Zimmer TM puts a smaller volume of bone at this risk, but the bone that is at risk experiences peak strains several times higher than reported for those designs, reflecting a different failure philosophy, trading distributed moderate overload for concentrated severe overload at one geometric weak point. Taken together, these findings position the Zimmer TM at the focal-overloading end of the central trade-off in TAR fixation design: fewer trabecular elements are mechanically compromised than with distributed keel- or peg-based systems, but those elements carry a disproportionate burden, and the medial compartment is left chronically under-stimulated. This is not evidence that the design is unsafe in general terms. The metallic and polyethylene components remain well within safe limits throughout. However, it could mean that the anterolateral rail edge and the medial bone stock are the two specific locations where this implant's long-term performance will most likely be decided, and where targeted attention, in surgical positioning, implant geometry, or follow-up imaging, is warranted. The talar interface, consistently identified in registries as the weaker mechanical link in TAR systems [49], shows this same spatial pattern of peripheral overloading and central underloading.

These findings are consistent with the clinical timeline reported across TAR systems: tibial

failures tend to occur early post-operatively, whereas talar failures typically present later [49]. This temporal pattern may help explain why the complications reported thus far for the Zimmer TM system have predominantly involved the tibial component. Although the present model predicts lower strain magnitudes in the talus than in the tibia, the talus is a smaller, more vascularly precarious bone with limited capacity to withstand sustained mechanical overload [49], meaning that talar failure, though not yet apparent within current follow-up windows, remains plausible over the longer term.

Translating the present finite element findings into clinical practice suggests several considerations regarding the use of the Zimmer TM prosthesis, though these should be interpreted in light of the inherent limitations of a static single-specimen model. The most severe mechanical demands predicted occur during dorsiflexion, the phase associated with peak push-off forces. This may suggest that patients implanted with this design could benefit from avoiding high-impact activities generating excessive cyclic push-off forces, such as running or jumping, as these would be expected to further amplify the strain concentrations identified here. Whether this translates into a clinically meaningful increase in failure risk cannot be determined from the present data alone.

Regarding postoperative surveillance, the distinct spatial patterns identified here suggest that targeted monitoring of the anterolateral tibial interface and the peripheral talar edges, for early signs of subsidence or radiolucency, may be more informative than a generalized assessment. Concurrent observation of the medial and central compartments for signs of reduced bone density could allow earlier detection of the dual-failure mechanisms predicted by this model, though prospective clinical validation would be required to confirm the relevance of these patterns in vivo.

Whether the Zimmer TM's fixation philosophy ultimately proves superior to the distributed-contact approaches of keel- and peg-based designs cannot be resolved by a single static model. What this study does establish is that the design successfully resolves some of the mechanical failure modes that have historically plagued total ankle arthroplasty, while simultaneously introducing specific, spatially predictable vulnerabilities – anterolateral overloading and medial underloading – that are not apparent from gross survivorship data alone. The strain magnitudes predicted at the anterolateral edge and the persistence of medial under-stimulation across all loading conditions suggest that these vulnerabilities are not incidental artifacts of the model, but a structural consequence of the implant's geometry that should be expected to progress with cyclic loading over time. Given that focal trabecular damage and chronic under-stimulation are cumulative processes, this implies that, even though no failure is evident over the mid-term follow-up window currently reported in the literature, a decline in fixation integrity at these specific sites is likely to emerge as more Zimmer TM implants reach their tenth and fifteenth years in service, and should become detectable as a measurable trend in long-term registry data.

Overall, the objectives established at the beginning of this dissertation were successfully achieved. A comprehensive biomechanical assessment of the currently available surgical solu-

tions for the treatment of ankle osteoarthritis was performed, and a finite element framework was developed to investigate the load-transfer mechanisms and structural response of a representative contemporary total ankle prosthesis. The obtained results provided valuable insight into the influence of implant geometry and fixation design on the long-term mechanical performance of total ankle arthroplasty.

