Haptics in Minimally Invasive Surgical Simulation and Training

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Haptics is a valuable tool in minimally invasive surgical simulation and training. This article discusses important aspects of haptics in MISST, such as haptic rendering and haptic recording and playback.

Minimally invasive surgery has revolutionized many surgical procedures over the last few decades. MIS is performed using a small video camera, a video display, and a few customized surgical tools. In procedures such as gall bladder removal (laparoscopic cholesystectomy), surgeons insert a camera and long slender tools into the abdomen through small skin incisions to explore the internal cavity and manipulate organs from outside the body as they view their actions on a video display.

Because the development of minimally invasive techniques has reduced the sense of touch compared to open surgery, surgeons must rely more on the feeling of net forces resulting from tool-tissue interactions and need more training to successfully operate on patients. Although tissue color and texture convey important anatomical information visually, touch is still critical in identifying otherwise obscure tissue planes, blood vessels, and abnormal tissues, and gauging optimal forces to be applied for tissue manipulation. Much of the art of MIS and training for a particular procedure depend on the education and refinement of the trainee's haptic sensorimotor system.

The benefits of using haptic devices in medical training through simulation[1–7] have already been recognized by several research groups and many of the companies working in this area (Immersion Medical, Surgical Science, Mentice, and Reachin Technologies, for example). The rapid increase in the number of papers on haptics and surgical simulation published in international conference proceedings in the last five years indicates haptics' growing importance. In spite of this growth, very little research objectively compares interface devices, algorithms, or even surgical training with and without haptics. We therefore describe a framework that includes most of the important aspects of haptics in minimally invasive surgical simulation and training (MISST), with examples drawn primarily from our work at the MIT Touch Lab. Although we refer to many of the important papers by other groups, because of space limitations the literature cited is by no means exhaustive.

Over the last seven years, we've explored the role of haptics in MISST in the following areas:

- *Haptic interfaces*—We've integrated commercially available haptic devices into a training system designed to simulate minimally invasive procedures.
- *Haptic rendering*—We've developed computational models of surgical instruments to detect collisions with geometric models of organs to compute and reflect interaction forces to a user and organ-force models that respond to user interactions in real time during simulation sessions.
- *Haptic recording*—We've used haptic devices to measure material properties of organs during a recording session.
- *Haptic playback*—We've developed user interaction techniques based on force feedback to guide a user during a training session.

The following sections discuss these areas in more detail, with particular emphasis on haptic rendering techniques.

Integration of force-reflecting robotic devices

Realistic simulation of a full surgical procedure is not a feasible goal with current technology. Work in this area needs to initially focus on simulating part-task procedures. As a first step, replacing mechanical laparoscopic training boxes, currently used in hospitals for part-task training, with computer-based systems can significantly improve training transfer. Our group at the MIT Touch Lab has developed a VR-based training system, shown in Figure 1, that looks similar to a mechanical training box from the outside, but can be customized to simulate various part-task procedures in virtual environments.[1] The

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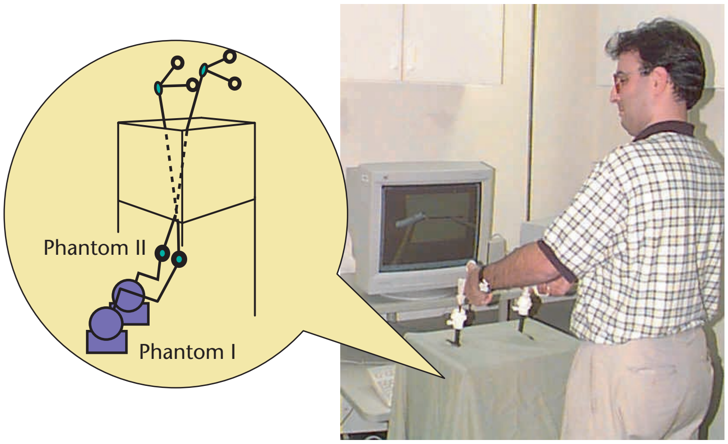
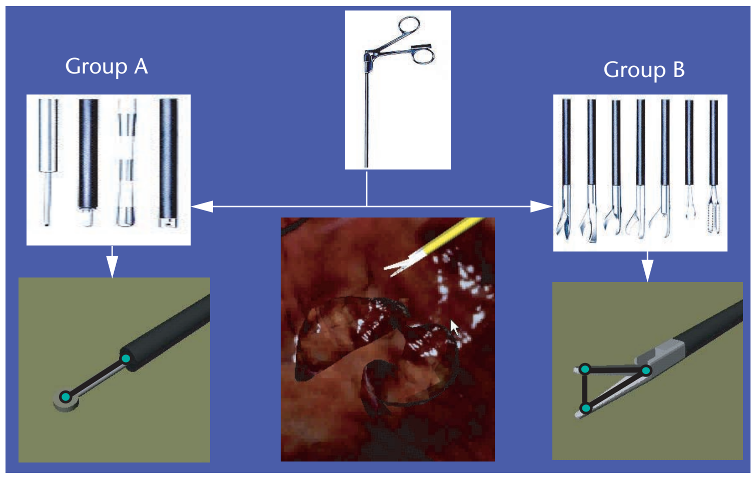


1 Our laparoscopic training system. As the user manipulates the actual instruments and interacts with virtual organs using the simulated instruments, the computer screen displays the associated deformations of the virtual organs, and a pair of commercially available haptic devices feed the reaction forces to the user. Sometimes, an additional actuator is attached to the handle of the laparoscopic instruments to supply force feedback to the user for simulating tissue grasping.

system's hardware components include a personal computer with a high-end graphics card and a box with instrumented laparoscopic tools that are interfaced with force-feedback devices hidden in the box and attached to the distal ends of the laparoscopic tools.



