Stress analysis of the standing foot following surgical plantar fascia release

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Abstract

Plantar fascia release is a surgical alternative for patients who suffer chronic heel pain due to plantar fasciitis and are unaffected by conservative treatment. A computational (finite element) model for analysis of the structural behavior of the human foot during standing was utilized to investigate the biomechanical effects of releasing the plantar fascia. The model integrates a system of five planar structures in the directions of the foot rays. It was built according to accurate geometric data of MRI, and includes linear and non-linear elements that represent bony, cartilaginous, ligamentous and fatty tissues. The model was successfully validated by comparing its resultant ground reactions with foot-ground pressure measurements and its predicted displacements with those observed in radiological tests. Simulation of plantar fascia release (partial or total) was accomplished by gradually removing parts of the fascia in the model. The results showed that total fascia release causes extensive arch deformation during standing, which is greater than normal deformation by more than 2.5 mm. Tension stresses carried by the long plantar ligaments increased significantly, and may exceed the normal average stress by more than 200%. Since the contribution of the plantar fascia to the foot’s load-bearing ability is of major importance, its release must be very carefully considered, and the present model may be used to help surgeons decide upon the desired degree of release. © 2002 Elsevier Science Ltd. All rights reserved.

Keywords: Computational model; Numerical analysis; Finite element method; Plantar pressure

1. Introduction

The most important functional role of the plantar fascia is to maintain the arched structure of the foot which provides a firm supportive base during standing and absorbs dynamic reaction forces during gait. In addition, the plantar fascia, together with other ligamentous structures, tends to reduce deformations of the foot’s arch under static and dynamic weight-bearing. Thus, skeletal loads, which act to flatten the arch, are balanced not only by the stiffness of the bones and joints, but also by tension forces which are developed in the plantar fascia and ligaments (Ker et al., 1987; Kim and Voloshin, 1995). Abnormal foot structures, such as those characterized by an excessively high arch, may overload the plantar fascia, and, thereby, lead to plantar fasciitis, an inflammatory condition that results from micro tears in the fascia and is suffered mostly by athletes. Conservative treatment of plantar fasciitis includes anti-inflammatory medications and prescription of appropriate footwear, heel pads or orthoses to reduce the repetitive weight-bearing stresses to the plantar fascia. In the worst cases, surgical intervention involving partial or total plantar fascia release is performed (Fig. 1), to detach the plantar fascia from the plantar calcaneal aspect (Lutter, 1986; Gormely and Kuwada, 1992; White, 1994).

The goal of the present study was to develop a computational model for analyzing the static structural behavior of the foot in the standing posture following surgical plantar fascia release. The effects of the plantar fascia degree of release (PFDR) on the distributions of stresses and deformations in the foot during standing were characterized, thus providing a biomechanical tool that can be applied clinically in a pre-surgical evaluation of the partial/total release results. Validation of the model was achieved by comparing computationally predicted reaction forces (and related contact stresses) to experimentally obtained static foot-ground pressure patterns, and predicted arch deformation to experimentally measured deformation demonstrated by lateral X-rays.
2. Methods

2.1. Geometry

The model comprises five planar cross-sections through the foot (Fig. 2), which together yield a convenient representation of its complex three-dimensional (3D) half-dome shaped structure. The intermetatarsal ligaments that firmly hold the five rays of the foot and form an integral structure during the standing posture, enabled the use of this approach, which significantly simplifies the computational procedures compared with solid modeling in the 3D space. The cross-sectioning planes, S₁–S₅, which align with the direction of the corresponding five foot rays, are all vertical to the ground. Each plane passes through the center of its respective metatarsal head and through the calcaneal base. Planes S₁ and S₅ pass through the two edge points of a virtual line (of ~2 cm in length), which crosses the calcaneus-ground contact surface at its widest area. Planes S₂–S₄ cross this line at equal intervals (Fig. 2a).

