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Non-linear finite element model to assess the effect of tendon forces on the foot-ankle complex

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ABSTRACT

A three-dimensional foot finite element model with actual geometry and non-linear behavior of tendons is presented. The model is intended for analysis of the lower limb tendon forces effect in the inner foot structure. The geometry of the model was obtained from computational tomographies and magnetic resonance images. Tendon tissue was characterized with the first order Ogden material model based on experimental data from human foot tendons. Kinetic data was employed to set the load conditions. After model validation, a force sensitivity study of the five major foot extrinsic tendons was conducted to evaluate the function of each tendon. A synergic work of the inversion-eversion tendons was predicted. Pulling from a peroneus or tibialis tendon stressed the antagonist tendons while reducing the stress in the agonist. Similar paired action was predicted for the Achilles tendon with the tibialis anterior. This behavior explains the complex control motion performed by the foot. Furthermore, the stress state at the plantar fascia, the talocrural joint cartilage, the plantar soft tissue and the tendons were estimated in the early and late midstance phase of walking. These estimations will help in the understanding of the functional role of the extrinsic muscle-tendon-units in foot pronation-supination.

1. Introduction

Foot finite element (FE) models have been developed during the last three decades improving their features as computational capacity and constitutive models for biological tissues were improving. Mechanical behavior of bone has been well addressed, but foot soft tissues approaches are still evolving [1]. Advances have been achieved in simulating the non-linear behavior of foot plantar soft tissue and refined constitutive models with real geometries of ligaments are currently used in foot modeling. However, muscle and tendon components have not been appropriately addressed yet [2].

Realistic tendon simulation provides refined estimation of the mechanical performance on the foot-ankle complex. Kinematic and dynamic tendon data can be found in the literature [3–5]. However, stress levels of tendons are rarely reported mainly due to the complexity of performing experimental measurements and the difficulty of modeling soft tissue. Fill this gap is relevant from a clinical perspective since stress helps to estimate pain and tissue damage and it is independent of the structural characteristics of the tendon.

In computational foot modeling, tendon representations have been limited to reaction forces in the tendon insertions [6,7] or the use of one-dimensional link elements [8,9]. Recent approaches included the realistic geometry of Achilles tendon (AT), but the remaining tendons that control foot motion were represented by truss elements or neglected [10]. The consideration of the real tendon geometry allows the study of the tendon itself and not only its reaction in the bone structure. This opens new avenues in the analysis of the foot tendon performance. Furthermore, the use of linear material models to assess the non-linear behavior of tendon tissue is other of the current boundaries in foot tendon simulation. Particularly, in FE foot modeling, only three different approaches have been used for this tissue [2]. The first approach was presented by Wu [11] in a 2D foot FE model where the tendon tissue was configured linear elastic transverse isotropic using a Young's modulus

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of 1200 MPa for the axial direction and a Poisson's ratio of 0.4. The second approach was given by Gu et al. [12] in a 3D model of the Achilles tendon. They simulated the Achilles tendon behavior using an incompressible hyperelastic two-parameter Mooney–Rivlin formulation. The third approach was the isotropic linear elastic material model with Young's modulus of 450 MPa and Poisson's ratio of 0.3 firstly used by Garcia-Aznar et al. [13]. Linear material models are based on the consideration that stresses and strains are proportional. This approximation disregards the initial elongation of the tendons at lower stress values, the so-called toe region [14,15]. As for non-linear material models, that initial strain is considered, as well as the non-linear transition previous to the linear region, providing more realistic stress estimations.

The purpose of this study was to establish a three-dimensional FE model of the human foot using detailed realistic geometry and non-linear behavior of tendons. The model was used to shed light on the role of each tendon in the mechanical response of the foot. For this, force sensitivity analyses of ankle stabilizer tendons, i.e. peroneus, tibialis, and Achilles tendons, were performed. Furthermore, the mechanical solicitations of the internal foot components were predicted at the beginning and the end of the midstance phase of walking. These estimations will help in the understanding of the functional role of the extrinsic muscle-tendon-units of the foot and in the quantification of its mechanical performance.

