Real Time Volumetric Deformable Models for Surgery Simulation

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Abstract. Surgical simulation increasingly appears to be a mandatory aspect of tomorrow's surgery. From today's gesture training, surgical simulators will evolve in order to perform surgical planning with the ability to rehearse the various steps of the operation, thus reducing operating time and risks. In this paper, we describe a virtual environment for surgical training and more specifically a model based on elasticity theory which conveniently links the shape of deformable bodies and the forces associated with the deformation while achieving real time performance.

1 Introduction

A recent advance in surgery is minimally-invasive techniques. In laparoscopic surgery, the basic technique is to make small incisions through which different surgical tools and a micro video camera can be inserted, avoiding the large incisions needed for open surgery. However, although this method has several advantages, the working conditions are substantially modified for the surgeon in comparison with a classical operation. The major difficulty lies in the hand-eye coordination. Therefore, training is a prerequisite and a simulator would assist surgeons by providing a new way to practice. Much recent work focuses on surgery simulation [4] [9] [2], etc. but none of them are able to simultaneously achieve the goals of real time computation and complex modeling of the biomechanical properties. Our simulator project attempts to develop a system for *gesture training* in hepatic surgery. The advantage of our method lies in the capacity to deform a model of an organ in real-time, using realistic volumetric models of non-homogeneous elastic tissues and including force feedback. This is achieved by using a volumetric mesh, which in addition is well suited for the simulation of tissue cutting.

2 Force Feedback

Recent work has shown that the sense of presence in virtual environments is highly correlated with the degree of perception of that environment [1]. In particular, the sensation of forces and the sensation of textures are very important in medical applications. Since in minimally invasive surgery the surgeon's hands remain outside of the patient's body, we do not need tactile sensing, but only force sensing. The realism of the sensation of forces is highly dependent on the physical realism of the model. Our system, uses a volumetric mesh with non-homogeneous elasticity. We believe this to be necessary for physical realism. The flow of information in the simulator should form a closed loop [6]: the model deforms according to the surgeon induced motion of the force feedback device, this deformation allows us to compute the force of contact and finally the loop is closed by generating this force through a mechanical transmission (see Fig. 3). The main difficulty is related to the real time constraint imposed by such a system: to remain realistic, the forces must be computed at a very high frequency, at least equal to 500 Hz and the deformations at about 50 Hz.

3 Linear elasticity

Most papers dealing with deformable models [11] [12] [8] use physically-based models or realistic dynamics [2] [3]. Linear elasticity is often used [10] [9] [3] as a way to obtain a good approximation of the behavior of a deformable body. The stress-strain relation provided by the theory of elasticity allows us to model various physical behaviors. These relations are importants in our situation since they will give a physical link between the deformation of the body and the force induced in the force feedback system. However, the use of a force feedback system for the interaction with the virtual organ implies several constraints. First, the boundary conditions are different than in a classical approach (a set of external forces applied to the model) since we cannot measure the forces exerted on the virtual organ. Second, the dynamic solving of the time-dependent differential equations do not allow for fast deformations with complex models.

3.1 Theory

We consider a volumetric deformable object and its configuration (shape) Ω before deformation. Under the action of a field of forces, this object will deform and will take a new configuration Ω^*. Then the problem consists of determining the displacement field $\overrightarrow{u}$ which associates to the position P of any point of the object before deformation, its position P^* in the final configuration. If we assume that the Euclidean space $\mathbb{R}^3$ is referred to an orthogonal system $(O, \overrightarrow{e_1}, \overrightarrow{e_2}, \overrightarrow{e_3})$, the displacement $\overrightarrow{u}(P)$, that describes the motion of any point P in the model, can be written as:

$$\overrightarrow{u}(P) = \overrightarrow{PP^*} = \sum_{i=1}^{3} u_i(x_1, x_2, x_3)\overrightarrow{e_i}. \qquad (3.1)$$

From the knowledge of the displacement field $\overrightarrow{u}$, we deduce the components $\varepsilon_{jk}(\overrightarrow{u})$ of the *strain* tensor ε in linear theory to be:

$$\varepsilon_{jk} = \frac{1}{2}\left(\frac{\partial u_j}{\partial x_k} + \frac{\partial u_k}{\partial x_j}\right) \qquad j = 1,2,3 \quad k = 1,2,3. \qquad (3.2)$$

