Biomechanical response of the plantar tissues of the foot in healthy and degenerative conditions

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Summary

Introduction: This work reports a set of results on the mechanical response of foot plantar tissues in healthy and degenerative conditions during the gait cycle, by means of a computational approach.

Methods: A three dimensional finite element model of the foot was developed starting from the analysis of biomedical images. Different constitutive models were defined to interpret the mechanical response of the biological tissues. In particular, a specific visco-hyperelastic constitutive formulation was provided for foot plantar soft tissue considering the general features of tissue mechanics. Degenerative phenomena induce histomorphological alterations and modification of the mechanical properties, as stiffening and lower damping capabilities. Different constitutive parameters for healthy and degenerative conditions were identified by the inverse analysis of experimental data from mechanical tests.

Results: The three dimensional numerical model interprets the capability of the plantar soft tissue to act under mechanical actions in different conditions during the phases of the gait cycle.

Conclusion: Numerical results highlight aspects of the different stress and strain distributions at the heel strike and the midstance of the healthy and degenerative conditions.

Level of evidence: V.

KEY WORDS: plantar soft tissue, degenerative phenomena, gait cycle, finite element analysis.

Introduction

Foot plantar soft tissue can be affected by various degenerative phenomena determined by traumas, congenital or inflammatory infections and neoplastic disorders. One of the major responsible of foot disorder is the diabetes mellitus and its complications that are associated with several structural and physiological abnormalities^{1,2}. A variation in mechanical functionality is determined with particular effects on the response during the gait cycle.

The plantar soft tissue is characterized by a specific configuration in which fibrous septa envelope adipose compartments^{1,3}. Collagen fibres within septa are spirally arranged and fixed to bone or inferiorly to dermis. The adipose chambers contain packed fat cells arranged in micro lobules. Lobules are grouped by further bundles of collagen and elastin fibres. This complex system, composed of connective and adipose components, operates for proper functionality, mostly to attenuate ground impact and to redistribute plantar pressures⁴. The degenerative effects, as aging, pathology or trauma, cause histo-morphological and mechanical changes^{1,3}. Degenerative phenomena can determine a gradual loss of collagen, a decrease in the quantity of elastin and a reduction of liquid content. This is associated with a local degradation of fat tissue and damage of fibrous tissue septa through the distortion and the fragmentation of the fibrous proteins. Furthermore, increased fat content correlated with obesity leads to a growing local stiffness. Pathologies, such as diabetic or atrophied phenomena, cause alterations in the ratio of saturated and unsaturated fatty acids. Fat globules decreases in size and number and also a reduction of the number of septal fibres take place. Besides, the connective septa become thicker. Experimental tests on foot tissues and structures in degenerative conditions confirm the presence of stiffer soft tissue⁵⁻⁹, a less smoothed plantar pressure distribution with higher local values and a reduction in the ability to absorb impulsive loads.

Numerical analyses can be adopted to obtain accurate and reliable information on foot mechanics. Computer modeling, particularly finite element modeling, has the capability to characterize internal stress states and their evolution within foot structure under different static and dynamic loading scenarios. Two-dimensional and three-dimensional finite element models of the foot have been developed over the years, to interpret stress-related injuries. Furthermore, numerical modelling approach is a robust tool to evaluate local interaction phenomena between foot and footwear with primary reference to insole. Some studies report the influence of specific footwear characteristics, recalling materials properties and geometrical configuration¹⁰⁻¹³. The adoption of two-dimensional models determines relevant approximations in the evaluation of the overall mechanical problem^{5,10}. With regard to the reported three-dimensional models, constitutive formulations, as linear elastic or hyperelastic models^{11,12}, have been usually assumed to characterize plantar soft tissue mechanics.

