Muscle-driven finite element simulation of human foot movements

L.A. Spyroua,b and N. Aravasa,b*

aDepartment of Mechanical Engineering, University of Thessaly, Volos 38334, Greece; bThe Mechatronics Institute, Centre for Research and Technology – Thessaly (CE.RE.TE.TH.), 1st Industrial Area, Volos 38500, Greece

(Received 20 December 2010; final version received 23 February 2011)

This paper describes a finite element scheme for realistic muscle-driven simulation of human foot movements. The scheme is used to simulate human ankle plantar flexion. A three-dimensional anatomically detailed finite element model of human foot and lower leg is developed and the idea of generating natural foot movement based entirely on the contraction of the plantar flexor muscles is used. The bones, ligaments, articular cartilage, muscles, tendons, as well as the rest soft tissues of human foot and lower leg are included in the model. A realistic three-dimensional continuum constitutive model that describes the biomechanical behaviour of muscles and tendons is used. Both the active and passive properties of muscle tissue are accounted for. The materials for bones and ligaments are considered as homogeneous, isotropic and linearly elastic, whereas the articular cartilage and the rest soft tissues (mainly fat) are defined as hyperelastic materials. The model is used to estimate muscle tissue deformations as well as stresses and strains that develop in the lower leg muscles during plantar flexion of the ankle. Stresses and strains that develop in Achilles tendon during such a movement are also investigated.

Keywords: biomechanics; human foot; foot and lower leg; plantar flexion; foot movement; finite element modelling

1. Introduction

Foot and lower leg structures, including bones, ligaments and muscle-tendon systems work together to develop various movements. Most available computer models of muscle-driven human locomotion use rigid-body dynamics and model muscles using a series of line segments (e.g. Delp et al. 1990; Anderson and Pandy 2001; Erdemir et al. 2007); such models are unable to predict local tissue loadings, shape changes of soft tissues, and interactions of musculotendinous units with the surrounding soft tissues and bones during human movement.

On the other hand, three-dimensional anatomically detailed finite element models of human foot are broadly used to estimate internal stresses and strains on hard and soft tissues during standing or gait, to study the effects of pathologies and injuries, and to design insoles and footwear (Gefen et al. 2000; Chen et al. 2001; Camacho et al. 2002; Cheung et al. 2005; Cheung and Zhang 2006; Goske et al. 2006; Spyrou 2006). The development of three-dimensional finite element models of biological structures including all the appropriate information about tissues' geometrical complexities is important to better estimate critical internal loadings and their exact location on tissues as well as tissue shape deformations.

In a three-dimensional finite element analysis, the detailed representation of geometry is accompanied by the appropriate constitutive equations that describe the mechanical behaviour of biological materials. Although, the mechanical properties of skeletal components or other passive materials can be well defined in an organ-level finite element model, the situation is not as clear for human muscle–tendon systems due to their complex active and passive behaviour; realistic, three-dimensional constitutive models that account for the active muscle behaviour have been developed only recently (Martins et al. 1998; Johansson et al. 2000; Yucesoy et al. 2002; Lemos et al. 2004; Tsui et al. 2004; Blemker et al. 2005; Liang et al. 2006; Ito et al. 2010; Spyrou and Aravas 2011). This seems to be a reason that muscles of the lower leg, whose active mechanical behaviour is responsible for various foot movements, are missing from the finite element models of the foot. Instead, kinematic or kinetic data are used and any motion and internal stressing is due to externally applied displacements or loads. The representation of muscles as three-dimensional volumes accompanied by their active behaviour to image-based models of biological structures is a basic step towards realistic modelling of the musculoskeletal system: it provides a better description of the whole structure loadings during movement, and leads to an accurate prediction of their behaviour (Blemker et al. 2007).

In this paper, an attempt is made to produce a natural human foot movement by incorporating realistic geometrical characteristics and material properties for...
muscles and tendons in a three-dimensional anatomically
detailed finite element model of foot and lower leg. To our
knowledge, this is the first time a simulation of foot
movement, in the context of the finite element method,
based entirely on the idea of considering muscles as
‘active’ materials and not by applying external forces on
tissues is presented.

