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Biomechanical Study of Tarsometatarsal Joint Fusion Using Finite Element Analysis

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Abstract

Complications of surgeries in foot and ankle bring patients with severe sufferings. Sufficient understanding of the internal biomechanical information such as stress distribution, contact pressure, and deformation is critical to estimate the effectiveness of surgical treatments and avoid complications. Foot and ankle is an intricate and synergetic system, and localized intervention may alter the functions to the adjacent components. The aim of this study was to estimate biomechanical effects of the TMT joint fusion using comprehensive finite element (FE) analysis. A foot and ankle model consists of 28 bones, 72 ligaments, and plantar fascia with soft tissues embracing all the segments. Kinematic information and ground reaction force during gait were obtained from motion analysis. Three gait instants namely the first peak, second peak and mid-stance were simulated in a normal foot and a foot with TMT joint fusion. It was found that contact pressure on plantar foot increased by 0.42%, 19% and 37% respectively after TMT fusion compared with normal foot walking. Navico-cuneiform and fifth meta-cuboid joints sustained 27% and 40% increase in contact pressure at second peak, implying potential risk of joint problems such as arthritis. Von Mises stress in the second metatarsal bone increased by 22% at midstance, making it susceptible to stress fracture. This study provides biomechanical information for understanding the possible consequences of TMT joint fusion.

Keywords:

Finite element analysis; Tarsometatarsal joints; Biomechanics; Contact pressure; Arthritis.

Introduction

Tarsometatarsal (TMT) joints comprise the base of five metatarsal bones and their articulation with three cuneiforms and cuboid bone. Reports showed that 0.2% of fractures and dislocations were sustained over this region [1-3]. The actual incidence maybe higher than reported because such injuries were often unreported [3-6]. Delayed treatment of injuries may lead to significant complications [7] and are eventually recommended to surgeries [8]. Open reduction and internal fixation are one of reliable means of securing and maintaining reductions of TMT joint factures and dislocations [1, 9-11]. However, outcomes of the operation may not always be positive. Because foot and ankle is an intricate and synergetic system and individual segments interacts interdependently, clinical modifications may have profound impact on its biomechanical functions. Long-term side effect of TMT joint fusion, one of the most common internal fixation treatments, are clinically reported as posttraumatic arthritis, flatfoot deformity and instability [10, 12, 13]

To avoid the occurrence of these complications, biomechanical understanding of effect of surgeries in foot and ankle is of great important. While it is not easy to obtain biomechanical information by experimental methods, computational approaches such as finite element (FE) analysis offer a feasible alternative. FE analyses can simulate *in vivo* conditions with complex geometries, material properties and boundary and loading conditions and offer insight into internal information including stress distribution, contact pressure, and deformation. FE methods have been used to assist in surgical decision in foot and ankle. An FE model comprised of major musculoskeletal structures without embracing soft tissue was developed to investigate the effect of plantar ligaments release on human [14] and indicated that the surgery may relieve focal stress associated with heel pain. To further understand injury mechanism in ankle and subtalar joints under impact loading, a simplified FE model with fixed fore- and mid-foot and ankle was developed [15]. An FE study was carried out to investigate the effect of foot postures on bone healing after surgery through analysis of peak strains in the fifth metatarsal [16]. The same loading conditions were applied for different foot postures. It was concluded that foot postures did not significantly influence the peak strain at fracture site but eversion of foot was more risky.

The aforementioned studies offer insight into the biomechanical environment, but over simplifications and limitations in terms of modeling of geometry, application of boundary and loading conditions, may affect accuracy of analysis. Moreover, there is limited biomechanical research in TMT joint fusion. A more accurate FE model of foot and ankle for TMT joint fusion is necessary. In this study, simulation of TMT joint fusion was carried out based on a comprehensive 3D FE model of foot and ankle to give a better understanding of its biomechanical performance.

Methods

An FE model of foot and ankle was developed using ABAQUS FE package (Dassault Systems Simulia Corp., Providence, USA). The geometrical information was obtained from 3D reconstruction using MIMICS v14 (Materialise, Leuven, Belgium) of magnetic resonance images (3.0T Siemens, Erlangen, Germany) of 2 mm intervals from the right foot of a normal female adult with the body height of 164 cm and mass of 54 kg. The model, as shown in Fig. 1, consists of 28 bony segments, 72 ligaments and plantar fascia, embraced by bulk soft tissue [17, 18]. The bony segments were modeled as separated individuals and interactions were set as surface to surface contact.