The aim of this work is to analyze aspects of the mechanical behavior of the healthy and degenerative plantar soft tissue, using a numerical approach and considering the main function of the plantar soft tissue during walking. The biomechanical role consists in an efficient shock absorption, mostly to redistribute the forces induced during gait. The evaluation of the compressive stress and strain fields, obtained by the numerical analysis, allows the identification of the variation of the functional response in dependence on degenerative conditions due to aging or pathologies. Recalling previous research activities in foot biomechanics¹³⁻¹⁵, a refined finite element model of the foot was developed, considering the real structural conformation and behavior of the biological system. Specific constitutive formulations for the different biological tissues were developed and identified. A visco-hyperelastic model was developed for plantar soft tissue, which plays a prominent role in foot biomechanics and whose degeneration may affect the overall functionality of the foot¹⁶. Subsequently, different sets of constitutive parameters of the visco-hyperelastic model were identified to describe the mechanical response of the healthy and degenerative conditions. Finite element analyses were performed to investigate the plantar soft tissue response during the gait cycle. Loading and boundary conditions were applied considering average results from experimental data reported in literature^{13,17,18}. The numerical results allowed to identify the different stress and strain fields within the tissue and to appreciate the variation of the mechanical performance of the overall system.

Materials and methods

Finite element model of the foot

The definition, development and validation of the solid and numerical models of the foot entail a relevant effort and represent part of a research activity described in previous papers^{13,17,18}, based on an av-

eraged configuration of the human foot. The solid model of the foot was developed based on MRI and CT data. Imaging density segmentation software¹³, which applies the density segmentation techniques on the DICOM images, was adopted to obtain 3D CAD solid models. Subsequently, the solid model was morphed according to anthropometric data to address an averaged configuration. The solid model of the foot was composed of the tibia and fibula and the full set of foot bones, cartilage, soft tissues (as muscles and connective tissues), plantar soft tissue and skin (Fig. 1). A fully three-dimensional representation of ligaments was adopted for the ankle structures and the plantar fascia. The virtual solid model was meshed by about one million and three hundreds four node tetrahedral elements, using the software MSC-Patran 2007 (MSC Software Corporation, Santa Ana, CA).

Constitutive formulation

The mechanical behavior of the bones was defined by an orthotropic, linear elastic model¹⁸, while the cartilage mechanics was described by a hyperelastic model considering data from literature¹⁹. The mechanical behavior of ankle ligaments and plantar fas-

Figure 1 a,b. Representation of the numerical model of the foot: overall bony components (a) and general sagittal section of the foot (b).

cia was defined by specific visco-hyperelastic formulations¹⁶. A further hyperelastic formulation was adopted to describe the mechanical behavior of soft tissues surrounding the bony segments in the surrounding regions⁵. A fiber-reinforced hyperelastic model was assumed to define the anisotropic and non-linear mechanical behavior of the skin²⁰.

The plantar soft tissue exhibits a non-linear visco-elastic behaviour, material and geometric non-linearity, almost incompressible and time dependent response, as deeply investigated in literature^{5,6,8,15,21}. The constitutive model adopted at the purpose, with the consideration of both healthy and degenerative conditions, is the visco-hyperelastic constitutive model, as already described in other works²². The formulation was developed considering the equilibrium behaviour of the tissue and the contribution of viscous terms:

$$\mathbf{S}(\mathbf{C}, \mathbf{q}^i) = \mathbf{S}^\infty(\mathbf{C}) + \sum_{i=1}^n (\mathbf{S}^i(\mathbf{C}) - \mathbf{q}^i(t)) \quad (1)$$

where $\mathbf{S}$ is the second Piola-Kirchhoff stress tensor, while $\mathbf{C}$ is the right Cauchy-Green strain tensor. $\mathbf{S}^\infty$ specifies the equilibrium response of the material and it is provided by an hyperelastic potential W^∞ , as $\mathbf{S}^\infty = 2\partial W^\infty$. Because of the almost incompressible behaviour of the tissue, the hyperelastic potential W^∞ can be split into volumetric U^∞ and iso-volumetric W^∞ terms:

$$U^\infty(I_3) = \frac{K_v}{2+r(r+1)} \left[(I_3^{1/2} - 1)^2 + I_3^{-r/2} + rI_3^{1/2} - (r+1) \right] \quad (2)$$