8.2 Future Work

To further advance the biomechanical understanding of total ankle arthroplasty, several promising avenues for future research could be explored. First, transitioning from the current static framework to a comprehensive dynamic analysis covering the entire continuous gait cycle is highly recommended. Incorporating detailed representations of the complex capsular structures, ligaments, and active muscle forces through musculoskeletal multibody dynamics (MSK MBD) could provide a more physiological evaluation of the joint's dynamic loading environment [64, 76]. Furthermore, given the unique lateral transfibular surgical approach required for the Zimmer TM prosthesis, explicitly modeling the fibula and its associated metallic fixation hardware could be particularly insightful. Physiologically, the fibula is not merely a stabilizing structure but actively transmits approximately 6% to 17% of the total axial load during gait [24]. Including the fibula and its associated ligaments in future computational models could effectively redistribute the applied forces, possibly reducing the lateral tibial stress concentrations predicted in the current isolated tibia model. Regarding the bone-implant interface, future computational models could attempt to simulate the temporal biological process of osseointegration. Rather than assuming an instantaneously bonded solid interface, modeling the progressive ingrowth of host bone into the highly porous Trabecular Metal lattice could reveal how interface stiffness and load transfer mechanisms evolve over years of clinical service [95]. Additionally, to address the critical challenges of focal trabecular crushing and talar aseptic loosening, the latter being consistently identified in registries as the weakest mechanical link in TAR systems [49], future studies should consider employing topological optimization algorithms. Refining the implant's geometry to achieve a more evenly distributed contact pressure could possibly mitigate localized bone yielding. Direct numerical comparisons involving a broader range of contemporary prosthetic designs under identical boundary conditions would also assist in identifying the most favorable geometric features for long-term survivorship [95]. Finally, integrating direct clinical feedback from orthopedic surgeons regarding realistic intraoperative component positioning, alongside prospective clinical validations and access to manufacturer CAD models, could effectively bridge the gap between *in silico* predictions and surgical reality. In particular, the use of native implant geometries would help reduce uncertainties associated with geometric reconstruction, which may influence the local stress and strain magnitudes predicted by the model, ultimately guiding the development of safer and more durable ankle implants.

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Appendix A

Conference Abstract Submitted to 45th CNOT

As part of the scientific dissemination activities arising from this dissertation, preliminary findings were communicated through the submission of a scientific abstract to the *45th National Congress of Orthopaedics and Traumatology (CNOT)*, organized by the Portuguese Society of Orthopaedics and Traumatology (SPOT).

The text reproduced below corresponds to the original version of the abstract submitted to the conference.

Estudo Computacional Da Transferência De Carga No Tornozelo Com A Prótese Trabecular Metal Total Ankle™

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Introdução / Introduction

A artroplastia total do tornozelo tem vindo a afirmar-se como uma solução eficaz no tratamento da artrose avançada desta articulação, permitindo aliviar a dor e restaurar a funcionalidade do membro inferior. Entre as opções disponíveis, a prótese Trabecular Metal Ankle destaca-se pela sua abordagem lateral transfibular e tecnologia inovadora, baseada numa estrutura trabecular metálica que favorece a osteointegração e a estabilidade do implante. Este trabalho computacional teve como objetivo analisar os princípios biomecânicos, características estruturais e previsão do desempenho clínico desta artroplastia total do tornozelo.

Materiais e Métodos / Material and Methods

O modelo CAD desenvolvido foi implantado segundo o protocolo cirúrgico da prótese. A prótese é constituída por três componentes principais: os componentes tibial e talar, fabricados, respetivamente, numa liga de titânio e de cromo-cobalto, e o componente de desgaste em polietileno de ultra-alto peso molecular. A presença de uma estrutura porosa em tântalo favorece a osseointegração e contribui para a estabilidade da prótese. A simulação computacional foi realizada através da análise pelo método dos elementos finitos considerando uma carga em posição neutra. Assumiu-se a integração total entre a prótese e o tecido ósseo, bem como a existência de atrito na interface articular.