Five sagittal cross-sectional images of a non-weight-bearing right foot were taken from a 27-years-old healthy female volunteer, using a 0.5T Open MRI scanner (General Electric Medical Systems Co., Milwaukee, USA). In order to locate the cross-sections as shown in Fig. 2, they were built on the subject’s transverse image of the plantar aspect of the foot (in correspondence to Fig. 2a). In addition to the above MRI scans, lateral X-rays of the non-weight-bearing and weight-bearing right foot were taken from the same subject (with a marked scale of dimensions) in order to experimentally measure skeletal displacements due to body load during standing, and thus, validate the computational predictions (the MRI clinical system does not allow scanning of a foot in the standing posture). The contours of the bony elements in each of the five MRI cross-sectional images were digitized by means of edge-detecting software, while allowing a maximal distance of 2 mm between two adjacent sampled points. A smoothing moving-average filter with a window width of 5 points was utilized to correct local errors of the automatic digitization without loss of

Fig. 1. Plantar fasciitis (a) is an inflammatory condition of the plantar fascia. When conservative treatments fail, a surgical procedure to release the plantar fascia may be carried out in which the entire plantar surface of the heel is displaced as a flap (b), the fascia is released, and the flap is sutured into place (modified from Duvries, 1978).

Fig. 2. The five selected planar cross-sections (marked as S₁–S₅) presented from (a) a plantar view, (b) an anterior view and (c) a posterior view.
meaningful geometrical information (differences between cross-sectional areas of bones in the original MRI and processed images did not exceed 1.5%). The geometric data of the bone’s contours were transferred to a commercial finite element (FE) analysis software package (ANSYS Co., Canonsburg, PA, USA) for construction of the five cross-sectional structures (Fig. 3).

Cartilage layers and ligaments forming the joints between adjacent bones were introduced to conform to the foot’s anatomy. The thickness of the ligaments and the locations of their insertions were based on anatomical descriptions (Gray, 1995). The cross-sectional shape of all the ligaments was assumed to be rectangular, their thickness was determined as being 1.8 mm and the thickness of the plantar fascia was taken as being 2 mm, according to measurements of Simkin (1982) on cadaveric normal feet. It should be noted that the 3D “fan-like” structure of the real fascia towards the forefoot could not be accounted for while using the present planar modeling approach. It was further assumed that no space exists between the bones and the attached plantar tissue pad, and, similarly, that no space exists between the fascia and the soft tissue above it.

2.2. Material properties

Bony elements and cartilage were idealized to linear, perfectly elastic and isotropic materials, while ligaments, fascia and the soft tissue fat pad were considered as being non-linear materials. The Young modulus and Poisson ratio for bone were taken from Nakamura et al. (1981) as being 7300 MPa (weighing cortical and trabecular elasticity values) and 0.3, respectively. For the cartilage, the Young modulus and Poisson ratio were taken from Patil et al. (1996) as being 10 MPa and 0.4, respectively. The typical experimental non-linear stress-deformation relation for the ligaments was adapted from Race and Amis (1994) who used an Instron system to test normal lower-limb ligaments under tension. The mechanical behavior of the soft tissue pad was taken from Nakamura et al. (1981) who obtained the non-linear stress-deformation curve of a specimen taken from the heel of a fresh cadaver. For the computational procedures, the above experimental data were fitted to polynomial expressions in the form of

$$\sigma = a_1 \lambda^3 + a_2 \lambda^4 + a_3 \lambda^5 + a_4 \lambda^2 + a_5 \lambda + a_6$$  \hspace{1cm} (1)