The purpose of this paper is to (i) simulate plantar
flexion of the human ankle, (ii) predict Achilles tendon
stress and strain distributions during foot movement and
(iii) estimate the stress and strain state of the contracted
plantar flexor muscles. Both active and passive behaviour
of muscles is considered and movement results exclusively
from muscle contraction without applying any external
loads. Bones, ligaments, articular cartilage and the rest soft
tissues (mainly fat) are included in the model. The
advantage of the current approach is that a simulation of
foot movement is accompanied by an estimation of local
internal loadings in each tissue and prediction of their
shape deformation.

Standard notation is used throughout. Boldface symbols
denote tensors the orders of which are indicated by the
context. The usual summation convention is used for
repeated Latin indices of tensor components with respect to a
fixed Cartesian coordinate system. Let \(a, b\) be vectors, \(A, B\)
second-order tensors and \(C\) fourth-order tensor; the
following products are used in the text:

\[
(ab)_{ij} = a_i b_j, \quad A : B = A_{ij} B_{ij}, \quad (AB)_{ijkl} = A_{ij} B_{kl}, \quad (C : A)_{ij} = C_{ijkl} A_{kl}.
\]

2. Methods

The process of developing biomechanical finite element
models begins with the acquisition of data that will be used
to define the three-dimensional geometry of the biological
tissues (Section 2.1). In the finite element analysis, the
tissues included in the model are assigned specific material
characteristics as described in Section 2.2. The formul-
lation is completed by describing the finite element
analysis and prescribing the appropriate boundary
conditions of the problem (Section 2.3).

2.1 Geometry representation

In this study, CT scans with intervals of 1 mm (nearly 500
images) were chosen to represent the geometry of the right
foot and lower leg of a normal female individual of age 28
in the neutral foot position. Three-dimensional data-sets
are acquired and segmented using the AMIRA v4.1
software to describe the boundaries of skeleton, muscles,
tendons and skin surface. Two characteristic segmented
CT images are shown in Figures 1 and 2.

Three-dimensional surfaces are calculated directly
from the reconstruction of the segmented images. Each
surface is described by a set of triangles in a three-
dimensional space. The desirable volumetric mesh results
by filling with tetrahedra the volume enclosed by each
surface (Figure 3). The model used in the current analysis
consists of 51,113 nodes and 271,622 four-node
tetrahedral first-order solid elements, with a total of 153,
339 degrees of freedom. For comparison purposes,
calculations with a smaller number of elements were
carried out; the results showed that the aforementioned
finite element mesh provides an accurate representation of
the tissues’ geometrical characteristics as well as
convergent numerical results.

2.2 Materials

In this study, muscles and tendons are considered as
continuum composite materials that consist of fibres
surrounded by connective tissue and biofluids. The
constitutive model used for these materials has been
developed recently by Spyrou and Aravas (2011) and is
described briefly in the following. The model is based on
an idea put forth by Liang et al. (2006) for the mechanical
behaviour of muscular hydrostats and expanded to skeletal
muscles and tendons.

Let \(\varepsilon^f\) be the part of the Eulerian logarithmic strain
tensor that is associated locally with the volume preserving
axial deformation in the current local direction of the fibre.
At every continuum point in the tissue \(\varepsilon^f\) is given by the
expression (Spyrou and Aravas 2011)

\[ \varepsilon^f = \frac{\varepsilon_m}{2} (3\mathbf{m}m - \mathbf{I}), \]  

(1)

where \( \mathbf{m} \) is the unit vector defined locally along the deformed fibre (Figure 4), \( \varepsilon_m \) is the local axial logarithmic strain in the fibre and \( \mathbf{I} \) is the second-order identity tensor.

The difference between the total strain \( \varepsilon \) and the fibre strain \( \varepsilon^f \) is associated with the connective tissue and the biofluids surrounding the fibres:

\[ \varepsilon^{ct} = \varepsilon - \varepsilon^f. \]  

(2)

In particular, \( \varepsilon^{ct} \) represents the extent to which the total strain is locally not axisymmetric with respect to the fibre and is not volume preserving, i.e. it represents any area change transverse to the muscle fibre due to a local volume change in the ‘composite’ material and any local transverse and axial shear relative to the fibre.