Fig. 1

Fig. 1. Finite element model of foot and ankle, consisting of 28 bones, 72 ligaments, plantar fascia and encapsulated soft tissue. Part of the soft tissue was removed for a better view of bone structures.

Solids elements were used to mesh foot and ankle structures. The bony and encapsulated soft tissue structures were meshed into 4-noded tetrahedral elements. As the ligaments were assumed to sustain tensile force only, they were represented by truss elements. Truss elements can only transmit force along the axis or the centre line of the element, and cannot resist loading perpendicular to the axis. No compressive stress was generated by choosing tension-only option. The distance between the two connecting nodes defines the length of each truss element and the cross-sectional area is specified. In this FE model, total 98 tension-only truss elements were used to represent the ligaments and the plantar fascia. Muscles in the FE model were represented by lines connecting the anatomical attachment points of muscles to bones, and external force can be applied. Achilles tendon was divided into five axial connector elements.

A number of material property models can be used, from the simple linear elasticity, to nonlinear elasticity, and even viscoelasticity. To reduce the complexity and the size of the problem, except for the encapsulated soft tissue, all tissues including bony, ligamentous and cartilaginous structures were idealized as homogeneous, isotropic and linearly elastic. The linearly elastic properties can be defined by providing any two constants of Young's modulus E, shear modulus G, and Poisson's ratio ν. The Young's modulus and Poisson's ratio were selected for bony structures and were assigned as 7300 MPa and 0.3, respectively [19]. These parameters were obtained by averaging the elasticity values of cortical and trabecular bones in terms of their volumetric contribution. The Young's modulii of the cartilage [20], ligaments [21] and the plantar fascia [22] were selected from literatures. The cartilage was assigned with a Poisson's ratio of 0.4. The ligaments and the plantar fascia were assumed to be incompressible.

To simulate barefoot stance, a horizontal plate consisting of an upper concrete layer and a rigid bottom layer was used to establish the foot-ground interface. The upper layer was set as linearly elastic property to represent ground, and the rigid bottom to facilitate applying boundary and loading conditions during gait. The horizontal ground support was meshed with hexahedral elements. The foot-ground interaction was simulated as frictional contacts. Sliding may occur when the shear force reached the maximum frictional force which is determined by the coefficient of friction. The coefficient of friction between the foot and ground was taken as 0.6 [23].

The FE model of foot and ankle was validated in our previous studies [17, 24, 25, 26] through comparison of the distribution and the peak value of the plantar pressures between FE prediction and experimental data. The results showed reasonably comparable between the FE prediction and experimental measurement [17, 26].

This model was modified to represent surgery of first and second TMT joint fusion. In surgical operation, the affected TMT joints are fixed to constrain the relative motion (Fig. 2a) using screws. To simulate the joint fixation in FE model, articulating interfaces of the two joints, and other contact surfaces among medial cuneiform, intermediate cuneiform, first and second metatarsal were tied together. Relative motion among these bones was totally constrained to simulate actual surgery, as shown in Fig. 2b.

Fig. 2

Fig. 2. Surgery of first and second tarsometatarsal joint fusion (a) and, four tied bones in model for simulation (b). Articular surfaces among the first and second metatarsal bones, and medial and intermediate cuneiforms were tied together to simulate the fixed joints.

For the simplicity, it was assumed that the TMT joints fusion did not change the gait pattern much based on the fact that the relative motion of TMT joints is quite limited in normal foot. Three instants during stance phase of gait, namely first peak, midstance and second peak, were simulated in normal foot and foot with TMT joint fusion.

Boundary and loading conditions were based on experimental data. The exact position of foot and ankle in each instant of gait was obtained from motion analysis system (Vicon, Oxford Metrics, Oxford, UK). Seven retro-reflective markers were attached to the right lower limb defining three segments, including fore-foot, hind-foot, and foot shank. The kinematic information, including the angle of foot and foot shank to ground, was collected and calculated as boundary conditions for the simulations. Ground reaction forces (GRFs) in vertical, antero-posterior and medial-lateral directions were recorded using force platforms (Advanced Mechanical Technology, Inc., Watertown, USA). The curve of GRFs during stance phase was obtained from the same subject of the foot model, marked with three instants simulated in this study. The instant of the first peak in terms of the vertical GRF (576N) is at about 25% and the second peak (598N) located around 70% of stance phase. The point of midstance (519N) was chosen at the valley of the curve between the first and second peak, around 40% of stance phase, as shown in Fig. 3. To simulate different instants, active extrinsic muscle forces, in addition to GRFs, were applied. Muscle forces were estimated from physiological cross-section areas of muscles and respective electromyography (EMG) data during gait with a linear EMG-force assumption from literatures [27,28].