2 Grouping of surgical instruments for simulating tool–tissue interactions. Group A includes long, thin, straight probes. Group B includes tools for pulling, clamping, and cutting soft tissue.

Haptic rendering for simulation of tool–tissue interactions

Haptic rendering algorithms detect collisions between surgical instruments and virtual organs and render organ-force responses to users through haptic interface devices[8] (see also the article by Salisbury et al. in this issue). For the purpose of haptic rendering, we've conceptually divided minimally invasive surgical tools into two generic groups based on their functions (see Figure 2):

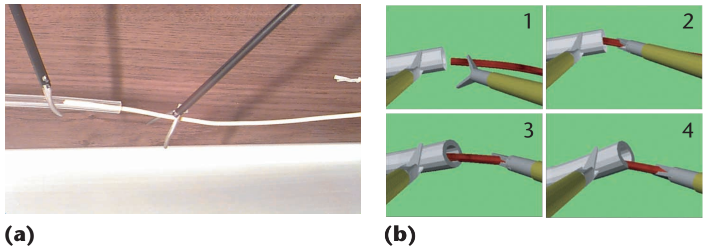
- long, thin, straight probes for palpating or puncturing the tissue and for injection (puncture and injection needles and palpation probes, for example)
- articulated tools for pulling, clamping, gripping, and cutting soft tissues (such as biopsy and punch forceps, hook scissors, and grasping forceps)

To demonstrate the concept, we've generated a 3D computer model of an instrument from each group (a probe from the first group and a forceps from the second) and displayed their behavior in a virtual environment. During real-time simulations, we display the 3D surface models of the probe and forceps to provide the user with realistic visual cues. For the purposes of haptic rendering of tool–tissue interactions, we've developed ray-based rendering, in which the probe and forceps are modeled as connected line segments.[1]

Modeling haptic interactions between a probe and objects using this line-object collision detection and response has several advantages over existing point-based techniques, in which only the tip point of a haptic device is considered for touch interactions[8]:

- Users feel torques if a proper haptic device is used. For example, the user can feel the coupling moments generated by the contact forces at the instrument tip and forces at the trocar pivot point.
- Users can detect side collisions between the simulated tool and 3D models of organs.
- Users can feel multiple layers of tissue if the ray representing the simulated surgical probe is virtually extended to detect collisions with an organ's internal layers. This is especially useful because soft tissues are typically layered, each layer has different material properties, and the forces/torques reflected to the user depend on the laparoscopic tool's orientation.
- Users can touch and feel multiple objects simultaneously. Because laparoscopic instruments are typically long slender structures and interact with multiple objects (organs, blood vessels, surrounding tissue, and so on) during a MIS, ray-based rendering provides a more natural way than a purely point-based rendering of tool-tissue interactions.

To simulate haptic interactions between surgical material held by a laparoscopic tool (for example, a catheter, needle, or suture) and a deformable body (such as an organ or vessel), we used a combination of point- and ray-based haptic rendering methods. In our simulation of a catheter insertion task[1]



3 Training for laparoscopic catheter insertion into a common bile duct using (a) a standard mechanical training box and (b) a VR-based simulation system. In the VR-based system, the user feels the interaction forces between the laparoscopic forceps and the catheter via ray-based haptic rendering, the laparoscopic forceps and the bile duct via ray-based haptic rendering, and the catheter and the bile duct via point-based rendering.

(see Figure 3), we modeled the surgical tools using line segments and the catheter using a set of points uniformly distributed along the catheter's center line and connected with springs and dampers. Using our point-based haptic rendering method, we detected the collisions between the flexible catheter and the inner surface of a flexible vessel to compute interaction forces.

The concept of distributed particles can be used in haptic rendering of organ–organ interactions (whereas a single point is insufficient for simulating organ–organ interactions, a group of points, distributed around the contact region, can be used) and other minimally invasive procedures, such as bronchoscopy and colonoscopy, involving the insertion of a flexible material into a tubular body .

Deformable objects

One of the most important components of computer-based surgical simulation and training systems is the development of realistic organ-force models. A good organ-force model must

- reflect stable forces to a user,
- display smooth deformations,
- handle various boundary conditions and constraints, and
- show physics-based realistic behavior in real time.

Although the computer graphics community has developed sophisticated models for real-time simulation of deformable objects,[9] integrating tissue properties into these models has been difficult. Developing real-time and realistic organ-force models is challenging because of viscoelasticity, anisotropy, nonlinearity, rate, and time dependence in material properties of organs. In addition, soft organ tissues are layered and nonhomogeneous. Tool–tissue interactions generate dynamical effects and cause nonlinear contact interactions of one organ with the others, which are quite difficult to simulate in real time. Furthermore, simulating surgical operations such as cutting and coagulation requires frequently updating the organ geometric database and can cause force singularities in the physics-based model at the boundaries.

