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Toward MR-Compatible Needle Tip Force Display using EAP Actuation

Amy Kyungwon Han, Shiquan Wang, Jung Hwa Bae, David L. Christensen and Mark R. Cutkosky

***Abstract*— We present a display based on electroactive polymer (EAP) diaphragms intended to provide tactile feedback for an instrumented, MR-compatible, force sensing biopsy needle. The device is pinched between the thumb and index fingers and provides a low-impedance stimulus to the fingertips, comparable to using a voice-coil, for frequencies below 100Hz. Future work will employ multi-layer EAPs for higher forces. User tests will evaluate the efficacy of the device to render events like membrane puncture for MR-guided procedures.**

I. INTRODUCTION

Magnetic resonance imaging (MRI) has become a powerful non-invasive tool for guided medical treatments and diagnosis. For biopsy and other needle-based procedures in MRI scanners, it can be desirable to have haptic display of dynamic forces experienced at the needle tip. In general, the physician cannot experience these forces directly because the patient is inside the MR bore. In previous work we demonstrated that a voice-coil actuator can provide sufficiently accurate display of tip forces to improve accuracy in detecting membrane puncture [1], however it was not suitable for use adjacent to an MR machine. In other work, we have shown that ultrasonic piezoelectric motors can provide steering guidance that helps people to locate targets within tissue [2], however, the ultrasonic motors are inherently stiff, position-based devices and not well suited for accurate rendering of dynamic needle tip forces. Among possible actuators that are both inherently suited for low-stiffness, back-driveable force display, and also MR-compatible, are electroactive polymers (EAPs). EAPs have large strain, high power-to-weight ratio, a moderately wide range of possible actuation frequencies (especially for small displacements), and low cost. Like biological tissue, they have compliance and damping, and their design and stiffness can be tuned for specific applications. EAPs are actuated by applying a high voltage and very low current, hence they are not affected by and do not interfere with high magnetic fields in the MRI environment. Previous investigations of EAP actuators adjacent to, or within, MR bores include a manipulator composed of bistable EAP mechanisms [3] and folded, linear actuators using silicone rubber as the dielectric material [4]. In both cases, it was demonstrated that the actuators produced no imaging artifacts. More generally, EAPs have been considered for use in haptic displays due to their potentially high power density, large strain, and fast response. A braille display with commercial tubular EAP actuator is presented in [5]. In other work, a single-axis commercial EAP actuator is compared to a voice coil actuator [6]; a multi-axis haptic display is presented in [7].

In view of the aforementioned advantages, we selected EAP actuators for the display of needle tip forces in MR-guided biopsy. In this preliminary paper we briefly review the EAP haptic display design and present force and displacement results comparing its performance to that obtained from a small voice coil actuator used in [1]. Ongoing work will improve the design with additional layers for higher force capability and apply it in user tests to determine whether it provides a measurable improvement in tasks such as membrane puncture or tissue stiffness identification for needle-based procedures.

II. ELECTROACTIVE POLYMER

An electroactive polymer (EAP) is a thin dielectric elastomer film sandwiched between stretchable, flexible electrodes on the upper and lower surfaces. When a large voltage is applied to the electrodes, they generate a Maxwell stress, which squeezes the elastomer in the thickness direction. Being incompressible, it expands in the planar direction. The haptic display presented here is based on a highly pre-stretched ($\approx 400\%$) diaphragm of 3M VHB 4910 acrylic film. The film is supported by a circular frame at its periphery; a disk is attached to the center for applying forces normal to the plane (fig. 1b). The electrodes are created by spraying a thin ($\approx 10\mu$m) layer of a very soft silicone rubber filled with carbon particles. The design and fabrication details are adapted from those in [8].

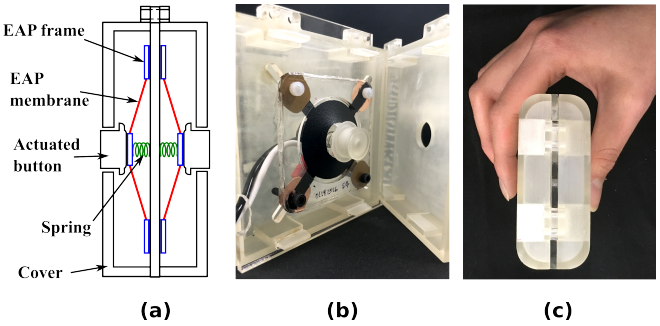

Fig. 1. (a) Cross-section side view of the haptic display, (b) EAP inside the device, (c) fingers placed on the actuated buttons.

