SIMULATING SOFT TISSUE WITH A TACTILE SHAPE DISPLAY

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INTRODUCTION

One of a surgeon's most important tools is a highly developed sense of touch. Surgeons rely on sensations from the fingertips to guide manipulation and to perceive a wide variety of anatomical structures and pathologies. One important property used to assess the health of organs and tissues is compliance, which surgeons normally obtain by squeezing or indenting with their fingers.

Unfortunately, new surgical techniques, such as minimally invasive procedures and those involving robotic manipulators, separate the surgeon's hands from the surgical site. In these situations the surgeon's perception is limited to visual feedback and force feedback through the handles of long instruments. However, vision provides very little information about the compliance of an object. Furthermore, Srinivasan and LaMotte [4] have shown, in direct manipulation experiments, that kinesthetic (force-position) sensing alone is also insufficient: their subjects were unable to determine the difference between even the hardest and softest rubber samples used in the absence of the distributed skin sensation of surface deformation. This suggests that even the most accurately relayed force feedback is inadequate for compliance discrimination.

DISPLAY SYSTEM

This problem can be addressed by providing tactile, as well as kinesthetic, feedback to the surgeon. One method of conveying the needed tactile information, for both remote surgery and virtual surgical simulators, is by using a tactile display to recreate the tactile stimulus directly on the surgeon's fingertip. A tactile display is a device consisting of an array of pins which are raised and lowered to provide a sensation of surface deformation. Previously, tactile displays have primarily been considered in terms of their utility in providing small scale shape, such as during tumor localization [3].

Here, we consider what is needed to make a display feel like a compliant object. For this we must understand how the finger would have interacted with the compliant object if it was in direct contact, and then how to relay this information through a rigid shape display. The principle behind the simulated tissue interaction is that the surgeon imparts the desired contact force to the display. The stiffness display system then produces the correct deformation profile and rigid body motion, corresponding to the expected result for the real object.

The proposed tissue stiffness display system would comprise a shape display mounted on an instrumented linear actuator. The shape display is an array of closely spaced (i.e., 2 mm apart) individually actuated pins which press against the fingerpad. Shape memory alloy wires raise and lower the pins, providing the necessary tactile information [3]. The linear actuator provides the kinesthetic (force-displacement) relationship and measures the total force exerted by the surgeon's finger.

CONTACT MODEL

In order to predict the response of the fingerpad to a remote object, we need a good model of the finger-tissue interaction. Experimentally, we have begun to examine this issue by considering the dynamic contact of the human fingerpad with a flat, rigid surface [2]. We have found that the quasi-static pressure response of the fingerpad can be successfully approximated by Hertzian contact [1] modified to allow for a nonlinear variation of the modulus of elasticity. Approximating the fingerpad as a sphere, the contact pressure distribution, $p(r,x)$, can be described by:

$$p(r,x) = \frac{2E_f^*(x)}{\pi R_f}\left[a(x)^2 - r^2\right]^{1/2} \qquad (1)$$

where r = radial distance from the center of contact,
x = maximum fingerpad deformation (i.e., at r=0),
E^*_f = modulus of elasticity of the fingerpad,
R_f = radius of curvature of the fingerpad, and
$a(x)$ = radius of the contact area.

From Hertz theory, the radius of contact, $a(x)$ is given by:

$$a(x) = \left(R_f\delta\right)^{1/2} = \left(\frac{3PR_f}{4E_f^*(x)}\right)^{1/3} \qquad (2)$$

where P = total contact force, and
δ = rigid body indentation.

The nonlinear varying modulus of elasticity is assumed to be given by:

$$E_f^*(x) = \frac{2b}{m}\left[e^{mx} - 1\right] \qquad (3)$$

where b and m are parameters of the model fit to experimental data [2].