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Physically Based Hybrid Approach in Real Time Surgical Simulation With Force Feedback

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Abstract. This paper describes a novel hybrid-modeling paradigm for the simulation of surgical tool-soft tissue interactions in real time medical simulations using force feedback. A local point collocation-based method of finite spheres is coupled with a global boundary element technique to capture local features of the interaction (e.g., nonlinearities of the soft tissue) without sacrificing global accuracy. The technique is demonstrated using realistic examples.

1.Introduction

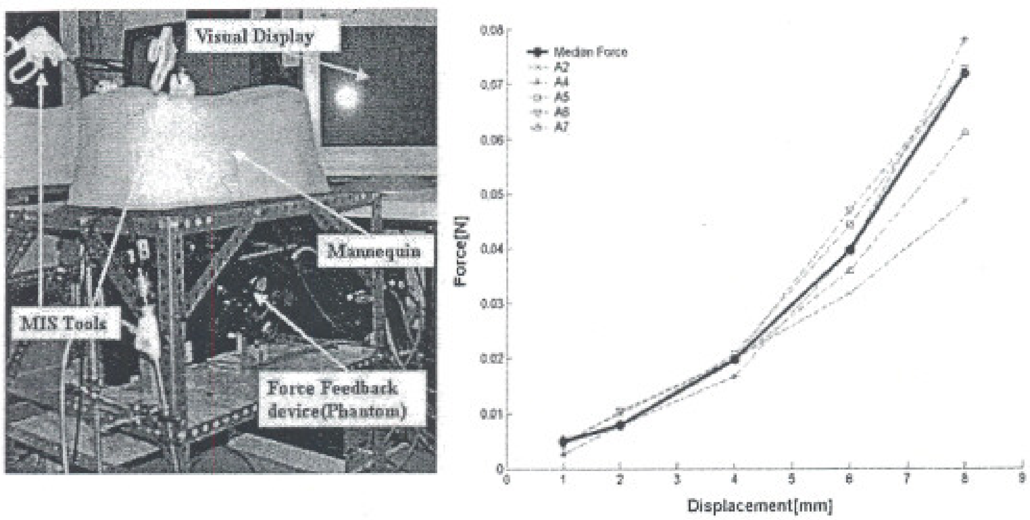
The success of a laparoscopic surgeon depends to a great extent on his or her training, since laparoscopic surgery has to be performed in the presence of several difficulties such as the limited field of view, improper hand-eye coordination and reduced haptic cues [1]. A successful multi-modal surgery simulator will not only provide customized practice environments for medical personnel, but it will also reduce the use of animals and cadavers currently used for such training [2].

An important issue in medical simulation is the modeling of tool-tissue interaction in real time. It is a very challenging task because it involves the simulation of complex mechanics of soft tissues in real time as well as real time display of multi-modal information.

In multimodal surgical simulation the visual loop must be updated at 30 Hz for real time graphics while a much higher update rate of approximately 1kHz is required for stable haptic interactions [3]. This imposes severe restrictions on the complexity of the models that can be rendered in real time. Therefore, developing an efficient computational model without sacrificing fidelity is one of the major issues in surgical simulation. Several techniques can be found in literature for the simulation and display of deformable objects. Among these physically based approaches, like the finite element method, which models the underlying physics of deformable objects have drawn the attention of researchers due to the fact that tool-tissue interactions may be modeled with accuracy and robustness. Bro-Nielson [4] and Cotin *et al.*[5] developed simulation systems to utilize three dimensional solid finite element models (FEM) based on linear elasticity. In these systems, real time performance was achieved by the use of condensation, and precomputation of the stiffness matrix governing tissue behaviors. James and Pai [6, 7] modeled real-time quasi-static deformations using the boundary element method (BEM) and achieved real time performance for haptic interaction using low rank updates of the stiffness matrix, although they did not apply their technique to surgical simulation.

The physically-based techniques found in literature are inherently computationally expensive and real time implementation is not possible without extensive precomputations or gross simplifications. It is well known that soft tissue behavior is highly nonlinear, especially in the vicinity of the surgical tooltip where the deformations are large, and surgical procedures almost always involve cutting. Figure 1(b), for example, shows the nonlinear force-displacement behavior of soft tissues during a typical *in vivo* experiment performed on a pig liver[8]. Precomputed data may be invalid when such situations are encountered.

In section 2 we describe a hybrid modeling approach to achieve real time nonlinear performance in surgical simulation. This technique is quite generic and is applicable to the modeling of localized nonlinearities. The global behavior of the soft tissue is modeled using an integral representation of the equations of linear elastostatics. However, the behavior of the tissue in the vicinity of the surgical tooltip is modeled using the meshfree method of finite spheres. The two techniques are then seamlessly coupled together for real-time interactions. In section 3 we discuss numerical implementation and some system issues.