Stretching the film increases its stiffness and reduces its thickness to approximately 60μm. It also produces a relatively high initial tension so that the diaphragm resists forces applied to the center disk, in the direction normal to the plane. Applying a high voltage relaxes the initial tension so that the stiffness drops nearly to zero for small deflections.

A.K. Han is supported by a Samsung Scholarship.
All Authors are with the Mech. Eng. Dept., Stanford University, Stanford, CA 94305 kwhan@stanford.edu

The current EAP diaphragm has an outer and inner diameter 31.5mm and 14mm, respectively and is designed for voltages up to 5.75kV (with currents on the order of microamps) and displacements of ± 5 mm in the normal direction. At 5.75kV, the device stiffness with respect to normal displacements decreases from 0.14N/mm to 0.06N/mm. Higher force and stiffness are possible with multi-layer devices.

III. HAPTIC DEVICE PROTOTYPE

A. Design

The haptic display, shown in fig. 1, employs two EAP diaphragms, preloaded by an inner compression spring, and has a height, length, and width of 80mm, 80mm, and 30mm, respectively. A user holds the device pinched between the thumb and index finger, pressing on buttons attached to the inner disks of the EAPs. The grip is intended to be similar to that used for holding the base of a needle or other surgical tool. When a voltage is applied, the EAPs relax, and the spring extends, producing a sensible force against the fingertips.

B. Characterization

Isometric force and displacement in the normal direction for a single EAP actuator and compression spring (one half of the system in fig. 1) were measured as a function of frequency at 5.5kV sinusoidal voltage. A commercial load cell (ATI®Gamma) and muscle lever (Aurora Scientific 305B) are used for the force and displacement measurements, respectively. The amplitude of oscillations of the moving part, and the maximum displacement from initial position at 0kV are shown in fig. 2 as (a) and (b), respectively. The displacement testing is performed without external load on the buttons; however, the motions are affected by a small amount of friction and moving mass in the device.

The maximum blocked, or isometric, force, with the buttons fixed at their rest position, is shown in fig. 2 as (c). The force amplitude is almost constant up to 80Hz, whereas the displacement decreases with frequency due to the relatively high viscoelasticy of the film. For comparison, curve (d) shows the force from the small voice coil used in [1]. The voice coil receives a controlled sinusoidal current, and the force at 1Hz has been scaled to match that from the EAP.

IV. CONCLUSIONS AND FUTURE WORK

As the results in fig. 2 indicate, an EAP-based haptic device that is gripped between the thumb and index fingers produces useful stimuli at frequencies below 100Hz, not unlike those from a lightly damped voice coil in parallel with a spring. In application, the performance will be in between the cases of free displacement and blocked-force, depending on how firmly users grip the device. It is likely that higher forces will be desired, which can be accomplished by using multi-layer EAP diaphragms. Prototypes with up to 4 layers have been fabricated with behavior that extrapolates directly from single-layer devices.

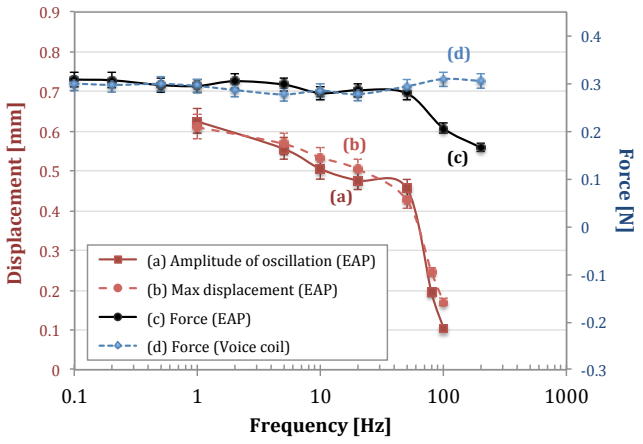

Fig. 2. Oscillation amplitude (a) and maximum displacement (b) of one EAP actuator as a function of frequency. Isometric force of the EAP actuator (c) compared to the force from a voice coil actuator (d) versus frequency.

Beyond increasing the force magnitude, a second possible concern is the viscoelasticity of VHB film, which reduces displacements at frequencies above 20Hz. An alternative is to use a silicone rubber film (e.g., Elastosil film, Wacker Inc.) with less viscoelasticity, but somewhat lower displacements.

In addition to refining the design, the next steps include testing with experienced needle users in simulated membrane puncture and tissue discrimination tests, using feedback from the instrumented biopsy needle presented in [1]. Ultimately, we aim to conduct in-vivo animal tests in an MRI scanner.

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