Resultados / Results

As tensões nos componentes metálicos mantiveram-se abaixo do limite de cedência elástica do material (79,4 MPa na componente tibial; 40,0 MPa na componente talar), concentrando-se nas cúpulas talaras, contorno lateral e rails de fixação. No tecido ósseo, as deformações trabeculares revelaram-se fisiologicamente normais ($\sim 1000 \mu\epsilon$). Contudo, o osso cortical peri-implantar na zona tibial (zona anterior, lateral e adjacente aos rails) mostrou picos extremos que vão para além do limiar crítico de microdano ($7961 \mu\epsilon$). No componente de polietileno de ultra-alto peso molecular, as tensões estão dentro da janela de segurança (tensão média de 2,5 MPa), embora se tenham registado picos de tensão ântero-laterais próximos do seu limite elástico (tensão máxima de 12,8 MPa; pressão de contacto de 10,2 MPa).

Discussão / Discussion

O modelo numérico preditivo confirma a estabilidade mecânica dos componentes protéticos, posicionando as estruturas de ancoragem (rails) e as cúpulas talaras como os principais vetores de dissipação de carga. Embora o padrão global de deformação no osso trabecular seja indutor de remodelação óssea positiva, os picos no osso cortical peri-implantar alertam para alguma concentração local de tensões sob carga estática (invariavelmente acumuladas nos rails, contorno lateral e bordo anterior). Ao constatar-se este risco de microrrotura localizada numa simples posição neutra, prevê-se que a dinâmica da marcha (flexão plantar e dorsiflexão) aumente o microdano na interface de fixação, predispondo o implante à micromobilidade e afundamento precoce e ao descolamento asséptico. De igual modo, ao notar-se que os picos ântero-laterais se aproximam do limite elástico do polietileno considerado, antecipa-se uma natural suscetibilidade à microdeformação e desgaste assimétrico a longo prazo.

Conclusão / Conclusion

O implante origina uma transferência de carga estruturalmente segura nos componentes metálicos e um estímulo mecânico adequado no osso trabecular. Contudo, os picos críticos de sobrecarga no osso cortical sob carga estática sugerem que as exigências dinâmicas da marcha agravarão o dano local, podendo levar à subsidência precoce e ao descolamento asséptico na prática clínica.

Appendix B

Zimmer TM Prosthetic Components

This appendix presents the technical drawings and principal dimensions of the prosthetic components considered in the finite element model. The dimensions were obtained from the reconstructed CAD geometries and are intended to provide a concise geometric characterization of the Zimmer Trabecular Metal™ Total Ankle System.

B.1 Technical Drawings

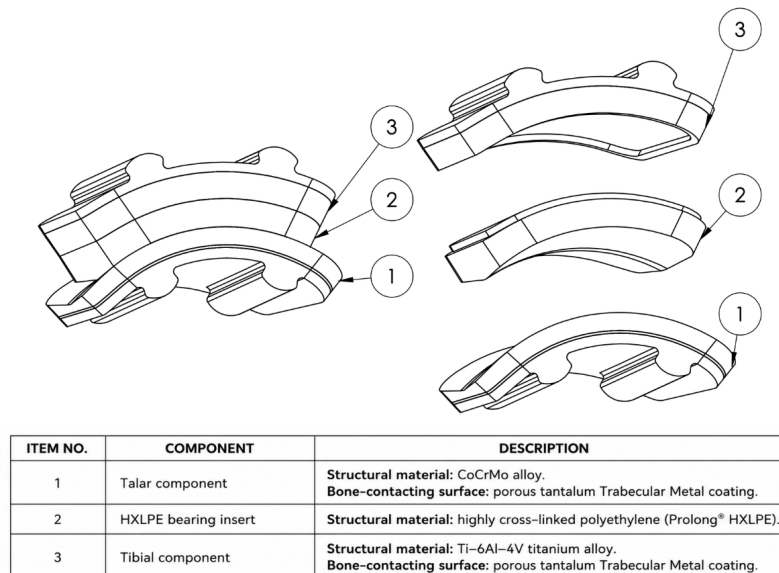


Figure B.1: Exploded view of the Zimmer Trabecular Metal™ Total Ankle System and corresponding component materials.