with a correlation coefficient of $R^2 = 0.998$, where $\sigma$ is the stress in MPa, $\lambda$ is the resultant stretch ratio and the constants $a_i$ (in MPa) are detailed in Table 1. The non-linear material properties of the plantar fascia were determined based on the results of Kitaoka et al. (1994), who found that the stiffness of the intact fascia is about 200 N/mm, i.e., about 1.2 times the stiffness of the lower-limb ligaments (Race and Amis, 1994). Applying these findings to the present study, it was assumed that the stresses required to induce a given deformation of the plantar fascia are 120% the values of corresponding stresses in the constitutive law for the ligaments (Table 1). The Poisson ratios were taken as 0.4 for the ligaments and fascia (Chu et al., 1995) and as 0.49 for the soft tissue pad (Nakamura et al., 1981). In order to prevent ligaments from bearing compressive loads, the model was initially solved for a state in which all ligaments behaved according to the constitutive law given in Eq. (1). Subsequently, ligaments in which some compression was evident were identified, and their stiffness was set to zero (this was the case with most of the dorsal ligaments). Finally, the model was resolved while assuring that (i) ligaments carry only tension stresses and that (ii) no tensile stresses are taken by the cartilaginous layer beneath bone-cartilage contact

Fig. 3. The computational model of the human foot in the standing posture with its five cross-sectional structures (marked as S1–S5). Arrows indicate body loads and forces in the Achilles tendon. The constraints limiting vertical displacements of discrete nodal points are marked by triangles. LPL = long plantar ligament.
regions of each of the foot joints, precisely as in the living foot (Woo et al., 1987).

2.3. Loads and constraints

The total load carried by the foot model was determined to be 300 N. The external force system, representing the body load on each of the five cross-sections of the model in the standing posture, was applied by weighing the loads to the five rays to be distributed as 25%–19%–19%–19%–18% for the first through the fifth ray, respectively (Simkin, 1982). In addition to the body-weight surface load, a concentrated force was applied at the posterior aspect of the calcaneus to represent the effect of the Achilles tendon force due to contraction of the triceps surae muscle group (Fig. 3). Simkin (1982) calculated that this force should be approximately 50% of the body load during standing. Assuming that the load at the Achilles tendon divides equally between the five cross-sections, a vertical force of 30 N was applied at the tendon’s insertion point in each of the cross-sectional structures.

Supports constraining the model’s vertical displacements (Fig. 3) were positioned under the calcaneal base, under each of the five metatarsal heads and under the lateral metatarsal and cuboid bodies (Saltzman and Nauwoczenski, 1995). The maximal distance between adjacent constraints was determined to be 12 mm, identically to the distance between sensors of the contact pressure display (CPD) method for foot-ground pressure (FGP) measurements (Arcan, 1990; Brosh and Arcan, 1994; Gefen et al., 2000, 2001), to enable validation of model predictions according to experimentally obtained FGP patterns. Sliding of the structures in the horizontal direction was prevented by constraining a point under the center of the calcaneal base in each cross-sectional structure against translation displacement in that direction. Consequently, this point was also used as a reference for movements of all other nodal points.

2.4. Numerical method

The FE analysis resulted in a quantitative structural stress distribution in terms of the principal tension \( \sigma_1 \), the principal compression \( \sigma_2 \), and the von Mises equivalent stress \( \sigma_{\text{v,M}} \), which weighs the effects of principal tension and compression stresses according to the relation:

\[
\sigma_{\text{v,M}} = \left( \sigma_1^2 + \sigma_2^2 - \sigma_1 \sigma_2 \right)^{1/2}. \tag{2}
\]

Automatic division was used to generate five optimal meshes containing 1100–2000 planar structural elements for describing the curved geometry of bones, cartilages, ligaments, fascia and soft tissue (Fig. 4). The above meshes were determined by a converging process in which the mesh density was gradually increased until the deviation in the produced stress values did not exceed 5%. The selected 8-node elements allowed for large deflections and large strains. Each node had two degrees of freedom, i.e., translation in both the \( x \) and \( y \) directions, a state which is compatible with bending stresses.

2.5. Model validation

The FGP during standing, a fundamental characteristic of this posture, was selected as a basic validation criterion for the computational predictions. Accordingly, the percentage of weight carried by each foot segment was experimentally measured from 20 feet of 10 normal, healthy subjects (age range: 21–30 years) during standing, using the optical CPD method. The model's predictions of percentage of weight carried by the foot segments were compared with the experimentally measured ones (by employing a statistical \( t \)-test with a significance level of 5%), and no statistically significant differences were evident (Table 2).

