Finite element analysis of biomechanical effects of total ankle arthroplasty on the foot

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Abstract Background: Total ankle arthroplasty is gaining popularity as an alternation to ankle arthrodesis for end-stage ankle arthritis. Owing to the complex anatomical characteristics of the ankle joint, total ankle arthroplasty has higher failure rates. Biomechanical exploration of the effects of total ankle arthroplasty on the foot and ankle is imperative for the precaution of postoperative complications. The objectives of this study are (1) to investigate the biomechanical differences of the foot and ankle between the foot with total ankle arthroplasty and the intact foot and (2) to investigate the performance of the three-component ankle prosthesis.

Methods: To understand the loading environment of the inner foot, comprehensive finite element models of an intact foot and a foot with total ankle arthroplasty were developed to simulate the stance phase of gait. Motion analysis on the model subject was conducted to obtain the boundary and loading conditions. The model was validated through comparison of plantar pressure and joint contact pressure at the talonavicular joint was measured in a cadaver foot.

Results: Plantar pressure, stress distribution in bones and implants and joint contact loading in the two models were compared, and motion of the prosthesis was analysed. Compared with the intact foot model, averaged contact pressure at the medial cuneonavicular joint increased

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Introduction

Total ankle arthroplasty (TAA) is gaining popularity due to the concept that it can provide more functional movements than ankle arthrodesis for reconstruction of degenerative ankles with end-stage arthritis. However, clinical reports have indicated a wide range of iatrogenic complications and a low success rate [1,2] in TAA surgeries. Failure rates were reported to range from 10% to 20% within 10 years after surgery [3–10]. Some failures required conversion to ankle arthrodesis [11], and in extreme cases, led to amputation [4].

Surgical failures may be a result of the fact that prostheses cannot totally resemble natural human ankles, which have complex anatomical structures, sophisticated kinematics and intimate interactions and stabilization mechanisms. Sufficient understanding of the biomechanics of TAA is imperative. Previous biomechanical investigations, including gait analyses [12–19], cadaveric experiments [17,20,21] and radiographic observations, provided useful, but insufficient, exploration of the inner foot. Computational methods are featured to provide insight into human bodies and have been widely used in biomechanical observations.

Finite element (FE) models of TAA have been developed and used to investigate the contact pressure and kinematics of the implants during gait [22]; to evaluate the effects of alignment of prosthesis components [23]; to postulate the bone-remodeling process after TAA [24]; to identify the failure mechanism of the polyethylene component [25] and to further investigate other clinical issues under physiological loading conditions [26]. Precisely, an FE model of TAA with the foot–ankle complex was used to investigate plantar pressure and stress distribution in bones in balanced standing [27]. Models constructed in these studies were based on partial foot segments, which were insufficient for observing the biomechanics of the entire foot and ankle. In this study, FE models of an intact foot and a foot with TAA were developed to (1) evaluate the influence of TAA on the foot biomechanics in terms of plantar pressure, joint contact pressure, bone stress distribution and force transmission and (2) investigate the motion and loading distribution of the ankle prosthesis.

Materials and methods

Ethical approval for this project was granted by The Hong Kong Polytechnic University Human Subject Ethics Sub-committee (reference number HSEARS20070115001). The participant who participated in the gait experiment was informed of the experimental procedures and gave written informed consent for the participation in the magnetic resonance image (MRI) scanning, gait experiment and for publishing the case details without disclosing the participant’s identity.

Development of finite element models

An FE model of the intact foot [28,29] involving 28 bones, 103 ligaments, plantar fascia, nine groups of extrinsic muscles and a bulk of encapsulated soft tissue was developed (Fig. 1). A female participant, aged 29 years with a body mass of 54 kg and a height of 165 cm, was recruited to acquire the MRI (2 mm slice interval, 3.0T, Siemens, Erlangen, Germany) of the right foot. She claimed to have no history of lower limb injuries or pathologies. Geometries of 28 bones and foot surface were reconstructed from the MRI using Mimics software (Materialise, Leuven, Belgium) and further edited into an FE model using Abaqus software (Dassault Systèmes Simulia Corp., Providence, RI, USA). The interphalangeal joints of the four lesser toes were simplified as a connection using a 2-mm thick soft layer, while other articulations were defined as frictionless surface-to-surface contact with nonlinear contact properties. Ligaments were constructed using tension-only truss elements,
and muscles were represented using axial connector elements that allowed force application.

