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A HYBRID FORMULATION FOR SOFT TISSUE MODELING ON REAL-TIME SURGERY SIMULATION

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Abstract. *On the improvement of surgical training and planning methods, virtual reality and computational mechanics have been applied for the development of technological tools which aim to reduce patient risk in surgical procedures.*

The main focus of this work is modeling and simulation of soft tissue behavior for virtual reality systems. Soft tissues can be consider as composite materials, formed by an isotropic matrix and a bundle of reticular fibers with anisotropic behavior. In order to simulate the material behavior for virtual applications, nonlinear deformations must be calculated on real-time.

In this article, a hybrid formulation called Equivalent Spring-Mass Model (ESMM) is proposed. ESMM is a construction based on a fusion of the Spring-Mass Model (SMM) and an Equivalent Strain Energy Density Function (ESEDF). In this formulation, parameters of one-dimensional elements can be estimated on real-time to reach the strain energy calculated with a complex constitutive model on a pre-computational stage.

ESMM was implemented on a virtual environment for laparoscopic surgery with a liver mesh segmented from medical images using InVesalius Software. The parameters were calculated using experimental data of uniaxial tensile tests of porcine liver parenchyma. ESMM elements shown good results and the estimations predict the tissue behavior.

Keywords: *Soft tissue modeling, Virtual reality, InVesalius, Surgery simulation.*

1 INTRODUCTION

In healthcare and medical area, guarantee security and integrity of the patient during medical procedures is a key issue. In surgical procedures, the degree of expertise, skills, and precision needed on a surgeon is very high, and therefore the learning process on training is a crucial factor on their professional performance, due its importance for increase experience and skills development.

With this in mind, the development of tools for training and surgical planning is a big research concern that can lead to improvement on the medical procedures of today. Virtual reality, among other computational and technological tools applied, allows the surgeon to acquire experience without contact with a living being, being then a risk free and ethical environment for practice of soon-to-be surgeons, or for planning and testing a procedure before being done in the operating room (SEYMOUR, 2008), (ISSENBERG, 2005).

A virtual training system involves the integration of several factors, including: medical imaging, computer graphics, material models, and mechatronic systems for interaction device. A key issue on this kind of devices is to achieve a proper model for prediction of the real behavior and mechanical properties of soft tissues, without affecting the computational performance of the real-time simulation (MORENO, et al. 2014).

In this article is exposed the development of open-source computational tools that allows virtual surgical training, based on interactive real-time simulation. It includes a brief description of geometry acquisition, mathematical model for soft tissue simulation and the generation of the virtual environment implemented in previous work.

2 METHODOLOGY

In order to create a virtual environment close to the reality, it is needed to represent the accurate geometry of the anatomical structures from 3D reconstructions of medical images. Using InVesalius software, a 3D model of anatomical structures can be reconstructed from CT (Computed Tomography) and MRI (Magnetic Resonance Imaging) two-dimensional images (AMORIM, et al., 2011), (CAMILO, et al., 2012). Once reconstructed, this 3D model is processed to remove noise and improve the surface, reduce the amount of vertex and have a uniform meshing, as shown in figure1a.

At this point, the 3D model is a tridimensional triangle mesh which represents the external structure of an organ. A volumetric mesh must be created from this model in order to calculate the behavior of the organ considering the internal stresses and deformation propagation. For this propose an internal skeleton structure, exoskeleton structure or tetrahedral mesh were proposed and applied as shown in figure1c (MORENO, et al., 2012), (MORENO, et al., 2014).

