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Numerical Modelling of Soft Tissue Injury Due to Impact

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Abstract— **Soft tissue injuries due to impact loading are a major health problem. The objective of this study was to simulate the impact of projectiles onto human body area and to show the validity of the model developed, based on a finite element model of the flash-ball impact on the human leg. To determine realistically the strain and stress in the biological soft tissues, anisotropic hyperelastic constitutive laws are necessary in the context of finite element analysis. The contact between impacting bodies is solved by the bi-potential method which consists of projecting the displacement equations onto the constraining directions associated to contact points. The time integration of the equation of motion is achieved by means of a first order algorithm. The algorithm is implemented into the finite element code FER/Impact using C++ Object Oriented Programming techniques.**

Keywords— **Impact, Soft tissue, Injury, Hyperelasticity, Finite element.**

I. INTRODUCTION

A quantitative evaluation of soft tissue injury due to impact loading is of general interest. Hemorrhage and edema are the usual sequels to traumatic pulmonary impact. To gain some quantitative understanding of the phenomena, we propose a numerical approach to simulate the impact of projectiles onto the human body area. In such a context of simulation, there is an ongoing quest for realistic visual deformation of soft tissues and faster algorithms to solve multiple contact forces while controlling the numerical stability of the solution. A computational efficient simulation relies on three optimized numerical “black boxes”, the first one to detect the potential contact points (collision detector), the second one to solve the nodal displacements of the soft tissues (displacement solver) and the third one to solve the multiple frictional contact forces (contact solver).

To determine realistically the strain and stress in the biological soft tissues such as ligaments, tendons or arterial walls, anisotropic hyperelastic constitutive laws are necessary in the context of finite element analysis. Most energy densities used to model transversely isotropic and orthotropic soft tissues take a power law form [1] or present an exponential behavior [2, 3]. It is usually assumed that anisotropy is due to the collagen fibers behavior [4] while the ground substance, or matrix, behaves in an isotropic manner.

The contact between impacting bodies is solved by the bi-potential method [5] which consists of projecting the displacement equations onto the constraining directions associated to contact points. The contact laws (Signorini and Coulomb) are formulated from an augmented Lagrangian formulation (bi-potential formulation) and computed by Uzawa or Newton techniques which lead to an iterative predictor/corrector process. The advantage of this method is to separate the contact solver from the displacement solver.

The time integration of the equation of motion is achieved by means of a first order algorithm. The algorithm is implemented into the finite element code FER/Impact [6] using C++ Object Oriented Programming techniques.

One numerical example is performed in this study to show the validity of the model developed. This example concerns the flash-ball impact on the human leg.

II. HGO HYPERELASTIC MODEL

To determine the strain and stress in the biological soft tissues such as ligaments, tendons, pelvic organs or arterial walls, anisotropic hyperelastic constitutive laws are often used in the context of finite element analysis. The most used strain-energy functions take a power law form or present an exponential behavior. More recently, Balzani et al. [7] have proposed polyconvex strain energy functions combining an exponential form with a power law to take care of the tissues behavior in the low load domain. More realistic models have been also recently developed to capture the inter-fiber angle change by adding to the strain energy the contribution of the fiber-matrix shear interaction [8]. In general, the anisotropy can be represented via the introduction of a so-called structural tensor, which allows a coordinate-invariant formulation of the constitutive [9–11]. It is usually assumed that anisotropy is due to the collagen fibers behavior [4], while the ground substance, or matrix, behaves in an isotropic manner, so the energy densities modeling transversely isotropic and orthotropic soft tissues are separated into isotropic and anisotropic parts [7,12].

$$W = W_{iso} + \sum_{a=1}^n W_{ani}^a \quad (1)$$

Each anisotropic density W_{ani}^a refers to a preferred direction of the material. The number fiber family. n is the number of fibers. To model the embedded collagen fibers of soft biological arterial tissues, the HGO (Holzapfel, Gasser, Ogden) constitutive law [3, 4] superposes two transversely isotropic energies corresponding to two fiber families:

$$W_{ani}^a = \begin{cases} \frac{k_1}{2k_2} [\exp(k_2(J_4^a - 1)^2) - 1] & \text{if } J_4^a \geq 1 \\ 0 & \text{if } J_4^a < 1 \end{cases} \quad (2)$$

where J_4 denotes the fibre stretch. The HGO model has been implemented into an in-house finite element code FER [13, 14].