$$W^\infty(I_1) = \frac{C_1}{\alpha} \left\{ \exp[\alpha(I_1 - 3)] - 1 \right\} \quad (3)$$

where I_1 is the first iso-volumetric invariant of the right Cauchy-Green strain tensor, as $I_1 = \text{tr}(I_3^{-1/3}\mathbf{C})$, and I_3 is the third invariant, as $I_3 = \det(\mathbf{C})$. Constitutive parameters K_v and C_1 are related with the equilibrium initial volumetric and shear stiffness, respectively. Parameters r and α characterize the evolution of material stiffness with strain because of the non-linearity of the mechanical response. The influence of viscous phenomena was defined by the stress components $\mathbf{S}^i$ and the viscous variables $\mathbf{q}^i$, which respectively specify the instantaneous stress contribution, as

$$\mathbf{S}^i(\mathbf{C}) = (\gamma^i / \gamma^\infty) \mathbf{S}^\infty(\mathbf{C}) \quad (4)$$

and the damping of the stress contribution itself, as:

$$\dot{\mathbf{q}}^i(t) = \frac{\gamma^i}{\gamma^\infty \tau^i} \int_0^t \exp\left(-\frac{t-s}{\tau^i}\right) \mathbf{S}^\infty(\mathbf{C}) ds \quad (5)$$

where τ^i are relaxation times, as measures of time for the viscous phenomena development, γ^i are relative stiffness defining the contribution of the viscous processes, and $\gamma^\infty = 1 - \sum_{i=1}^n \gamma^i$ is the balance equation.

The identification of the different sets of constitutive parameters for healthy and degenerative conditions is performed in accordance with the procedure already proposed in^{8,16}. The constitutive parameters of the plantar soft tissue were identified by analysing *in vitro* and *in vivo* experimental tests reported in literature²², as described in²². Viscous constitutive parameters were assumed according to¹⁵ for both healthy and degenerative conditions, because experimental tests showed that degenerative processes do not significantly affect viscous phenomena⁶. The reliability of the foot model has been previously assessed by comparing results from numerical analyses and data from experimental investigations, as reported in other works^{15,21}.

Numerical analyses

Specific numerical analyses were developed to evaluate the overall mechanical response of the foot structures during different phases of the gait cycle^{13,16,18}. The first analysis interpreted the initial phase of the gait cycle up to the heel strike. The boundary and loading conditions of the numerical analysis were evaluated considering results from experimental activities^{13,17}. A coordinate system was defined considering the X-axis and the Z-axis as two perpendicular axis on the plane and the Y-axis as the axis perpendicular to the plane (Fig. 1). The sagittal-plane angle between foot and plane was about of 14 degrees. A relative motion between the foot and the plane was imposed along Y direction.

The second analysis was performed to simulate the midstance. The numerical analysis concluded when the reaction force evaluated on the tibia and fibula reached the 105% of the body weight.

The transient peak of the heel strike and of the midstance was estimated considering a gait velocity of 1.44 m/s for healthy condition and of 1.35 m/s for degenerative plantar soft tissue²³. The interaction phenomena between ground and foot skin were specified by a Coulomb contact condition, according to a 0.42 friction coefficient²⁴.

The analyses were developed using the general-purpose finite element software ABAQUS 6.14 (Dassault Systèmes Simulia Corp., Providence, RI). The specific constitutive models were implemented with the use of *ad hoc* routines capable to interpret the behaviour of the tissues in consideration of the specific formulations developed.

Results

The mechanical response of degenerative plantar tissue is characterized by an increment in stiffness, correlated with a variation in plantar pressure distribution and a reduction in the ability to absorb impulsive loads. The visco-hyperelastic constitutive model defined with adoption of specific constitutive parameters interprets the degenerative mechanical behavior that was compared with healthy conditions. Numerical results were reported for the evaluation of the stiffness

variation due to degenerative condition, during the gait cycle loading. For the evaluation of the mechanical response of foot tissues, two steps of the gait cycle were chosen: the heel strike and the midstance (Fig. 2). The contours of numerical results with regard to the minimum principal stretch field were reported for the heel strike and the midstance (Fig. 2). The results were reported over a sagittal section of the foot in correspondence of the second metatarsal head for healthy condition. The results show the increase of strain from the heel strike to the midstance. The contours of numerical results with regard to the minimum