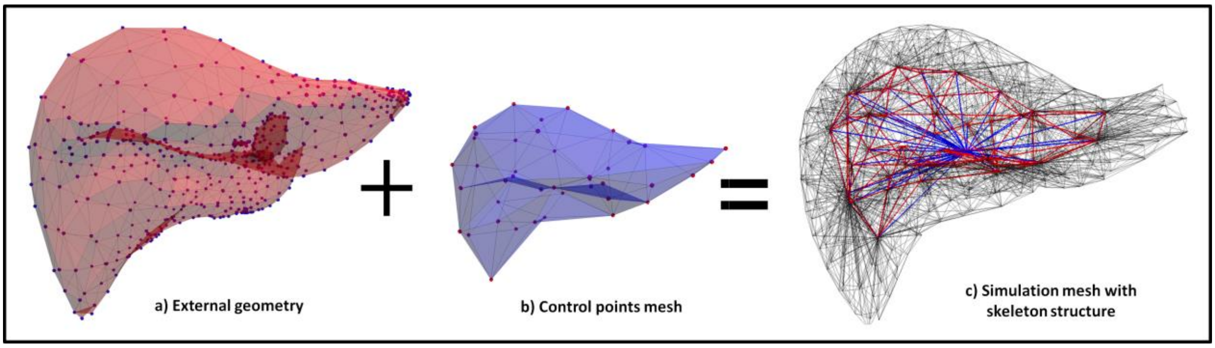

Figure 1. a) External geometry b) Control points mesh c) Simulation mesh with skeleton structure

2.1 Soft tissue modeling

In order to simulate an organ and its response to the user interaction, it is needed the application of a mathematical model that represent the material mechanical properties and the implementation of a computational methods to solve the equations for the particular geometrical model. Due to the application requirement, an appropriate selection of those factors is critical, considering that the algorithms must respond for real-time interaction in the simulation (DELINGETTE, et al., 2004), (ZHANG, et al., 2006), (HALIC, et al., 2009).

A first approach was developed using a construction based on a Spring-Mass Model (SMM) with a multi-level of stiffness. The SMM is a simplification where the object is considered as a set of mass points (vertex) connected by springs (edges) with defined stiffness parameters, as shown in Figure 2 (MORENO, et al., 2012).

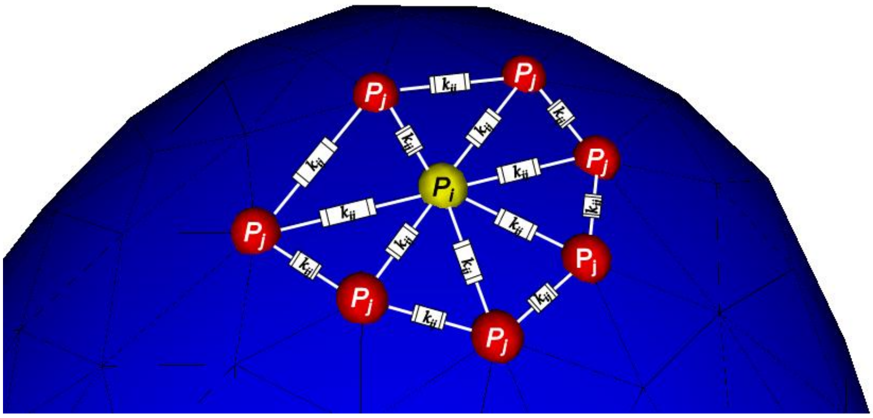

Figure 2. SMM illustration, with point Pi in yellow and neighbors Pj in red

Then, the force F_i on a point P_i is calculated by considering the contribution of all the spring formed between the point P_i and each point P_j in its neighborhood $N(P_i)$. At each time step, the force F_i is calculated using the position of the points at that time, the length of each spring at rest position (L_{ij}) and a stiffness parameter assigned (k_{ij}) regarding the spring level, as shows the following equation:

$$\mathrm{F_i} = \sum_{\mathrm{j} \in \mathrm{N(P_i)}} k_{ij} \left(\left\| P_i P_j \right\| - L_{ij}^0 \right) \frac{P_i P_j}{\left\| P_i P_j \right\|} \tag{1}$$

Although SMM allows calculating deformation of an object considering one or more stiffness parameters (proposed on multi-level SMM), it can not represent precisely the behavior of soft tissues, because it is a linear one-dimensional element and the stiffness parameters are constants.