(a) (b)

Figure 1 (a) The multi-modal VR simulator setup in the TouchLab/MIT (b) Nonlinearity in *in-vivo* force-displacement relationship from the experiments on a pig liver [8]

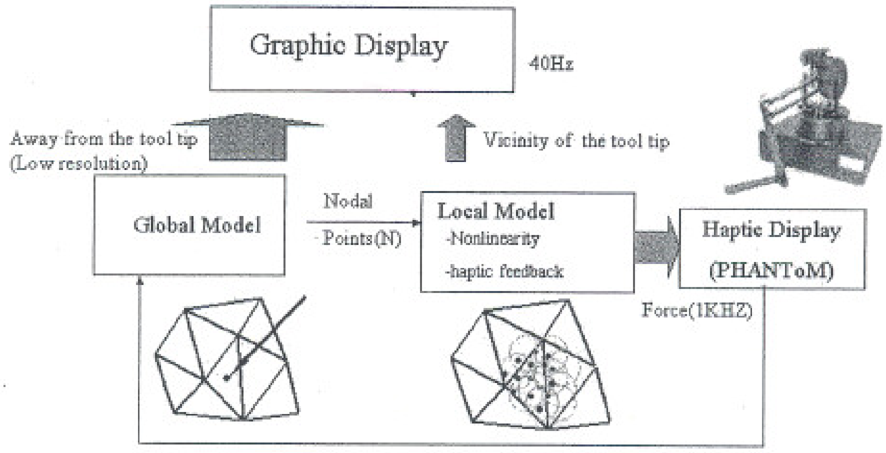


Figure 2 Schematic of our modeling strategy for real time rendering of deformable objects

2. Modeling strategy

We observe that for laparoscopic surgical simulation, accurately computing the deformation field in the vicinity of the tooltip is essential, while this is not so for the rest of the domain where the deformations are relatively small. Based on this observation, we had developed a localized point collocation-based method of finite spheres technique in [9] which only the vicinity of the tooltip was discretized using a novel meshfree technique. The rest of the domain was, however, assumed to be rigid. This technique is quite appealing when the size of the organ is rather large and surgical cutting is not performed. The major advantage of this technique is that it is not limited to linear tissue behavior and real time performance may be obtained without using any precomputations.

However, when the size of the organ is small and surgical cutting needs to be performed, and then the assumption of zero deformation outside the "region of influence" of the tool is not very accurate. We have therefore developed a novel hybrid modeling paradigm which allows us to incorporate local nonlinearities as well as local changes in mesh topology by coupling the localized point collocation-based method of finite spheres with a linear global model. This is a multi-rate simulation approach and the global model is updated at a lower rate than the localized point collocation model (see Figure 2).

For the global model, we use an integral formulation of the equations of linear elastostatics posed on the surface of the computational domain, discretized using the boundary element method (BEM) [9]. The integral equation formulation reduces the dimensionality of the problem by one and only the surface, instead of the volume, is discretized. The same triangular surface mesh used for rendering the geometry of the organ can be used as the boundary element mesh. The main problem associated with the integral equation formulation is that dense global stiffness matrices are generated.

The displacement ($\mathbf{u}$) and traction ($\mathbf{p}$) vectors at a point $\mathbf{x}$ in the domain or on the boundary may be represented by their three Cartesian components

$$\mathbf{u} = \mathbf{u}(\mathbf{x}) = (u_x, u_y, u_z)^T$$
$$\mathbf{p} = \mathbf{p}(\mathbf{x}) = (p_x, p_y, p_z)^T$$

Using piece-wise constant elements (i.e., the displacements and tractions are assumed to be constant over each element), the BEM equations of linear elastostatics with 'E' elements is given by [9]

$$\mathbf{c}\mathbf{u}_i + \sum_{i=1}^{E} \left(\int_{\Delta_i} \mathbf{p}^* d\Gamma \right) \mathbf{u} = \sum_{i=1}^{E} \left(\int_{\Delta_i} \mathbf{u}^* d\Gamma \right) \mathbf{p}$$

where Δ_i is the surface of the ith element, u^* and p^* are the 'fundamental solutions'(see [9]). The coefficient 'c' depends on the smoothness of the boundaries and can be found in literature (for a Lipschitz boundary, $c = 0.5$). Satisfying this equation at the centroids of each of the elements and incorporating the boundary conditions, we obtain the following system of linear algebraic equations:

$$\mathbf{AY} = \mathbf{F}$$

where **Y** is a vector of length 'N' and contains the unknown deformations and tractions at the centroids of the boundary elements. **A** is 'N' by 'N' dense. **F** is the known right hand side vector which includes both the loading and boundary conditions. The solution of this system is symbolically represented as,

$$\mathbf{Y} = \mathbf{A}^{-1}\mathbf{F}$$

Usually, computing the inverse of **A** is an $\mathbf{O(N^3)}$ process therefore it is expedient to precompute it. A structural reanalysis technique using the Sherman-Morrison-Woodbury formula is used (a similar technique has been described in ref. [6, 7]) to incorporate the changes in loading conditions.