The lateral X-rays of the weight-bearing and non-weight-bearing foot of the subject used for model geometry reconstruction were digitized, and the positions of the talus and metatarsal heads were detected on each radiograph. Distances between the talus apex and the calcaneus plantar aspect as well as those between the talus apex and the metatarsal heads were measured on both the weight-bearing and non-weight-bearing X-rays using image-processing software. The obtained lengths were used to calculate the deflection of the talus and the lengthening of the foot for comparison to the computational predictions. Weight-bearing feet of healthy, normal subjects generally show very small displacements during standing, i.e., mean talus depression of 0.15 mm and even smaller elongation of the foot’s rays (Carlsoo, 1964). The model simulations of displacement

<table>
<thead>
<tr>
<th>Tissue</th>
<th>( a_1 )</th>
<th>( a_2 )</th>
<th>( a_3 )</th>
<th>( a_4 )</th>
<th>( a_5 )</th>
<th>( a_6 )</th>
</tr>
</thead>
<tbody>
<tr>
<td>Ligaments</td>
<td>-412640.5</td>
<td>2235967.7</td>
<td>-4841544.8</td>
<td>5236972.7</td>
<td>-2829945.7</td>
<td>611190.6</td>
</tr>
<tr>
<td>Plantar fascia</td>
<td>-488737.9</td>
<td>2648898.5</td>
<td>-5736967.6</td>
<td>6206986.7</td>
<td>-3354935.1</td>
<td>724755.5</td>
</tr>
<tr>
<td>Soft tissue fat pad</td>
<td>0</td>
<td>59.2</td>
<td>-275.5</td>
<td>480.4</td>
<td>-371.9</td>
<td>107.7</td>
</tr>
</tbody>
</table>

Table 1

Material properties for the ligaments, plantar fascia and soft tissue fat pad of the model, specified as coefficients \( a_i \) (in MPa) of a polynomial stress-deformation relation, \( \sigma = a_1 \lambda^2 + a_2 \lambda^2 + a_3 \lambda^2 + a_4 \lambda + a_5 + a_6 \), where \( \sigma \) is the stress in MPa and \( \lambda \) is the stretch ratio.
distributions in the normal foot predicted a talus depression of 0.28 mm. The maximal lengthening of the foot was predicted to occur at the first ray where the first metatarsal head was displaced anteriorly by 0.13 mm. The radiological measurements also demonstrated a small talus depression of 0.08 mm, and an increase of 0.05 mm in length for the medial rays.

2.6. Simulation of surgical plantar fascia release

Simulation of partial and total release of the plantar fascia was carried out by gradually decreasing the fascia’s thickness in intervals of 25%, until it was completely detached. The distributions of structural stresses and displacements were calculated for these different simulated post-operative conditions. Reduction in thickness of the plantar fascia is very likely to increase the loads that are developed in the plantar ligaments in order to compensate for its decreased stiffness. In order to quantitatively characterize the biomechanical effects of surgical procedures for release of the fascia, the averaged stress values in two of the most important elements forming the foot’s arch, i.e., the deep long plantar ligament (LPL) and the superficial LPL (Fig. 3), were examined. The average stress developing in each of these ligaments was defined as

$$\bar{\sigma} = \frac{1}{S} \int_0^S \sigma_{v,M} \, d\xi,$$

where the linear course of length, $S$, is bounded by the centers of the insertion surfaces of the ligament into the bones to which it attaches.

The foot model was tested in a sensitivity analysis to quantify the influence of variations in soft tissue mechanical properties on the stability of the numerical predictions, in terms of the resulted foot displacements following simulated surgical release of the fascia. For any particular soft tissue component, i.e., cartilage, ligaments, and plantar fat pad, it was decided to carry out the sensitivity analysis by increasing or decreasing the stiffness of that component, keeping the material properties of all others at their neutral state (Table 1). For each tissue component, stiffness was varied by assuming that the stresses required to induce a given deformation alter by 25% and 50% in respect to the neutral cases.