Figure 2 a,b. Indication of the vertical component of the overall ground reaction force (a). Results from the numerical analysis at the heel strike (a) and at the midstance (b): contours of the distributions of the minimum principal strain on a sagittal section at the second metatarsal.

principal stretch and stress fields over a transversal section A-A in correspondence of the calcaneus were reported for the heelstrike (Fig. 3). The results highlight the decrease of strain with the progression of the degeneration and, however, the improvement of the stress intensity, to confirm the increase of local tissue stiffness.

The minimum principal stretch and stress in the plantar soft tissue for healthy and degenerative conditions was reported in correspondence with the foot extension from left to right on the plantar soft tissue (Fig. 4). The plot compares the distribution of stretch and of the stress during the midstance and highlights a reduction of the strain and an increase of the stress because of the major stiffness of the degenerative plantar soft tissue.

Discussion

The investigation of foot biomechanical response is faced with relevant problems, related to the complex structural configuration and the proper identification of tissues mechanics. In general, three-dimensional numerical models allow for a reliable representation of variable loading conditions and tissue properties with improved extension and accuracy with regard to the even repetitive and difficult action requested by experimental testing. The results in terms of strain and stress confirm the potentiality of the numerical approach that must be supported by basic information of mechanical testing data.

The onset of degenerative phenomena due to aging, pathology or trauma determines size reduction of adipose chambers, septa thickness increase, fragmentation of collagen bundles and elastic strands. Such histological changes of the tissue structural conformation entail a modification of mechanical properties. In literature, the tissue stiffening because of degenerative phenomena was evaluated by the heel pad compressibility index (HPCI). Results from experimental tests^{8,9} and from computational activities^{25,26} showed an increase of HPCI for degenerative conditions. The results achieved with numerical analysis developed during the heel strike and the midstance confirm the data reported in literature.

The relevant efforts to develop appropriate models of all the foot structures and tissues, as bones, cartilage, soft tissues and skin determine some limitations of the numerical model.

A note must be reported also on the loading and boundary conditions adopted. The comparison between the gait cycle for healthy and pathological conditions was deeply investigated and reported in Hazari et al.²⁷. Some differences were highlighted in terms of gait velocity, stance period, stride length and plantar pressure. Pathological patients showed a lower gait velocity. At the same time, the ground reaction forces (GRF) at initial contact and during the gait cycle did not show significantly differences. Numerical results were reported in terms of minimum principal stretch at the same GRF with a different gait cycle,

showing the growing stiffness of degenerative plantar soft tissue.

This investigation is part of a more extended research activity developed by the Authors and reported in different works. Preliminary, in²¹ the specific constitutive parameters of plantar soft tissue in healthy condition were defined starting from *in vitro* and *in vivo* investigations by different Authors^{5-7,22} on tissue samples from a large set of subjects. Subsequently, the numerical model of the foot was defined and numerical

analyses that interpreted specific experimental activities were developed¹⁵. The influence of degenerative phenomena on the mechanical behaviour of the plantar soft tissue were investigated by the correlation between histological data, experimental tests and numerical analyses¹⁶. The present work represents an additional contribute, in spite of the limits assumed, considering the influence of degenerative phenomena on the plantar soft tissue during the gait cycle, joining the studies developed on the foot over time²⁰. The

Figure 4 a-c. Results from the numerical analysis: distributions of the minimum principal stretch (a) and stress (b) for the midstance of the gait cycle, along a path from a sagittal section at the second metatarsal.

procedure can be associated also to a computational tool for the investigation of the biomechanics of the healthy and degenerative conditions in relation to the interaction phenomena occurring between foot and footwear products, to estimate reliability and comfort as relevant tasks for industrial applications.

Ethic

The Authors declare that this research was conducted following basic ethical aspects and international standards as required by the journal and recently update in²⁸.

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