*Medicine Meets Virtual Reality*
*J.D. Westwood et al. (Eds.)*
*IOS Press, 1999*

# Thin Walled Models for Haptic and Graphical Rendering of Soft Tissues in Surgical Simulations

*Suvranu De, Mandayam A. Srinivasan*
*Laboratory for Human and Machine Haptics, Department of Mechanical Engineering and Research Laboratory of Electronics,*
*Massachusetts Institute of Technology, Cambridge, MA, 02139*

## 1.Introduction

This paper presents a novel modeling paradigm for physically based, real-time rendering of interactions between surgical tools and soft tissues for surgical simulations in multimodal virtual environments (VEs). Such VE systems require graphical rendering of organ motion and deformation together with haptic rendering of tool-tissue interaction forces [1]. We will concentrate on minimally invasive surgery where the surgeon uses long slender tools in conjunction with visual feedback from a video monitor. Because such surgical procedures have to be performed with obstructed vision and very little haptic cues, the surgeon typically has to undergo considerable training. Surgical simulators are designed to substitute for and eventually replace cadavers and animals which are currently used for such training. To prevent negative training transfer, the models used in such simulations need to be as physically based as possible. On the other hand, for the simulation to appear realistic, it should be in real time. This requires a visual update rate of at least around 30Hz and a haptic update rate of around 1 kHz. Accurate simulation of tool-tissue mechanics is computationally very intensive due to inherent complexities of the governing partial differential equations and the nonlinearities resulting from large deformations and material behavior [2]. Therefore, optimal choice of organ models with respect to computational speed and accuracy, satisfying the criterion of real time rendering is crucial. We propose here models and algorithms for physically based rapid graphical and haptic rendering, especially those encountered during palpation, piercing or incision of soft tissues.

By "physically based" we mean that the models are based on (1) adequate representations of the geometry of the organ, (2) suitable material constitutive relations which capture the salient traits of the tissue behavior, and (3) boundary conditions that are properly imposed. One of the most difficult issues in physically based modeling of soft tissues is a proper representation of tissue properties. Living tissues exhibit a highly nonlinear force-displacement behavior. Another salient feature of living tissues is that they can undergo very large deformations. Whereas in metals elastic strains hardly exceed a few percent, strains of the order of 100 percent are not uncommon in soft tissue systems. Living tissues also have a rate-dependent viscous behavior. Moreover, during the tool-tissue interactions, the conditions of contact change (e.g., the region of contact depends on the net force of contact) giving rise to contact nonlinearities. Over and above this is the problem of dealing with change in the topology of the organ and the boundary conditions during singular processes such as cutting or incision. Choice of a physically realistic simulation process for such an operation is quite a challenging problem.

## 2.Thin Walled Models

The three dimensional nature of the real world results in considerable computational burdens when it comes to modeling the objects we see around us. Moreover, if we consider dynamics, the fourth dimension of time adds to the complexity. But with respect to haptic and visual interactions with objects, we make the following observation. While touching an object through a tool, we *see* the surface deforming under pressure and *feel* the net force due to the traction distribution within the area of contact between the tool and the object. Hence, if by some means, we could reflect the properties of the material constituents "inside" the solid to its surface, the computational burden could be reduced significantly. This is the basic idea behind the approach developed in this paper.

We view volumetric solid objects as being represented by "thin-walled" structures for the computation of surface deformations and interaction forces. Thin walled structures are found all around us. The simplest example is a balloon. When inflated, it is a thin walled membrane filled with a gas. A more complex example of a thin walled structure is an auto-body or the fuselage of an aircraft. The important point is that such structures are common and we have efficient mechanistic techniques for solving their mechanical behavior. The novelty of our approach is that we model general three dimensional deformable bodies as "thin-walled" structures so far as visual and haptic interaction with them are concerned. In the language of Mechanics, thin walled structures are broadly classified into *membranes*, structures with essentially no bending stiffness compared to the in-plane stiffness, and *shells*, structures in which bending behavior is also important.