To obtain the FE model of TAA, the ankle joint in the FE model of the intact foot was resected by the prosthetic ankle, Scandinavian Total Ankle Replacement (STAR, Waldemar Link, Hamburg) prosthesis. Three implant components including a tibia plate, a talar component and a mobile bearing were digitalized and aligned in corresponding positions in the ankle joint based on the guidelines of the surgery, cutting the overlapping ankle bones using Boolean operation, and were fixed to the interface. The bearing component slid between the tibia plate and the talar component with a coefficient of friction of 0.07 [30]. Meshes and material properties of foot segments and implant structures are listed in Table 1.

**Boundary and loading conditions**

Boundary and loading conditions were obtained from gait analysis of the model subject using Vicon system (Oxford Metrics, Oxford, UK). Reflective markers were attached to the lower limb defining seven body segments to record the motion of the segments and the connecting joints. The participant walked at her natural speed for eight trials, with the feet stepping on individual force platforms (AMTI, Advanced Mechanical Technology, Inc., Watertown, MA, USA) in each trial to record ground reaction forces. Muscle forces were calculated based on electromyography (EMG) signal [39] and muscle cross-section area [40]. Plantar pressure was measured using F-scan (Tekscan Inc., Boston, MA, USA) during the gait analysis for model validation.

Fig. 2 shows ground reaction forces in vertical, medio-lateral and anteroposterior directions and the averaged foot-shank angle. Two peaks and a valley occurred in the curve of the vertical ground reaction force, representing the maximum force impacted on the hind and forefoot and full body weight supporting during single-foot support, respectively. These three characteristic instants, namely first-peak (17.5% of the stance phase), mid-stance (48% of the stance phase) and second-peak (76% of the stance phase), were chosen for simulation. To ensure TAA surgery as the exclusive factor in the simulation, the same boundary and loading conditions were applied to the models of both the TAA foot (Fig. 1) and the intact foot.

**Table 1**  Element type and material properties for segments of the finite element models.

| Component                  | Element type                  | Young’s modulus $E$ (MPa) | Poisson’s ratio, $v$ | Cross-section area (mm$^2$) |
|----------------------------|-------------------------------|---------------------------|----------------------|----------------------------|
| Tibia/talar implants [31]  | 4-node linear tetrahedron     | 116,000                   | 0.32                 | —                          |
| Mobile bearing implant [32]| 4-node linear tetrahedron     | 8100                      | 0.46                 | —                          |
| Bone [33,34]               | 4-node linear tetrahedron     | 7300                      | 0.3                   | —                          |
| Cartilage [35]             | 4-node linear tetrahedron     | 1                         | 0.4                   | —                          |
| Ligaments [36]             | 2-node linear 3-D truss       | 260                       | —                     | 18.4                       |
| Plantar fascia [37]        | 2-node linear 3-D truss       | 350                       | —                     | 58.6                       |
| Ground                    | 8-node linear brick           | 17,000                    | 0.1                   | —                          |
| Encapsulated soft tissue [38]| 4-node linear tetrahedron  | $C_{10}$: 0.085 $C_{01}$: -0.058 $C_{20}$: 0.039 $C_{11}$: -0.023 $C_{02}$: 0.009 $D_1$: 3.652 $D_2$: 0.000 | — | — |
Model validation

The computational model was validated by plantar pressure measurement and cadaveric experiments. Plantar pressure was measured during the gait trials. A pressure sensor was attached to the plantar foot using double-sided tape. The participant stood upright with the feet apart by a shoulder width for 5 seconds to record the plantar pressure distribution in balanced standing. Plantar pressure distribution during stance phase was recorded during the participant walking. FE-predicted plantar pressure was obtained through the application of boundary and loading conditions to the finite element foot model. Fig. 3 shows the comparison between the two measurements during balanced standing, and at the first-peak and the second-peak instants during gait. There was no observable variation of peak pressure location between the two measurements. In balanced standing, the peak pressure of the forefoot located beneath the heads of the second and third metatarsals and that in the hindfoot located beneath the heel. In the first-peak instant, it located beneath the heel and transferred to the forefoot beneath the first to third metatarsal heads. Deviation of the averaged pressure of the forefoot in balanced standing was less than 15%, and in other cases, it was less than 10%.

A fresh cadaveric right foot and ankle complex was adopted for a mechanical test. The foot and ankle was fixed on a tensile testing machine at the resected end of the distal tibia and fibula bones and was supported by a rotatable plate on the plantar foot. Body weight was applied through a compression force from the tension machine. Muscle forces were applied by adding weight to the corresponding tendons. A K-scan (Tekscan Inc) sensor was inserted into the talonavicular joint to measure the joint contact pressure. The cadaver foot was applied with the same loading condition as in FE simulation. The deviation between the experimental measurement and computational prediction was less than 5%.

Results

TAA increased the contact pressure at the medial cuneonavicular joint and bone stress (maximum von Mises stress) in the second and third metatarsals. Forces that transmitted in the medial aspect of the foot were also increased; however, the peak plantar pressure decreased.