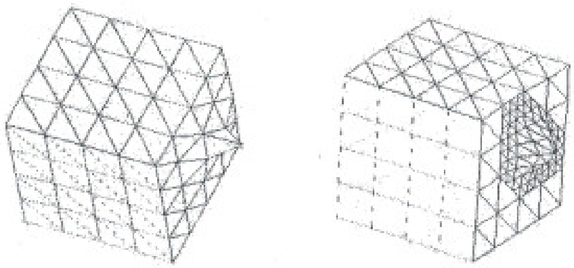
The global model is linear and it is not possible to achieve real time haptic updates. We therefore pre-compute the global boundary element model and couple it with the localized point collocation based method of finite spheres **[10]**. This latter technique uses a sprinkled set of nodal points around the tooltip to perform discretization. An "influence zone" is associated with each nodal point. The approximation u_h of a variable u (e.g. displacement), using 'M' spheres, may be written as

$$u_h(\mathbf{x}) = \sum_{J=1}^{M} h_J(\mathbf{x}) \alpha_J$$

where α_J is the nodal unknown at node J. The nodal shape function $h_J(\mathbf{x})$ at node J is generated using a moving least squares technique (see [10]). We satisfy the equations of motion only at the nodal points to obtain the discretized set of equations

$$\mathbf{KU} = \mathbf{f}$$

where $\mathbf{K}$ is the stiffness matrix and $\mathbf{f}$ is the vector containing nodal loads. A point to note is that the formulation is very general, where $\mathbf{K}$ represents the tangent stiffness matrix if an incremental analysis is performed for nonlinear problems. The number of nodal points in the local model is much smaller than the number of nodal point in the global model and therefore only a small set of equations have to be solved at every time step. The coupling of this local model with the global boundary element model is shown in Figure 3(b).



(a) (b)

Figure 3 The deformation of a cube (a) using a global boundary element model and (b) the hybrid model.

3. Simulator setup

We have developed a multimodal training system for simulating laparoscopic surgical procedures. The hardware components of our simulator include a Windows NT based personal computer (a Pentium III 900 MHz processor) with a high-end graphics accelerator (NVDIA TNT M64) and two Phantom force feedback devices from SensAble Technologies, Inc. Each Phantom is equipped with a laparoscopic surgical instrument (Figure 1(a)). The source code is written based on OpenGL library for graphics rendering and GHOST for haptic rendering with C++ environments. A visual feedback loop of 40 Hz and a haptic feedback loop of 1 kHz are implemented. Figure 4 shows the simulated deformation of a human kidney pulled by a virtual laparoscopic tool.

4. Concluding Remarks

In this paper, we have proposed a hybrid approach to compute the deformations and tooltip reaction forces of tissues and organs in multimodal medical simulations using two complementary physically based modeling schemes. While the global model is linear and can be updated rapidly using a structural reanalysis technique, the local model can handle nonlinearities in tissue behavior quite effectively. We are able to model the complex behavior of biological tissues in real time using this approach. The hybrid technique described in this paper is a significant step towards the development of general simulators, since it provides a clear framework for the display of accurate local deformations and interaction forces without incurring any significant performance penalties.

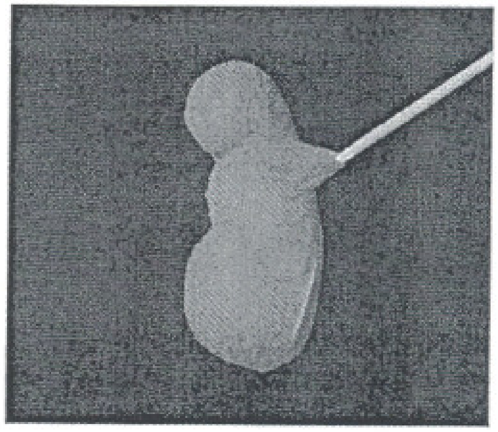


Figure 4 The deformation of a kidney model that had 688 polygons and 2068 degrees of freedom, computed using the proposed hybrid computational scheme.

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