REAL TIME SIMULATION OF DEFORMABLE OBJECTS WITH FORCE FEEDBACK

Application to surgery simulation

Moulay Brahim El Khalil Ghembaza, Karim Djouani and Yacine Amirat
LISSI Laboratory, Paris 12 University, 120-122 rue Paul Armangot, 94400 Vitry-sur-Seine, France

Keywords: Medical simulators, Physically-based models, Deformable objects, Virtual reality, Continuous collision detection, Haptic interaction.

Abstract: This paper presents some issues in the simulation of deformable objects with force feedback. It presents an overview of our approach for the conception of a virtual reality medical simulator. We describe a new base finite element method (Extended Tensor-Mass Model) suitable for soft tissue simulation under real time constraints. Our approach allows fast computation of non-linear and viscoelastic mechanical deformations and forces. As far as real-time interactions are concerned, we present our work on collision detection and haptic interaction. Thus, for contact management, a continuous collision detection method based on cubic spline parametric approximation is proposed. Finally, interactive endovascular simulator is described.

1 INTRODUCTION

The variety and the complexity of medicine have made it for a long time force of progress for many scientific and technical fields. The medical domain is among the main application areas for numerical imaging and vision since their beginnings. In parallel, the graphical tools, computer science and robotics have become central for modern medicine. These assistance tools are part of what is called Surgetics (a new generation of Computer- and Robot-Assisted Surgery systems). The surgery simulation, which constitutes actually an active research field, concerns some practical tool designs that allow to offer experts at the same time the possibility to practise intensive training for operative gestures, unrestricted by ethical problems, and the ability to plan with precision some interventions and surgical procedures.

During the 1990s, a great interest for medical procedures simulation has been developed. The earlier simulators had been developed for navigation within 3d-anatomical data bases and found many applications in education and training. These simulators used only geometrical models for the anatomical structures, without taking into account their physical reality.

Therefore, new simulators have been proposed in order to overcome the drawbacks cited above, by using more realistic physical modelling of the various anatomical structures and their interactions (Moline, 1997), (Gross, 1999). Taking into account the physical phenomena should not only allow to improve quality of the medical simulators, but also to widen considerably their field of application.

The main components of an advanced surgery simulation environment can be summarised by the following elements:

1. High precision of 3d-data acquisition is accomplished by medical imaging and/or vision systems. A pre-treatment extracts the anatomical structures and creates geometrical models;
2. Accuracy of deformation modelling of the biological tissues and surgical tools, allowing the surgeon to modify geometry and topology of various Virtual Environment “VE” objects for incisions simulating, tissues repositioning, transplantation, cutting, perforation, etc;
3. Continuous collision detection algorithm between deformable and/or rigid VE objects and contact-friction management;
4. Realism by haptic rendering synthesis (contact efforts of medical tools and biological tissues), and real-time insuring (Parallel computing), using adequate interaction devices to feel the haptic sensations met in a conventional surgery are required;
5. Specific problems for some interventions: assistance by Virtual Fixtures during navigation

Brahim El Khalil Ghembaza M., Djouani K. and Amirat Y. (2005).
REAL TIME SIMULATION OF DEFORMABLE OBJECTS WITH FORCE FEEDBACK - Application to surgery simulation.
In *Proceedings of the Second International Conference on Informatics in Control, Automation and Robotics - Robotics and Automation*, pages 310-315
DOI: 10.5220/0001185203100315

of tools in hollow bodies, in order to improve the precision and the security of the operative gesture, particularly at the time of guidance and placement of the surgical tools.

Thus, the design of a surgery simulation environment requires finding a compromise between complexity of the adopted models and fast calculation of the algorithms.

The search for an efficient method for real-time computation of nonlinear elastic mechanical deformations is only at its beginning. We can distinguish two main communities. On the one hand the biomechanics community is interested into the precise characterization of the behaviour laws of certain biological tissues, but without being concerned with computation time. On the other hand, the computer graphics expert's community is committed to the development of deformable object simulators for biomedical applications, but adopts very simple mechanical models, in generally linear elastic ones, and without being concerned with matching these mechanical models to the experimental behaviour of biological tissues.