For this case, soft tissues are considered as a composite material. This is because tissues mechanical properties are related to its cellular composition, which is non-uniform. Soft tissues are conformed by a matrix with an isotropic behavior, and a bundle of collagen fibers with an anisotropic behavior (HOLZAPFEL, 2001), (ELÍAS-ZUÑIGA, et al., 2014).

In the literature, there are different models that consider anisotropy and hyperelasticity properties of soft tissues, most based on continuum mechanics to define their Strain Energy Density Function (SEDF). However, the amounts of parameters that need to be calibrated to characterize a particular tissue are considerably high, and some of them cannot represent a wide range of different tissues behaviors. In addition, the main problem still is the computational resources to solve the equations on real-time for a large amount of elements, since most of them are quite complex. Due these issues, a new hybrid model is proposed in this work, constructed by a SMM model combined by an energetic equivalence to a more sophisticated SEDF, which is solved on a pre-computation stage.

For modeling the behavior of soft tissue and biological materials, it is possible to use a construction based on the rule of mixtures of strain energy, proposed by Elías-Zuñiga (2014). In this constitutive model, it was considered finite deformations of hyperelastic materials by applying the rule of mixtures to create an Equivalent Strain Energy Density Function (ESEDF) model, which includes energy contributions from isotropic and anisotropic volumetric fractions of the material (CANTOURNET, et al. 2007), (ELÍAS-ZUÑIGA, et al., 2014).

The ESEDF model defines the strain energy of a material as follow:

$$\mathrm{W_T} = (1 - \mathrm{f})W_{iso} + fW_{aniso} \tag{2}$$

Where W_{iso} represent the strain energy of the isotropic part, and W_{aniso} is the anisotropic strain energy contribution of f, which is the anisotropic volumetric fraction. Both energies can be estimated trough formulations based on Arruda-Boyce and Spencer models, by:

$$\mathrm{W_{iso}} = \mu\left[N\left(\beta\lambda_r + \ln\left(\frac{\beta}{\sinh\beta}\right)\right) - \ln\left(\frac{\beta}{\lambda_r}\right)\right] + c \tag{3}$$

Where N is the chain number, μ represent the material shear modulus, c is an energy constant, β is the inverse Langevin function of λ_r, and $\lambda_r = \sqrt{\frac{I_1}{3N}}$ (ELÍAS-ZUÑIGA, et al., 2002).

And:

$$\mathrm{W_{aniso}} \cong \frac{A_1}{3}(I_{1i} - 3) + \frac{A_2}{9}(I_{1i} - 3)^2 - \frac{2A_1}{3}\ln\sqrt{I_{3i}} \tag{4}$$

Where A_1 and A_2 are energy density fitting parameters, and I_{ki} represent the principal invariants *izotropized* by the consideration of the fibers at (±1,±1,±1) directions (ELÍAS-ZUÑIGA, et al., 2014).

2.2 Equivalent Spring-Mass Model

In the new hybrid model proposed in this work, a SMM model stiffness parameter was defined in function of an energetic equivalence of the ESEDF model, which is pre-computed on a deformation range to avoid time response problems. The strain energy of a linear spring is based on a case of Neo-Hooke model for one-dimensional elements, and will be found the parameter to make it equivalent to ESEDF:

$$W = \frac{\mu}{2}(I_1 - 3) = \frac{\mu_e}{2}\Delta\lambda^2 = W_T \tag{5}$$

Then, it is possible to define parameters for nonlinear springs that will represent individually the behavior of soft tissue in a more accurate way. These stiffness parameters are in function of the pre-computed ESEDF, by energy equivalence, and are calibrated with data of mechanical tests of tissue samples. The stiffness parameter of an element in function of its stretch ratio is:

$$\mu_e = \frac{2}{\Delta\lambda^2}\left[(1 - \mathrm{f})W_{iso} + fW_{aniso}\right] \tag{6}$$

$$\mu_e = \frac{2}{\Delta\lambda^2}\left[(1\text{-}f)\left(\mu\left[N\left(\beta\lambda_r + \ln\left(\frac{\beta}{\sinh\beta}\right)\right) - \ln\left(\frac{\beta}{\lambda_r}\right)\right]\right) + f\left(\frac{A_1}{3}(I_{1i} - 3) + \frac{A_2}{9}(I_{1i} - 3)^2 - \frac{2A_1}{3}\ln\sqrt{I_{3i}}\right)\right] \tag{7}$$