III. CONTACT MODELING

Without going into details, after spatial and temporal discretization, non linear problems involving contacts are governed by the following nodal algebraic equations defined at each time step:

$$F_i(U) + F_e(t) + F_c = 0 \quad (3)$$

where F_i is the vector of internal forces, F_e denotes the vector of external loads and F_c the vector of contact action/reaction forces. These equations are strongly non-linear with respect to the nodal displacements U , because of finite strains, large displacements of solids and the contact phenomenon (irreversibility of frictional effects).

A typical solution procedure for this type of non-linear analysis is obtained by using the Newton–Raphson iterative procedure:

$$\begin{cases} K_T^i \Delta U = F + F_c \\ U^{i+1} = U^i + \Delta U \end{cases} \quad (4)$$

$K_T^i = -\frac{\partial F_i}{\partial U}$ is the tangential matrix (including mass and stiffness matrix) and $F = F_i(U_i) + F_e(t)$.

The gap vector between two bodies Ω_1 and Ω_2 in the global coordinates system is defined by:

$$X^{i+1} = U_{c1}^{i+1} - U_{c2}^{i+1} + X_0 \quad (5)$$

where U_{c1} (resp. U_{c2}) is the displacement vector of the contact node of Ω_1 (resp. Ω_2) and X_0 is the initial gap vector. The equation (7) can be easily transformed as follows:

$$X^{i+1} = CU^{i+1} + X_0 \quad (6)$$

Where C is a condensation matrix.

By the virtual work principle, we have:

$$R^T \delta X = F_c^T \delta U \Rightarrow F_c = C^T R \quad (7)$$

where R represents the nodal contact forces in the global reference frame (X, Y, Z) . By combining equations (4), (6) and (7), we obtain :

$$X^{i+1} = W R + U_F \quad (8)$$

with

$$W = CK^{-1}C^T; \quad X_F = CK^{-1}F + CU^i + X_0$$

Let Q be the rotation matrix between the local frame (T_1, T_2, N) (Fig.1).and (X, Y, Z) . Let x and r be respectively the gap vector and the contact force vector in the local frame. Equation (8) may be written in the local frame:

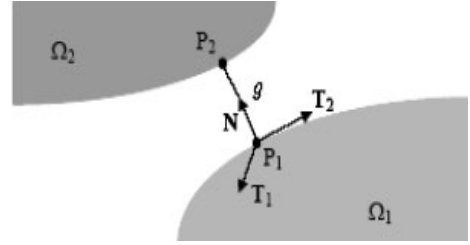


Fig. 1 The gap between two contact points

$$x = wr + x_f \quad (9)$$

where : $x = QX^{i+1}$; $r = QR$; $x_f = QX_f$; $w = Q^T W Q$.

For the following, let x_n be the algebraic value of the normal gap, x_t the tangential part of the gap vector, r_n the algebraic value of the normal contact force and r_t the tangential part of the contact force vector.

The complete contact law (Signorini conditions + Coulomb friction laws) is a complex non-smooth dissipative law including three statuses:

No contact : $x_n > 0$ and $r=0$

Contact with sticking : $\|x_t\| = 0$ and $r \in \text{int}(K_\mu)$ (10)

Contact with sliding : $r \in \text{bd}(K_\mu)$ with $r_t = -\mu r_n \frac{x_t}{\|x_t\|}$

where $\text{int}(K_\mu)$ and $\text{bd}(K_\mu)$ denote the interior and the boundary of the so-called coulomb cone respectively.

DeSaxe and Feng [15] have proposed an augmented Lagrangian formulation of the contact derived from a bipotential function as follows:

$$r^* = r - \rho x^*; x^* = x + \mu \|x_t\| N \quad (11)$$

They have demonstrated that the three possible contact statuses as mentioned in Equation (10) can be stated from the following projection operator: if $r^* \in K_\mu$ (contact with sticking) then $r = r^*$, if $r^* \in K_\mu^*$ (separating) then $r = 0$ and if $r^* \in \mathbb{R}^3 - (K_\mu \cup K_\mu^*)$ (contact with sliding) then r is the orthogonal projection of r^* onto K_μ . K_μ^* is the polar cone of K_μ (Fig.2).