If the fibre strain (1) were the only strain in the tissue, then the corresponding local true stress tensor would be a uniaxial stress in the local direction of the fibre \( \mathbf{m} \), i.e.

\[ \sigma^f = \sigma^m \mathbf{mm}, \]  

(3)

where \( \sigma^m \) is the true stress in the fibre direction.

The nominal longitudinal stress in the muscle fibre \( \sigma_0^m \) has an active and a passive part and is written in the form

\[ \sigma_0^m = \sigma_0^{m(\text{act})} + \sigma_0^{m(\text{pas})}, \]  

(4)

where

\[ \sigma_0^{m(\text{act})} = \sigma_{\text{max}} f_s f_c (\varepsilon_0^m, f_a \varepsilon_c), \]  

(5)

Figure 2. (a) CT scan of the lower leg. (b) Segmented CT scan of the lower leg.

Figure 3. (a) Visualisation of the foot and lower leg. Volume finite element meshes: (b) soft tissues (mainly fat), (c) skeleton and (d) skeleton and muscles with tendons.
the nominal longitudinal strain rate
stress on the nominal longitudinal strain rate
fibre stress $s$
on calcium level
Figure 4. Tissue fibre in the deformed configuration with its
depth of $f_0$ at the activation level, $f_0$ describes the dependence of the active muscle stress to the nominal longitudinal strain rate $e^{m}_0$, $f_0$ describes the dependence of the passive stress on the nominal longitudinal strain $e^{m}_0$ and the activation level, and $f_{pr}$ is the function that relates the passive muscle stress to the nominal longitudinal strain rate $e^{m}_0$. The functions ($f_c, f_r, f_p, f_{pr}$) are all dimensionless and normalised so that

$$
\max f_c(e^{m}_0, f_a) = f_c(e_{0a}, f_a) = 1 \text{ and } f_t(0) = 1,
$$

where $e_{0a}$ is the value of the nominal strain that corresponds to the optimal fibre length at the activation level $f_a$. Lloyd and Besier (2003) suggest a relationship of the following form for $e_{0a}$

$$
e^{m}_{0a}(f_a) = k(1 - f_a),
$$

where $k$ is a constant. The dependence of passive tension $f_p$ on calcium level $f_a$ is included in the model. Spyrou and Aravas (2011) suggest a relationship of the following form for $f_p$,

$$
f_p(e^{m}_0) = f_p(e^{m}_{0a}, f_a) = (1 + cf_a)f_p(e^{m}_{0a}),
$$

where $c$ is a constant. The true stress $\sigma^m$ that appears in (3) is determined from the corresponding nominal stress $\sigma^m_0$ as

$$
\sigma^m = \exp(\epsilon_m)\sigma^m_0.
$$

In the general case, additional stresses $\sigma^{ct}$ develop due to the deformation $\epsilon^{ct}$ of the connective tissues and the biofluids. As a first approximation, we assume that the total true stress $\sigma$ in the tissue can be written as the sum of $\sigma^f$ and $\sigma^{ct}$:

$$
\sigma = \sigma^f + \sigma^{ct}.
$$

In the present model, an isotropic linear hyperelastic model is used for the non-fibre part of the muscle and the constitutive equation for $\sigma^{ct}$ is written in the form

$$
\sigma^{ct} = \frac{1}{J} C^e : \epsilon^{ct},
$$

where $J$ is the determinant of the deformation gradient tensor and $C^e$ is the fourth-order isotropic elasticity tensor.

The constitutive model described above is non-linear, rate-dependent and anisotropic. The corresponding constitutive model for the tendons is the same as that of the muscle described above, with the nominal fibre stress having only a passive part, i.e.

$$
\sigma^m_0 = \sigma^{tend}(e^{m}_0),
$$

where $\sigma^{tend}(e^{m}_0)$ describes the dependence of the passive stress on the nominal longitudinal strain in the tendon $e^{m}_0$.