A wide class of compliant organs like the stomach, gall bladder, etc., may be modeled as membranes enclosing a fluid, much like "water-beds" [3]. The degree of compressibility of the organ can be controlled by defining an appropriate Bulk Modulus for the fluid inside. When bending stiffnesses are not negligible compared to the in-plane membrane stiffnesses, the model can be extended by replacing the membrane with a shell structure with or without fluid inside. Figure 1 shows the boundary value problem associated with the modeling of a generic organ as a membrane enclosing a fluid of Bulk Modulus $\kappa$. In the reference configuration (corresponding to time 0) the geometry of the organ is expressed with respect to an orthogonal reference frame and displacements on a region $S_u$ of the boundary are specified. The tool tip is modeled as a point. Three components of displacement are specified along three orthogonal directions at the point where the tool tip contacts the surface. This results in a deformation field which maps the reference configuration to the final configuration. The problem is to compute this deformation field and the reaction forces at the tool tip. A Total Lagrangian [4] description is adopted and the Virtual Work Principle is written as follows:

***Virtual Work Principle*** :

$$\int_{V^0} {}_0^tS_{ij}\delta\ {}_0^t\varepsilon_{ij}dV = {}_0^t\Re$$

${}_0^tS_{ij}$ = Second Piola Kirchhoff Stress.

${}_0^t\varepsilon_{ij}$ = Green Lagrange Strain.

${}_0^t\Re$ = Total Virtual Work done by all external loads (e.g., body force, surface tractions, inertia, etc.)

***Boundary Conditions:***
$U = U^{sp}$ on $S_u$ of the boundary.

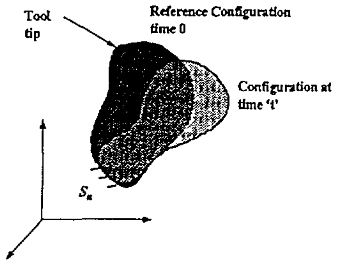


**Figure 1** A generic organ is modeled as a membrane filled with an incompressible or compressible fluid. It is subjected to zero displacements along the hatched boundary and specified displacements at the point where the tool tip contacts the membrane surface.

It is to be noted that this formulation is very general. It is true for nonlinear strain-displacement as well as nonlinear stress-strain relations of the material of the membrane and for both static and dynamic analysis. Once the mathematical model is defined, we discretize the membrane surface using finite elements. The "surface model" of the organ, used to define its geometry in computer graphics algorithms, is adopted as the surface of the thin-walled structure. But, unlike the graphical surface models, we endow the organ surface with a thickness that can be variable across the surface. Finite element analysis is performed by discretizing the membrane with the same triangular elements used in representing the organ geometry graphically (see Fig. 2). The virtual work principle is used to derive the incremental equations of motion, thereby transforming the nonlinear problem into a sequence of simpler linear problems. The response at time 't' is used to obtain the response at time 't + Δt' by solving the following relationship:

$$ {}^t[\mathbf{K}]\{\Delta^t\mathbf{u}\}=\{{}^t\mathbf{R}\} $$

${}^t[\mathbf{K}]$ = Tangent Stiffness Matrix.

$\{\Delta^t\mathbf{u}\}$ = Incremental displacement vector.

$\{{}^t\mathbf{R}\}$ = Nodal Load vector.

***Displacement Update***

$$ {}^{t+\Delta t}\mathbf{u} = {}^t\mathbf{u} + \Delta^t\mathbf{u} $$

The effect of the internal fluid pressure manifests itself in two ways. First, it adds extra terms to the tangent stiffness matrix and secondly, it shows up as an applied force term in the equilibrium equations. The choice of simple triangular elements results in closed form computation of the global tangent stiffness matrices, resulting in a substantially accelerated computational procedure.

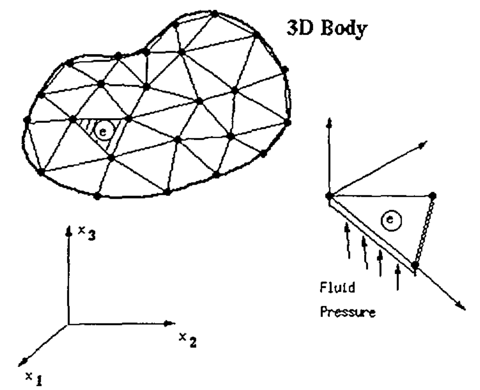


**Figure 2** A general three dimensional body is modeled as a membrane filled with a fluid and discretized with linear triangular elements. One such triangular element is also shown in its local coordinates. The fluid pressure acts on one face of the triangle.

