A 3-Dimensional Finite Element Model of the Human Foot and Ankle for Insole Design

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Objective: To investigate the effect of material stiffness of flat and custom-molded insoles on plantar pressures and stress distribution in the bony and ligamentous structures during balanced standing.

Design: A 3-dimensional (3-D) finite element model of the human ankle-foot complex and a custom-molded insole were developed from 3-D reconstruction of magnetic resonance images and surface digitization. The distal tibia and fibula, together with 26 foot bones and 72 major ligaments and the plantar fascia, were embedded in a volume of soft tissues.

Setting: Computational laboratory in a rehabilitation engineering center.

Participant: A healthy man in his mid twenties (weight, 70kg).

Interventions: Not applicable.

Main Outcome Measures: Foot-support interfacial pressure, von Mises stress in bony structures, and strain of the plantar fascia were predicted using the finite element model.

Results: A custom-molded, soft (Young modulus, E=0.3MPa) insole reduced the peak plantar pressure by 40.7% and 31.6% at the metatarsal and heel region, respectively, compared with those under a flat, rigid (E=1000MPa) insole. Meanwhile, a 59.7% increase in the contact area of the plantar foot was predicted with a corresponding peak plantar pressure increase of 22.2% in the midfoot.

Conclusions: The finite element analysis implies that the custom-molded shape is more important in reducing peak plantar pressure than the stiffness of the insole material.

Key Words: Ankle; Finite element analysis; Foot; Pressure; Rehabilitation.

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RHEUMATOID FOOT PAIN and diabetic ulceration are closely related to abnormal plantar pressure distributions. Increasing evidence suggests that these painful foot syndromes or diseases can be successfully resolved or relieved by fitting a proper foot insole, helping to relieve elevated plantar pressures. To achieve an optimal foot support design for subjects with a specific foot deformity or functional requirement, it is essential to explore stress distribution across the plantar foot surface and bony structures.

Pressure distributions between the foot and different supports have been measured by experimental methods with the use of in-shoe pressure sensors and a pedobarograph. Because of inherent difficulties and lack of better technology for that measurement, the load transfer mechanism and internal stress states within the soft tissues and the bony structures were not well addressed. Previous rationales behind the insole’s functional role load distribution and foot stabilization depended merely on subjective views or interfacial pressure measurements.

Today, computational modeling, such as the finite element method, is a complementary tool to enhance our knowledge of foot biomechanics. Finite element analyses can predict the load distribution between the foot and supports, and provide information on the internal stress and strain states of the ankle-foot complex. The finite element analyses enable efficient parametric evaluations to be made for the outcomes of insole shape and material modifications, without needing to fabricate and test orthoses in a series of patient trials.

Several finite element models of the foot or footwear have been developed, based on certain assumptions. These assumptions include simplified geometry, limited relative joint movement, ignorance of certain ligamentous structures, and simplified material properties. Early models were based on a simplified or partial foot shape. Analyses were conducted under assumptions of linear material properties, infinitesimal deformation, and linear boundary conditions, without considering friction and slip. Recent models have improved in selected aspects by incorporating geometric, material, or boundary non-linearity (eg, large model deformation, nonlinear material properties, slip/friction contact conditions). Previous finite element analyses have contributed to the understanding of biomechanic behavior and performance of foot supports. Researchers have used it to study what effects orthotic thickness and stiffness have on plantar soft tissue and plantar pressure distribution. In general, custom-molded, thicker and softer orthoses have been shown to reduce the peak plantar pressure and redistribute it in a more uniform pattern. However, a more detailed model of the human foot and ankle—a model that incorporates realistic geometric and material properties of both bony and soft tissue components—is needed to enhance the reliability of the quantitative evaluations of different orthotic designs.

The objective of the present study was to establish a 3-dimensional (3-D) finite element model of the foot and ankle, using actual geometry of the foot skeleton and soft tissues in contact with an insole support. The finite element model can be employed to predict the mechanical interaction between the foot and different types of insole. This technology will help researchers design an optimal orthosis for a desirable plantar pressure distribution and stabilizing support of the foot.

