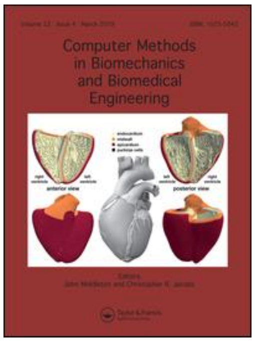

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Finite element model of a below-knee amputation: a feasibility study

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KEYWORDS Trans-tibial amputation; lower limb prosthesis; optimization; socket pressures; finite element model

1. Introduction

In 2005, the number of lower limb amputees was 1.6 million in the USA and is projected to reach 3.6 in 2050. Among causes for amputations, dysvascular diseases, trauma and diabetes are the main one and dysvascular conditions are increasing because of an ageing population (Ziegler-Graham et al. 2008).

The manufacturing of lower limb prosthesis is currently mainly artisanal while the use of computer assistance is still limited. As a consequence, the quality of the prosthesis will greatly depend on the prosthetist know-how. However, this is not the only factor influencing prosthesis quality, and the type of the prosthesis, its design, or the materials for both liner and socket should also be taken into account. Although prosthesis quality assessment is subjective, quantitative measurements such as pressures or temperatures in the liner and patient feedback through questionnaires can be assessed.

Mechanical interactions between a stump and the prosthesis were accurately predicted using finite element method (FEM, Colombo et al. 2011; Goh et al. 2005). Existing models mainly aim to develop a realistic model with no or limited validation. Complete optimization of the prosthesis using numerical analysis has, to the best of our knowledge, not been fully performed.

Thus, the main objective of this project is to define a new FEM method to fully optimize prosthesis shape in order to improve the subject's comfort. The definition of this method will first need a validation of the FEM by comparing experiments and simulations results together with measuring the method reproducibility.

2. Methods

2.1. Clinical prosthesis assessment

Nine FSR pressure sensors were placed on the subject's stump (51 years old, male, 7 years since left leg below knee traumatic amputation) as shown in the Figure 1.

Testing consists into two phases: static testing in which the subject, initially seated, stood for 5 s before to seat; and dynamic testing when the subject, initially seated, stood and walk for a predefined distance before to seat back.

Both pressure measurements were analysed with a python script to compute pressures (min, max, mean, std) during 5 repetitions of standing position (static) or during five steps (dynamic) respectively.

2.2. Development of the model

The existing full leg LLMS model (detailed in Arnoux et al. 2005) was used to develop the amputee model. Briefly, this model consists in a full human body including simulation of the components in the leg (flesh, knee joints …). Only one leg is kept from the model (from the femur to below the fibula and tibia heads) in order to reduce computation time.

Subject anatomy (tibia, fibula and skin) was segmented from an MRI sequence using Mimics (Materialize NV, Belgium) and 3D optic scanner (Artec 3D Studio, Luxembourg) for smoothing. The LLMS derived model was scaled by comparing the size of the bones' heads. Bones and skin relative positions were kept as well as the whole skin in order to save the stump anatomy. Lower parts of the bones were sectioned, and the junction between LLMS sectioned bones and segmented ones was then created while the gap between the skin and the bones was filled with soft tissues as tetrameshes. Finally, properties and materials were added to the new part as previously defined in LLMS (see Table 1).

In order to validate the model, several simulations were made. For every simulation, the load was set to 375 N, representing half of the patient weight, applied on the femur's head; the boundary conditions were set on the faces of a box starting above the knee with a stump shaped hole in it and chosen as a first approach of a prosthesis (Figure 2).

The three last simulations were as followed:

- Simulation 1: The box material was defined as soft tissues of the model.

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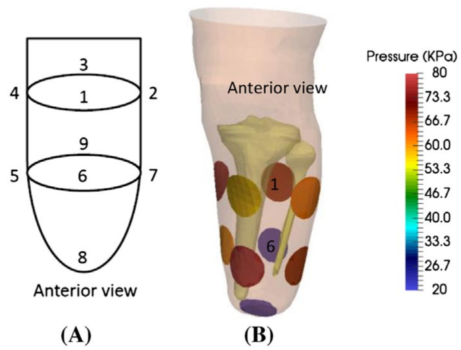

Figure 1. (A) Sensors locations on the subject's stump and (B) pressure measurements at max values averaged on 5 steps.