There are currently two main approaches for developing force-reflecting organ models: particle-based methods[2,4,7] and finite-element methods (FEM).[1,3,5,6,9] In particle-based models, an organ's nodes are connected to each other with springs and dampers. Each node (or *particle*) is represented by its own position, velocity, and acceleration and moves under the influence of forces applied by the surgical instrument. In finite-element modeling, the geometric model of an organ is divided into surface or volumetric elements, properties of each element are formulated, and the elements are assembled together to compute the deformation states of the organ for the forces applied by the surgical instruments.

Particle systems have been used extensively in computer graphics to simulate the dynamic behavior of clothes, fluid flow, and deformable objects. This technique is easy to implement because the developer doesn't necessarily need to construct the equations of motion explicitly. However, the integration of realistic tissue properties into particle models isn't a trivial task. The construction of an optimal network of springs in 3D is a complicated process and particle systems can become oscillatory or even unstable under certain conditions.

During the last few years, our research has focused more on the development of real-time finite-element models[1] and a new mesh-free modeling concept[10] for simulating the force-reflecting behavior of organs. In finite-element models, the behavior of soft biological tissues is governed by differential equations of continuum mechanics, and techniques based on such equations, though computationally demanding, seem more promising than particle-based methods in modeling tissue characteristics.

One of the major advantages of continuum-based approaches is that only a few material parameters are required to describe the response of a physical system, which can be obtained by performing in vivo experiments (see our discussion on haptic recording of tissue properties). Moreover, modeling the response of multilayered tissue exhibiting complex nonlinear, viscoelastic, and anisotropic behavior can be implemented in a more unified framework using FEM-based tissue models.

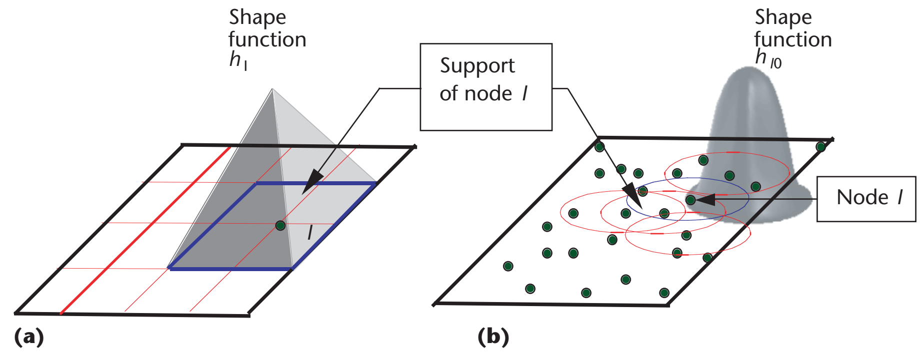
Mesh-based finite-element modeling techniques

We've developed a mesh-based finite-element model to simulate real-time visual and haptic rendering of soft tissues.[1] Simulating the real-time behavior of a deformable 3D organ using finite-element models becomes more difficult as the total number of nodes or degrees of freedom (DOF) increases. Although finite-element models have been developed for medical applications, less attention has been paid to displaying time-dependent deformations of large-size models in real time.

We've developed computationally efficient techniques for real-time simulation of dynamically deformable (that is, time-dependent deformations) 3D organs modeled by finite-element equations. We've implemented a modal analysis approach such that only the most significant vibration modes of an organ model were selected to compute the dynamics of deformations and the interaction forces. The dynamic equilibrium equations for a deformable organ using finite-element modeling can be written as

$$M\ddot{U} + B\dot{U} + KU = F \qquad (1)$$

where M, B, and K represent the mass, damping, and stiffness matrices, and F and U are force and displacement vectors. After deriving the equations of motion for a deformable organ, the solution is typically obtained using numerical techniques. However, the solution of these equations becomes computationally intensive as the number of elements is increased. We can make modeling simplifications by assuming that high-frequency deformation modes contribute little to the overall computation of deformations and forces. For this purpose, the dynamical equations are transformed into a more effective form (in a process known as *modal analysis*) and real-time solutions are obtained with reasonable accuracy via modal reduction (see the "Modal Analysis of Soft Tissue Dynamics" sidebar for details). The errors introduced by the modal reduction are insignificant compare to the computational advantage gained through the approximation.[1]



4 Discretization of a domain in R^2 by (a) the finite-element method and (b) the method of finite spheres. A finite-element model discretizes the domain by quadrilateral elements with a node at each vertex point. The finite-element shape function h_I is shown at node I. MFS discretizes the domain using a set of nodes only. Corresponding to each node I, there is a sphere (a disk in R^2) centered at the node. The sphere supports a set of shape functions (h_{I0}, for example) corresponding to node I.