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METHODS

The geometry of the finite element model was obtained from 3-D reconstruction of magnetic resonance (MR) images from the right foot of a healthy man in his mid twenties (height, 174cm; weight, 70kg). Coronal MR images were taken at intervals of 2mm in the neutral unloaded position. The images were segmented using MIMICS, version 7.10, to obtain the boundaries of skeleton and skin surface. The boundary surfaces of the skeletal and skin components were processed using SolidWorks 2001 to form solid models for each bone and the whole foot surface. The solid model was then imported and assembled in the finite element package ABAQUS, version 6.3.

The finite element model, as shown in figures 1A and B, consisted of 28 bony segments, including the distal segments of the tibia and fibula and 26 foot bones: talus, calcaneus, cuboid, navicular, 3 cuneiforms, 5 metatarsals, and 14 components of the phalanges. The phalanges were fused together with 2-mm-thick solid elements, which simulated the connection of the cartilage and other connective tissues. The interactions among the metatarsals, cuneiforms, cuboid, navicular, talus, calcaneus, tibia, and fibula were defined as contact surfaces, which allow relative articulating movement. To simulate the frictionless contact between the joint surfaces, the ABAQUS automated surface-to-surface contact option was used. Except for the collateral ligaments of the phalanges and other connective tissue, a total number of 72 ligaments and the plantar fascia were included and were defined by connecting the corresponding attachment points on the bones. The bony structures were merged with the encapsulated soft tissues, while the ligamentous structures were superimposed within the volume of soft tissue. The bony and soft tissue structures were meshed with a total of 50,964 tetrahedral elements and the ligaments were defined with 103 tension-only truss elements.

All tissues were idealized as homogeneous, isotropic, and linearly elastic (table 1). The Young modulus and Poisson ratio for the bony structures were assigned as 7300 and 0.3MPa, respectively. These values were selected by weighing cortical and trabecular elasticity values. The elastic modulus of the soft tissues was measured using the ultrasonic indentation system. The Young modulus and Poisson ratio were assigned to foot soft tissues as 15 and 0.45MPa, respectively. The mechanical properties of the cartilage, ligaments, and the plantar fascia were selected from the literature.

Flat and custom-molded insole supports (fig 1C) were simulated. The custom-molded insole was made from the unloaded shape of the subject’s bare foot. The barefoot shape was obtained by an impression cast with the subject sitting in the neutral position. The positive cast was digitized and imported to SolidWorks to form the solid models of the insole. A 5-mm-thick insole was meshed into 3-D brick elements with a Poisson ratio of 0.4 and a varied Young modulus of 0.3 (soft), 1 (firmer), and 1000MPa (rigid) for simulation of (1) open-cell polyurethane foams, such as Professional Protective Technology’s PPT material; (2) high-density ethylene vinyl acetate; and (3) polypropylene materials, respectively. A very rigid, 1-mm-thick bottom layer was used to simulate the ground support and to facilitate the application of concentrated ground reaction forces. The foot-insole interface was modeled using contact surfaces with a friction coefficient of 0.6. The insole was properly aligned in a way that permitted an initial foot-ground contact to be established, with minimal induced stress and contact pressure, before loading was applied. Interfacial pressures between the foot and flat support during balanced standing were measured by the F-scan system. The measurement was conducted on the same subject who volunteered for the MR scanning. To validate the model, we used the plantar pressures measured for this subject during barefoot standing to calculate the foot’s center of pressure (COP) and to compare the plantar pressure distribution predicted by the finite element technique.
A person with a body mass of 70kg applies a vertical force of 350N on each foot during balanced standing. Force vectors corresponding to half of the body weight and the reaction of the Achilles’ tendon were applied. The vertically upward force of the Achilles’ tendon, with a magnitude of 175N, was represented by 5 equivalent force vectors at the posterior extreme of the calcaneus. This value of Achilles’ tendon loading was based on the study of Simkin,23 who calculated that the Achilles’ tendon force was approximately 50% of the force applied on the foot during balanced standing. A net normal vertical force of 350N was applied at the COP of the inferior surface of the foot support. For our subject, the COP was about 90mm from the posterior extreme of the foot and 30mm from the medial heel extreme. The superior surface of the soft tissue, distal tibia, and fibula was fixed throughout the analysis, while the point of load application at the COP was allowed to move in the vertical direction only. The prescribed loading and boundary conditions allowed the equilibrium condition of the plantar foot and ankle to be established with unconstrained motion of the ankle joint and the insole support during weight bearing.

