

# Force Interactions in Laparoscopic Simulations: Haptic Rendering of Soft Tissues

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**Abstract.** Research in the area of computer assisted surgery and surgical simulation has mainly focused on developing 3D geometrical models of the human body from 2D medical images, visualization of internal structures for educational and preoperative surgical planning purposes, and graphical display of soft tissue behavior in real time. Conveying to the surgeon the touch and force sensations with the use of haptic interfaces has not been investigated in detail. We have developed a set of haptic rendering algorithms for simulating "surgical instrument - soft tissue" interactions. Although the focus of the study is the development of algorithms for simulation of laparoscopic procedures, the developed techniques are also useful in simulating other medical procedures involving touch and feel of soft tissues. The proposed force-reflecting soft tissue models are in various fidelities and have been developed to simulate the behavior of elastically deformable objects in virtual environments. The developed algorithms deal directly with geometry of anatomical organs, surface and compliance characteristics of tissues, and the estimation of appropriate reaction forces to convey to the user a feeling of touch and force sensations.

## 1. Introduction

In collaboration with a team from Massachusetts General Hospital (MGH), we are developing a laparoscopic surgery simulator that can be used to improve practical skills of medical personnel in executing a laparoscopic procedure. The primary focus of this work is the development of physically based models to simulate real-time realistic interactions of surgical instruments with computer models of internal organs for laparoscopic procedures. The models display the visco-elastic behavior of soft tissues in real-time on the computer screen and provide the user with haptic feedback through a force reflecting device. The hardware components of the set-up include an IBM compatible PC (200 MHz, Pentium-Pro) with a high-end 3D graphics accelerator, a force-feedback device (PHANToM from SensAble Technologies Inc.) to simulate haptic sensations, and stereo glasses for 3D visualization of virtual objects. During the simulations, the user manipulates the generic stylus of the force-feedback device to simulate the movements of a surgical instrument and to feel its interactions with the computer generated anatomical organs. The associated deformations of the organs are displayed stereoscopically on the computer monitor and reaction forces are fed back to the user through the haptic interface. The software was written in C/C++, using multi-threading techniques to create separate visual and haptic control loops, thereby increasing the haptics servo rate (varies from 500 Hz to 2 kHz) while simultaneously satisfying the requirements of graphics update rate of at least 30 Hz.

## 2. Surgical Instrument-Soft Tissue Interactions

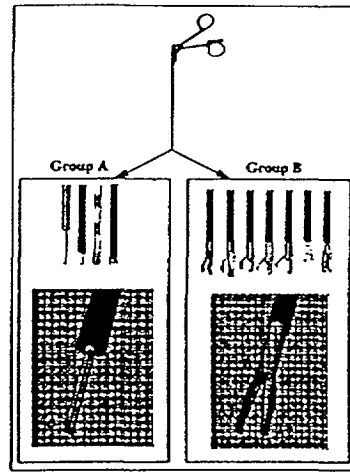


Figure 1. Grouping of laparoscopic instruments according to their mechanical actions: Group A includes the instruments that are mainly used to palpate and deform soft tissues whereas the instruments in Group B are utilized to grasp or cut soft tissues during laparoscopic interventions.

In laparoscopic surgery, a long thin laparoscope is inserted into body through a small incision and the surgeon interacts with internal organs through a set of standard mechanical instruments (e.g. forceps, curved needle, scissors) that can be inserted through trocars. Figure 1 depicts a set of typical laparoscopic instruments. In order to simulate their mechanical actions, we have divided the laparoscopic instruments into two groups based on their functions. The first group includes the long thin straight probes that are mainly used for palpating or puncturing the tissue and for injection (e.g. puncture needle, palpation probe, injection needle). Surgeons also use the closed laparoscopic forceps as a straight probe to explore the internal structures or to open the way for the other structures or to open the way for the other

structures. The second group includes the tools for pulling, clamping, gripping, and cutting soft tissues such as biopsy and punch forceps, hook scissors, and grasping forceps (see Figure 1). Following the grouping of surgical instruments, we have generated a 3D computer model of an instrument from each group to display their behavior in virtual environments. During the real-time simulations, we display the complete 3D models of laparoscopic instruments and their mechanical actions to provide the user with the visual cues. For the purposes of detecting collisions between the surgical instruments and the anatomical models of internal organs, the laparoscope and the surgical instruments are modeled as simple 2D geometric primitives such as line segments and polygons. For example, the long rigid tube of the laparoscope, laparoscopic needle, and probe are modeled as line segments (see Group A in Figure 1: The line segment AB typically represents the computational model of a probe or a needle for the purposes of fast collision detection) whose position and orientation are provided by the encoder signals of the haptic device. Similarly, laparoscopic forceps and scissors are modeled as a combination of line segments and a triangle (see Group B in Figure 1: The triangle  $A_1A_2B$  typically represents the arms of the forceps or scissors for purposes of fast collision detection).