The Hooke's law gives the components $\sigma_{jk}(u)$ of the *stress* tensor σ as:

$$\sigma_{jk} = \sum_{l=1}^{3} \sum_{m=1}^{3} E_{jklm} \varepsilon_{lm}(u). \tag{3.3}$$

For an isotropic material, the coefficients E_{jklm} are given by:

$$E_{jklm} = \lambda \delta_{jk} \delta_{lm} + \mu(\delta_{jl}\delta_{km} + \delta_{jm}\delta_{kl}) \quad \forall \, \varepsilon_{jk} = \varepsilon_{kj} \tag{3.4}$$

where the scalars λ and μ are the *Lamé coefficients* and δ the Kronecker symbol.

3.2 The finite element method

More and more researchers tend to use finite element methods to solve the equations governing deformable models [5] [8]. In our approach, we have used a classical finite elements scheme, i.e. with Lagrange tetrahedral elements of type P_1 (linear elements with zero order continuity, only involving the values of the function at the vertices of the element). The use of this class of elements implies the decomposition of the domain Ω into a set of tetrahedral elements. Through variational principles, the resolution of the problem of elasticity theory becomes equivalent to the resolution of a linear system $\boldsymbol{K}\,\boldsymbol{U} = \boldsymbol{F}$ where $\boldsymbol{K}$ is the stiffness matrix, $\boldsymbol{U}$ is the unknown displacement field and $\boldsymbol{F}$ the external forces. The size of the matrix $\boldsymbol{K}$ is $3\,N \times 3\,N$, where N is the number of vertices (or nodes) of the mesh, each having 3 degrees of freedom. Clearly, the size of the mesh (in terms of number of nodes) is an important parameter which influences the computation time. Since we want to model organs with anatomical precision, it appears that the resolution of such a linear system cannot be computed in real time without the implementation of a speed up algorithm (see Section 4).

3.3 Boundary conditions

Let u be the displacement field and suppose we know the value of the displacement of the node i on the boundary: $\overrightarrow{U} = \{U_1, U_2, U_3\}$. Practically, $\overrightarrow{U}$ corresponds to the displacement of the node i of the model in contact with the virtual tool. Then, to take into account this boundary condition in the resolution step of the linear system, we modify the matrix $\boldsymbol{K}$ in the following way: $K_{3i+j,k} = 0$, for $j \in \{1,2,3\}$, $k \in \{1,..,n\}$ and $k \neq 3i+j$. $K_{3i+j,3i+j} = 1$ for $j \in \{1,2,3\}$. In matrix $\boldsymbol{F}$, the following substitution is done: $F_{3i} = U_1$, $F_{3i+1} = U_2$ and $F_{3i+2} = U_3$. However, although the previous boundary conditions are well suited for mechanical systems, they are usually too much restrictive for anatomical modeling. For this reason, we have added another boundary condition that can be viewed as attaching a spring of infinite length to the node i (see Fig. 2). According to the stiffness k of the spring, we can simulate *weak* or *strong* constraints, i.e. the node i can move off freely or not from its equilibrum position. Many of the ligaments attached to an organ can be modelled this way. This constraint, expressed as an external force $\overrightarrow{f}$, can be written: $\overrightarrow{f} = k(\overrightarrow{v} \cdot \overrightarrow{n})\overrightarrow{n}$, where k is the spring stiffness, $\overrightarrow{n}$ is a normalized vector in the direction of the spring and $\overrightarrow{v}$ the displacement of the node i. This relation is valid under the assumption $\|\overrightarrow{l_0}\| \gg \|\overrightarrow{u}\|$ ($\|\overrightarrow{l_0}\|$ is the spring rest length) and governed by the necessary linearity of the previous expression in order to achieve real time deformations (see Section 4).

3.4 Non homogeneous material

Although we have poor knowledge at the biomechanical properties of most organs, we are aware that linear elasticity is just an approximation. However, to take into account some anatomical characteristics in the deformation process, we use non-homogeneous elasticity. At the present time, we use *a priori* knowledge to set the elastic coefficient of any node in the volumetric model, according to the grey levels in the 3D medical image. Nevertheless, some works [7] suggest it will be possible to compute these parameters from medical image modalities.