Table 1. Component properties.

Part	Young's modulus	Poisson ratio	Number of elements	Type of elements
Skin	1 MPa	0.3	3036	Shell
Soft tissues	2.5 MPa	0.45	13,947	Tetra-mesh
Bones	10 GPa	0.3	452	Shell

- Simulation 2: The box material was defined as rigid material, with properties similar to the bones ones. This is the model shown in Figure 2.
- Simulation 3: Same material defined in simulation 2, the box was shortened and starts under the knee.

Parameters were defined to assess model quality.

- Numerical stability: the deformation of the meshing, the energy and mass balance of the system.
- Physical behaviour: general deformation of the leg, behaviour of the bones with the soft tissues.

3. Results and discussion

The dynamic pressure measurements showed that the maximum of pressure during a step were in area 3, 4 and 6 (Figure 1). These measurements were confirmed by the subject's feelings.

The ratio between hourglass energy and internal energy is important though. Simulation 3 has the lowest ratio, with 15% (Sim1 = 28%, Sim2 = 37%). Those values, superior than 10%, show a numerical instability in the model, which needs to be corrected.

The general behaviour, i.e. kinematic, tissue constriction, and bones relative displacements, were assessed quantitatively.

These measurements show that simplifications were too important first, and a more refined model is needed in order to obtain reliable results.

A further validation with a realistic model of the prosthesis will be required for validation of prosthesis model used. Additionally, to enable larger inclusion number and generalization of the FEM method, the skin geometry will be acquired with optical scanner information instead of MRI.

Physical and numerical reliabilities will be furtherly explored in order to improve the quality assessment and robustness of the model. Other materials and interfaces will be tested through an experimental design. Finally, comparison with local pressure values will be performed.

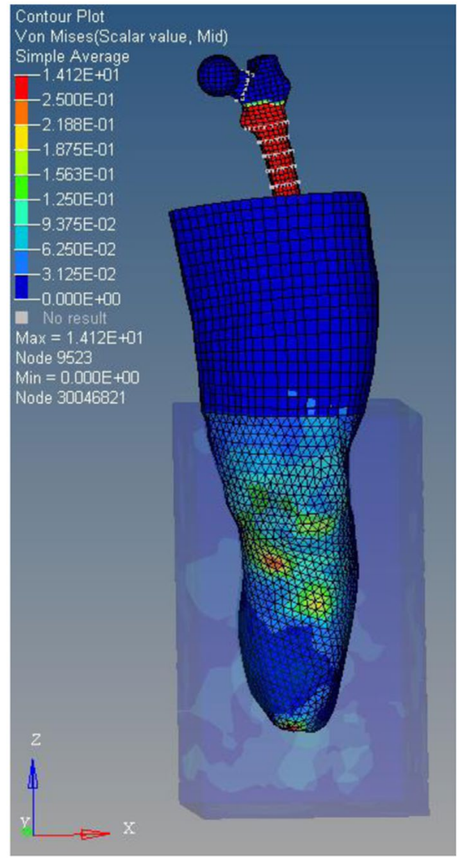

Figure 2. FE model, box as a boundary condition.

For a better simulation, a socket needs to be added between the stump and the prosthesis, with adequate friction coefficient and its material properties.

4. Conclusions

The method developed for any below-knee stump representation shown promising results, while such modelization could still be improved to match experimental results. Improvements of the model are still needed to accurately represent the reality. The pressures measurements together with patient specific model and future prosthesis optimization is a step forward in comparison with existing literature.

The final validation of this model will be full numerical and experimental comparison between an FEM optimized prosthesis and the initial one. The feedback of the subject will also be important to judge the quality of the prosthesis.

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