The constitutive model for muscles and tendons is implemented in the ABAQUS general purpose finite element programme (ABAQUS 2007). This code provides a general interface so that a particular constitutive model can be introduced as a ‘user subroutine’ (UMAT). The ‘implicit’ version of ABAQUS is used, in which the finite element formulation is based on the weak form of the balance of linear momentum, the solution is carried out incrementally, and the discretised nonlinear equations are solved using Newton’s method. The Jacobian of the equilibrium Newton-loop requires the so-called ‘linearisation moduli’ of the algorithm that handles the constitutive equations for the muscles and tendons; these moduli are defined in terms of a fourth-order tensor $C$ that relates the variation of stress $\partial \sigma$ to the variation of strain $\partial \epsilon$ over the increment under consideration:

$$
\partial \sigma = C : \partial \epsilon.
$$

The derivation of $C$ is lengthy and the details of the derivation are reported in Spyrou (2009). Here, we note that $C$ can be approximated by

$$
C = C^f + C^{ct},
$$

Figure 4. Tissue fibre in the deformed configuration with its
 direction defined locally by the unit vector $m$. Also shown is the
 fibre stress $\sigma^m$ acting on an infinitesimal fibre segment of length $ds$. 

The nominal longitudinal strain rate $f_a$ is the activation
 state which describes the pattern of the activation signal as
 a function of time and takes values in the range 0 $\leq f_a \leq$ 1, $f_c$ describes the dependence of the active stress on the nominal longitudinal strain $e^{m}_0 = \exp(e^{m}_0) - 1$ and accounts for the variation of the optimal fibre length on the activation level, $f_r$ is the function that relates the active muscle stress to the nominal longitudinal strain rate $e^{m}_0$, $f_p$ describes the dependence of the passive stress on the nominal longitudinal strain $e^{m}_0$ and the activation level, and $f_{pr}$ is the function that relates the passive muscle stress to the nominal longitudinal strain rate $e^{m}_0$. The functions ($f_c, f_r, f_p, f_{pr}$) are all dimensionless and normalised so that

$$
\max f_c(e^{m}_0, f_a) = f_c(e_{0a}, f_a) = 1 \text{ and } f_t(0) = 1,
$$

where $e_{0a}$ is the value of the nominal strain that corresponds to the optimal fibre length at the activation level $f_a$. Lloyd and Besier (2003) suggest a relationship of the following form for $e_{0a}$

$$
e^{m}_{0a}(f_a) = k(1 - f_a),
$$

where $k$ is a constant. The dependence of passive tension $f_p$ on calcium level $f_a$ is included in the model. Spyrou and Aravas (2011) suggest a relationship of the following form for $f_p$,

$$
f_p(e^{m}_0) = f_p(e^{m}_{0a}, f_a) = (1 + cf_a)f_p(e^{m}_{0a}),
$$

where $c$ is a constant. The true stress $\sigma^m$ that appears in (3) is determined from the corresponding nominal stress $\sigma^m_0$ as

$$
\sigma^m = \exp(\epsilon_m)\sigma^m_0.
$$

In the general case, additional stresses $\sigma^{ct}$ develop due to the deformation $\epsilon^{ct}$ of the connective tissues and the biofluids. As a first approximation, we assume that the total true stress $\sigma$ in the tissue can be written as the sum of $\sigma^f$ and $\sigma^{ct}$:

$$
\sigma = \sigma^f + \sigma^{ct}.
$$

In the present model, an isotropic linear hyperelastic model is used for the non-fibre part of the muscle and the constitutive equation for $\sigma^{ct}$ is written in the form

$$
\sigma^{ct} = \frac{1}{J} C^e : \epsilon^{ct},
$$

where $J$ is the determinant of the deformation gradient tensor and $C^e$ is the fourth-order isotropic elasticity tensor.

The constitutive model described above is non-linear, rate-dependent and anisotropic. The corresponding constitutive model for the tendons is the same as that of the muscle described above, with the nominal fibre stress having only a passive part, i.e.

$$
\sigma^m_0 = \sigma^{tend}(e^{m}_0),
$$

where $\sigma^{tend}(e^{m}_0)$ describes the dependence of the passive stress on the nominal longitudinal strain in the tendon $e^{m}_0$.

