Stresses in the ankle joint and total ankle replacement design

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\textbf{A B S T R A C T}

The ankle is a highly congruent joint with a surface area of 11–13 cm\textsuperscript{2}. Total ankle replacements have been attempted since the early 1970s and design has continually evolved as the early designs were a failure. This was because the stresses involved and the mutiaxial motion of the ankle has not been understood until recently. It has been shown that the talus slides as well as rolls during the ankle arc of motion from plantarflexion to dorsiflexion. Furthermore, the articular surfaces and the calcaneofibular and tibiocalcaneal ligaments have been shown to form a four bar linkage dictating ankle motion. A new design ankle replacement has been suggested recently which allows multi-axial motion at the ankle while maintaining congruency throughout the arc of motion. The early results of this ankle replacement have been encouraging without any reported failures due to mechanical loosening.

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\textbf{1. Introduction}

The ankle joint acts as a link between the leg and the foot and plays a role in transferring load from the leg to the foot. The most common pathologies which can disrupt ankle function are osteoarthritis, rheumatoid arthritis and posttraumatic arthritis. Saltzman et al. [40] showed that posttraumatic arthritis was the most common and tends to occur more commonly in younger patients. The classical treatment for ankle arthritis has been arthrodesis but total ankle replacement has become popular over the last few decades.

The success of the total hip and knee replacements in the 1970s prompted attempts at total ankle replacement (TAR) [12]. The short and intermediate term results of the TARs were quite satisfactory, but the long-term results were unsatisfactory [48]. Hence there was a need to improve the performance as well as the survival rate of the existing TAR by improving upon design, constraint and function. In this article, the biomechanics of the ankle joint is analysed and current designs of total ankle replacement explored to determine if the ideal requirements for ankle replacement have been met.
2. The native ankle joint

2.1. Anatomy and stability

2.1.1. Bony configuration

The ankle consists of three bones (tibia, fibula and talus), collateral and syndesmotic ligaments and is a dynamic and highly congruent joint. The talar body articulates superiorly with the tibial plafond, medially with the medial malleolus and laterally with the lateral malleolus. Inman [24] showed that the talus is not a cylinder, but rather a section of a frustum of a cone, the apex of which is directed medially and thus has a smaller medial and larger lateral radius. The tibial plafond serves as the most superior aspect of the ankle joint and articulates with the dome of the talus. It is concave in the anteroposterior plane and elevated slightly on the medial side. This shape dictates an orientation of the distal tibial plafond that is slightly oblique from distal lateral to proximal medial [24].

2.1.2. Ligamentous configuration

The deltoid ligament stabilizes the ankle medially [10]. The contribution of the deltoid ligament to ankle joint contact has been reported by Earll et al. [13]. They report that deltoid ligament sectioning produces the greatest changes in both contact area size (decreased up to 43%) and peak pressure values (increased up to 30%) and this emphasizes the fundamental role in ankle mobility played by the deltoid ligament (Fig. 1).

The syndesmotic ligaments join the tibia to fibula and consist of an anterior and posterior tibiofibular ligament in addition to the transverse tibiofibular ligament and the interosseous ligament.

The anterior talofibular ligament (ATFL), calcaneofibular ligament (CFL), and posterior talofibular ligament (PTFL) all contribute to the lateral stability of the ankle (Fig. 2). The position of the ankle determines the tautness or looseness of the lateral ligaments. The PTFL is maximally stressed and CFL taut in dorsiflexion while ATFL is taut and PTFL, CFL loose in plantarflexion [11,39].

2.2. Biomechanics

2.2.1. Mechanical properties of bone and cartilage

The average thickness of the ankle cartilage is approximately 1.6 mm whereas the thickness of the knee cartilage is 6–8 mm [43]. The yield strain of human trabecular tibial bone is 1.6 mm whereas the thickness of the knee cartilage is 6–8 mm [21,36,39]. Biomechanics is taut and PTFL, CFL loose in plantarflexion [11,39]. PTFL is maximally stressed and CFL taut in dorsiflexion while ATFL determines the tautness or looseness of the lateral ligaments. The position of the ankle must be achieved. In particular, valgus malalignment should be corrected [20].