In this case, μ_e is calculated for a specific deformation range calibrated from tensile tests data. It is possible to define μ_e for different directions by considering fibers aligned to main axis on the *isotropized* form on Eq. (4), using tensile test data of the oriented tissue.

With the stiffness function defined using an energetic equivalence, the deformation of each element can be calculated with a parameter based on its specific stretch ratio at that time. The angle between the element and a defined dominant fiber orientation (direction cosines of the ith family fiber) of the tissue is used to calculate the equivalence parameter for each direction:

$$\mathrm{k_{ij}} = v_{ij}\left[\mu_{ex}\cos^2\alpha, \mu_{ey}\cos^2\beta, \mu_{ez}\cos^2\gamma\right] \quad (8)$$

Where μ_{ex}, μ_{ey}, μ_{ez}, represent the stiffness functions defined for each direction, v_{ij} a parameter for volumetric compensation and α, β and γ represent the angles between the fiber direction θ and the element orientation $\o_{ij}$ for each axis as shown in figure 3.

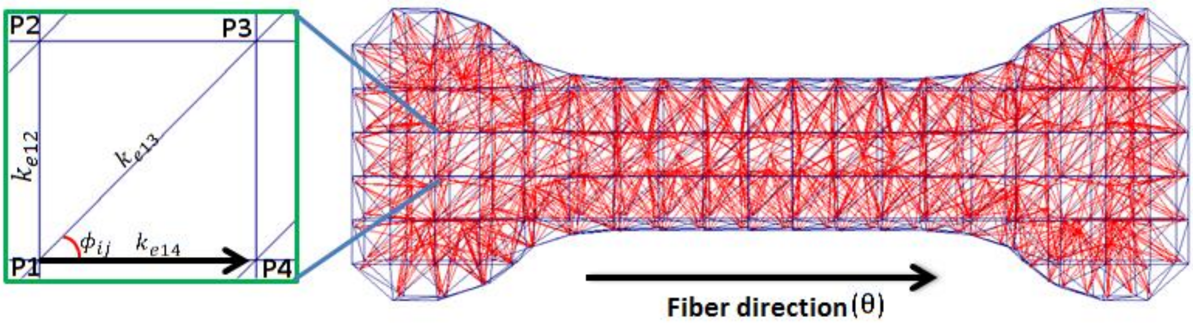

Figure 3. Mesh representation of "dog bone" sample with internal elements

Therefore, applying Eq. (8) on Eq. (1) it is possible to redefine the force F_i on a point P_i as:

$$\mathrm{F_i} = \sum_{\mathrm{j}\in \mathrm{N(P_i)}} v_{ij}\left[\mu_{ex}\cos^2\alpha, \mu_{ey}\cos^2\beta, \mu_{ez}\cos^2\gamma\right]\left(\left\|P_iP_j\right\| - L_{ij}^0\right)\frac{P_iP_j}{\left\|P_iP_j\right\|} \quad (9)$$

This hybrid model will referenced as Equivalent Spring-Mass Model (ESMM). Having established an energetic equivalence, it is possible to calculate μ_e, which represent a stiffness function that make a deformed element achieve the same strain energy than the soft tissue at the same stretch ratio.

3 RESULTS

The ESMM was implemented as material model for a real-time simulation of the behavior of soft tissues using Python and the Visualization Toolkit (VTK). Calibration parameters for ESMM were found using experimental data from uniaxial mechanical tensile tests of porcine liver parenchyma samples.