The constitutive model for muscles and tendons is implemented in the ABAQUS general purpose finite element programme (ABAQUS 2007). This code provides a general interface so that a particular constitutive model can be introduced as a ‘user subroutine’ (UMAT). The ‘implicit’ version of ABAQUS is used, in which the finite element formulation is based on the weak form of the balance of linear momentum, the solution is carried out incrementally, and the discretised nonlinear equations are solved using Newton’s method. The Jacobian of the equilibrium Newton-loop requires the so-called ‘linearisation moduli’ of the algorithm that handles the constitutive equations for the muscles and tendons; these moduli are defined in terms of a fourth-order tensor $C$ that relates the variation of stress $\partial \sigma$ to the variation of strain $\partial \epsilon$ over the increment under consideration:

$$
\partial \sigma = C : \partial \epsilon.
$$

The derivation of $C$ is lengthy and the details of the derivation are reported in Spyrou (2009). Here, we note that $C$ can be approximated by

$$
C = C^f + C^{ct},
$$
where

\[
C^f = (1 + \varepsilon_0^m)(\sigma_0^m + \Sigma)\mathbf{H} + \sigma^m \mathbf{A}, \quad (16)
\]

\[
C^{ct} = \frac{1}{f} \mathbf{L}^c : \left( \mathbf{I} - \frac{3}{2} \mathbf{H} + \frac{1}{2} \delta \mathbf{m} - \frac{3}{2} \varepsilon \mathbf{m} \right) - \sigma^{ct} \delta,
\]

with

\[
I_{ijkl} = \frac{1}{2}(\delta_{ik}\delta_{jl} + \delta_{il}\delta_{jk}), \quad H_{ijkl} = m_im_jm_km_l, \quad (18)
\]

\[
\Sigma = (1 + \varepsilon_0^m) \left( \frac{\partial \sigma_0^m}{\partial \varepsilon_0^m} + \frac{1}{\Delta t} \frac{\partial \sigma_0^m}{\partial \varepsilon_0^m} \right), \quad (19)
\]

\[
A_{ijkl} = \frac{1}{2}(\delta_{ik}m_j + \delta_{il}m_j)m_l + \frac{1}{2}(\delta_{ij}m_k + \delta_{jk}m_k)m_l - 2H_{ijkl}, \quad (20)
\]

where \(\Delta t\) is the corresponding time increment.

In the finite element calculations, the activation function \(f_a\) increases linearly from 0 to 1 over a time period of 1 s and then decreases linearly from 1 to 0 over a time period of 1 s. The maximum active isometric stress at optimum fibre length \(\sigma_{max}\) takes the value \(\sigma_{max} = 300\) kPa. The constitutive functions \(f_e(\varepsilon_0^m), f_r(\varepsilon_0^m)\) and \(f_p(\varepsilon_0^m)\) for the muscles are taken from the work of (Delp et al. 1990) and are shown in Figures 5–7. The constants \(k\) and \(c\) in Equations (8) and (9), respectively, take the values \(k = 0.15\) and \(c = 0.4\).

The isotropic elasticity tensor \(\mathbf{L}^c\) in (12) is defined in terms of Young’s modulus \(E = 15\) kPa (Yucesoy et al. 2002) and the Poisson ratio \(\nu = 0.49995\) (nearly incompressible material).

The constitutive function \(\sigma_{max}^{puls}(\varepsilon_0^m)\) for the tendon is taken from the work of Zajac (1989) and is shown in Figure 8.

The bony structures shown in Figure 3 are modelled as homogeneous, isotropic and linearly elastic, with Young’s modulus \(E = 7300\) MPa and Poisson’s ratio \(\nu = 0.3\) (Cheung et al. 2005).