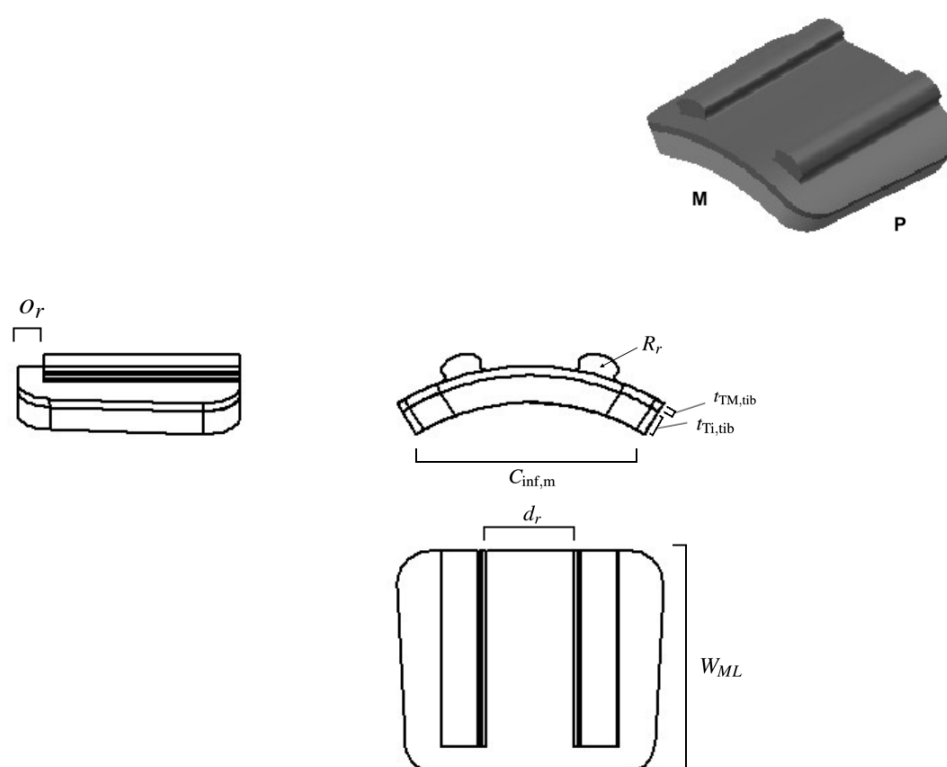


Figure B.2: Technical drawings of the tibial component.

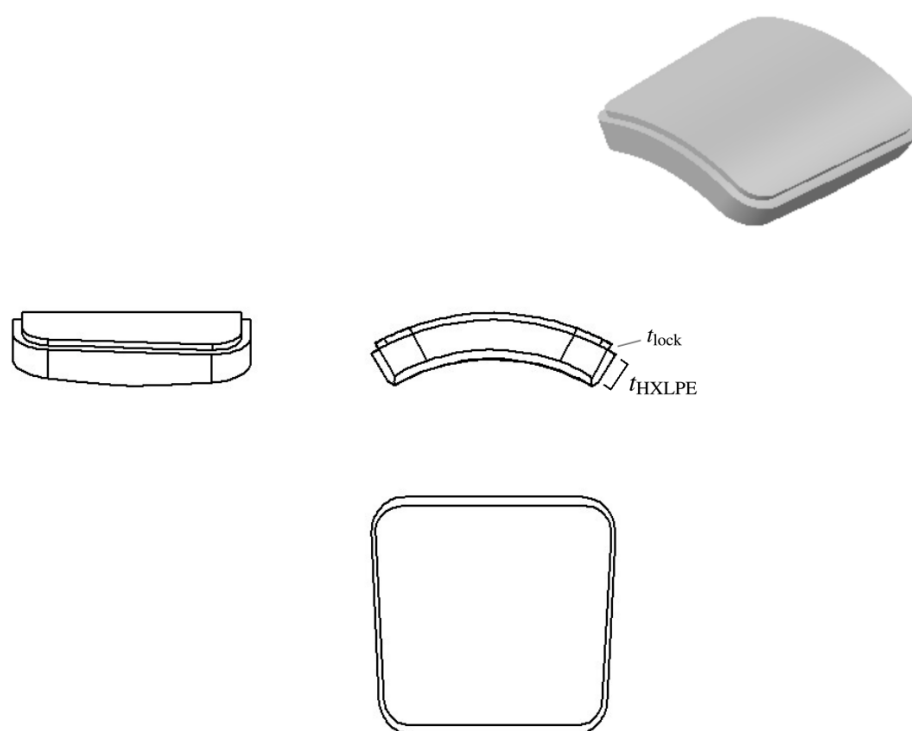


Figure B.3: Technical drawings of the HXLPE bearing insert.