2. Methods

2.1. Finite element model

2.1.1. Geometry reconstruction

The right foot of a 49 years old male volunteer, with weight of 70 kg and height of 170 cm, was scanned to obtain the geometry of the FE model. Two different tomographies were utilized to capture the geometry details of all tissues in the foot [16]. First, x-ray 0.6 mm slice distance computed tomographic images were segmented to define bone volumes (cortical and trabecular) and to sketch a primary distribution of the tendons. Then, magnetic resonance images, which provide a better definition of soft tissues, were used to refine the initial segmentation, especially the deeper layers of muscles. MIMICS software (Materialise, Leuven, Belgium) was employed to generate the three-dimensional surfaces (STL files) from the two-dimensional stacked image data. This image processing software, frequently cited in three-dimensional reconstruction from medical images [2], use an adapted marching cubes algorithm to create three-dimensional surfaces based on the masks, i.e. region of interest, selected during the segmentation process. Several anatomy references were additionally consulted to help in the identification and delimitation of each individual tendon geometry [17–19].

Due to the complexity of the internal foot components, mostly the intrinsic muscles and tendons concentrated under the foot arch, the reconstructed geometry was processed to avoid volume intersection and other geometrical disruptions before meshing. The detailed segmentation of the foot-ankle complex consists of 102 independent segments corresponding to 30 cortical bone segments, 18 trabecular bone segments, 22 cartilage segments, 29 tendon and muscle segments, 2 fascia segments and the soft tissue volume surrounding the complex (Fig. 1).

2.1.2. Meshing

ANSYS ICEM CFD (ANSYS Inc., Canonsburg, PA, USA) was chosen because of its efficiency in meshing large and complex models and its extended mesh diagnosis. A trial-error approach was employed to optimize the mesh size of each segment. The conditions to achieve a reasonable mesh size without compromising the calculation time were:

- A minimum mesh size sufficiently small to fit into the tightest segments, particularly in the forefoot where many different minor components are concentrated such as proximal, medial and distal phalanges, toes joint cartilages and the thinnest tendons.
- A maximum mesh size consistent with the minimum to avoid large differences in element size between regions and to ensure that the results were independent of the mesh density.
- The mesh accuracy had to achieve more than 99% of the elements better than 0.2 mesh quality (Jacobians) and check that the poor elements were located away from the region of highest interest, i.e. bone, tendon and fascia structures.
- The number of elements below a million. With our current computational capacity (Intel Core i5 3.2 GHz CPU and 4 GB RAM), meshes higher than one million elements increase disproportionally the computational time.

The equilibrium was found with 806.475 linear tetrahedral elements with element sizes as follows: 1 mm for smallest cartilages between phalanges, 2 mm for phalanges, the thinnest tendons and the rest of the cartilages, 3 mm for metatarsals and the rest of the tendons, 4 mm for intrinsic muscles and AT and 5 mm for the large bones in the hindfoot and the fat tissue. This configuration is similar to the 4 mm element size reported by Isvilanonda et al. [8] but optimized for geometry requirements. The quality of the mesh was checked taking as reference the recommendations of Burkhart et al. [20]. All parameters were within good mesh quality ratios (Table 1).

2.1.3. Boundary conditions

The pre-processing of the mesh previous to calculus was conducted with the software I-DEAS (SDRC, Milford, CT, USA). A rigid plane under the foot was included in the model in order to calculate ground reaction forces. A node-to-surface contact between the model and the rigid plane was defined with a friction coefficient of 0.6 [6]. Loads, boundary conditions, calculus, and post-processing were processed using ABAQUS (ABAQUS Inc., Pawtucket, RI, USA). The loads and boundary conditions varied for each analysis and are described in Section 2.4.

2.2. Material properties: tendon characterization

All the material properties used in the model were taken from the literature except for tendon tissue. The constitutive model, the material parameters, and the references are detailed in Table 2. The material model for tendons was determined based on experimental data collected from 100 uniaxial tensile tests of different human foot tendons [15].

Several hyperelastic formulations predefined in ABAQUS were chosen to simulate the actual behavior of human foot tendons. Constitutive models with a smaller number of material parameters were preferred because less experimental tests are required to determine their parameters. Ogden first and second order, Polynomial $n = 1$ and 2, and Polynomial reduced order 1, 2 and 3 formulations were included in the analysis. The material models that fitted the average experimental curve and also performed a real physiological behavior (compression stress with negative strains) were then used to simulate uniaxial tendon tests. The model that showed better agreement with experimental data was chosen.

Once the parameters of the material model for foot tendon tissue were defined, the model performance was compared with previous foot tendon material models reported in the literature: the linear models used by Wu [11] and Garcia-Aznar et al. [13] and the hyperelastic model used by Gu et al. [12]. Four simple one-dimensional models were created with the structural properties of four randomly selected tendon samples from a previous experimental study [15] to replicate the tests. Each model was run four