Recently, some approaches have been proposed (Liu et al., 2004), (Schwartz et al., 2004) in order to take into account the requirements of the two communities cited above: biomechanical modelling of soft tissues under real-time constraint. Our work joints the same point of view while aiming at proposing an approach which is sufficiently simple and quick to be compatible with real-time applications. We try to reproduce as exactly as possible the real biological tissue behaviour obtained by biomechanical experimentation.

The present paper is organized as follows. In section 2, dynamic simulation of biological tissues has been addressed and a new approach is presented for deformation and forces modelling. In section 3, we lay out how we handle some aspects of real-time interaction that are inevitable in current medical simulators such as collision detection and haptic interaction. In section 4, interactive endovascular procedures are described and our simulator workbench is presented. Finally, some conclusions and perspectives are given.

2 DYNAMIC SIMULATION OF SOFT TISSUE

Modelling soft tissues consists of formulating constitutive equations related to their deformation. A survey of deformable object modelling was done by (Gibson and Mirtich, 1997). In brief, they divided the work done on deformable objects into two parts: non-physically based models and physically based models. Physically based models can further be divided into discrete object models and other models based on continuum mechanics. The latter is usually solved using the finite element method (FEM).

In advanced simulation environments, accurate process modelling at the geometrical, physical and physiological level is required. The biological tissue's mechanical deformation modelling constitutes an essential part of surgery simulation (Delingette, 1998). It runs up against two fundamental and antagonistic constraints: on one hand the realism of numerical modelling and, on the other hand, the computation speed.

2.1 Linear Tensor-Mass Model

The most promising approach towards real-time computation of nonlinear viscoelastic deformation appears to be the tensor-mass model introduced by (Cotin et al., 2000). The tensor-mass algorithm for linear elasticity is both time-efficient and physically accurate. It also allows local topological changes of mesh elements, so that simulation of cutting or perforation is possible.

The force $F_{T_i(j)}$ applied to a summit $P_{T_i(j)}$ of tetrahedron T_i is defined as follow:

$$F_{T_i(j)} = \sum_{k=0}^{3} \left[K_{jk}^{T_i}\right] \cdot P_{T_i(k)}^0 P_{T_i(k)} \qquad (1)$$

where $P_{T_i(k)}^0$ are the rest positions of the four vertices of tetrahedron T_i, $P_{T_i(k)}$ are the current positions of the vertices, and $\left[K_{jk}^{T_i}\right]$ are 3×3 stiffness tensors depending only on the rest geometry of tetrahedron T_i and on the Lamé coefficients. These tensors can be pre-computed; therefore computation at run-time is restricted to matrix-vector multiplications and matrix summations. Given a complete mesh, the total elastic force F_i applied on a vertex P_i is obtained by summing the forces contributed by all adjacent tetrahedrons of T_i:

$$F_i = \left[K_{ii}\right] \cdot P_i^0 P_i + \sum_{j \in N(P_i)} \left[K_{ij}\right] \cdot P_j^0 P_j \qquad (2)$$

where $\left[K_{ii}\right]$ is the sum of tensors $\left[K_{ii}^{T_k}\right]$ associated with the tetrahedron adjacent to P_i, $\left[K_{ij}\right]$ is the sum of tensors $\left[K_{ij}^{T_k}\right]$ associated with the tetrahedron adjacent to edge (i, j), and $N(P_i)$ is the neighbourhood of vertex P_i. The resulting system has to be solved dynamically.

Based on FEM theory, we have proposed a new approach for physically-based deformable modelling as an extension of the linear elastic tensor-mass method. Our approach allows fast computation of non-linear and viscoelastic mechanical deformations and forces. Its principle consists of pre-computing a certain number of tensors depending on the geometrical and mechanical characteristics of each finite element, which are combined dynamically during the simulation phase. The proposed method is sufficiently generic to be applied to a large variety of behaviours and objects, as soft biological tissue deformation under real-time computation requirements.

2.2 Non-linear viscoelastic Tensor-Mass Model

We now present the non-linear and viscoelastic Tensor-Mass Model, (Ghembaza et al., 2005), for soft tissue simulation.