The ligaments and plantar fascia are shown in Figure 9(a) and are taken into account in the finite element model using ‘tension-only’ truss elements. The ligaments are modelled as linearly elastic materials with Young’s modulus in the range \(E = 100–512\) MPa and cross-sectional areas in the range \(A = 7.1–45.2\) mm\(^2\) (Siegler

---

**Figure 5.** Dimensionless function \(f_e\) versus \(\varepsilon_0^m\) (Delp et al. 1990) at various activation levels.

**Figure 6.** Dimensionless function \(f_r\) versus \(\sigma_0^m/\varepsilon_{max}\), where \(\varepsilon_{max} = 5s^{-1}\) (Delp et al. 1990).

**Figure 7.** Passive muscle fibre force–strain relationship (Delp et al. 1990) at various activation levels.
et al. 1988). The plantar fascia is modelled also as a linearly elastic material with Young’s modulus $E = 350$ MPa and a cross-sectional area $A = 290.7$ mm$^2$ (Cheung et al. 2005).

The heel pad and the rest soft tissues in the finite element model, including the regions of articular cartilage shown in Figure 9(b), are modelled using the material data provided by Lemmon et al. (1997). The non-linear and nearly incompressible nature of the soft tissues is modelled as a hyperelastic material with a second-order polynomial strain energy function of the form

$$
U = \sum_{i+j=1}^{i+j=2} C_{ij} \bar{I}_i^2 - 3 \bar{I}_i + D_1 (J - 1)^2, \quad (21)
$$

where $U$ is the elastic strain energy per unit undeformed volume, $C_{ij}$ and $D_1$ are material parameters, $J$ is the determinant of the deformation gradient tensor,

$$
\bar{I}_1 = \bar{\lambda}_1 \bar{\lambda}_2 \bar{\lambda}_3, \quad \bar{I}_2 = \frac{1}{\bar{\lambda}_1} \frac{1}{\bar{\lambda}_2} + \frac{1}{\bar{\lambda}_3},
$$

$\bar{\lambda}_i = \lambda_i / J^{1/3}$, and $\lambda_i$ are the principal stretches. The coefficients of the hyperelastic material model $C_{ij}$ and $D_1$ are calculated based on the uniaxial stress–strain data of Lemmon et al. (1997) and take the values $C_{10} = 85.56$ kPa, $C_{01} = -58.41$ kPa, $C_{20} = 39.0$ kPa, $C_{11} = -23.19$ kPa, $C_{02} = 8.51$ kPa and $D_1 = 271.5$ kPa.

### 2.3 Finite element analysis

The purpose of the current simulation is to produce plantar flexion of the ankle by activating the appropriate muscles (gastrocnemius and soleus) without applying any external forces on the foot and to estimate the stress and deformation state of the contracting muscles and Achilles tendon during foot movement.

Plantar flexor muscles follow one cycle of activation and deactivation. The activation function $f_a$ is applied to the plantar flexor muscles for 2 s; i.e. $f_a$ increases linearly from 0 to 1 over a period of 1 s and then decreases linearly from 1 to 0 over a period of 1 s. Initially, the fibre directions of the lower leg muscles and the Achilles tendon are assumed to be parallel to the long axis of the tibia bone (vertical with regard to the plantar surface of the foot). A quasi-static analysis that accounts for the geometry changes (‘large strain’ analysis) is carried out using the ‘implicit’ version of ABAQUS. The upper bound of the lower leg is kept fixed throughout the analysis.

![Figure 9](image)
3. Results

Figure 10 shows the initial state and the deformed configuration of foot and lower leg as predicted by the finite element solution. The contraction of the plantar flexor muscles causes the motion of the ankle joint and the toes to point downward. An animation of the motion can be found at http://www.youtube.com/watch?v=-ItZyJiSdVms and in the supplementary online material of this article. Contours of the axial fibre strain $\varepsilon_m$ at the posterior compartment of the lower leg (Figure 11) show a logarithmic strain of about $-0.60$ in the gastrocnemius shortened muscle fibres (blue region in Figure 11).