3. Results

The resultant displacements of the normal, healthy foot (Fig. 5a) were shown to be generally small, reaching no more than 0.3 mm. The highest vertical deflection due to flattening of the foot’s arch under the body weight was found in the talonavicular joint (0.28 mm). The largest horizontal displacement appeared in the talus apex, and could be associated with the calcaneal rotation and shear deformation of the subtalar joint during body-load bearing. Maximal lengthening of the normal foot was found to occur at the first ray, where
the first metatarsal head was displaced anteriorly by 0.13 mm. The smallest elongation (of 0.08 mm) was shown to occur at the fifth ray. The long plantar ligaments were shown to exert significant local tension stresses (ranging between 28 and 45 kPa) in the regions of their anterior and posterior insertion surfaces (see cross-section S2 in Fig. 5b).

The isostatics that develop in the second ray of the normal foot (cross-section S2) are shown in Fig. 6. The isostatic representation uses a vector plot to describe how stresses are transferred in the structure. The bright arrows in the isostatic diagrams represent tension stress flow, while the dark arrows represent transfer of compression stresses. Several regions of interest, such as the talus, calcaneus, metatarsals and joints are magnified for a clear view (Fig. 6). Cartilages were shown to transfer only compression stress and ligaments were loaded only in tension, as expected.

Computational predictions of the effects of partial and total release of the plantar fascia were achieved by gradually removing parts of it in the model and then, calculating the resultant distributions of stresses and displacements. The vertical displacements at the apex of the talus and the resultant averaged stresses at the deep and superficial LPLs were obtained for different PFDR-simulated procedures (Table 3). Comparison of the results obtained for the normal foot with those of a simulated total fascia release predicted an additional 2.57 mm lowering of the arch under body weight following surgery. The predicted arch deformation \( \delta \) (mm) was shown to be exponentially correlated with the PFDR (expressed in percentage), according to the relation

\[
\delta = a \exp(b \text{PFDR})
\]

with a correlation coefficient of \( R^2 = 0.99 \), where the constant values are \( a = 0.2812 \) mm and \( b = 2.3691 \). As a result of a total surgical release, the predicted averaged stress in the superficial and deep LPLs increased by as much as 204% and 123%, respectively (Table 3). This procedure also amplified the tension stresses that were transferred to the calcaneus, cuneiform and metatarsal bones, thereby exerting stress concentrations in the vicinity of the insertion surfaces of these ligaments (Fig. 7).

The sensitivity of numerically predicted foot displacements to variations in material properties of the model’s soft tissue components was analyzed, and the results are detailed in Fig. 8. The decrease in arch height post-surgery was shown to range from 0.5 to 8 mm, and the elongation of the first ray of the foot varied between 0.03 and 1.2 mm, while stiffness of each soft tissue component was decreased from 1.5 to 0.5 times its neutral stiffness (Table 1). Alterations in cartilage and soft tissue pad stiffness were shown to have the most significant effect on the numerically predicted arch displacements. Increasing the stiffness of articular cartilage directly elevated the stiffness of joints, and caused decreased arch flattening during load-bearing. Thus, the maximal talus depression appeared to decrease significantly, by as much as 65%, as the cartilage stiffness increased to 1.5 times its neutral value. Lengthening of foot (measured at the first ray) was also mostly affected by the stiffness of the cartilage, and decreased by 77% as the cartilage’s stiffness was raised to 1.5 times its neutral stiffness. Increase of ligament stiffness had a similar, although less predominant effect on foot flattening during standing, inducing a decrease of about 38% in the predicted foot lengthening for ligaments that are 1.5 stiffer than the neutral ones.