The characterization of liver parenchyma was achieved with the proposed hybrid model, allowing to predict the behavior of liver tissue on the virtual environment. A comparison between ESMM predictions and experimental data is shown in figure 4.

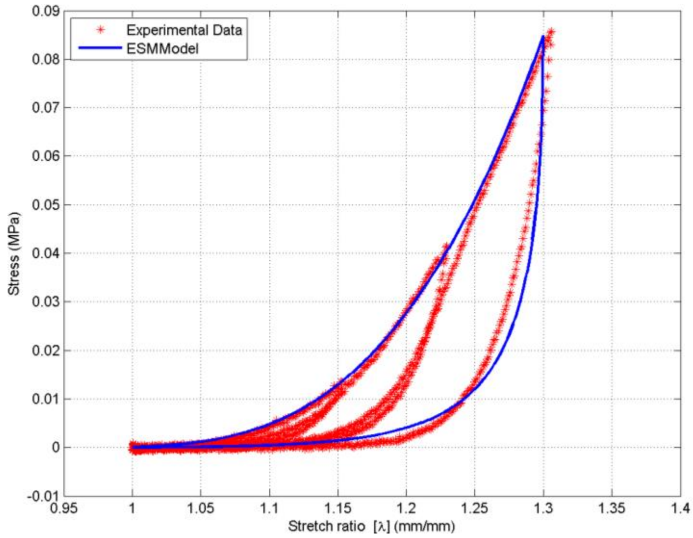

Figure 4. Graphic that show the prediction of the model using μ = 2.973e-06 MPa, N = 1.1074, A1 = 3.9106, A2 = 451.9193, f = 0.0019, b = 5.2022, C = 0.0699. Young's modulus E = 0.67712 MPa (exp data).

Using this ESMM implementation on an initialization stage, the stiffness parameters μ_e were calculated and k_{ij} was defined for each element at each stretch ratio on the 3D geometrical mesh of liver (regardless the orientation, considering $\mu_e = \mu_{ex} = \mu_{ey} = \mu_{ez}$ for this test case). Pre-calculated data was stored to use on the real-time simulation, and the result for the stiffness parameter and the force F_i is shown in figure 5.

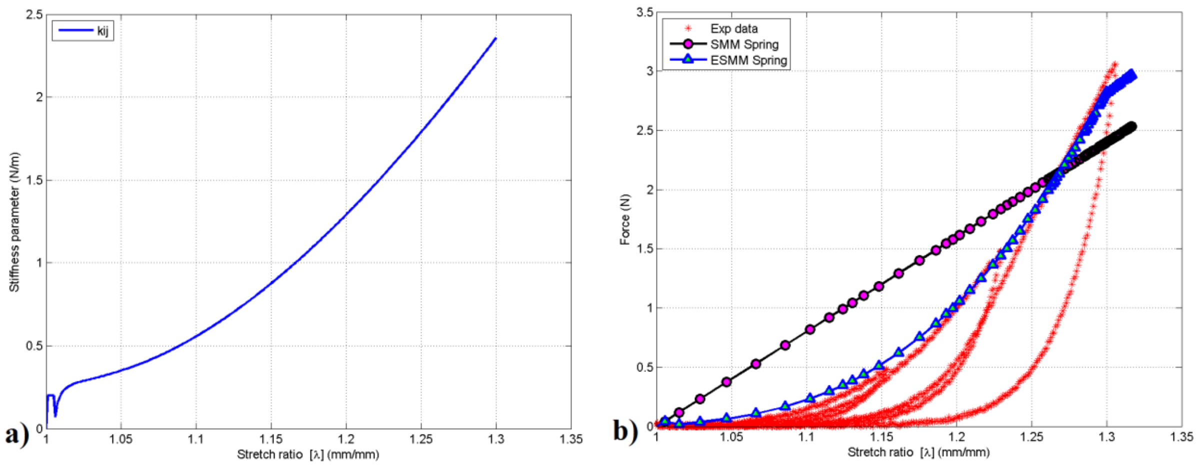

Figure 5. a) Graphic that shows the behavior of μe in function of the stretch ratio, b) Graphic that compares the behavior of a spring of SMM and ESMM with experimental data.