The joint motion of the ankle prosthesis in the frontal plane was constrained, which induced a limited range of motion in the sagittal plane. Asymmetric load was predicted such that the lateral aspect of the implants sustained much higher stress than the medial.

Plantar pressure

Fig. 4 shows the plantar pressure distribution in the intact foot model and TAA surgical model. The peak plantar pressure at the first-peak, mid-stance and second-peak instants was 0.260 MPa, 0.553 MPa and 0.605 MPa, respectively, in the TAA foot, which were 21.7%, 19.0% and 11.4%
Figure 3  Comparison of plantar pressure between finite element prediction and gait analysis measurement for validation. CPRESS, contact pressure distribution.

Figure 4  Comparison of the plantar pressure distributions between models of the intact foot and total ankle arthroplasty foot at the first-peak, mid-stance and second-peak instants. COP, center of pressure.
lower than that in the intact foot model. In both anteroposterior and mediolateral directions, the translation of the location of the center of pressure was less than 8 mm at three instants.

**Joint contact pressure**

Averaged contact pressures in 11 joints in each model were compared, as shown in Fig. 5. The highest pressure occurred in the medial cuneonavicular joint of the TAA foot at the second-peak instant, reaching 3.17 MPa, which was 67.4% higher than that of the intact foot.

The averaged contact pressure at the subtalar joint was 0.35 MPa, 0.52 MPa and 0.72 MPa, respectively, at the three instants in the TAA foot. It decreased by 11%, 3.5% and 7.7% compared with that of the intact foot. In the articulations of the hindfoot and midfoot, consisting of the calcaneocuboid and talonavicular joints, the increase of pressure at the talonavicular joint was less than 1% at the first-peak and midstance instants; however, it was 20.5% higher in the TAA foot than that in the intact foot at the second-peak instant (2.00 MPa vs. 2.41 MPa). In the five tarsometatarsal joints, connecting the midfoot and forefoot, the TAA foot sustained 44.0% higher pressure than the intact foot at the second-peak instant except for the fifth tarsometatarsal joint.

**Force transmission**

In the articulations between the hindfoot and midfoot in the TAA model, the majority of forces transmitted through the talonavicular joint, which were 177 N, 285 N and 618 N, respectively, at the three instants. They were 2.4% lower and 10.1% and 20.3% higher than the corresponding values in the intact foot at the three instants, respectively. The forces transmitted from the midfoot to the forefoot mainly through the first three tarsometatarsal joints, among which the first tarsometatarsal joint sustained larger contact forces in the TAA foot model than in the model of the intact foot. The contact force in this joint was 39 N, 70 N and 236 N, respectively, at the three instants, which was 36.9% and 25.0% lower at the first-peak and mid-stance instants and 18.3% higher at the second-peak instant than that in the intact foot model.

Among the three cuneonavicular joints in the midfoot of the TAA model, the medial one supported the largest contact force throughout the gait. The values of the force were 33 N, 71 N and 249 N, respectively, at the three instants and 45.2% and 25.9% lower and 33.9% higher than those of the intact foot, respectively. In the hindfoot of the TAA model, contact force in the subtalar joint was 134 N, 201 N and 304 N, respectively, at the three instants. It was 18.0%,
13.3% and 18.5% lower than the corresponding values in the intact foot model, respectively, at the three instants.

Fig. 6 shows the force transmission (in terms of times of body weight) from the hindfoot to the forefoot. Forces transmitted through the lateral path did not vary apparently between the two foot models, whereas more obvious changes occurred in the medial path at the mid-stance and second-peak instants. In total, forces that were 0.48 and 0.95 times the body weight transmitted through the medial way in the intact foot, respectively, at the mid-stance and second-peak instants, and this increased to 0.53 and 1.15 times the body weight in the TAA foot.

Stress in metatarsal bones

The maximum von Mises stress (Fig. 7) was located in the second metatarsal, and was 20.4 MPa, 30.6 MPa and 55.3 MPa, respectively, at three instants in the TAA foot. Compared with the model of the intact foot, the variation was subtle at the first-peak instant but increased by 19.8% and 31.2%, at the mid-stance and second-peak instants, respectively. Another notable deviation between the foot models was that the first metatarsal sustained higher stress in the TAA foot than in the intact foot at the second-peak instant.

Prosthetic joint motion and loading

Motions of the prosthetic ankle were investigated (Fig. 8). In the sagittal plane, it rotated by 2.5°, 4° and 3.5°, respectively, at the first-peak, mid-stance and the second-peak instants. Rotation did occur in the transverse plane but restricted in the frontal plane. Asymmetric loading was exerted in the prosthetic joints, such that the lateral side of the bearing component sustained higher stress than the medial side. The stress at the lateral side of the bearing component was 17.2 MPa, 25.4 MPa and 67.6 MPa at the three instants, respectively.