Fig. 3

Fig. 3. Curve of vertical and antero-posterior components of ground reaction force during stance phase of gait and simulated points including the first and second peaks, and midstance.

Muscle forces were applied to the corresponding muscle structures represented by dotted lines. Achilles tendon force was represented by five equivalent force vectors, applied individually to five connector elements. The vertical and antero-posterior components of GRFs were applied as concentrated forces to the bottom of the rigid layer at the centre of pressure obtained from the force Platform. The superior surfaces of the encapsulated soft tissue, distal tibia and fibula were fixed throughout the simulation and the foot shank positions were represented by turning the rigid plate to the same angles. The applied boundary conditions are shown in Fig. 4.

Fig. 4

Fig. 4. Boundary and loading conditions for simulation of gait instants. The superior surfaces of soft tissue, tibia and talus bones were fixed. Ground reaction forces of antero-posterior and vertical directions were applied. Muscle forces were applied to muscle representatives.

In addition to gait instants, balanced standing was simulated to understand the effect of fusion on the arch stiffness. Half of the body weight (270N) was applied vertically to the rigid plate as GRF. The same force was applied to Achilles tendon, with neglecting the other muscles [29].

Results

In order to reveal the effect of fusion of TMT joints, contact pressures at the articulating interfaces in the mid- and hind-foot, von Mises stress in the five metatarsal bones, and contact pressure distribution on the plantar foot were analyzed. Results in normal and joint fusion models were compared.

Fig. 5 shows the plantar pressure distributions in the normal and fusion models. The distribution patterns were similar, while the peak pressure increased after fusion. In the normal model, the peak pressures at the first peak, mid-stance and second peak were 0.50 MPa, 0.60 MPa, and 0.64 MPa, respectively. After fusion, they increased by 0.42%, 19% and 37% and reached 0.51 MPa, 0.72 MPa, and 0.88 MPa. The peak pressure had little change in the instant of the first peak and increased afterwards. Arch height stiffness and contact area are factors related to contact pressure. Arch height stiffness was estimated by the flexibility of arch height. The arch height was measured by the distance between the dorsal peak of the intermediate cuneiform and the plantar peak of the calcaneus bone in the superior-interior direction and was found to differ between the normal foot and foot with fused joints. The FE simulated results showed that the fused foot had 24% less variation in arch height the normal foot during balanced standing.

Fig. 5

Fig. 5. Plantar pressure distributions in the normal and fused tarsometatarsal joint models in three instants.

Contact pressures at the joints of ankle, subtalar, talonavicular, calcaneocuboid, navico-mcuneiform, navico-icuneiform, navico-lcuneiform, lcunecuboid, and the third, fourth and fifth TMT were investigated. It was found that joint fusion increased the contact pressure at joints of ankle, talonavicular, navico-icuneiform, navico-cuboid and fifth meta-cuboid. In order to deliver direct expression, ratios of contact pressures between the five joints in all simulated instants to ankle joint during first peak of normal foot were shown in Fig. 6.

Fig. 6

Fig. 6. Comparison of normalized contact pressure at five joints between normal foot and foot with two tarsometatarsal joint fusion. These five joints showed increased contact pressure after the joints fusion. All contact pressures were divided by that of ankle joint during the first peak instant.

As shown in Fig. 6, the maximum contact pressure at these five joints mainly increased as the gait cycle progressing, among which ankle joint showed more obvious changes and sustained higher pressure than others. But the variation caused by fusion was not as obvious as by gait stage, increased by 12% and 14% and 0.58% in three instants. The talonavicular joint was subjected to increased contact pressure by 5.2%, 1.7% and 11% and in navico-icuneiform joint it was recorded as 12%, 16%, 27% increase. Among all the fluctuations, navicular and cuboid contact pair showed the most considerable variation during midstance, but with a relatively small magnitude. The following two were the fifth meta-cuboid and navico-icuneiform joints during second peak, 40% and 27% respectively.