RESULTS

We constructed a 3-D finite element model of the human foot and ankle to study the effects that stiffness and shape of foot insole have on plantar pressure distribution, and to study the internal stresses in the bones during balanced standing. Figure 2 depicts the plantar pressure distribution obtained from the F-scan measurements and Figure 3 gives the pressure distribution predicted by finite element simulation during balanced standing. Both the measured and predicted values showed high pressures around the soft tissue beneath the calcaneus and the metatarsal heads, especially the second metatarsal head (figs 2, 3). The model predicted a peak pressure of .266 and .194MPa at the heel and metatarsal region, respectively, with a flat, rigid (E=1000MPa) insole. With a flat, soft (E=0.3MPa) insole, the peak pressures were .214 and .162MPa, respectively. The corresponding peak pressure measured with the F-scan sensors were .14 and .09MPa with the polypropylene platform and .13 and .07MPa with the PPT platform.

Figure 3 shows the predicted plantar pressure patterns with flat and custom-molded insoles of different Young modulus ratings of 0.3, 1.0, and 1000MPa. Under the same loading condition, use of a custom-molded and a softer insole would reduce the peak plantar pressure and increase the contact area between the plantar foot and the insole. The efficacy of specific insoles in redistributing the plantar foot pressure can be further seen in figure 4.

Figure 4 displays 3 features: (1) the effects of stiffness of the flat and custom-molded insoles on the peak pressure (fig 4A), (2) effects on the contact area between the plantar foot and the insoles (fig 4B), and (3) the von Mises stress distribution in the bony structures (fig 4C). During balanced standing, the heel region experienced the highest plantar pressure. Compared with a flat, rigid (E=1000MPa) insole, reductions of about 16.5% and 19.5% in peak plantar pressure over the metatarsal and heel regions were predicted with the use of a flat and soft (E=0.3MPa) insole. Use of the custom-molded insole had a resounding effect on pressure reduction. The custom-molded, rigid insole reduced the peak plantar pressure by about 23.2% and 24.4% over the metatarsal and heel regions, respectively. The soft custom-molded insoles provided pressure reduction of 40.7% and 31.6%, respectively. The rigid and soft custom-molded insoles enabled increases of peak plantar pressures of about 68.8% and 22.2% in the midfoot, respectively, compared with the use of a flat, rigid insole.

The contact area between the plantar foot and the insole increased significantly with use of the custom-molded insole (fig 4B). This increase mainly resulted from the increase over the midfoot contact. Compared with a flat, rigid insole, in the custom-molded insoles the contact area of the plantar foot increased by 51.5% for rigid and 59.7% for soft orthotics. A 13.5% increase in contact area was noted with a softer insole, but the effect was less pronounced than that found with custom-molded insoles.

During balanced standing, the highest von Mises stress in the bony structures was predicted to occur in the forefoot region in all the cases calculated (fig 4C). Over the forefoot, relatively high von Mises stresses were predicted at the midshaft of the second and third metatarsals. The calcaneocuboid joints and the posterior articulation of the subtalar joint sustained relatively large stresses in the midfoot and rearfoot regions, respectively. In general, use of the softer and custom-molded insoles of different Young modulus ratings of 0.3, 1.0, and 1000MPa.
insole reduced the stress on the bony structures. The stress reduction was more pronounced in the forefoot and midfoot and with use of the custom-molded insoles (fig 4C). Compared with the flat, rigid insole, the forefoot experienced a 5.6% and 11.7% peak stress reduction with use of soft, flat and soft, custom-molded insoles, respectively. The corresponding decreases in the midfoot were 3.4% and 16.8%. Interestingly, the rearfoot experienced a small increase in peak von Mises stress with use of softer insoles.