Next we examine the stress and deformation state at the gastrocnemius muscle region where the maximum contraction is observed (Figure 10(b)). We define the hydrostatic stress $p$ (a measure of normal stress) and the von Mises equivalent stress $\sigma_{eq}$ (a measure of shear stress) as follows

$$p = \frac{1}{3} \sigma_{kk} \quad \text{and} \quad \sigma_{eq} = \sqrt{\frac{3}{2} s : s},$$

(23)

where $s = \sigma - p\delta$ is the stress deviator. The corresponding quantities are defined also for the stress $\sigma^c_{ct}$ associated with the connective tissue. A summary of the stress state in the region of maximum contraction is shown in Table 1.

We also determine the volumetric strain $\varepsilon_v$ (a measure of normal strain) and equivalent strain $\varepsilon_{eq}$ (a measure of shear strain):

$$\varepsilon_v = \varepsilon_{kk} \quad \text{and} \quad \varepsilon_{eq} = \sqrt{\frac{2}{3} e : e},$$

(24)

where $e = e - \frac{1}{3} e, \delta$ is the strain deviator. The corresponding quantities are defined also for the strain $e^c_{ct}$ in the connective tissue. A summary of the strain state in the region of maximum contraction is shown in Table 2.

We also examine the change of shape and the stress and strain distribution in the Achilles tendon (Figure 12). Figure 12(a) presents the way Achilles tendon is deformed in comparison with its initial state. Figure 12(b),(c) shows the contours of the maximum principal stress $\sigma_1$ and the axial fibre strain $\varepsilon_m$ in the Achilles tendon, respectively; the maximum value of $\sigma_1$ and $\varepsilon_m$ is found to be $29.91$ MPa and $2.99\%$ and appear in the same location as shown in Figure 12(b),(c).

4. Discussion

The main objective of this work was to introduce a three-dimensional finite element model of the human foot and lower leg to simulate human ankle plantar flexion and estimate internal stresses and strains as well as shape changes of the deformed tissues during human movement.
The finite element solution provides a realistic representation of the foot movement during the plantar flexion of the ankle as shown in the animation (see Video in Supplementary Material).

The stress developed in tendons by voluntary muscle contraction is known to be about 30% of their maximum tensile strength, which is approximately 100 MPa (Maganaris and Paul 1999; Stone and Karatzafiri 2003; Zajac 1989). The calculated value of $\sigma = 29.9$ MPa agrees well with the aforementioned results. Also, the calculated tendon strain of 2.99% is in agreement with previous reports (Maganaris and Paul 1999). As shown in Figure 12(b,c), the stresses and strains are distributed non-uniformly in Achilles tendon and the maximum principal stress and maximum fibre strain appear at the lateral portion of Achilles tendon, which agrees with the findings of Wallenböck et al. (1995). Figure 12(c) shows that fibres in the region of maximum fibre strain are stretched, whereas those at the opposite side of the tendon are compressed.

There are several limitations to this work. First, the construction of the three-dimensional finite element model of foot and lower leg was based on computed tomography (CT) imaging data. By this method, the bony structures are easily distinguishable and can be well defined during the segmentation procedure, but the ability to perform a detailed segmentation of the soft tissues is limited. For example, the muscles of the lower leg were all segmented as one structure (Figure 2(b)), thus being unable to account for interactions among them and define shape deformations for each muscle independently. In such cases, magnetic resonance (MR) imaging data should be preferred for the development of three-dimensional anatomically based finite element models. Second, all fibres of the lower leg muscles were assumed to be in the same direction (vertical with regard to the plantar surface of the foot) and have the same material properties; a more accurate modelling approach should account for the exact fibre orientations and consider the proportions of fast and slow fibres in each muscle tissue.

Despite the limitations outlined above, the results of the modelling analysis show that the computational approach followed here is promising for realistic three-dimensional finite element modelling of complete human body parts, such as the foot and lower leg, that contain many tissues interacting together, where the simulation of natural human movement is accompanied by the examination of local tissue loadings and soft tissue shape.
deformations. Also, there is good evidence to suggest that the current modelling approach is in the right direction.

Acknowledgements
This work was carried out while the first author was supported by the Mechatronics Institute of the ‘Centre for Research and Technology – Thessaly’ (CE.RE.TE.TH.). The authors would like to thank Prof. K. Malizos of the Medical School of the University of Thessaly for fruitful discussions.

References