Mesh-free techniques

A potential solution to some of the problems faced by the finite-element techniques (for example, remeshing after a surgical cut) is to use a numerical technique that does not use a mesh. The *method of finite spheres* is one such mesh-free computational technique.[10] MFS discretizes the computational domain using a scattered set of points, or nodes. The displacement field is approximated using functions that are nonzero over small spherical neighborhoods of the nodes, as Figure 4b shows. As in the finite-element scheme, MFS uses a Galerkin formulation to generate the discretized versions of the partial differential equations governing the deformable medium's behavior. In this respect, MFS is a generalization of the finite-element scheme. Figure 4 compares the two techniques. The "Meshless Method of Finite Spheres"

Modal Analysis of Soft Tissue Dynamics

To implement the modal analysis, we define the following transformation:

$$U(t)_{nx1} = \Phi_{nxn} X(t)_{nx1} \qquad (1)$$

where Φ is the modal matrix and U and X represent the original and modal coordinates. The modal matrix is obtained by solving the eigen problem for free undamped equilibrium equations:

$$K\phi = \omega^2 M\phi \qquad (2)$$

where ω and ϕ represent the eigenvalues (the vibration frequencies) and eigenvectors (the mode shapes) of the matrix ($M^{-1}K$). The modal matrix, Φ, is constructed by sorting the frequencies in ascending order and then placing the corresponding eigenvectors into the modal matrix in column-wise format ($0 \le \omega_1 \le \omega_2 \le \omega_3 \ldots \le \omega_n, \Phi = [\phi_1, \phi_2, \phi_3, \ldots \phi_n]$).

Finally, a set of decoupled differential equations (the modal system) is obtained using the modal matrix and the transformation defined by Equation 1:

$$\ddot{X}_i + \alpha_i \dot{X}_i + \omega_i^2 X = f_i \qquad i = 1,\ldots,n \qquad (3)$$

where n is the system's degrees of freedom (DOF) and $\alpha_i = 2\omega_i\zeta_i$, and $f_i = \phi_i^T F$ are the modal damping and force respectively. (Note that ζ is known as the damping ratio or modal damping factor.)

To implement the modal reduction, we first reduce the modal matrix by picking only a few significant modes (that is, the first r columns of the modal matrix). Our differential system for modal coordinates is reduced to r number of equations, which are then solved using a numerical integration technique:

$$\ddot{X}_i^R + \alpha \dot{X}_i^R + \omega_i^2 X^R = f_i^R \qquad i = 1,\ldots,r \qquad (4)$$

where the superscript R represents the reduced system. We then transfer the modal coordinates back to the original coordinates using the following transformation and display the nodal deformations:

$$U(t)_{nx1} = \Phi_{nxr}^R X^R(t)_{rx1} \qquad (5)$$

Similarly, forces are transferred back to the original coordinates and reflected back to the user through force-feedback devices in real time:

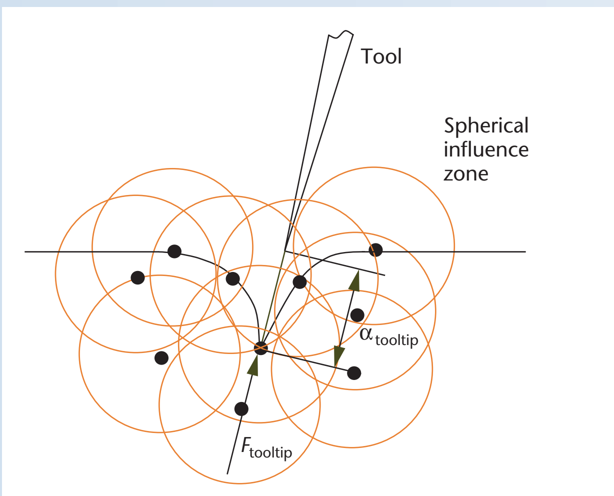
$$F(t)_{nx1} = \Phi_{nxr}^R f^R(t)_{rx1} \qquad (6)$$

Meshless Method of Finite Spheres

For simulating the force-reflecting behavior of organs, we've developed a specialized version of the MFS. In this technique, matter is represented as a collection of nodes. The nodes possess finite (spherical) *influence zones*, as illustrated in Figure A. The interlocking of these influence zones allows the nodes to move in a coordinated fashion under elastic force fields (just as magnetic particles would move under the influence of each others' magnetic fields). An important aspect of the mesh-free idea is the flexibility of defining a spherical influence zone for each node on an arbitrary domain in 3D.

In this technique the approximation u_h of the component u of the displacement field using N nodes can be written as

$$u_h(\mathbf{x}) = \sum_{J=1}^{N} h_J(\mathbf{x})\alpha_J \tag{7}$$



A Schematic of the MFS technique. α_{tooltip} and F_{tooltip} are the prescribed displacement and computed reaction force at the tool tip, respectively.

where α_J is the nodal unknown at particle J. The nodal shape function $h_J(\mathbf{x})$ at particle J is generated using a moving least squares method:

$$h_J(\mathbf{x}) = W_J(\mathbf{x})P(\mathbf{x})^T C^{-1}(\mathbf{x})P(\mathbf{x}_j)$$

where

$$\mathbf{C}(\mathbf{x}) = \sum_{I=1}^{N} W_I(\mathbf{x})P(\mathbf{x}_I)P(\mathbf{x}_I)^T, J = 1,\ldots,N \tag{8}$$

The vector P($\mathbf{x}$) contains polynomials ensuring consistency up to a desired order. (If we choose P(x) = $\{1, x, y, z\}^T$ for example, a first-order accurate scheme is ensured in 3D, similar to bilinear finite elements.) W_J is a radial weighting function at node J and is nonzero only on the sphere at node J (for example, quartic spline).