As a first step we show that adequate real-time correction of linear elasticity parameters allows to model different types of non-linear elastic deformations. In our model, expression of force $F_{T_i(j)}$ applied on vertex $P_{T_i(j)}$ within a tetrahedral mesh element Ti is:

$$F_{T_i(j)} = \sum_{k=0}^{3}\left(\left[K_{jk}^{T_i}\right] + \delta\lambda(T_i)\left[A_{jk}^{T_i}\right] + \delta\mu(T_i)\left[B_{jk}^{T_i}\right]\right)\cdot P_{T_i(k)}^{0} P_{T_i(k)} \quad (3)$$

where $\left[K_{jk}^{T_i}\right]$, $\left[A_{jk}^{T_i}\right]$ and $\left[B_{jk}^{T_i}\right]$ are 3×3 tensors, λ_i and μ_i are the Lamé coefficients of the material, and $\delta\lambda_i$ and $\delta\mu_i$ are non-linear corrections. Tensors only depend on the geometry at rest so that pre-computation is possible.

The linear tensor-mass method can typically deal with meshes of a few thousand nodes in real-time. The non-linear extension increases computing time by a factor 4. However the method remains suitable for real-time applications, due to the fact that the non-linear computational overhead is restricted to a limited number of mesh elements where the highest deformation rates occur (Ghembaza et al., 2005).

The tissue's mechanical properties are defined locally for every finite element by a stiffness tensor associated with this element. Thus, it is possible to build inhomogeneous models, composed of various structures with different mechanical properties.

Viscosity can easily be introduced into the tensor-mass model, under assumption that the behaviour is restricted to a simple linear viscous relation. We have introduced a viscous force that is proportional to the deformation speed and to a viscosity coefficient η. After discretisation into a tetrahedral mesh, the expression obtained is very similar to (1), except that deformation speed replaces deformation, and a viscosity tensor replaces the stiffness tensor. Expression of the viscous force $F_{T_i(j)}^{v}$ applied to a vertex $P_{T_i(j)}$ of tetrahedron T_i is given by:

$$F_{T_i(j)}^{v} = \sum_{k=0}^{3}\left[K_{jk}^{T_i}\right]^{(v)} \cdot \frac{d}{dt} P_{T_i(k)}^{0} P_{T_i(k)} \quad (4)$$

where $\left[K_{jk}^{T_i}\right]^{(v)}$ are 3×3 tensors depending only on the rest geometry of tetrahedron T_i and on the viscosity coefficient η.

We apply the complete algorithm presented in (Ghembaza et al., 2005) to simulate dynamically the different behaviours.

3 REAL-TIME INTERACTIONS

The surgeon's assistance for preoperative, as well as for intra-operative gestures, requires both a real-time realistic visualization (soft tissue deformation modelling) and force feeling (haptic rendering interface).

The goal of a medical simulator is to allow real-time interactions with realistic modelling. It is well known that during a simulation, given any physical model, the most difficult aspects, in terms of computational time or updating data structures, are collision detection, the different rates for haptic interaction between graphical updates and physical simulation and topology modification during specific surgical procedures. We address these problems in the following sections.

3.1 Contact Management

The collision detection is the most critical step in dynamic simulation, because it requires a very significant calculating time compared to that necessary to the calculation of movement and deformations of the objects. Figure 1 shows the contact's management procedure.

When dealing with contact characterization, the goal is to detect if, when, and where objects collide. To deal with this computationally demanding problem in our simulation, we use an inspired continuous collision detection method (i.e. collision detection must be able to return the first collision time) from (Redon et al., 2002) without taking into account of friction between a deformable and a rigid bodies.

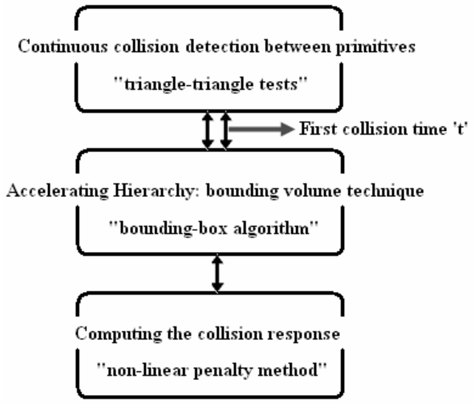

Figure 1: Algorithm to detect and manage all contact elements

Indeed, one of the major drawbacks of the discrete collision detection methods is that it can "miss" the collisions if the speed of object is very high. Moreover, discrete collision detection requires backtracking methods to compute the first contact time, which is necessary in constraint-based analytical dynamics simulations. Continuous collision detection methods overcome such a problem, because they interpolate the trajectory of every element (triangle in our case) between two sampling time and thus calculate the minimal time corresponding to the first collision. Very few continuous methods have been proposed in the literature. The approach developed by (Redon et al. 2002) is well adapted to treat the collisions between rigid objects. We propose an extension of this approach to treat collisions between deformable objects, using parametric approximation method (Cubic Spline representative the deformation trajectory) to interpolate the mesh elements (figure 2).

