An Analysis of Soft Tissue Loading in the Foot—a Preliminary Report

ABSTRACT—The foot's response to load during weightbearing is investigated using finite element stress analysis of a two-dimensional model. The analysis predicts the stress states within the plantar aspect tissue during a variety of shoe conditions. Shoe soles with a range of elastic properties are modeled in an attempt to find a shoe sole that minimizes peak stresses within the foot's soft tissue. This presently simplified and idealized model of the foot and shoe during mid-stance demonstrates significant dependency of stress development within the tissue on shoe elastic properties. These results would seem to justify further more detailed and realistic modeling of shoeing mechanics, and systematic efforts to correlate physical measurements and clinical experience with the trends indicated by finite element analyses.

INTRODUCTION

Excessively high stresses on the foot for prolonged periods cause many foot problems. Diabetic dermatological complications are among the most serious of foot problems. The tissue of diabetics heals poorly and the vascular impairment often associated with diabetes causes ischemia and neuropathy (5). Treatment of clinical problems secondary to the disease, such as vascular insufficiency, neuropathy, ulcers, and gangrene, are of increasing importance with diabetic patients. The foot, due to its remoteness within the circulatory system, its position while sitting or standing, and its role in weightbearing, is the major site of these disease manifestations (5).

The importance of understanding skin response to load was outlined by Murphy (9). Husain (8) experimentally showed that excessively high stresses applied for prolonged periods of time cause mechanical damage to skin and soft tissue by cutting off the blood supply to the tissue.

A study of transferring load to soft tissue using the elasticity theory was done by Bennett (1-3), who investigated soft tissue failure and dermatological problems experienced in fitting prosthetic devices to amputees. Aspects of prosthetic socket flexibility and shape were investigated. Although he realized that there are numerous complexities in real tissues, and variations with anatomical locations, in order to simplify the analysis to manageable proportions he assumed that soft tissue possessed a constant Young's modulus and Poisson's ratio.

Chow and Odell (6) analyzed a cushion which distributed the pressure more uniformly over the skin of a sitting person. Their study was motivated by the need to improve understanding of wheelchair cushion design and to prevent pressure sores (decubitus ulcers). They presented a simplified buttock model consisting of a deformable hemisphere (soft tissue) with a rigid core representing the ischial tuberosity. The finite element analysis used 12 noded, curved, isoparametric quadrilateral elements. Each element...
was a torus within the hemisphere. The pressure distribution was assumed to be symmetric about a vertical axis through the core, and loading was applied incrementally to the linearly elastic model. (The soft tissue was assumed to be linearly elastic and nearly incompressible.)

We have investigated the stress and deformation of soft tissue in the foot resulting from skeleton-to-shoe-sole load transfer with the foot flat on the ground in mid-stance. This study was motivated by the need for a better understanding of pressure distribution in the diseased foot (and in particular, the diabetic foot) accurately fitted with a molded foam shoe. A design analysis of the shoe sole in a two-dimensional simplified model, with the aim of preventing or retarding this debilitation process, is presented as a step toward further analysis of clinical trials. This study may also have implications for more general shoe problems, such as those associated with normal shoes and with athletic footwear.

**MATERIALS AND METHOD**

**Material Properties**

The nonlinear finite element model representing the foot and shoe sole utilized in this investigation requires the estimation of the mechanical properties of the skeleton, soft tissue, and shoe sole. The literature contains a number of papers (4,7,10,11) reporting the material properties of the soft tissues and skin of the body. These investigations, however, usually involve extension tests in a transverse direction, within the plane of the skin, and as such are not useful in the present analysis of the predominantly compression-loaded foot.

A soft tissue specimen was extracted from the heel of a fresh cadaver foot. It was peeled from the calcaneous, with the plantar skin preserved. The specimen (measuring approximately 30X30X15 mm.) was compressed between two flat platens at a strain rate of 0.16 per second using an Instron compression and extension test machine. Some fluid escaped, probably accounting for some of the hysteresis observed. The resulting data of force vs. displacement are plotted in Figure 1, along with the calculated stress versus strain and the Young's modulus versus strain curves. The soft tissue demonstrates a highly nonlinear response with Young's modulus approaching 10^7 Pa with a 40 percent strain.

A similar shape of curve was obtained on a cadaver entire foot, but because of irregular shape there was difficulty in defining area, stress, strain, thickness, etc. The constraints of adjoining tissues.
A Nonlinear Planar Finite Element Method

In linear finite element analysis (12) the B-matrix, which represents the strain-displacement relationship, is computed from the geometry of the unloaded state. However, if the loading of the system has a significant effect on the element geometry, the B-matrix must be computed from the geometry of the deformed element. Thus, the B-matrix becomes a function of the nodal displacements. The structural response is then geometrically nonlinear. Due to the relatively low Young’s modulus of the foot’s soft tissue and that of many potential shoe sole materials, regions of the foot-shoe-sole structure experience large nonlinear displacements. The analysis of the foot’s structural response is also complicated by material nonlinearities (Fig. 1) potentially of both the soft tissue and shoe sole materials.