The equations of linear elasticity as well as boundary conditions are then discretized using Equation 7 to give the following compact set of equations:

$$\mathbf{KU} = \mathbf{F} \tag{9}$$

where **K** is the system stiffness matrix (not symmetric, but banded), **U** is the vector of displacements at the nodes, and **F** is the RHS vector. These equations are solved to obtain the body's displacement field. The stiffness matrix in Equation 9 can then be partitioned as

$$\mathbf{K} = \begin{bmatrix} \mathbf{K}_{aa} & \mathbf{K}_{ab} \\ \mathbf{K}_{ba} & \mathbf{K}_{bb} \end{bmatrix} \tag{10}$$

corresponding to a partitioning of the vector of nodal parameters as $\mathbf{U} = [\mathbf{U}_{\text{tooltip}} \ \mathbf{U}_b]^T$ where $\mathbf{U}_b$ is the vector of nodal unknowns that can be obtained as $\mathbf{U}_b = -\mathbf{K}_{bb}^{-1}\mathbf{K}_{ba}\mathbf{U}_{\text{tooltip}}$. The reaction force that will be reflected to a user via a haptic device is then obtained by $\mathbf{F}_{\text{tooltip}} = \mathbf{K}_{aa}\mathbf{U}_{\text{tooltip}} + \mathbf{K}_{ab}\mathbf{U}_b$. Our approach for the automated distribution of nodes improves the computational complexity of the mesh-free method.

sidebar describes our implementation of the MFS for simulating the force-reflecting behavior of organs.

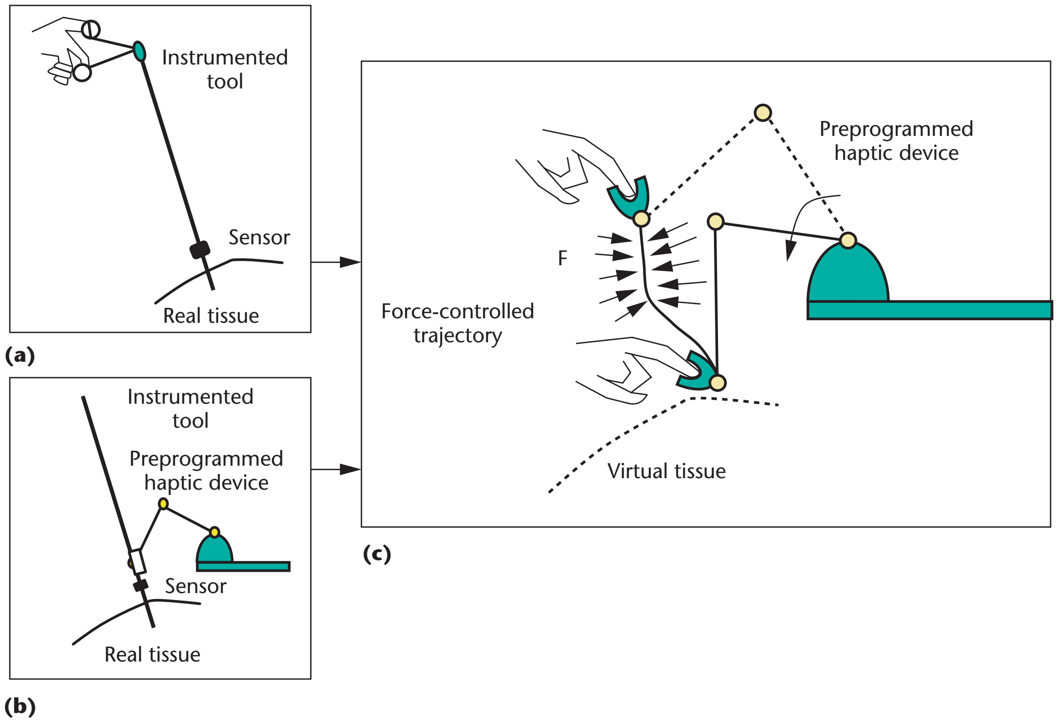
Haptic recording of tissue properties

The concept of haptic recording involves the use of a haptic device in recording surface (such as texture), material (such as softness), and geometrical (such as shape) properties of objects that can be played back to a user through the same device (see Figure 5). We divide haptic recording methods into two groups:

- *free-form*, in which the user holds a probe equipped with sensors; and
- *controlled*, in which a programmed robotic arm uses a probe.