4. Discussion

A computational FE model of the foot in the standing posture is presented, for simulation of post-surgical conditions of plantar fascia release. The results of the
simulation of a total surgical release of the fascia demonstrated significant rise in stresses carried by the LPLs. Extensive deformation of the arch (2.85 mm), i.e., to more than 10 times the predicted normal arch deformation, was observed as well. These computational results are in good agreement with the measurements of Sharkey et al. (1998), who demonstrated that complete fascia release in cadaveric feet caused a 2.5 mm decrease in the mean arch height under loading conditions which simulated the push-off stage of gait. Kitaoka et al. (1997) measured larger decreases in arch height of cadaveric feet, ranging between 3.3 and 11.5 mm, and
reported that the changes in arch height were more pronounced in the unstable or destabilized feet. This observation may be well correlated with the present sensitivity analysis, which demonstrated that the decrease in arch height following plantar fascia release is much more predominant in a foot with less stiff joints or ligaments, where it may reach up to 8 mm deformation (Fig. 8). In view of these experimental and computational findings, it is suggested to avoid surgical plantar fascia release in patients with tendency to a more flexible arch (e.g., with evidence of pes planus deformity), because of the likelihood of further significant deformation.

Plantar fasciitis is mainly associated with athletes, but it can affect anyone involved in intensive physical activities (Lutter, 1986). A significant decrease of the arch height post-surgery, as predicted by the model, may reduce the dynamic shock-absorbing abilities of the foot, and cause further musculoskeletal damage, as shown in clinical studies where subjects with flat feet could not sustain long marches, and were at higher risks for developing stress fractures (Simkin and Liechter, 1990; Brosh and Arcan, 1994; Kim and Voloshin, 1995). Moreover, the plantar fascia has an important role in relieving metatarsal stresses. The dorsal aspects of the medial metatarsals are normally loaded in compression

Table 3
Predicted deformation of the foot’s arch and long plantar ligament (LPL) averaged stresses for different simulated plantar fascia degrees of release (PFDR)

<table>
<thead>
<tr>
<th>PFDR (percentage)</th>
<th>Arch deformation (mm)</th>
<th>Deep LPL stress (kPa)</th>
<th>Superficial LPL stress (kPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>0 (normal foot)</td>
<td>0.28</td>
<td>35</td>
<td>25</td>
</tr>
<tr>
<td>25</td>
<td>0.48</td>
<td>39</td>
<td>30</td>
</tr>
<tr>
<td>50</td>
<td>0.99</td>
<td>47</td>
<td>42</td>
</tr>
<tr>
<td>75</td>
<td>1.73</td>
<td>60</td>
<td>57</td>
</tr>
<tr>
<td>100 (total release)</td>
<td>2.85</td>
<td>78</td>
<td>76</td>
</tr>
</tbody>
</table>

Fig. 7. Simulated distribution of principal tension stresses following total release of the plantar fascia.

Fig. 8. The dependency of post-surgical decrease of arch height (circles) and lengthening of the foot (squares) in alteration of the stress-strain behavior of the articular cartilage (top), ligaments, (center) and plantar soft tissue (bottom). For each tissue component, stiffness was varied by assuming that the stresses required to induce a given deformation alter by 25% and 50% in respect to the stress-strain relation that was defined as being the neutral case.
Removal of the fascia elevated the bending loads on these bones, and, thereby, increased the dorsal compression stresses by as much as 65%. This suggests that release of the fascia will accelerate fatigue damage to these bones during intensive activity such as marching and hence, when surgical removal or significant release of the fascia is considered, a decrease in the foot’s abilities of load-bearing and shock-attenuation should be taken into account.

The model presented in this work may be further applied to investigate numerous acquired foot deformities or traumatic injuries, and the mechanisms acting in such conditions could be studied by altering its geometrical or material properties. Moreover, various new or conservative surgical interventions could be evaluated by removing or adding elements to the computational simulation. Orthotics and supportive devices may also be assessed, and their effect on the internal and foot-ground contact stress distributions can be studied. When used together with FGP experimental measurements, the present computational foot model can be a highly effective biomechanical tool with clinical applications in pre- and post-treatment evaluations.

Acknowledgements

With this paper, I wish to honor the memory of the late Prof. Mircea Arcan (deceased June, 2000), my academic supervisor, who motivated and guided my work in the field of musculoskeletal biomechanics.

References