During balanced standing, we found that the plantar fascia sustained a maximum strain of about .49% and .37%, respectively, with use of flat insoles and custom-molded, rigid insoles. Use of softer material had a negligible effect on strain of the plantar fascia.

**DISCUSSION**

The predicted plantar pressure distribution pattern was, in general, comparable to the F-scan measurement. However, the
From the finite element prediction, the change in supporting conditions would strongly affect the pressure distribution of the plantar foot. This hypothesis is supported by experimental findings. Kato et al. reported a range of 19% to 80% mean peak plantar pressure reduction with the use of foot orthoses in 13 diabetic patients. Novick et al. reported mean peak plantar pressures of .164, .011, and .142MPa for the heel, midfoot, and metatarsal regions, respectively, during walking with a flat, soft insole. The corresponding values for the custom-molded, rigid insole were .18, .019, and .143MPa. These measurements are consistent with the finite element prediction, reflecting the corresponding pressure-relieving and pressure-redistributing abilities of the soft, custom-molded insole. Kato reported a mean increase in postorthotic contact area of about 63%, which is comparable to the finite element prediction (59.7%). Both a softer material and a custom-molded shape have a role in the reduction of peak plantar pressure. These insoles aided in assimilating the plantar pressure in a more uniform manner than a rigid, flat insole, which tended to concentrate the load at the heel and beneath the first and second metatarsal heads. The custom-molded insoles enable a more uniform distribution of pressures, whereas soft, flat insoles provide localized pressure relief. The custom-molded insoles are more effective in redistributing the plantar pressure to the midfoot region than the soft, flat insole. The finite element prediction indicated that an appropriate insole can reduce high plantar pressure and may relieve foot pain, especially for persons with especially stiff plantar tissue. The finite element model may help researchers design an optimal foot support with appropriate insole shape and material properties to suit patients’ individual needs.

In addition to providing a more uniform plantar pressure pattern, use of custom-molded insoles reduced the strain of the plantar fascia. The arch support provided by the custom insoles reduced foot lengthening and tension on the fascia during load bearing. The results thus suggested a possible therapeutic effect of custom orthoses in terms of stress relief of the plantar fascia and relief of associated foot problems, such as plantar fasciitis or insertional painful heel syndrome. The plantar fascia strains (37%–49%) predicted by finite element analysis were consistent with the strain of about 0.5% reported by Kogler et al., who used a microstrain transducer to measure the plantar fascia strain of cadaveric specimens with a similar magnitude of compressive loading.

Finite element analysis supports the use of soft, custom-molded insoles for redistributing plantar foot pressure. Particular care should be taken in prescribing the insoles to avoid complications from redistributed pressures. In fact, the foot’s soft-tissue compliance should be among the factors used to decide which type of orthosis and the material prescribed. For instance, diabetic patients with neuropathic ulcers may need a semirigid, custom-fitted insole to redistribute the localized pressures, whereas an elderly patient with a lack of plantar fatty pads may require use of a softer insole. Localized tissue sensitivity of the plantar foot should be inspected and the insole should be modified and evaluated as appropriate. To achieve a balance between pressure relief and control of foot motion, it may be necessary to provide a properly rigid functional orthosis that can achieve optimal efficacy. This consideration particularly applies to patients requiring multipurpose treatments.

Limitations

To simplify the analysis in this study, we assigned homogeneous and linearly elastic material properties to the model. The ligaments within the toes and other connective tissue, such as the joint capsules, were not considered. Use of homogeneous material and linearity of the encapsulated soft tissue stiffness

predicted values of peak pressure were higher than the F-scan measurements. The difference may be caused by resolution differences between the F-scan measurement and the finite element analysis. Having a spatial resolution of about 4 sensors per cm², the F-scan sensors recorded an average pressure for an area of 25mm². By contrast, the finite element analysis provided solutions of nodal contact pressure rather than an average pressure calculated from nodal force per element surface area. The measured peak plantar pressure was therefore expected to be smaller than the predicted values.