Figure 6  Comparison of force transmission in models of the intact foot and total ankle arthroplasty foot at the first-peak, mid-stance and second-peak instants. Force is depicted in terms of times of body weight. Black arrows are for intact foot, and blue ones are for total ankle arthroplasty foot.
Figure 7  Comparison of von Mises stress distribution in metatarsals between models of the intact foot and total ankle arthroplasty foot at the first-peak, mid-stance and second-peak instants.

Figure 8  Motion of the prosthetic ankle joint in sagittal planes and the stress distribution in the bearing component.
Discussion

The emergence of TAA allows the retention of the ankle joint for patients with end-stage ankle arthritis, but its complications may hinder the advantages of this intervention. In this study, the biomechanical effects of this surgery on the foot were evaluated based on validated computational models of the foot and ankle, and the joint motion and loading distribution of the ankle prosthesis were also investigated. This surgery technique has predicted and demonstrated alteration of the joint contact pressure, bone stress and force transmission patterns of the foot. Asymmetric loading was induced in the implants.

The three-component design of the ankle prosthesis allows joint motion in the sagittal and transverse planes, while inversion/eversion motion in the frontal plane is structurally constrained. As found in this study, transition and internal/external rotation in the transverse plane occurred, but the range of motion was smaller than that in an intact ankle [41]. The prosthetic ankle is theoretically capable to cover the full range of motion in the sagittal plane as found in an intact foot. However, this rotation in this study was found to be smaller than in the intact foot, which may limit the motion of dorsiflexion due to the coupling effect. This finding verified results in gait analysis studies that dorsiflexion was reduced in TAA patients [11–13,42], which was clinically compensated by Achilles tendon lengthening [11].

Eversion moment at the ankle joint existed during most part of the stance phase [43]; Resistance of this motion resulted in mediolateral asymmetric loading distribution in the bearing component, such that the lateral side sustained much higher loading than the medial side. The asymmetric loading might be a potential indication of fracture of the bearing component [44] and/or talus subsidence or migration [45–47]. Optimization of prosthetic ankle designs to permit rotation in the frontal plane could be a fundamental solution for insufficient dorsiflexion.

Plantar pressure is employed in clinical practice and rehabilitation for the identification of plantar foot disorders. The plantar pressure distribution was found not to be affected by TAA surgery. This was consistent with the results of other biomechanical studies [48]. The anteromedial displacement of the center of pressure (COP) can possibly be interpreted as a consequence of variations in the force transmission pattern.

Contact pressure at the talonavicular and the medial cuneonavicular joints increased in the TAA model at the second-peak instant and were higher than that at other joints. Excessive contact pressure at articular interfaces of joints was believed to be a predominant factor of osteoarthritis [49,50]. These two joints might have a potential risk of degeneration, but until now, no clinical reports have clearly pointed out osteoarthritis at these joints.

More forces transmitted medially in the TAA foot at the mid-stance and second-peak instants, which could explain the phenomenon of increased stress in the medial metatarsals.

The second and third metatarsal bones are most commonly affected by stress fractures, and the fracture of the second metatarsal is one of the most common complaints after foot and ankle surgeries [51]. In this study, the two bones bore much higher von Mises stress than the other metatarsal bones. Although the first metatarsal sustained higher stress in the TAA foot than in the intact foot at the second-peak instant, this stress was much lower than that in the second metatarsal.

This study had several limitations. First, computational models were based on simplifications and assumptions. Bones of the FE model were reconstructed without separation of cortical and trabecular components and were assigned as homogeneous, isotropic and linear elastic material. Second, boundary and loading conditions applied to TAA foot were same as those of the intact foot. Considering these limitations, results from this study were expected to qualitatively analyse the biomechanical effects of TAA from a theoretical perspective, rather than an exact representation of this surgery. Further studies should include motion analysis on TAA patients and application to FE simulations to improve this condition.

Total ankle arthroplasty induced increased loadings in the medial cuneonavicular joint and the second and third metatarsals and forces that transmitted from the first ray. These findings have implications for more extensive attention to patients with foot problems in these regions. The ankle prosthesis bore asymmetric loading, such that the stress on the lateral aspect was much higher than the medial. Prosthesis optimization in terms of joint motion in the frontal plane might be beneficial for a more accurate representation of human ankle joint. All these findings should be further validated by clinical evidence.

Conflicts of interest

All authors declare that they have no conflict of interest.

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