An "Accelerating Hierarchy" approach (bounding volume technique) is implemented in order to decrease the number of "triangle-triangle tests" and thus to increase the speed of the algorithm. Thus, a recursive division of the space containing the whole of the triangles, based on a "bounding-box" algorithm (Axis-Aligned Bounding Boxes: AABB) (van den Bergen, 1997) has been implemented.

However, computing the collision response requires us to evaluate the involved local deformation of the colliding objects, using a non-linear penalty method (Deguet et al., 1998) (Moore et al., 1988) to ensure the separation of the objects in collision. This can be done by determining the fictitious interpenetration of the objects. Figure 3 shows these stages.

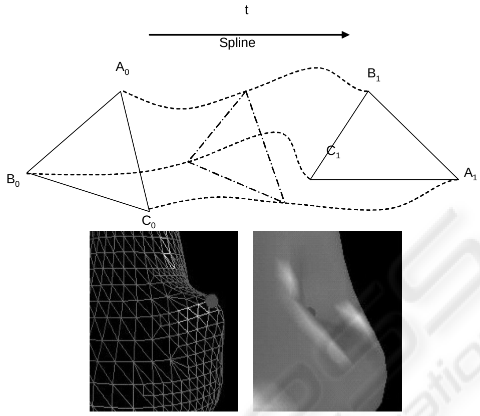

Figure 2: Cubic Spline based deformation trajectory

3.2 Haptic Rendering

Haptic systems gives people the sensation of touching objects in virtual environments or teleoperation applications. Including haptic technology improves the perception of a surgeon leading to a deeper sense of immersion. Many problems arise in haptic applications especially in the case of deformable objects manipulation, for instance computational time, numerical instability in the integration of the body dynamics, time delays, etc. Lengthy computations are forbidden in haptic systems which need high simulation rates (about 300 Hz to 1 KHz) to obtain realistic force feedback. The update rates of the physical objects being simulated is normally of the order of 25 to 60 Hz. This difference in simulation rates can cause an oscillatory behaviour in the haptic device that can become highly unstable (Adams et al., 1999). Several numerical approaches (Cavusoglu et al., 2000) have been proposed to solve the difference rate problem.

For our purpose, the objective is to develop robust and rapid algorithms which allow haptics feedback for deformable objects. The reaction calculation is ensured by a compliance method (interaction between a flexible or rigid body, the surgical tool for instance, and deformable body as soft biological tissues).

The reaction force ($\vec{F}_c$) is calculated using the minimal distance dist between the local model (soft biological tissue) and the haptic tool.

Thus, the force vector $\vec{F}_c$ is given by:

$$\vec{F}_c = \begin{cases} -k \cdot dist \cdot \vec{n} - b \cdot (\vec{v} \cdot \vec{n}) \cdot \vec{n} & if\ dist < 0 \\ \vec{0} & otherwise \end{cases} \quad (5)$$

Where k is the rigidity coefficient, b is a damping coefficient, dist is the penetration depth between the two bodies, $\vec{v}$ is the relative linear velocity of these two objects at collision. $\vec{n}$ is the normal direction of contact.

4 APPLICATION

The application concerns endovascular procedure simulation with physical deformation modelling of the Abdominal Aorta and Aneurysm (AAA). The latter is an arterial wall pathology involving a permanent dilation of the abdominal aorta which can be life-threatening. The endovascular procedure is mainly used for the treatment of AAA. The goal of the intervention is therefore to repair the swelling and prevent the rupture of the aneurysm. The prosthesis is hooked from inside the aorta into its wall with a stent. Figure 3 shows the prosthesis deployment process in an AAA endovascular procedure.

