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Conception and evaluation of a 3D musculoskeletal finite element foot model

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1. Introduction

Modelling the foot accurately is essential in understanding its behaviour under healthy and pathological conditions. For example, it has helped analysing the risks of injuries in diabetic foot (Chen et al. 2010). This work aims at developing and evaluating a patient-specific finite element (FE) model of the foot in the context of pressure ulcer prevention, orthopaedic and motion analysis. Using the most recent functional knowledge about foot anatomy is necessary to simulate functions such as support, weight bearing, locomotion or foot surgery and its consequences.

2. Methods

2.1. Modelling

The model has been developed using the 3D biomechanical simulation platform ArtiSynth (artisynth.org). Starting from a CT and an MRI exam of a single patient, 30 bones have been modelled as articulated rigid bodies connected with cables that simulate the 210 segmented ligaments in their actual positions and therefore define the articulations with contact. The aponeurosis is modelled with five parallel multipoint ligaments connected by transversal ligaments. Fifteen extrinsic and intrinsic Hill's model muscles have been positioned according to their anatomical course and can be independently activated in order to allow a natural movement of the foot.

A FE mesh of the soft tissue was created by applying a new automatic FE mesh generator, TexiMesh (texisense.com), to the surfaces resulting from MRI and CT segmentation. The FE mesh has 142,060 elements (mainly hexahedrons) and 66,362 nodes. Three soft tissue layers with Neo-Hookean materials (Young moduli, Poisson Ratio) were created to represent a 1-mm skin layer (200 kPa, 0.485), the fat (30 kPa, 0.49) and muscle (60 kPa, 0.495) tissues, Figure 2(A). A fourth layer represents the heel anatomical soft structure (100 kPa, 0.4998).

2.2. Weight-bearing evaluation

The foot model was first evaluated for static weight-bearing position. We compared (Figure 1) simulated plantar pressure (SPP) with the real plantar pressure (RPP) collected from the same patient standing onto a Zebris FDM-SX platform. For this simulation, half of the patient's weight was applied onto the foot model while in contact with a horizontal FE plate (Figure 1(B)) having the same number of 'sensors' as the Zebris platform (one element per sensor). In order to represent dynamic loading, the muscles first make a dorsiflexion before foot contact. We compare the mean pressure (MP) and peak pressure (PP) at platform surface after regionalization (Gefen et al. 2000) using a dedicated software, TexiLab (texisense.com). The RPPs are the mean of four trials.

2.3. Dynamic evaluation

This evaluation was made during an adduction (ADD)/abduction (ABD) movement: a 3D motion analysis of the patient's foot was performed using the Leardini's marker set (Leardini et al. 1999). A needle and surface EMG monitoring of the Tibialis Posterior, Tibialis Anterior, Triceps and Peroneus muscles was used as input for the simulation. Local referential and regression equations were created to have 3D point comparison between virtual markers from motion analysis and real anatomical markers from CT reconstruction, to assess our model's motions resulting from simulated muscle activations (Figure 2). For the simulated and real kinematics (five trials), we compared the 3D angle formed by the mass centre of the tibia, the mass centre of the talus and the centre of the second metatarsal head, considered as an anatomical axis of the foot. Angle measurements were also performed in 2D after projecting the reference points on the three anatomical planes.

Movements were split into
- Dorsal/plantar flexion in the sagittal plane.
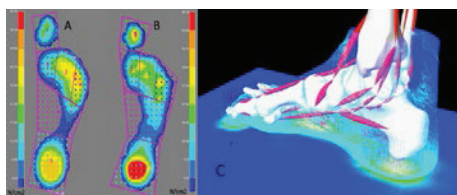- Pronation/supination in the frontal plane.
- ABD/ADD in the horizontal plane.

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Figure 1. (A) Real PP, (B) simulated PP and (C) stress visualization of the FE model in weight-bearing position.
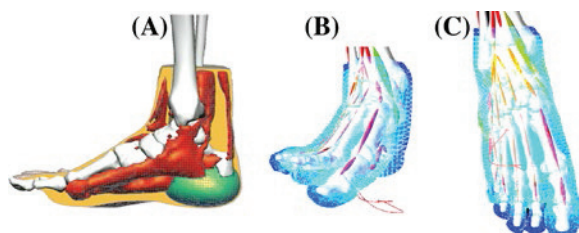

Figure 2. (A) FE model, (B) ABD kinematic and (C) ADD kinematic.

These categories were made to facilitate a clinical interpretation of the results.

In this article, only the ABD/ADD kinematics is studied.

3. Results and discussion

3.1. Weight-bearing simulation

Table 1 compares the simulated and real peak and MPs under the second (region 2) and the fourth/fifth (region 3) metatarsal heads. The simulated pressures are close to the real ones considering pressure measurement and subject weight uncertainties. In the second metatarsal area (region 2), the differences between RPP and SPP are about 0.6 N/cm² for the MP and 1.6 N/cm² for the PP (Table 1). In the fourth and fifth metatarsal area (region 3), the differences are 1.9 N/cm² for the MP and 0.9 N/cm² for the PP. We assumed that our precise results regarding the literature are due to the use of the vertical component of the simulated platform's Von Mises Stress (VMS) instead of foot's skin plantar VMS in others studies.

3.2. Dynamic EMG simulation

Table 1 provides simulated kinematics with muscle activation using real EMG input. The corresponding 3D angle between the tibia and second metatarsal bone is 1.6% greater than real anatomical measurements for the maximum 3D angle and 3.9% smaller for the minimal 3D angle. More than 120° could be considered as a complex motion with ADD and plantar flexion. Less than 120° could be a synergic function of the dorsiflexion and the ABD. The projection angle, in the horizontal plane, is 0.4%

Table 1. Static and dynamic results.

	Real pressure	Simulated pressure	Difference
	Region 2	*Region 2*	
Mean pressure	2.9 N/cm²	2.3 N/cm²	0.6 N/cm²
Peak pressure	14.1 N/cm²	15.9 N/cm²	1.6 N/cm²
	Region 3	*Region 3*	
Mean pressure	4.3 N/cm²	2.4 N/cm²	1.9 N/cm²
Peak pressure	16.6 N/cm²	17.5 N/cm²	0.9 N/cm²
	Real kinematic	*Simulated kinematic*	*Difference*
Angle 3D MAX	146.7°	149.1°	1.6%
Angle 3D MIN	114.9°	110.6°	−3.9%
Angle 2D ADD MAX	148.8°	149.4°	0.4%
Angle 2D ABD MAX	115°	113°	−1.8%

greater in ADD and 1.8% smaller in ABD. These excellent results should be carefully analysed since the reference frame is fixed to the tibia during simulation, while it has a skin marker dependency for the real motion analysis. The projected angles could thus be misestimated and the difference could be greater.

These results are preliminary and cannot be compared to other biomechanical FE studies since it is, to our knowledge, the first time real EMG input was used for assessing foot motion during swing phase or unloading.

Our work was limited to only one case since we wanted to evaluate the model with real data. It is important to note that our group has already proposed a methodology to easily adapt this single model to other patients' anatomy. Other patient-specific simulations could thus be performed.

4. Conclusions

This article has introduced a new musculoskeletal and FE foot model providing realistic simulations in both static and dynamic frameworks. The ranges of motion in dynamic, and plantar pressure in loading condition, are anatomically and clinically realistic.

Other studies using this model will simulate ankle arthrodesis or foot orthotics. The model could also become relevant for the simulation of neuro-orthopaedic surgical interventions, orthotic devices analysis or educational purposes like functional anatomy.

References

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