The predicted peak von Mises stress showed that the mid-shaft of the second and third metatarsals were the most vulnerable regions. The confined positions of these metatarsals, especially with tissue stiffening, are probably the cause of stress concentration. Apart from the mid-shaft of the metatarsals, the junctions of the subartalar and calcaneocuboid joints were also possible sites of fracture or lesion under weight bearing. Use of the softer insole, especially the custom-molded insoles, is effective in reducing the bone stress in the forefoot and midfoot region.

From the finite element prediction, the change in supporting conditions would strongly affect the pressure distribution of the plantar foot. This hypothesis is supported by experimental findings. Kato et al. reported a range of 19% to 80% mean peak plantar pressure reduction with the use of foot orthoses in 13 diabetic patients. Novick et al. reported mean peak plantar pressures of .164, .011, and .142MPa for the heel, midfoot, and metatarsal regions, respectively, during walking with a flat, soft insole. The corresponding values for the custom-molded, rigid insole were .18, .019, and .143MPa. These measurements are consistent with the finite element prediction, reflecting the corresponding pressure-relieving and pressure-redistributing abilities of the soft, custom-molded insole. Kato reported a mean increase in postorthotic contact area of about 63%, which is comparable to the finite element prediction (59.7%). Both a softer material and a custom-molded shape have a role in the reduction of peak plantar pressure. These insoles aided in assimilating the plantar pressure in a more uniform manner than a rigid, flat insole, which tended to concentrate the load at the heel and beneath the first and second metatarsal heads. The custom-molded insoles enable a more uniform distribution of pressures, whereas soft, flat insoles provide localized pressure relief. The custom-molded insoles are more effective in redistributing the plantar pressure to the midfoot region than the soft, flat insole. The finite element prediction indicated that an appropriate insole can reduce high plantar pressure and may relieve foot pain, especially for persons with especially stiff plantar tissue. The finite element model may help researchers design an optimal foot support with appropriate insole shape and material properties to suit patients’ individual needs.

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Limitations

To simplify the analysis in this study, we assigned homogeneous and linearly elastic material properties to the model. The ligaments within the toes and other connective tissue, such as the joint capsules, were not considered. Use of homogeneous material and linearity of the encapsulated soft tissue stiffness
was a simplification of the real situation. Only the Achilles’ tendionloading was considered, whereas other intrinsic and extrinsic muscle forces were not simulated. The location of COP was assumed to be unchanged with the simulations of different insole supports.

Future Considerations

The finite element model we developed can be refined to simulate the actual situations more realistically by incorporating nonlinear material properties for the ligamentous and soft tissue structures and simulating the whole shoe structure. The load-bearing characteristic of the ankle-foot structures under different stance phases requires the incorporation of detailed load-bearing characteristic of the ankle-foot structures under various loading and supporting conditions. Finite element analysis indicated that the insole’s custom-molded shape is more important in reducing peak plantar pressure than the stiffness of the material from which it is made. A comprehensive, finite element, ankle-foot model makes monitoring the parametric effect of different insole materials and appropriate simulations, the finite element model may aid in developing and prescribing a pressure-relieving orthosis pertinent to individual needs.

CONCLUSIONS

A geometric, detailed 3-D finite element model of the human foot and ankle was developed to estimate the plantar pressure and the internal stress and strain in the bony and soft tissue structures under various loading and supporting conditions. Finite element analysis indicated that the insole’s custom-molded shape is more important in reducing peak plantar pressure than the stiffness of the material from which it is made. A comprehensive, finite element, ankle-foot model makes monitoring the parametric effect of different insole shapes and material properties more efficient. The finite element model is an ideal clinical tool to investigate foot behavior under different supports and to explore the design of various forms of foot support.

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References