Using the ESMM implementation in this work, it was possible to make a simulation of non-linear elements based on an energetic equivalence with a SEDF in a pre-computed phase.

Stored k_{ij} data for elements on the 3D mesh was used on the virtual environment, and the results for one P_iP_j element were plotted on figure 5b. The interactive simulation of the liver mesh with the ESMM elements is shown on figure 6, running on a virtual environment for laparoscopic surgery developed on previous works (MORENO, et al., 2012), (MORENO, et al., 2014).

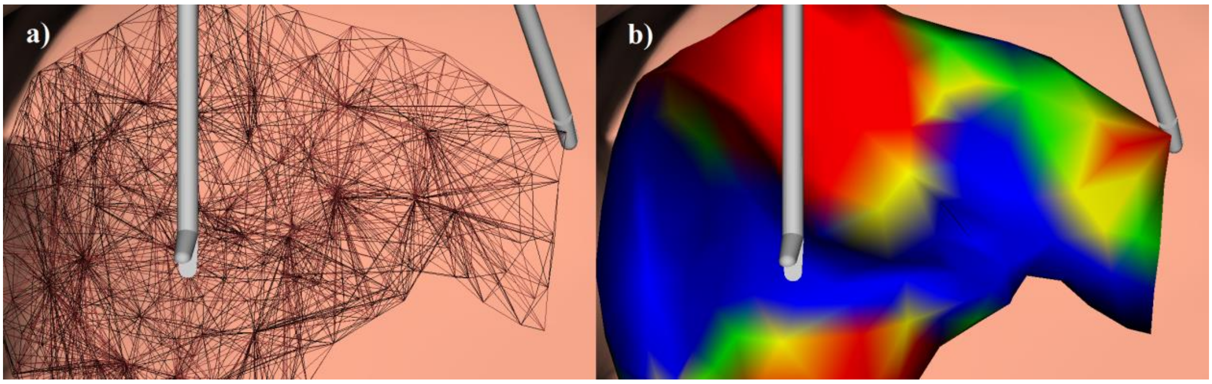

Figure 6. Virtual environment with liver mesh using ESMM springs a) ESMM springs formed with the liver mesh b) Deformation color map of the liver mesh

Since the main objective is to run on a real-time virtual simulation system, the implementation performance of this model was measured. Running the calculation process on an Intel® Core™ i7-4510U @ 2.00GHz with 8GB RAM for a test mesh of 5322 elements, a initialization phase to get the parameters from $\lambda = 1$ to λ_{max} needed 1.43 *s* to read, solve and store data of ESMM. On the simulation, the process required 3.517 *ms* to calculate the deformation of the mesh trough gravity and user interaction. This time lead to a 28 *frames/s* refresh frequency, allowing the interaction between user and virtual environment without significant delay.

4 CONCLUSIONS

The proposed hybrid construction (ESMM) was implemented on a virtual environment and was able to calculate the stiffness parameters for elements to represent an anisotropic non-linear behavior. Each element represents the behavior that a sample of porcine tissue had on the uniaxial tensile test used to calibrate the model.

Although ESMM had good results on this work, the fact that each one-dimensional element has an approximated behavior of the real tissue does not mean that an object composed by these elements simulate the whole organ behavior. This is because the one-dimensional spring elements model the whole organ in just segments, without considering the volume that each one represents. That is why a volumetric compensation parameter was added to the model, in order to considerate the amount of volume that each spring is representing, in function of the simulation mesh construction.

The consideration of volumetric compensations and the implementation of volumetric elements would be researched in future works. Also, the study of tissue properties would continue after the porcine tissue mechanical tests done to calibrate the model on this work, with focus on biaxial tensile test of porcine tissue samples. Applicability of microtomography images of tissue samples will be researched in order to study principal fiber orientation vectors, tissue composition and the anisotropic fraction (*f*).

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