There are several numerical methods which permit a nonlinear analysis. The Newton-Raphson method, the modified Newton-Raphson, direct iteration, and incremental loading have been used in such situations (12). The Newton-Raphson type methods are computationally more efficient than incremental loading methods—however, they are potentially unstable. The material nonlinearity of the present problem (i.e., increasing Young’s modulus with increasing strain) is unfortunately of a nature that can cause a divergence in the solution obtained by the Newton-Raphson method.

To assure numerical stability in the solution, an incremental loading method was chosen for this analysis. In this method, load is applied to the structure in a series of small increments. During each load increment, the deformation of the structure is small and assumed to be linear. The summation of all structural displacements produces a large displacement and nonlinear response to full load. The most straightforward application of this incremental loading method is to increase the load by a fixed fraction of the full load. The accuracy of analysis of the fully-loaded structure is improved with a reduction of this fraction of full load; i.e., with more load increments. The size of the load increments should be such that the finite elements within the structure experience small strains during any loading increment. If, however, a material has a low Young’s modulus or a region of a low Young’s modulus (as a function of strain), excessively large numbers of constant load increments may be required to assure reasonable accuracy.

In the case of soft tissue (Figure 1) the Young’s modulus changes in magnitude up to $10^7$ Pa. At the beginning of the loading process, the stiffness of each element is extremely low and large element strains are possible even for a small structural load. Due to this potentially large change in Young’s modulus, a variable incremental loading method was adopted. This method improves the efficiency of the incremental loading method by using current values of element Young’s modulus to estimate the size of the next load increment while assuring a predetermined small strain limit in each element. An initial small load is applied to the structure and the elements are searched (at each step) to find the element of greatest strain, the “pilot” element. The strain state of each element is used to calculate the current stiffness of that element using the appropriate Young’s modulus from Figure 1. The magnitude of load in the next load increment is determined by the strain in the pilot element such that the strain in this element during the next load increment does not exceed a preset small value (e.g., 1.0 percent). As the stiffness of the pilot element increases, the magnitude of the incremental load will correspondingly increase. Incrementing the load in this manner guarantees that during the load increment only a small strain occurs in each element and that full structural load is applied with a minimum number of increments. Figure 2 demonstrates the algorithm for this nonlinear large displacement analysis.

Convergence Study

The solutions determined by this incremental loading method are sensitive to the number of load increments; i.e., the incremental strain limit magnitude. Smaller strain magnitudes can be expected to produce more accurate results. To check the convergence of the numerical solution, the simple problem shown in Figure 3 was analyzed. This model represents a composite block of bone, skin, and...
shoe sole. The lowest layer is an arbitrarily chosen open-cell foam rubber with the elastic property shown in Figure 1 and a measured Poisson’s ratio of 0.22. The second layer represents the soft tissue of the heel which has the highly nonlinearly elastic property of Figure 1 and an assumed Poisson’s ratio of 0.49 (i.e., nearly incompressible). The top layer is bone with a Young’s modulus of $7.3 \times 10^3$ MPa and a Poisson’s ratio of 0.3, frequently used in the literature. These three layers are modeled as being bonded to each other, without gliding fascia or a sock. The convergence of displacement in this model (Figure 4), up to the compression force of 600 N, was studied for incremental strain limit magnitudes within pilot element 24 of 0.3 to 0.02. Satisfactory convergence occurred in this model with the use of about 20 load increments, corresponding to an incremental strain limit of about .05.

**FIGURE 2.**
Computer algorithm used in the incrementing of applied load.
FIGURE 3.—Simplified model used in the convergence study.

Finite Element Idealization of the Foot

The structure of the foot is complicated in both anatomical geometry and material properties. Any analysis of this structure requires some degree of simplification. A planar nonlinear finite element model of the foot (Fig. 5) was developed to investigate the trends in the dependence of stresses in the foot on shoe sole material properties. The complex bony structure of the foot, with articulated joints stabilized by muscles, is represented by a single elastic body with the exception (especially important in case 2) of a region of low Young's modulus about the metatarsal head region (indicated by cross-hatching) to permit forefoot flexibility. The layers of muscle, fascia, and skin of the foot's plantar aspect are represented by a homogeneous, isotropic, nonlinearly elastic material (Fig. 1) with Poisson's ratio of 0.49. The layers were assumed to be bonded together, a condition approximated by a snugly laced shoe without a sock.