In the past, we have developed novel approaches for detecting the collisions between the simulated stylus of the force feedback device and the virtual objects in the scene. The developed algorithms convey to the user the touch and feel of object shapes and surface details [1, 11]. We have extended these algorithms to simulate the mechanical functions of the surgical instruments and their interactions with soft tissues in virtual environments. The developed algorithms check the collisions of virtual instruments with 3D objects in the scene as the generic stylus of the force feedback device (and hence, the selected virtual instrument) is manipulated by the user. We then send the force commands to the haptic device for displaying the tactual feeling of deformable soft tissue surfaces. The collision detection algorithms for simulating the instruments of Groups A and B slightly differ in implementation. For instruments in Group A, collisions are detected between the stylus,

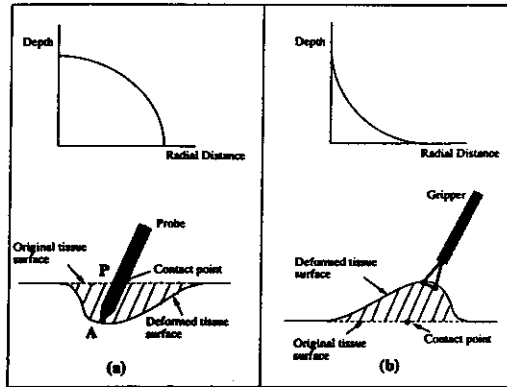
modeled as a line segment, and the geometrical models of 3D objects. These collision detection algorithms are modified for simulating the mechanical actions of laparoscopic instruments of Group B. In order to determine if the object is between the mechanical arms of the laparoscopic forceps or scissors, we first detect the collisions between the 3D objects and the triangle representing the open laparoscopic forceps (triangle  $A_1A_2B$  in Figure 1). The user can close the arms of the simulated forceps and scissors by pressing a switch located on the generic probe of the haptic device to grasp the tissue surface. Then, the simulated arms of the forceps (see the line segments  $A_1B$  and  $A_2B$  in Figure 1) are rotated about the junction point B incrementally until collisions are detected between the each arm of forceps and the 3D object. Once the collisions are detected, the user can pull the tissue and feel the reaction forces via the haptic device.

### 3. Force-reflecting deformable soft tissue models

Once the collision between the simulated instrument and the 3D model of the anatomical structure is detected and the collision point is identified, the problem centers around the instrument-tissue interaction. This involves a realistic haptic feedback to the user as well as a realistic graphical display of tissue deformation and/or piercing or incision, depending on what operation the user chooses to perform on the tissue. This is a nontrivial problem which requires the integration of physics of tissue behavior and computer graphics techniques (also known as physically-based modeling) to create a make-believe world that is realistic enough to mimic reality and efficient enough to be executable in real time.

Graphical simulation of deformable surfaces have been extensively studied in computer graphics and computer aided engineering. Various *direct* or *indirect* deformation techniques have been developed to manipulate the surfaces and to modify the *local* or *global* shape of the objects. One way to categorize the deformation techniques is according to the approach followed by the researchers to deform the surfaces: *geometric* or *physically-based* deformations. In geometric deformations, the object or the surrounding space is deformed based purely on geometric manipulations [4, 7, 9]. In general, the user manipulates the control points that surround the 3D object to modify the shape of the object. On the other hand, physically-based deformation techniques aim to model the physics involved in the motion and dynamics of interactions [12, 14]. Models simulate physical behavior of objects under the affect of external and internal forces. Dynamical equations with constraint relations are solved to describe the behavior of movement. Geometric-based deformation techniques are faster, and are relatively easier to implement. But they do not simulate the underlying mechanics of deformations. Hence, the emphasis is on visual display and the goal is to make deformations appear smoother to the end-user. Sophisticated physically-based models, although necessary for simulating the dynamics of realistic interactions, are not suitable for fully interactive, real-time simulation of multiple objects in virtual environments due to the current limitations in computational power. Yet, it is obvious that internal organs of the human body show physically-based behavior when they are manipulated with surgical instruments. Soft tissues are deformed visco-elastically when they are pushed or pulled by surgical instruments. Depending on the type of surgical instrument and the anatomical structure, soft tissues deform locally or globally. In order to simulate deformations of 3D objects as well as to provide the user with the tactual sensation of interaction forces in virtual environments, we have developed hybrid models that take advantage of geometric and physically-based modeling techniques. We loosely couple the deformation model with the force model to achieve real-time rendering rates. The deformation model estimates the

direction and the amount of deformation (displacement vector) of each node (i.e. a vertex) of the soft tissue surface when it is manipulated with a surgical instrument.