4 Real time deformations

The degree of realism required in surgical simulation necessitates a complex model of the organ. The complexity (number of vertices) of the mesh has a direct impact on the size of the matrices involved in the linear system $\boldsymbol{KU} = \boldsymbol{F}$. Then the computation time required for resolution of the system becomes too high for real-time deformation of the mesh. To speed up the interactivity rate, we take advantage of the linearity of the equations combined with a pre-processing algorithm. This pre-processing algorithm can be described as follow:

- set all the constrained nodes of the model
- for each free node n on the surface of the mesh, apply an elementary displacement (dx, dy, dz) and compute:
  - the volumetric displacement field corresponding to this deformation.
  - store the displacement of every free node in the mesh as a set of 3×3 tensors T_{nk} expressing the relation between the displacement of node k $(k \neq n)$ in the mesh and the elementary displacement of the node n.
  - compute the components of the elementary force f_n.
  - store this result as a 3×3 tensor F_n.

To decrease the memory required for storage of the tensors, we keep only the significant displacements i.e. tensors such that: $\lambda >$ threshold, with $\lambda = max\{|\lambda_1|, |\lambda_2|, |\lambda_3|\}$ where $\lambda_1, \lambda_3, \lambda_3$ are the eigenvalues of the tensor T_{nk}. In the case of a liver model, composed of 940 nodes, a reduction of 60% was obtained this way. Finally, the computation of the displacement field and the reaction forces are reduced to a linear combination, according to the displacement $(\delta x, \delta y, \delta z)$ imposed at a set of nodes on the surface of the mesh. After that, it becomes possible to deform, *interactively*, the model of any object, whatever its complexity (see Fig. 1).

5 Experiments

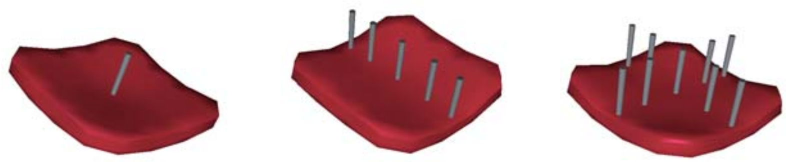

Figure 1. Deformation of a plate. The nodes on corners are rigidly fixed. (*Left*) the plate is deformed under a simple contact. (*Middle*) and (*right*) several constraints are combined to give the final deformation, in real-time.

a) high spring stiffness b) low spring stiffness

Figure 2. The left and right sides of the plate are linked to springs of various stiffness. In *a)* the deformation is close to what occurs with fixed nodes whereas in *b)* the deformation is smaller but coupled with a translation in the direction of the tool.

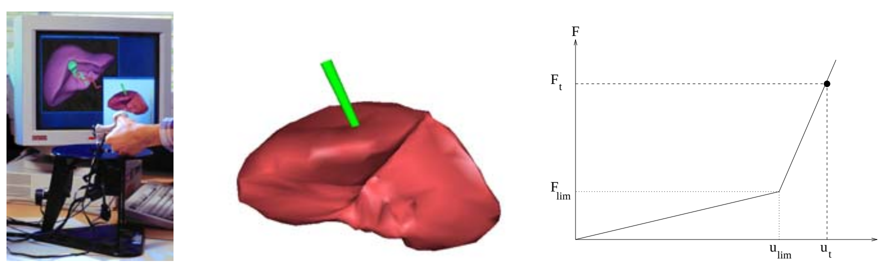

Figure 3. (*Left*) the surgeon moves a virtual tool via the force feedback device (*middle*) after collision detection the model of the organ deforms in real-time and (*right*) a non-linear reaction force is computed and sent back to the force feedback system.

6 Conclusion - Perspectives

In this paper we have proposed a method for deforming in real time a volumetric model of an object. Because our models are physically-based, they respond in a natural way to the interactions provided by the user. These interactions are guided via a force feedback device, and elasticity theory allows computation of the contact forces which are input in the mechanical system.

We are now working on dynamical tissue cutting simulation, using spring-masses models in combination with the method presented here. We also plan to add a dynamical behavior to our deformations in order to incorporate the velocity of the tool when it gets into touch with the organ in order to increase the realism of deformations.

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