2.2.2. Axis and range of motion

The normal range of motion in the ankle ranges from 23° to 56° of plantarflexion, and from 13° to 33° of dorsiflexion [31,41]. Three distinct axes of the ankle joint have been reported during various motions based on the curvature of the talar trochlea [3] with the axis inclined upwards medially during Plantarflexion (PF) and upwards laterally during Dorsiflexion (DF) (Fig. 3).

Although the ankle joint axis at the neutral position is often regarded as a single axis, dorsiflexion–plantarflexion hinge, the axis orientation may vary [26]. The axis has been described as a changing axis or changing instant centers of rotation due to the shape of the talar trochlea and the action of soft tissues [5]. In cadaveric and gait studies, the rotation has been shown to range between 10° and 12° [44]. It is the varying center of rotation that allows the talus to glide and slide within the ankle mortise during PF and DF [4,33]. Also, the curvature of the talus and the distal tibia show varying radii [4,50] that allow horizontal rotations to occur in the foot or leg with movements of the ankle and thus transverse plane motion is coupled with sagittal plane motion [33,41]. The axial rotation of the talus with respect to the radius of the talus is reported to be between 6° and 12° [37,44]. Lundberg et al. [33] observed 8.9° of external rotation of the talus as the ankle moved from neutral position to 30° of dorsiflexion, whereas a small amount of internal...
rotation occurred with plantarflexion from neutral to $10^\circ$, followed by external rotation at terminal plantarflexion [33]. Michelson and Helgemo [35] reported that dorsiflexion resulted in an average of $7.2^\circ \pm 3.8^\circ$ of external rotation of the foot relative to the leg with ankle dorsiflexion, and $1.9^\circ \pm 4.12^\circ$ of internal rotation with plantarflexion.

Leardini et al. [27] developed a mathematical model to explain the multiaxial motion of the ankle in the sagittal plane. A four-bar linkage model was described (Fig. 4) showing the talus/calcaneus and tibia/fibula rotating about one another on inextensible line segments that represent the calcaneofibular (line AB in Fig. 4) and tibiocalcaneal (deltoid) (line DC in Fig. 4) ligaments without resistance. Motion between the polycentric, polyradial trochlea consisted of a combination of “rolling” and “sliding” motions. In this model, rotation was dictated by the most anterior fibers of the anterior talofibular and calcaneo-fibular ligaments. Leardini [28] also showed that these specific fiber bundles were isometric through the range of sagittal motion of the ankle. The instant centre of rotation (shown by “star” in Fig. 4) translated from a posteroinferior to a superoanterior position, which is consistent with several studies that suggest that the ankle is congruent and rotates about a transient centre of rotation [42].

2.2.3. Restraints of ankle motion

The tibiotalar articulating surface contributes to 70% of the antero-posterior stability, 50% of inversion/eversion stability and 30% of internal/external rotation stability [47]. The rest of the stability is provided by the ligaments. Renstrom et al. [39] found that during various motions of the ankle joint, the anterior talofibular and calcaneo-fibular ligaments were synergistic, that is when one ligament is strained the other one is relaxed and vice versa and thus providing stability.

2.2.4. Forces and stresses in the ankle

The ankle has a load bearing surface area of 11–13 cm$^2$ [45]. The tibiotalar area, however, accounts for only approximately 7 cm$^2$ [45] while Calhoun et al. [8] found that during weight bearing 77–90% of the load is transmitted through the tibial plafond to the talar dome. With an interface area of 7 cm$^2$, the average compressive load per unit area at the interface during gait would be approximately 3.5 MPa in a patient of 700 N body weight [20]. A vertical load on the ankle of 5.2 times body weight was found during gait [45]. The peak resultant force acting at the ankle joint during the stance phase during running was 9.0–13.3 BW [6]. Landing from a jump generated 2–12 BW [34] while heel-toe running at 4.5 m/s generated a force of 2.8 BW [9,15].

In diseased ankles, the joint load decreased to approximately three times body weight; the same values were noted in replaced ankles [45] and anteroposterior and lateral shear forces during gait were estimated to reach levels of two and three times body weight, respectively. The vertical load that is transmitted to the trabecular bone at the prosthesis-bone interface may exceed the inherent trabecular bone strength in normal daily activities [20].