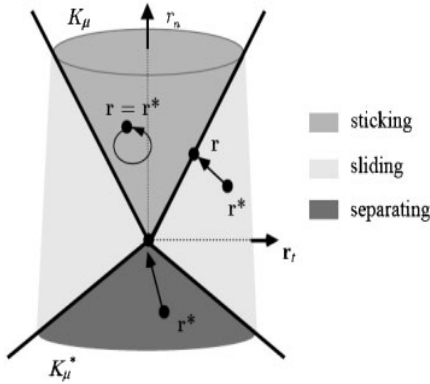


Fig. 2 The Coulomb cone and contact projection operators

Consequently, the projection operation can be explicitly defined by:

$$\text{Proj}_{K_\mu}(r^*) = r^* \quad \text{if } \|r_t^*\| < \mu r_n^* \quad (12)$$

$$\text{Proj}_{K_\mu}(r^*) = 0 \quad \text{if } \mu \|r_t^*\| < -r_n^*$$

$$\text{Proj}_{K_\mu}(r^*) = r^* - \left(\frac{\|r_t^*\| - \mu r_n^*}{1 + \mu} \right) \left(\frac{r_t^*}{\|r_t^*\|} - \mu N \right) \text{ otherwise}$$

The contact solving problem is defined by:

$$\begin{cases} x = wr + x_f \\ r = \text{Proj}_{K_\mu}(r^*) \end{cases} \quad (13)$$

where the unknowns are x and r . The solution can be done by using Uzawa or Newton techniques. Interested readers could find more details in [5].

IV. FINITE ELEMENT MODELLING

As a prerequisite of the 3D FE model, a pre-processing step is accomplished including the main following tasks: first, the constitutive laws of the soft tissues are identified.

The 3D human leg geometries are next identified: the data are acquired with CT scans. The CT images are further delineated in order to identify the leg's boundaries, the curves and 3D surfaces are generated.

The finite element discretization includes 2384 eight-node hexahedral elements and 3418 nodes (see Fig.1); thereafter, the set of nodes and elements is transferred to the finite element code FER/Impact, the results are showed in the post-processor FER/View.

This problem concerns the impact of a hollow flash-ball made of Mooney-Rivlin rubber materials into the human leg of HGO model. The radius of the ball is: $R = 0.015\text{m}$. The initial impact velocity of the ball is: $v = 100 \text{ m/s}$. Fig. 2 shows the distribution of the von Mises stress during the impact. It is noted that the concentration is localized in the impact zone as expected.

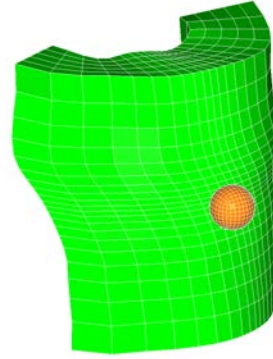


Fig. 3 Initial configuration and mesh

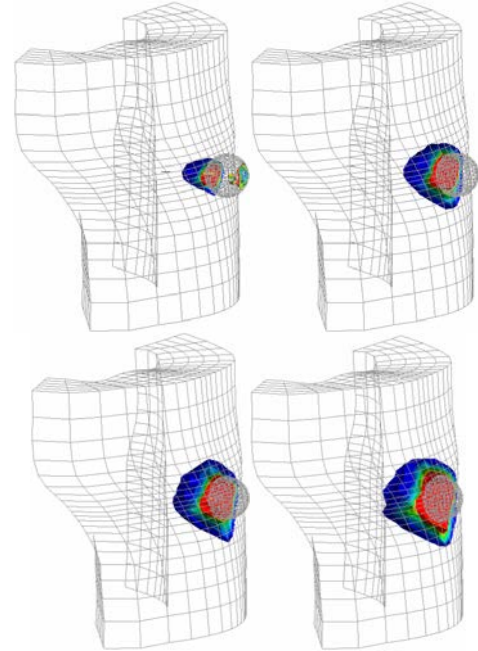


Fig. 4 Distribution of the von Mises stress ($t=2.5e-5$; $t=5e-5$; $t=7.5e-5$; $t=1e-4$)

V. CONCLUSIONS

As we can see in figure 4, we have succeeded to simulate efficiently an impact between two hyperelastic bodies. One is a soft tissue and the other is a rubber material. It is possible to simulate a great variety of injuries such as a flash ball impact in different parts of the human body. This is a first step which opens many perspectives and can propose a tool to help physicians to characterize injuries such as bruises. With this tool, it is possible to have a good idea of the impact force, the dimension of impact surface, the deepness of significant stresses, the energy absorbed by the soft tissue. Of course a lot of problems remain to be solved to have simulations close to reality such as :

- having quantitative measurements of the mechanical properties of biological tissues in vivo.
- improving the modelling of soft tissue by taking into account the water content in the tissue which affects its mechanical properties. The water content in the tissue is also a major factor which determines the thickness of the top layer of the skin.

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