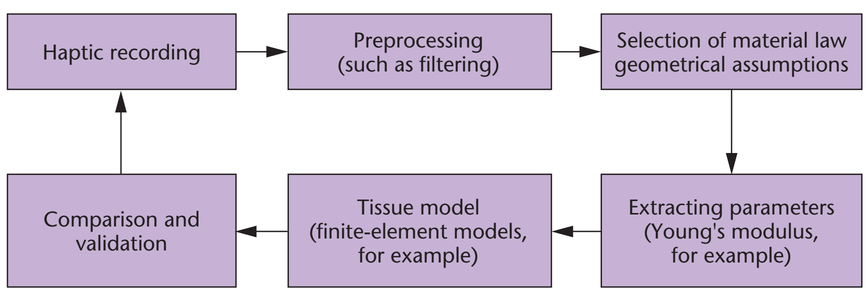
In MISST, accurate rendering of forces is highly dependent on the material properties of the tissue being simulated. Recording of tissue properties for characterization, especially of the living state in a body, is an important issue for MISST. For example, the linear homogeneous tissue models we've discussed require two independent parameters to compute interaction forces on the virtual instruments: Young's modulus and Poisson's ratio. Although material properties of various biological tissues have been measured and recorded in the past, the tissue samples were typically taken post mortem and tested under devices and procedures similar to those used for engineering materials. In addition to testing difficulties, the collected data would be unsuitable for MISST because the mechanical properties of soft tissues change rapidly after death.

Recently, considerable research[11,12] has been done on recording mechanical properties of tissues in a living state (in vivo) and within a body (in situ). Our group at MIT (in collaboration with Harvard Medical School) has also



5 Haptic recording and playback for MISST: (a) A human operator manipulates a laparoscopic tool equipped with sensors, making indentations for measuring and recording interaction forces; (b) a haptic device interfaced with a probe and sensors can be programmed to make controlled indentations for measuring and recording interaction forces; and (c) haptic playback involves the display of programmed forces to a user for guidance and control during training.

conducted controlled experiments on live animals to record the abdominal tissue properties using an instrumented haptic device.[13] A haptic device interfaced with a probe and force sensor was programmed to generate indenting motion, like ramp-and-hold or frequency sweeping signals, to record the force and displacement response of soft tissues for various indentations. From the force and displacement data recorded at various locations on the tissue surface, Young's modulus and Poisson's ratio can be obtained and integrated into tissue models for simulation. Figure 6 diagrams this process.



6 Haptic recording of organ properties: tissue properties are extracted from force-displacement data recorded via a haptic device equipped with force and position sensors. The properties are inserted into the physics-based tissue model for haptic rendering of the organ response in virtual environments.

Playing back haptic stimuli for training

Although researchers generally agree on the importance of training and performance assessment of MIS, they've yet to reach consensus on how best to measure and quantify performance. Most available assessment methods are subjective and the advantages that computer simulation offers could be easily exploited to define and standardize objective performance measures. Our goal in this section is not to cover this subject exhaustively, but to discuss briefly the role of haptics in training.

The benefits of using force feedback in surgical training through simulation have already been demonstrated in a suturing simulation using a performance metric that includes the force profile of users as one of the parameters.[7] Moreover, the force and torque profiles of human users operating on a porcine model have been recorded via sensors to quantify their skill level using Markov models.[11] The results of both studies show that the force and torque profiles of expert users are significantly different from those of naive users.

Playing back prerecorded haptic stimuli to a user through a force-feedback device could also be useful in medical training. A haptic device can be programmed to provide controlled forces to users along a predefined trajectory for training their motor-control skills. As trainees move out of the trajectory, force feedback can return them to it. We've implemented and tested this concept in our training system designed for simulating epidural injection.[14]

When guiding a needle into epidural space, a physician relies heavily on haptic cues. The appreciation of forces at each layer is important for the proper guidance of the needle. We've experimented with two modes of haptic guidance in our training setup. In the first, the simulator displays a virtual guiding needle on the screen that moves along the same path and with the same speed as an expert in a prerecorded trial. If the user's needle position exactly matches that of the guiding virtual needle, the user feels the same forces that the expert felt. In the event of a mismatch, the virtual instructor applies a force to pull the trainee back to the prerecorded trajectory.

In the second mode, or *tunnel guidance*, we disregard the time-dependency of the recorded data such that users perform the task at their own speed. The needle's movement is limited to the prerecorded trajectory, allowing users to concentrate solely on the forces encountered at each layer along the needle's insertion path.

Challenges

A critical issue in the design of simulators for medical training is the relationship between technology and training effectiveness. The focus of this article, however, is narrower and is concerned with technological issues of integrating haptic rendering into medical simulations. At the current state of simulator technology, fully realistic simulation isn't possible, and the main challenge is achieving adequate realism in part-tasks so that the trainee is better prepared to treat patients. Even with the limited objective of part-task simulations, integrating haptics poses technical challenges on all fronts: haptic interface hardware design, tissue and organ model development, tool–tissue interactions, real-time graphical and haptic rendering, and haptic recording and playback.

Haptic devices

One of the constant challenges in integrating haptics into virtual environments is the need for haptic interface devices with the requisite DOF, range, resolution, and frequency bandwidth, both in terms of forces and displacements. Medical simulations involving soft tissues are generally bimanual, and typically require six to seven DOF for each hand. The forces and motions involved are small but require the resolutions to be fine and for the device to be mechanically transparent so its characteristics minimally affect the intended force feedback. With current technology, it's hard to imagine a single universal device for all medical procedures, and it's likely that groups of similar medical procedures can be simulated with specific devices.

Organ models

The fidelity of force-reflecting organ models is an important issue and its relationship to training effectiveness is largely unknown. Although the Visible Human Project makes available geometric models of organs, empirical investigations of in vivo tissue mechanics are critical to measure material parameters needed for realistic simulations. In addition, current organ-force models are linear, homogeneous, and isotropic. Such models are inadequate to represent the inherent nonlinearities, anisotropy, and rate dependence of soft tissues.