B.2 Principal Dimensions of the Prosthetic Components

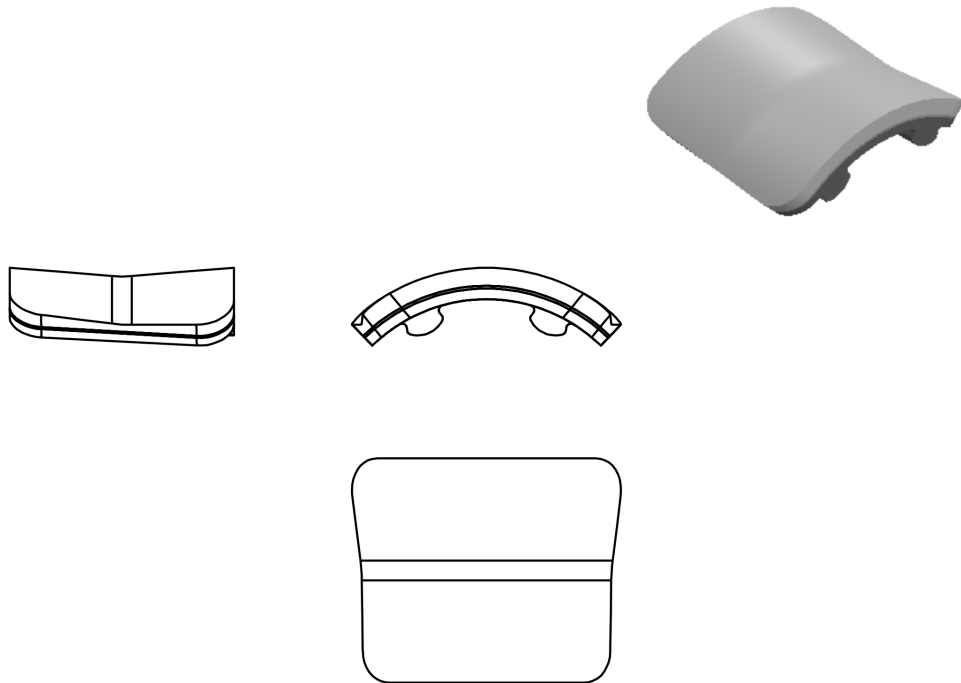


Figure B.4: Technical drawings of the talar component.

Table B.1: Principal dimensions of the Zimmer Trabecular Metal Total Ankle System components.

Parameter	Description	Value (mm)
<i>Common features</i>		
R_r	Rail radius	3,20
d_r	Distance between rail centres	9,80
o_r	Anterior and posterior rail offset	3,00
W_{ML}	Medial–lateral width	26,00
R	Radius of the articular surface	22,74
<i>Talar component</i>		
$C_{inf,m}$	Medial inferior chord length	23,94
$C_{inf,l}$	Lateral inferior chord length	27,00
$t_{Ti,tal}$	Titanium layer thickness	0,30
t_{CoCr}	CoCrMo layer thickness	2,10
$t_{TM,tal}$	Trabecular Metal coating thickness	1,25
<i>HXLPE bearing insert</i>		
$C_{inf,m}$	Medial inferior chord length	23,09
$C_{inf,l}$	Lateral inferior chord length	21,62
t_{HXLPE}	Insert thickness	3,30
t_{lock}	Superior locking feature thickness	0,88
<i>Tibial component</i>		
$C_{inf,m}$	Medial inferior chord length	24,68
$C_{inf,l}$	Lateral inferior chord length	27,18
$t_{Ti,tib}$	Titanium layer thickness	3,15
$t_{TM,tib}$	Trabecular Metal coating thickness	1,10