Having knowledge of the normal ankle joint, a discussion can now be carried about the various Total Ankle Replacement designs.

![Fig. 3. Axes of ankle motion.](image)

![Fig. 4. Four bar linkage model developed by Leardini et al. [27].](image)

![Fig. 5. First generation ankle replacement, which consisted of all polyethylene tibial component and metallic talar component (similar to hip replacement systems) fixed with cement.](image)
3. First generation ankle replacements

Around twenty three total ankle replacements were developed during early 1970s [2]. The first generation TARs were of two types namely constrained and unconstrained. Constrained prostheses limited motion to the sagittal plane. They were typically spherical, spheroidal, conical, cylindrical or sliding cylindrical in shape [31]. They provided greater stability [50], minimized impingement of the malleoli against the talus [12], and decreased wear of the polyethylene insert due to larger contacting surfaces. The only disadvantage of the prostheses was increased stresses at the bone cement implant interfaces leading to early failure. Examples of this prostheses type were Imperial College London Hospital (ICLH), Conaxial, St. George/Buchholz, Thompson Parkridge Richards (TPR), TNK and Mayo designs. Unconstrained prostheses were typically trochlear, bispherical, concaveconvex, and, convexconvex in shape [31]. They provided improved range of motion but stability was greatly reduced. They also minimized the bone cement implant interface forces. Examples of unconstrained prostheses were Bath and Wessex, Irvine, Smith and Newton implants. Both the constrained and unconstrained prostheses were cement fixed (Fig. 5) during the 1970s [50]. The results of unconstrained prostheses proved to be a total failure due to instability and impingement [14].

4. Second generation total ankle replacements

The second generation prostheses were developed to overcome the disadvantages and complications associated with the first generation prostheses [46]. These implants were intended to reproduce the anatomical characteristics of the ankle joint, joint kinematics, ligament stability and mechanical alignment. The sliding and rotational motions of the ankle joint were achieved by the two or three component designs [Fig. 6] [18].

*Agility ankle* was the prototype of the two component semiconstrained design. It replaced the medial, lateral and superior articulating surfaces of the ankle joint as fusion of the distal tibiofibular syndesmosis was carried out to stabilize the tibial component. The articular surface of the tibial component was larger than the surface of the talar component. The incongruency between the talar and tibial articulations theoretically allowed sliding as well as the rotational motions that supposedly mimic ankle kinematics [40] (Fig. 7).

It was not successful as this semiconstrained design does not replicate normal ankle motion, as the ankle ‘slides’ from side to side during rotation and dorsi- and plantarflexion motion. This raised concerns regarding increased wear debris production and early loosening [18]. Furthermore, a non-union of the attempted tibio-fibular arthrodesis, risks loss of fixation on the proximal side, thus compromising stability [17].

Examples of three component total ankle replacements are: *Scandinavian total ankle replacement (STAR)* has a convex talar component with a longitudinal ridge, which is congruent with the distal surface of the mobile meniscus. Dorsi- and plantarflexion at the menisco-tibial interface, but no talar tilt, are allowed while rotation is allowed at the (flat) menisco-tibial interface [20]. This presents a lower surface area for stress distribution in the distal tibia compared to some other ankle replacement designs but might also reduce stress shielding from a stem [17] (Fig. 8).

The concerns regarding the small dimension of the tibial component and the lack of circumferential bone support makes the tibial component prone to subsidence (sinking) in the distal tibia cancellous bone. Furthermore, motion only in one axis, due to the non-anatomic cylindrical shape of the talus, may produce load transfer to the medial side and overstress the medial ligaments and capsule [20]. Valderabano et al. [49] reported tilting of the tibial component in 9 of 68 ankles (13%) within the first 3 months, without progression thereafter. This can be the result of shear forces and imperfect apposition of tibial prosthesis and bone.

*Rames total ankle replacement*: required wide talar bone resection therefore subsidence was a problem and frontal plane stability relies entirely on the medial and lateral ligamentous structures. This does not correspond to normal ankle biomechanics, where frontal plane stability relies mainly on joint congruency [20].