Complex tissue models, on the other hand, are computationally expensive. Moreover, it's difficult to obtain the various parameters arising in such models by performing in vivo experiments. The accuracy requirements of organ models must also be investigated from the perspective of human haptic perception. For example, the results of our preliminary study show that the subjects cannot differentiate the forces generated by linear and nonlinear elastic tissue models in some surgical tasks. In addition, our earlier studies on human haptic perception show that visual cues significantly affect the haptic perception of the softness. This finding suggests that, for example, organ A will be perceived to be softer than organ B if its graphical display deforms more than organ B, even though they both reflect the same forces in magnitude to the user for a unit displacement of the indenting probe held in the hand.

Tool–tissue interactions

Most of the work in surgical simulation focuses on the development of force-reflecting deformable organ models. However, many other complex interactions between a surgical instrument and an organ occur during a MIS. For example, realistic visual and haptic rendering of tissue cutting, bleeding, and coagulation are important components of a surgical simulator. The physics of the dynamic phenomena underlying surgical cutting or tearing of soft tissues is extraordinarily complex. Accurately modeling the interactions between the sharp edge of a tool and the soft tissue has been proven to be rather difficult. Although researchers have developed graphics-based techniques to visually render virtual cutting, the development of a meaningful force model that is tightly coupled with experimental studies[15] remains an active research issue.

In this regard, mesh-free methods offer significant promise. The remeshing issue, which plagues any cutting operation using finite-element-based techniques vanishes completely because no mesh is used. The cut opens because of pretension in the tissue, which is incorporated into the constitutive model. As the cut opens, the computational nodes are redistributed only in the vicinity of the cut.

Real-time rendering

The number of computations required for real-time visual and haptic rendering of organ-force models is another bottleneck in our simulations. Because the update rates for sending the force commands to a haptic interface need to be in the order of several hundred Hz for organ-force models, we use fast numerical techniques and make modeling simplifications to reduce the number of computations. Investigators in the field, including our group, have suggested the following optimizations.

We can compute the deformations and forces in the local neighborhood of the tool–tissue interactions only. Similarly, a simplified intermediate model or buffering can be considered to handle situations where haptic data cannot be delivered at required rates.

Organ-force models can exploit single-point interactions. For example, if a force is applied to a single node of a static finite-element model, the nodal deformations can be easily computed from the index of the application point using $U = K_i^{-1}F_i$, where i is the ith column of the K_i^{-1} matrix and ith entry of force vector.

If the procedure doesn't require a high-fidelity model such as finite elements for tissue simulation, we recommend using a computationally more efficient but low-fidelity model such as a network of particles connected to each other via springs and dampers. This hybrid approach reduces the number of computations. In our catheter insertion simulation,[1] for example, we achieve real-time update rates by modeling the catheter using particles and the bile duct (into which the catheter is inserted) using a finite-element model.

The stiffness matrix in both mesh-based and meshless methods can be condensed to remove unwanted DOF. For example, because rotational DOF in finite-element models are required for the continuity of solution but aren't necessary for the display of deformations and forces, they can be removed from the stiffness matrix through condensation.

Constructing a multilayered computational architecture is highly useful. For example, a real-time dynamic analysis of force-reflecting deformable objects using finite-element techniques is quite challenging with the available computational power. Adding a computational layer—in which we can extrapolate new forces based on their previous values and rate of change—between the physics-based model and the display modules can meet this challenge. Using this approach, forces can be computed at 200 Hz using a finite-element technique, extrapolated between the computation cycles, and displayed to the user at rates close to 1 kHz.

Adaptive meshing techniques, in which the mesh becomes denser in the region of interest, and adaptive time steps, in which one obtains the numerical solution of the differential equations faster, have been suggested for the simulation of organ-force models. However, methods based on adaptive time steps may lead to vibrations and instabilities in the haptic devices because of unequal time steps. Similarly, methods based on adaptive meshing can return unrealistic force values and directions if the mesh surface is subdivided adaptively without paying attention to the underlying tissue model.

Haptic recording and playback

Also important are the benefits of using the same haptic device to simultaneously record and display haptic stimuli for MISST. Although considerable interest in haptic display of compliant objects for MISST exists, the haptic recording and playback concepts require further exploration. For example, force sensors are typically attached to haptic devices for measuring and recording material properties of soft objects, which can be achieved without additional sensors. Robotics researchers have succesfully obtained the compliant properties of objects to some extent by using the kinematic and dynamic properties of a robotic arm and the tip position of its end-effector only. This suggests that commercially available haptic devices today can be used for dual purposes without any major modifications in the design. In addition, the concept of force fields developed by robotics researchers for guiding robot arms and navigating mobile robots in the presence of obstacles can be used in haptic playback. A user's movements can be guided using force fields or grids for medical training. The use of force grids/fields, for example, can bring flexibility to the movements of a user who is asked to follow a prerecorded trajectory during a training session in MISST.