## 3. Results

One of the major strengths of this modeling scheme is that it is capable of simulating the nonlinear force-displacement behavior observed in actual *in vivo* experiments performed on biological tissues. To illustrate this point, we have shown in Figure 3 the force-displacement data (dashed lines) obtained when a human fingerpad is indented by a point load [5]. The fingerpad has been modeled as a semicylindrical membrane in plane strain, enclosing an incompressible fluid. The solid line indicates the result of the simulation.

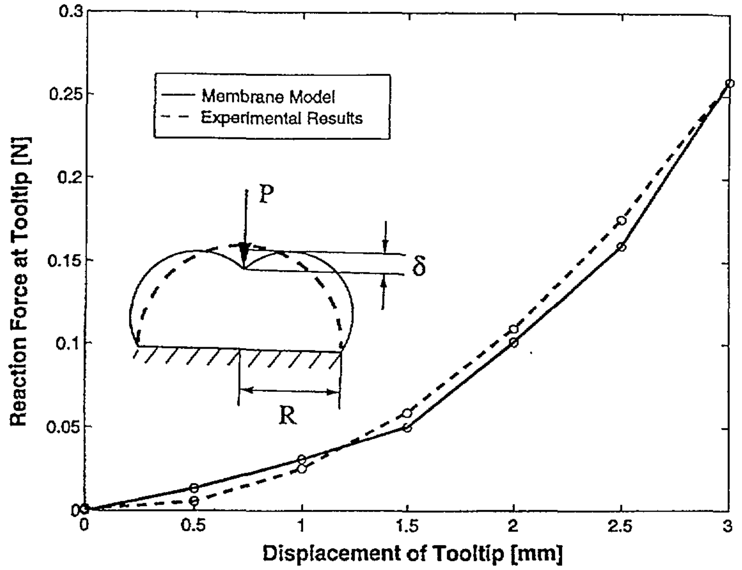


**Figure 3** This figure shows the match in force-displacement relationship between *in vivo* experiments performed on the human fingerpad by a point load and numerical simulations performed on a fluid filled membrane model of the fingerpad. The dashed curve represents the force (N) versus displacement (mm) response of a human fingerpad under steady state conditions when indented by a point load to various depths of indentation. The fingerpad is modeled as a semicylindrical membrane of radius R = 10 mm, filled with an incompressible fluid and subjected to the same depths of indentation , δ, by a pointed tooltip. The solid curve shows the corresponding force-displacement relationship at the tooltip obtained from the model.

So far as the speed of the computations involved, we have performed a benchmark test on a hemispherical domain. The domain was first modeled as a membrane enclosing a fluid. The membrane was discretized using 64 triangular finite elements (total number of nodes = 41). The computational cost was of the order of 2Mflops. We discretized the same domain using 3D tetrahedral finite elements (328 nodes). The computational costs were of the order of 900Mflops, roughly 450 times more. This saving in computational costs arises purely from the reduction in dimensionality due to a surface model replacing a 3D volumetric model.

## 4. Conclusions

In this paper we have presented a new way of modeling soft tissue behavior during surgical simulations. The novelty lies in modeling 3D continua as thin-walled membrane structures filled with fluid. These simple models reduce the dimensionality of the problem from 3D to 2D and are therefore computationally extremely efficient. Moreover, they have the power to predict the non-linear force-displacement response as well as the surface deformation profiles as observed in *in vivo* experimental data on soft tissues. Among other benefits of using this approach are the flexibility to model inhomogeneous and viscoelastic tissue behavior, ability to deal with realistic three-dimensional organs with relatively low computational overheads and a unified approach to haptic and graphical rendering of general deformable media.

**References**

[1] Basdogan, C., Ho and Srinivasan, M.A., "Force Interactions in Laparoscopic Simulations: Haptic Rendering of Soft Tissues." Medicine Meets Virtual Reality, pp 385-391, 1998.
[2] De, S. and Srinivasan, M.A., "A Finite Element Model of the Human Fingerpad Incorporating Viscoelasticity". 1998 (Manuscript in preparation).
[3] Srinivasan, M.A., "Surface Deformation of Primate Fingerpad Under Line Load", J. Biomechanics, Vol. 22, No. 4, pp. 343-349, 1989.
[4] Bathe K.J., "Finite Element Procedures". Prentice Hall, NJ, 1996.
[5] Srinivasan M.A., Gulati, R.J., and De, S., "Force Response of the Human Fingerpad to Dynamic Indentations", 1998 (Manuscript in preparation).