The model of Figure 5 has 198 nodal points with 342 triangular elements. Two loaded cases were

FIGURE 4. The convergence of displacement (model of Figure 3) for increasing number of load increments.
investigated (Fig. 5) with the load incrementally applied to the model by the loading method previously described. In load case 1, an element under the calcaneous was found to be the pilot element, that element experiencing the greatest strain. During the incremental application of load case 2 the pilot element was found to move anteriorly to a region under the metatarsal head. After the full load was applied to the model, the stress states of all elements representing the thick skin on the foot’s plantar aspect were searched for the maximum values of compressive and shear stresses in the tissues. This process of modeling and stress recording was repeated for a variety of elastic properties of shoe sole materials.

RESULTS

The nonlinear finite element analysis previously described was performed with varying representations of shoe sole elastic properties. Twenty different representations of linearly elastic shoe sole materials were modeled. The Young’s modulus of the shoe sole was varied from a value of 0.08 MPa to 1.00 $\times 10^3$ MPa. Presented in Figure 6 are the maximum values of compressive stress (principal stress) and maximum shear stress predicted to occur within the soft tissue of the foot subject to the two loading conditions of Figure 5 on a linearly elastic shoe sole. The magnitudes of both these stress quantities demonstrate a significant dependence upon the Young’s modulus of the shoe sole material.

In general, these stress values are seen to be relatively large with both extremely low and high values of shoe-sole Young’s modulus, while at intermediate values of shoe-sole Young’s modulus, there appears to be a relative minimum in both of these stress quantities, though the minimal values for compression and shear in the two cases did not completely coincide. This minimization of compressive and shear stress within the soft tissue of the foot occurred within the range of constant shoe sole Young’s modulus values of .1 to 1.0 MPa. The unloaded and fully deformed shapes of the foot shoe sole model, with a constant shoe sole Young’s modulus of .2 MPa, is shown in Figure 7.

The nonlinear shoe sole material (Fig. 1), with characteristics of a foam rubber material, was found to produce maximum compressive and shear stress, during load case 1, of 0.52 and 0.212 MPa respectively. These stress values are close to those obtained with the optimum linearly elastic material. During the first load case the maximum compression and shear within the skin occurred under the calcaneous, while during the second load case these maxima occurred under the metatarsal head.

DISCUSSION

The numerical model used in this investigation is that of a simplified and idealized planar section of the foot. The interpretation and use of this data should be performed with knowledge of the following limitations of the method. The bony skeleton of the foot is represented as a linearly elastic and homogeneous material. The soft tissue of the foot’s plantar aspect is represented as a highly nonlinear and nearly incompressible material. The foot and shoe are assumed to be in continuous contact; i.e., a conforming shoe. Additionally, the foot and shoe surfaces are assumed to have no relative motion; i.e., the foot will not slip on the shoe sole. Most of the shoe sole materials considered were linearly elastic materials of a single Poisson’s ratio. Only one
A geometric shoe sole is modeled; additionally, only two loading situations are considered. Although it is appreciated that this method possesses significant limitations, it is felt that the analysis offers the following insight to the understanding of the effect that shoe sole design has upon the development of stress within the foot.

This study of linearly elastic shoe sole materials showed the existence of optimum values of shoe sole Young's modulus. These optimum values occur subject to the constraints and modeling assumptions discussed above and as such may change with changes in those assumptions. In particular, they may change with changes in shoe sole geometry or thickness. At this optimum value, the maximum compressive stress and shear stress within the soft tissue of the foot was minimized. High-modulus shoe sole materials (i.e., stiff materials) result in high stresses within the foot as the soft tissue is loaded between the relatively stiff skeletal core and the stiff shoe sole. With exceedingly low moduli shoe soles (i.e., soft materials) the shoe sole effectively collapses under load, subjecting the soft tissue of the foot to high stresses induced by the underlying rigid foundation. Minimization of the two stress quantities considered during the two loading conditions occurred in the range of shoe sole Young's modulus of 0.1 to 1.0 MPa. All of the shoe sole materials investigated resulted in compressive stress development in the plantar skin about the heel that exceeded normal arterial pressure. These stress levels will result in an interruption of blood flow within the region during loading. A prolonged loading would result in tissue damage although cyclic loading, as in gait, is well tolerated in the normal foot.
SUMMARY

A planar finite element analysis, suitable for the analysis of structures involving both geometrical and material nonlinearity, is presented. The analysis is applied to a model of the foot and shoe sole subject to stance-phase loading, with a number of simplifying assumptions. The role of shoe sole elastic properties in the development of stress within the thick skin of the foot’s plantar aspect is investigated. The results show a significant dependence of stress within the foot on the elastic properties of the shoe sole. Assuming the existence of optimum linearly elastic shoe material, a range of values that minimizes both the peak compressive stress and the shear stress in the thick skin is shown. Representative nonlinear shoe sole materials are also analyzed which show peak compressive stress close to those of the optimal linearly elastic shoe material.

REFERENCES


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