*Hintegra ankle replacement*: it consists of a flat tibial component, a polyethylene inlay and a convex conic talar component with a smaller medial radius. Both the talar and tibial components have ventral shields for screw fixation [20] (Fig. 9).

The disadvantages are that it uses screws to obtain fixation in tibia which can loosen before bony ingrowth has occurred due to micromotion and inversion–eversion and rotational motion is not allowed, possibly causing higher stresses at the bone–implant interface [18].

*Buechel Pappas low contact stress (LCS) TAR*: its upper surface is flat, whereas its lower surface conforms to the trochlear surface, thereby providing unconstrained, sliding cylindrical motion with
LCS on the bearing surfaces, allowing inversion, eversion motion [5] (Fig. 10).

The main concern regarding the Buechel Pappas implant and all implants having a tibial stem (SALTO, AES, Mobility) to facilitate fixation is weakening of the anterior tibial cortex, as an anterior cortical window has to be produced for insertion of the component. Load transmission in the weakened anterior cortex in the supramalleolar area is a concern as the removal of the subchondral plate lowers the compressive resistance by 30–50%, and with sectioning of the subchondral bone 1 cm proximal to the subchondral plate, by 70–90% [32] and therefore early failures are a risk of this prosthesis. Furthermore, stress shielding is possible due to the stem as compared to other designs without stems [17].

But, a broad comparison of their outcomes reveals that the results of most of the existing total ankle replacement models are unsatisfactory because of the inability to restore adequately the critical stabilizing role of the ligaments, poor reproduction of the normal mechanics of the ankle joint and the lack of involvement of the underlying subtalar joint in the coupled pattern of motion of the entire ankle complex [19,16]. In order to overcome the unsatisfactory outcomes of the existing TAR models, a novel TAR model was required which would ideally not have a tibial stem and mimic the mutiaxial motion of ankle while maintaining congruency throughout the range of motion.

Leardini et al. [27,28] had developed the four bar linkage system and using computer animation demonstrated that rolling as well as sliding occurred in ankle motion, that is the talus slid forward on the tibial mortise while rolling backwards during plantarflexion and vice versa in dorsiflexion. From the model it was deduced that the shape of the articulation surface compatible with the ligament rotation during various ankle motions was an arc of a circle which is polycentric and polyradial in nature. The demonstrated mutual action of the articular surfaces and the ligaments in the intact ankle joint was used to enhance the total ankle replacement design with the help of computer aided design [28].

A new design convex-tibia fully congruent three-component prosthesis was eventually preferred among all the possible pairs of articular surfaces tested by Leardini and Moschell [29]. This work also pointed out that the flat-tibia plus natural-talus designs [5,25] were unlikely to restore the original characteristic pattern of ligament tensioning and thus likely to fail because of abnormal stresses.

As a result the BOX (Bologna Oxford (BOX)) total ankle replacement was developed as shown in Fig. 11. This Ankle replacement has a convex tibial component with bars to prevent

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Fig. 8. The STAR prosthesis a three component design but no inversion/eversion is allowed.

Fig. 9. Hintegra uses screw fixation which can loosen before bony ingrowth occurs.

Fig. 10. The tibial stem and the deep sulcus of the talar component accommodating a matching polyethylene surface, allowing inversion/eversion motion, are characteristic features of the Buechel–Pappas ankle replacement.

Fig. 11. The BOX ankle replacement (figure from [1]).
stress shielding, which articulates with a biconcave meniscal bearing with a groove on the talar component to provide mediolateral stability. The shapes of the three components are compatible with physiologic ankle mobility and with the natural role of the ligaments. This prosthesis maintains full congruence at the articulating surfaces of the meniscal bearing over the entire motion arc (Fig. 12). There have been pilot studies done and initial results have shown no failures due to mechanical loosening [30]. Ingrosso et al. [23] have shown that the BOX prosthesis seems to contribute to an early functional recovery at 6 months which is maintained at one year. With reduction of pain and recovery of joint control and gait, variables of high clinical interest, such as stance balance and ability in propulsion, improve considerably.

It remains to be seen whether the BOX meets the ideal requirements for a total ankle replacement and becomes successful. The search for the ideal ankle replacement continues.

References