Based on comparison between two models, the maximum change of von Mises stress was observed in the second metatarsal bone during midstance, showing a 22% increase from 26 MPa to 31 MPa, after fusion. The increase was 16 % in the first peak and 14% in the second peak. The fifth metatarsal bone increased by 5.1% and 9.5% in the first peak and midstance after fusion. The stress in the first and forth metatarsal bones did not change substantially after fusion. Fig. 7 shows the von Mises stress during midstance.

Fig. 7

Fig. 7. Von Mises stress in the five metatarsal bones (a) in normal foot model and (b) model with the first and second tarsometatarsal joint fusion in midstance. The second metatarsal showed the maximum change during midstance instant.

Discussion

The FE model of foot and ankle predicted changes of several biomechanical parameters after TMT joint fusion. The information obtained can help in understanding of current treatment protocols and precaution of potential risks of complications. It was found that joint contact pressures, von Mises stress in metatarsal bones, and plantar pressure in several parts increased after fusion.

The plantar pressure increased in the late instants of stance phase after fusion, possibly due to arch stiffness changes. Normal arch deforms to interact with the environment in a most effective way, protecting the segments from excessive loads. Structural modification, like TMT joint fusion could partially restrict the interaction and adjustment among joints. Smaller change in arch height of foot with joint fusion indicates that the joint fusion increases the arch stiffness.

The results indicated that joint fusion resulted in a limited range of movement of the first metatarsal bone and a stiffer arch. This could be attributed to that the fused foot is more capable to resistant the arch deformation, because the relative motion among the four fused bones was totally limited. This is consistent with a previous study in the outcomes of fixation treatment in tarsal bones, involving TMT joints. Minor restriction in the range of motion, particularly in talonavicular and medial TMT joints of mid-foot, were found with the patients [30]. The first metatarsal bone was observed to be significantly dorsiflexed in flatfoot relative to the talus bone [31, 32]. Fusion of TMT joints could be a way for correction of flatfoot, but may cause increasing in plantar pressure and other joint stress, especially during push-off stage.

In the analysis of the contact pressure in joints of hind- and mid-foot, the navicular and cuboid contact pair showed the most considerable variation in midstance, but with small magnitude. Navico-cuneiform and fifth meta-cuboid joints increased 27% and 40% at second peak, with much higher pressure than navicular and cuboid contact pair. Ankle and talonavicular joints also sustained increased contact pressure, but with slight changes. Limited motion of fixed bones induced higher contact pressure in mid-and hind-foot joints. These joints were subject to greater risk of deformation from a normal anatomical position under a continual and long-term excessive loading condition. This could be regarded as a predictor of foot pain and malalignment, as common clinical complications Malalignment of foot segments could further affect normal functioning of parts upper foot and ankle, for example the knee joint. A disordered mechanical environment also contributes to disturbing the maintenance of the articular cartilage and underlying bones. Heightened pressure on joints may leave them more susceptible to fatigue wear of the contact surfaces over a prolonged period. Unspecific pain and arthritis in foot were quite common after surgical treatments [3, 12, 13, 30, 33]. Clinical observations showed early degenerative signs of arthritis at fifth meta-cuboid joint and other TMT joints [30] and arthritis as long term side effect [33]. The predictions from this FE analysis indicate that navico-cuneiform and fifth meta-cuboid joints have the potentials to succumb to arthritis.

Von Mises stress is often considered as one predictor for bony stress fracture [34]. The five metatarsal bones are thought to be most susceptible for recurring stress fracture because of the long and thin shape and the function of loading transfer. Metatarsal stress fractures are most commonly seen in the second and the third metatarsals and fracture of the second metatarsal is reported to be one of the most common problems after surgeries in foot and ankle [35]. Among the five metatarsal bones, von Mises stress in the second metatarsal varied the most in the midstance. Considering the 22% increase of von Mises stress, the second metatarsal bone is more likely to sustain a fracture.