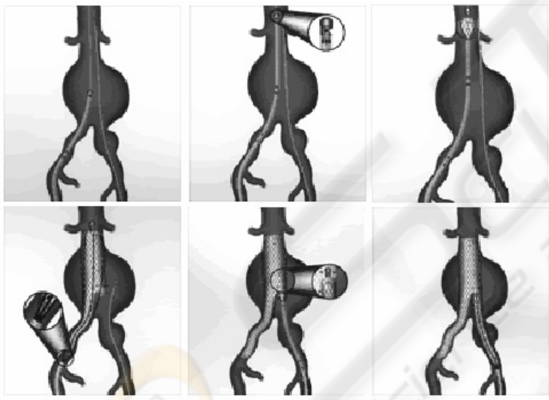

Figure 3: Prosthesis deployment procedure

The endovascular treatment is a complex surgery (Hausegger et al., 2001) which has not been deeply investigated and no simulators are available at the moment. The currently used catheters are passive. They cause important frictions with the aorta, leading to risk of damage, making the prosthesis delivering difficult and may even lead to failure. This phenomenon is also amplified by the lack of tactile sensation. In order to overcome these drawbacks, an active compliant micro-mechanism, (figure 4), is used to help the surgeon (Djouani et al. 2002). Therefore it is interesting to build a simulator allowing the surgeon to practice such a technique which offers a rich environment to practice a variety of aneurysms models and to manage different possible complications.

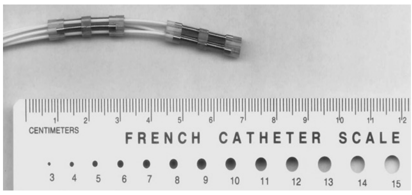

Figure 4: Experimental device in scale 1

The dynamic simulator and the haptic interface are designed as independent processes and they are connected via a local model. Figure 5 shows the basic visual-haptic platform. It has been implemented using C++/OpenGL and it provides force feedback through means of an haptic interface of type 6 dof PHANToM. The visual-haptic platform uses a PC - 2.8 GHz Pentium IV Intel processor with 512 MB of RAM.

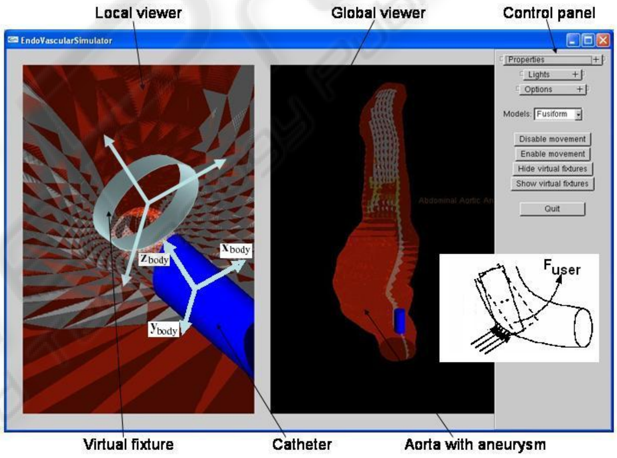

Figure 5: Haptic interaction environment

The dynamic simulator carries out the physical simulation, collision detection and the graphical rendering of different deformable objects in the virtual environment. It receives the haptic position from the PHANToM (and eventually, the distance between the haptic point and the local model) and sends to the haptic process the different parameters (i.e. set of colliding facets between the deformable virtual object and the tool) to update the local model. This update process is repeated at a rate of 30 Hz.

For the soft biological tissues modelling, we must deal with mechanical deformation of the aortic aneurysm tissues under contact with the catheter. These tissues present complex behaviours, showing viscoelasticity and anisotropy among other things. In (Watton et al., 2004), the authors provide rare experimental data on the aneurysm behaviour.

As the objective is to adapt the shape of the active catheter to the geometry of inspected vessel, we implement the algorithm proposed in section 3 for contact management. When an effort is sensed, an order is reported to the surgeon so as to decrease the interaction effort. This method is particularly well adapted in inspections of human vascular networks and allows, even if there are contact zones, to strongly limit the importance of the interaction efforts.