**Figure 2.** Simulation of local soft tissue deformations. A second order polynomial is used to deform the tissues and only the vertices that are within the radius of influence are transformed. (a) Pushing soft tissues: vertices of the 3D model are transformed along the longitudinal axis of the surgical instrument whose mechanical behavior is classified as a "Group A" instrument. Collision point is updated each time the instrument is manipulated. (b) Pulling soft tissues using the instruments in "Group B": once the conditions for grabbing soft tissues are satisfied, the collision point is

pushing or cutting. A mass-spring model (e.g. virtual springs between the nodes), combined with simple energy minimization techniques is utilized to compute the force vectors at the point of contact. The energy minimization approach ensures that each deformed node returns to its low energy state (i.e. original position) when released. The force response of each deformed node is calculated based on its original position, new position, and the viscoelastic law that governs its behavior as suggested by Cover et al. (1993) [5].

#### **Deformation models:**

In *local* deformations, a region of the tissue surface in the close vicinity of the surgical instrument deforms [2, 6]. In general, tissues that are highly visco-elastic deform locally when they are subjected to the external forces. Deformation of the human skin due to external forces is a good example of the local deformation and it has been shown that a second degree polynomial fits the empirical data very well [10]. The local deformation techniques are computationally less expensive and can be used to simulate tissue behavior in regions where the tissues are highly damped. For example, local deformation techniques can be utilized to simulate the insertion of a surgical needle into the crico thyroid membrane in order to oxygenate the virtual patient. In order to deform soft tissues locally, we translate all of the vertices within a certain distance (called the *radius of influence*) of the collision point, along the direction of the surgical instrument. The magnitude of translation is determined using a second order polynomial. The degree and the coefficients of the polynomial define the shape of the deformations. For example, if a second degree polynomial with no linear deformation term is assumed ( $a_1 = 0$ ), then the deformation function takes the following form

We utilize a second order polynomial model for local deformations and a 3D spline model for global deformations (i.e. nodes are connected to each other by means of smooth mathematical functions). The size of the region that is deformed depends on the material properties of the tissue, and the amount of the deformation depends on the tip location of the instrument. For displaying local deformations, the nodes which are only in the close vicinity of interaction are updated.

This approach significantly reduces the number of computations, and therefore enables real-time simulations. The second component, the force model, takes into account the force reflecting behavior of soft tissues when they are subjected to mechanical loads as a result of manipulations such as pulling,

$$\text{Depth} = a_0 + a_2(\text{Radial Distance})^2 \quad (1)$$

where,  $a_0 = AP$  and  $a_2 = -AP/(\text{radius of influence})^2$ . The vector  $AP$  is constructed from the coordinates of the stylus tip and the contact point. The *radial distance* is the distance of each neighboring vertex, within the radius of influence, to the collision point. When soft tissue is pushed by a surgical probe, the contact point (point  $P$ ) and the position of the instrument tip (point  $A$ ) are updated each time the generic stylus of haptic device is reoriented (see Figure 2a). When the soft tissue is pulled by a laparoscopic forceps, the contact point is determined only once and kept constant until the tissue is released (see Figure 2b).

In *global* deformations, soft tissue can deform in all directions when an external force is applied. Its behavior is similar to the behavior of an elastic balloon, filled partially with water. Inertial effects due to internal and external forces may cause global changes in the geometrical shape of the object. Highly elastic soft tissues such as our internal organs undergo these types of global deformations when they are manipulated. We have adapted the technique developed by Sederberg and Parry (1986) and Hsu et al. (1992) to simulate global deformations [9, 7]. Sederberg and Parry (1986) suggested a free-form deformation (FFD) technique for deforming the space that encloses the object. FFD enables the user to interactively modify the object shape by repositioning the lattice of control points that surround the 3D object. Any point within the lattice is defined as:

$$Q(u, v, w) = \sum_{i=0}^3 \sum_{j=0}^3 \sum_{k=0}^3 P_{ijk} B_i(u) B_j(v) B_k(w) \quad (2)$$

or, in matrix form

$$Q = BP \quad (3)$$

where,  $P_{ijk}$  are the control points, and  $B_i(u)$ ,  $B_j(v)$ ,  $B_k(w)$  are known as the third degree Bernstein polynomials or Bezier basis functions. The volume enclosing the 3D object can simply be parametrized ( $u$ ,  $v$ , and  $w$ ) using object coordinates ( $x$ ,  $y$ , and  $z$ ). Although this technique is useful for sculpting general 3D objects in virtual environments, it is obviously not practical to be used in our simulations, because, in real life, the surgeon directly interacts with the organs, not with the control points. Hsu et al. (1992) suggested a method for direct manipulation of free-form surfaces. In this method, control points are moved such that the resulting surface smoothly reaches its intended position by means of a least squares solution. Assume that a single point of the 3D object is translated an amount of  $\Delta Q$  and moved to a new location ( $Q + \Delta Q$ ), then Eq. (3) can be rewritten in the following form:

$$(Q + \Delta Q)_{1 \times 3} = B_{1 \times 64} (P + \Delta P)_{64 \times 3} \quad (4)$$

where,  $\Delta Q$  and  $\Delta P$  represent the changes in the position of object point and the control points (recall from Eq. (2) that there are 64 control points). Eq. (4) reduces to:

$$\Delta Q_{1 \times 3} = B_{1 \times 64} \Delta P_{64 \times 3} \quad (5)$$

Now, the goal is to calculate the change in the control points for a given  $\Delta Q$ . This can be achieved through the use of pseudoinverse solution:

$$\Delta P = (B^T B)^{-1} B^T \Delta Q \quad (6)$$

For single point manipulation, this solution reduces to the following simple form

$$\Delta P = \frac{B^T}{\sum_i (b_i)^2} \Delta Q \quad (7)$$

where,  $b_i$ 's are the elements of the B matrix [3]. Once the changes in the positions of control points are known, the deformed positions of the object can be calculated from  $Q_{new} = B(P + \Delta P)$ .

#### **Force model:**

The force model computes the interaction forces that are transferred to the user's hand via a haptic device. We have opted to use models composed of a network of springs together with a viscous element to estimate the interaction forces. Similar models have been extensively used in computer graphics to simulate deformable objects [14]. In our model, 3D objects are constructed from a mesh of vertices linked to its neighbors by massless springs. In addition, the viscoelastic force on each vertex that slowly moves the vertex back to its home position is calculated based on a simple energy minimization approach. We assume that there exists a massless virtual spring between the original position of each vertex and its updated position and between each vertex and its neighboring vertices. We used this approach to calculate the direction and magnitude of the interaction forces between the simulated instruments and the deformable organs. For each iteration, the force model calculates the reaction force that is sent to the haptic device as a force command to convey the tactual feeling of soft surfaces. Following the computation of the collision point, we first determine the closest vertex to the collision point and assume that the force is applied to this vertex directly. Then, the total force acting on the vertex is calculated as

$$\vec{F}_{total} = k_1 * (\vec{P}_{current} - \vec{P}_{original}) + \sum_{j=1}^N k_2 * (|\vec{L}_j| - d_j) \vec{L}_j / |\vec{L}_j| \quad (8)$$

where,  $\vec{L}_j = \vec{P}_{current} - \vec{P}_j$  and  $d_j = |\vec{P}_{original} - \vec{P}_j|$ .  $\vec{P}_{current}$  and  $\vec{P}_{original}$  represent the current and original coordinates of the vertex closest to the collision point, and  $\vec{P}_j$  represents the original coordinates of the neighboring vertex. The first component of the total force represents the home force that pulls the vertex back to its original position, and the second one represents the forces due to neighbors (where, N is the number of neighboring vertices). The spring coefficients,  $k_1$  and  $k_2$  can be selected properly depending on the application to simulate the surfaces with different material properties. For example, setting the coefficient  $k_1$  higher than  $k_2$  will make the surfaces easy to indent, but difficult to shear.

#### **Conclusions**

In this paper, we have proposed a set of algorithms for simulating surgical instrument - soft tissue interactions (e.g. tissue palpating and grasping) via a haptic interface device in virtual

environments (see Figure 3). Although the focus of our study is the simulation of laparoscopic procedures, the proposed techniques are general enough to be used in other types of medical simulations.

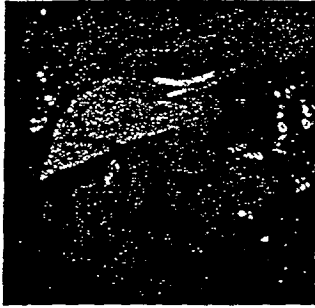


Figure 3. Simulation of soft tissue grasping using forceps.

The developed soft tissue models convey to the user a tactual feeling of pushing or pulling soft tissues. The proposed visual deformation models are decoupled from the force model, and the reaction forces are based on simplified viscoelastic models of soft tissues. Our organ-force models take into account the changes in the shape of the object, but do not consider the internal dynamics of objects in simulating deformations. In the future, we will further improve our models to include the forces that arise from these effects. Moreover, deformations generated using the free-form technique look smoother than the proposed local deformation technique but need to be improved to provide the user with more sophisticated

manipulation of control points.

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