Parametrical analyses using our FE can reveal biomechanical performance of foot and ankle complex after TMT joint fusion during the first, second peak and midstance. Stance phase of gait can be analyzed in the way by applying different boundary and loading conditions. Dynamic activities such as impact on foot during landing can also be simulated to find how foot segments adapt to shocks. This is much more difficult using experimental methods. Current biomechanical studies are mainly clinical follow-up investigations, gait analysis, and cadaveric experiments. Follow-up investigations involving radiography estimation and score systems are generally combined together to evaluate reductions, arthritis and fractures [36-38]. Gait analysis provides kinematic and kinetic information of foot and ankle to evaluate surgical outcomes and rehabilitations [39-43]. Cadaveric experiments could detail contact pressure, stress/strain in some regions of foot and ankle, but the measurements are still technically and ethically limited[44-46]. For joints with complex contours embraced by plenty of ligaments, it is difficult to obtain any measurements without destructive operation. These studies cannot provide enough biomechanical information such as the internal stress distribution and contact pressure which contribute to complications.

In this study, the FE model response to geometrical modification in TMT joints, and offered clear pictures in stress distribution in bones, contact pressure at joints and plantar foot, and arch deformation. Potential risks of this surgery were predicted. Optimal surgeries are expected to decrease the complications and negative long term outcomes, permitting effective surgical intervention to address foot problems. Based on the prediction in this study, it is speculated that rather than directly fusing the bones, wedge shaped osteotomy opening towards dorsi-aspect can be made at the joints before fixation. The first and second metatarsal bones could be fixed with more dorsiflexion, which could possibly distribute part of loads to other bones and thus relieve the load in second metatarsal bone. The decreased height of the longitude arch might transfer some forces from lateral foot to others and alleviate overloading condition of the cuboid. The biomechanical information obtained enabling surgeons with more low-risk, sophisticated treatment options that are currently not well known or considered too risky to undertake. FE analysis could be an effective method to explore the rationale of biomechanical changes undergone after surgery and might be beneficial to surgeons by providing direct guidelines for surgery planning.

The model analysis is based on the assumption that the TMT joints fusion did not change the gait pattern, which may not be proper for all individual subjects. The joint fusion was simulated through tying bones together without using screws. This simplification may influence the load distribution mainly close to the screws. The medial-lateral component of GRFs was not considered in applied loading conditions for simplification reason. In this model, the muscles and ligaments were simplified as connection elements rather than real geometrical solids. The effects of these simplifications are worthy to be analyzed in further study.

To draw a conclusion, surgery of the first and second TMT joint fusion changed biomechanical performance of foot and ankle complex. Long-term consequences of this change could potentially be arthritis of navico-cuneiform and fifth meta-cuboid joints and stress fracture of second metatarsal bone. Increased plantar pressure could be possible as a contributor of plantar foot pain. Flexible flatfoot maybe benefit from this modification in consideration of increased arch stiffness. Considering these changes due to fusion, surgical protocol could be possibly improved by making wedge osteotomy at the involved joints before fixation to avoid some negative effects. However, further study is needed for verification.

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Conflict of interest

There is no conflict of interest.

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Figure Captions

Fig. 1. Finite element model of foot and ankle, consisting of 28 bones, 72 ligaments, plantar fascia and encapsulated soft tissue. Part of the soft tissue was removed for a better view of bone structures.

Fig. 2. Surgery of first and second tarsometatarsal joint fusion (a) and, four tied bones in model for simulation (b). Articular surfaces among the first and second metatarsal bones, and medial and intermediate cuneiforms were tied together to simulate the fixed joints.

Fig. 3. Curve of vertical and antero-posterior components of ground reaction force during stance phase of gait and simulated points including the first and second peaks, and midstance.

Fig. 4. Boundary and loading conditions for simulation of gait instants. The superior surfaces of soft tissue, tibia and talus bones were fixed. Ground reaction forces of antero-posterior and vertical directions were applied. Muscle forces were applied to muscle representatives.

Fig. 5. Plantar pressure distributions in the normal and fused tarsometatarsal joint models in three instants.

Fig. 6. Comparison of normalized contact pressure at five joints between normal foot and foot with two tarsometatarsal joint fusion. These five joints showed increased contact pressure after the joints fusion. All contact pressures were divided by that of ankle joint during the first peak instant.

Fig. 7. Von Mises stress in the five metatarsal bones (a) in normal foot model and (b) model with the first and second tarsometatarsal joint fusion in midstance. The second metatarsal showed the maximum change during midstance instant.