Conclusion

It's generally accepted that just as flight simulators help train pilots, VR-based medical simulators have the potential to more effectively train medical personnel than the current practice of learning largely on the patient. Because MIS procedures involve touching, feeling, and manipulating the organs through instruments, integrating haptics into MISST seems essential. Although there are technical challenges in creating realistic simulations, recent improvements in electromechanical hardware and computational capabilities have created exciting opportunities. As we've described, the field has seen rapid progress in applying haptic technologies to medical simulations for training. Within a few years, many research laboratories and companies have developed simulators for a number of medical procedures. Successful integration of haptics into medical simulators will also help in its integration into telediagnosis or telesurgery systems in which a master robot allows the surgeon to feel the forces resulting from interactions between the slave robot and a patient. Although the current cost of such systems is quite high, it should decrease over time.

Development of haptics-integrated systems that are actually used by the medical community requires close collaboration between medical personnel and technologists. We hope that use of medical simulators for training will be widely accepted by the medical educators in the near future and cause a quantum leap in the quality of patient care. ■

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References

1. C. Basdogan, C. Ho, and M.A. Srinivasan, "Virtual Environments for Medical Training: Graphical and Haptic Simulation of Common Bile Duct Exploration," *IEEE-ASME Trans. Mechatronics*, vol. 6, no. 3, 2001, pp. 267-285.
2. H.K. Çakmak and U. Kühnapfel, "Animation and Simulation Techniques for VR-Training Systems in Endoscopic Surgery," *Proc. Eurographics Workshop Animation and Simulation (EGCAS 2000)*, 2000, pp. 173-185.

3. S. Cotin, H. Delingette, and N. Ayache, "Real-Time Elastic Deformations of Soft Tissues for Surgery Simulation," *IEEE Trans. Visualization and Computer Graphics*, vol. 5, no. 1, 1999, pp. 62-73.
4. J. Brown et al., "Algorithmic Tools for Real-Time Microsurgery Simulation," *Medical Image Analysis*, vol. 6, no. 3, 2002, pp. 289-300.
5. X. Wu et al., "Adaptive Nonlinear Finite Elements for Deformable Body Simulation Using Dynamic Progressive Meshes," *Proc. Eurographics*, *Computer Graphics Forum,* A. Chalmers and T.-M. Rhyne, eds., vol. 20, no. 3, Sept. 2001, pp. 349-358.
6. G. Székely et al., "Modeling of Soft Tissue Deformation for Laparoscopic Surgery Simulation," *Proc. 1st Int'l Conf. Medical Image Computing and Computer-Assisted Intervention* (MICCAI), LNCS 1496, Springer-Verlag, 1998, pp. 550-561.
7. R. O'Toole et al., "Measuring and Developing Suturing Technique with a Virtual Reality Surgical Simulator," *J. Am. College of Surgery*, vol. 189, 1999, pp. 114-128.
8. C. Basdogan and M.A. Srinivasan, "Haptic Rendering In Virtual Environments," *Handbook of Virtual Environments*, K. Stanney, ed., 2002, pp. 117-134; available at http://www.ku.edu.tr/~cbasdogan.
9. D.L. James and D.K. Pai, "Multiresolution Green's Function Methods for Interactive Simulation of Large-Scale Elastostatic Objects," *ACM Trans. Graphics*, vol. 22, no. 1, Jan. 2003, pp. 47-82.
10. S. De, J. Kim, and M.A. Srinivasan, "A Meshless Numerical Technique for Physically Based Real Time Medical Simulations," *Proc. Medicine Meets Virtual Reality* (MMVR 01), vol. 9, ISO Press, 2001.
11. J. Rosen et al., "Markov Modeling of Minimally Invasive Surgery Based on Tool/Tissue Interaction and Force/Torque Signatures for Evaluating Surgical Skills," *IEEE Trans. Biomedical Eng.*, vol. 48, 2001, pp. 579-591.
12. M.P. Ottensmeyer, E. Ben-Ur, and K.J. Salisbury, "Input and Output for Surgical Simulation: Devices to Measure Tissue Properties In Vivo and a Haptic Interface for Laparoscopy Simulators," *Proc. Medicine Meets Virtual Reality* (MMVR 00), ISO Press, 2000, pp. 236-242.
13. B. Tay et al., "Measurement of In Vivo Force Response of Intra-abdominal Soft Tissues for Surgical Simulation," *Proc. Medicine Meets Virtual Reality* (MMVR02), vol. 10, IOS Press, 2002.
14. T. Dang, T.M. Annaswamy, and M.A. Srinivasan, "Development and Evaluation of an Epidural Injection Simulator with Force Feedback for Medical Training," *Proc. Medicine Meets Virtual Reality* (MMVR 01), IOS Press, 2001, pp. 97-102.
15. S. Greenish et al., "Measurement, Analysis and Display of Haptic Signals During Surgical Cutting," *Presence*, vol. 11, no. 6, 2002, pp. 626-651.

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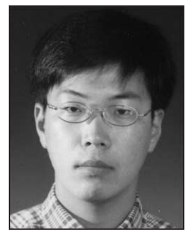

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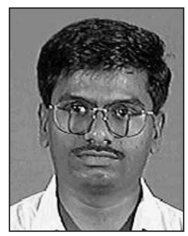

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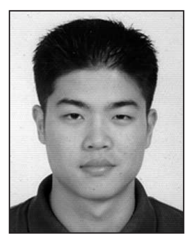

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