5 CONCLUSION

We have addressed some important issues in the conception of a realistic virtual medical simulator. We have presented some theoretical aspects in order to ensure real-time computation with realistic biomechanical modelling. Thus, we have described two main computational aspects to deal with deformable virtual objects simulation. The first aspect concerns the formulation of the deformation model that meets both fast graphics/haptics rendering rates and actual physical law accuracy. The second aspect concerns the contact management in the case of deformable and/or rigid object interaction. For continuous collision detection we use bounding volume techniques which we believe to be suitable for deformable objects like virtual organs in medical simulators. Finally, for haptic rendering, a non-linear penalty method has been used for the reaction force computation. Based on our approach, finally our endovascular simulator has been presented. Complexity analysis, serial and parallel algorithms are under study for the soft biological tissues simulation.

REFERENCES

Adams, R., and Hannaford, B., 1999. Stable haptic interaction with virtual environments. *In IEEE Transactions on Robotics & Automation 99*.

Cavusoglu, M., and Tendick, F., 2000. Multirate simulation for high fidelity haptic interaction with deformable objects in virtual environments. *In Int. Conf. on Robotics and Automation '00*.

Cotin, S., Delingette, H., and Ayache, N., 2000. A Hybrid Elastic Model for Real-Time Cutting, Deformations, and Force Feedback for Surgery Training and Simulation. *Visual Computer*, Vol. 16, No. 8, pp. 437-452.

Deguet, A., Joukhadar, A., and Laugier, C., 1998. Models and algorithms for the collision of rigid and deformable bodies. *In Robotics: the algorithm perspective, Proc. of the Workshop on the Algorithmic Foundations of Robotics*. Houston, USA.

Delingette, H., 1998. Toward realistic soft tissue modeling in medical simulation. *Proc. of the IEEE Special Issue in on Virtual and Augmented Reality in Medicine*, Vol. 86, No. 3, pp. 512-523.

Djouani, K., Bailly, Y., Amirat, Y., and Francois, C., 2002. A micro device design and control for endovascular stent-grafts delivering in aortic aneurysm treatments. *CARS'02 Computer Assisted Radiology and Surgery*.

Ghembaza, M.B.K., Amirat, Y., Djouani, K., and Daachi, D., 2005. Deformation Modelling for Surgery Simulation. *IEEE SMC'05 International Conference on Systems, Man and Cybernetics*. Hawaii, USA, to appear.

Gibson, F., and Mirtich, B., 1997. A survey of deformable models in computer graphics. *Tech. Rep.* MERL, Cambridge, MA, http://www.merl.com/reports/TR97-19/index.html.

Gross, M.H., 1999. Surgery simulation - a challenge for graphics and vision. *VMV '99 (Vision, Modeling and Visualization Workshop)*. Erlangen, Germany.

Hausegger, K.A., Schedlbauer, P., Deutschmann H.A., and Tiesenhausen, K., 2001. Complications in endoluminal repair of abdominal aortic aneurysms, *European Journal of Radiology*. 39 (1), pp. 22-33.

Liu, Y., Kerdok, A.E., and Howe, R.D., 2004. A nonlinear finite element model of soft tissue indentation. *2nd International Symposium on Medical Simulation.* Springer Verlag, Boston, MA, pp. 67-76.

May, J., White, G.H., and Harris, J.P., 2001. Endoluminal repair of abdominal aortic aneurysms - state of the art. *European Journal of Radiology*. 39 (1), pp. 16-21.

Moline, J., 1997. Virtual reality for health care: a survey. *In Virtual Reality in Neuro-Psycho-Physiology*. G. Riva Editor, Ios Press, Amsterdam.

Moore, M., and Wilhems, J., 1988. Collision detection and response for computer animation. *In Proceedings of SIGGRAPH*.

Redon, S., Kheddar, A., and Coquillart, S., 2002. Fast Continuous Collision Detection between Rigid Bodies. *EUROGRAPHICS*, Vol. 21, No. 3.

Schwartz, J.M., Deniger, M., Rancourt, D., Moisan, C., and Laurendeau, D., 2004. Modeling Liver Tissue Properties Using a Non-Linear Visco-Elastic Model for Surgery Simulation. *Medical Image Analysis Journal*.

van den Bergen, G., 1997. Efficient Collision Detection of Complex Deformable Models using AABB Trees. *Journal of Graphics Tools*. 2 (4), 1-13.

Watton, P.N., Hill, N.A., and Heil, M., 2004. A Mathematical Model for the Growth of the Abdominal Aortic Aneurysm. *Biomechanics and Modeling in Mechanobiology*, Vol. 3, pp. 98-113.