Tactor-Induced Skin Stretch as a Sensory Substitution Method in Teleoperated Palpation

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***Abstract*—When we use a tool to explore or manipulate an object, friction between the surface of the tool and the fingerpads generates skin stretch cues that are related to the interaction forces between the tool and the object. In this study, we emulate these naturally occurring skin stretch cues in order to convey force direction and magnitude information to users during teleoperation. We hypothesize that skin stretch feedback is a useful substitute for kinesthetic force feedback in force-sensitive teleoperated tasks. In this study, ten participants performed teleoperated palpation to determine the orientation of a stiff region in a surrounding artificial tissue using five feedback conditions: skin stretch, force, reduced gain force, graphic, and vibration. When participants received skin stretch feedback, they localized the stiff region as well as with force feedback, with no increase in task completion time. Additionally, participants receiving skin-stretch feedback localized the stiff region statistically significantly more accurately than those using vibration feedback. Although participants using skin stretch exhibited higher interaction forces than when using force, vibration, and graphical feedback, skin stretch statistically significantly decreased interaction forces compared with reduced gain force feedback. Thus, skin-stretch feedback is a compelling substitute for force feedback and may be useful in scenarios where force feedback is reduced or infeasible.**

***Index Terms*—Cutaneous feedback, haptic devices, sensory substitution, skin stretch, stiffness perception.**

I. Introduction

OUR sense of touch provides important haptic (kinesthetic and cutaneous) information when interacting with objects in the world around us. During teleoperation, where a user controls a remote robot by manipulating a local master device, this haptic information is reduced or lost completely. The loss of haptic information during teleoperation can be detrimental, reducing perceptual ability [1] and increasing environment interaction forces [2]. In some teleoperation scenarios, such as robot-assisted minimally invasive surgery, these deficiencies can make it difficult to perform delicate tasks, such as suturing and tissue exploration/characterization [3], [4]. Despite these deficiencies, teleoperation is desirable because it provides numerous advantages to direct manipulation, such as the ability to perform tasks in remote and hazardous locations, scale human motions, and remove hand tremor, all while retaining human judgment and supervision.

In surgical teleoperation, force feedback has been shown to be helpful for some tasks [2], [5], but the use of force feedback is limited due to stability issues caused by time delays or inaccuracies in robot modeling and force estimation. Thus, researchers have used sensory substitution methods to replace kinesthetic force feedback with graphic, auditory, or vibration cues [6]–[9]. These methods enable the display of important force information while preserving the stability of a telemanipulated system. However, graphic and auditory feedback requires the use of alternative senses for displaying force information, and in some applications, these senses are frequently saturated with important information. While vibration feedback uses the sense of touch, it cannot convey direction information without multiple actuators. Alternative forms of cutaneous feedback, such as pressure and skin stretch, are recently gaining attention and may provide a solution to naturally convey force magnitude and direction information during force-sensitive teloperated tasks.

We propose a tactor-induced skin-stretch device for sensory substitution that emulates naturally occurring tactile feedback to convey force information to the user. When users interact with the environment via a tool, they experience shear forces that cause local skin deformation of the fingerpad. For the same interaction configuration, larger force magnitude causes a greater amount of skin stretch. Prior work suggests that tactile cues, similar to the skin stretch created by tool-mediated external forces, are important for stiffness discrimination ability [10], [11]. In this paper, we control the motions of tactors placed against the skin to create tactile sensations consistent with the shear forces that occur naturally during tool-mediated interaction. We describe the design and implementation of a 1-degree-of-freedom (DoF) skin-stretch feedback device and use it to convey force information in a teleoperated palpation task. While this skin-stretch feedback emulates part of the tactile sensations associated with stiffness discrimination, it does not include all of the natural tactile information such as contact area spread rate or pressure distribution. In a previous study using a similar skin-stretch device, stiffness discrimination with tactor-induced

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skin-stretch cues was similar to receiving feedback from a kinesthetic force feedback device [12].

In this study, we compare skin-stretch feedback to four other methods for portraying force information in a teleoperated palpation task. In order to determine the efficacy of skin stretch when compared with force feedback and common force sensory substitutes, participants received skin stretch, force, reduced gain force, graphic, and vibration feedback, in addition to visual feedback of the remote environment. For this task, participants attempted to determine the orientation of a stiff region in a slab of artificial tissue. We found that participants using skin-stretch feedback performed similarly to those using traditional force feedback. Additionally, participant angular error using skin-stretch feedback was statistically significantly lower than when using vibration feedback.

II. Background

A. Force Feedback

Many studies have shown the benefits of using force feedback in teleoperation tasks. For example, Tholey *et al.* [3], [13] showed the importance of force feedback in tissue stiffness characterization, and Wagner *et al.* [4] showed that force feedback reduced interaction force and errors for both experts and novices in a blunt dissection task. While there are documented advantages to force feedback, state-of-the-art robot-assisted minimally invasive surgery systems, in particular, the commercially available da Vinci Surgical System (Intuitive Surgical, Inc.), have greatly attenuated force feedback.

To convey force feedback in teleoperated systems, force acting on the slave robot is measured or estimated and relayed back to the master manipulator. This can be achieved by directly measuring the forces acting on the slave robot, using a force sensor attached to the end effector, or by examining the position error between the master and slave robots to estimate the force. The lack of quality force feedback in teleoperation systems stems from a variety of causes. Accurate force sensing is challenging to implement in a clinical setting due to biocompatibility, sterilizability, and cost [14]. Bilateral teleoperation controllers that approximate the forces that occur at the slave robot are also challenging to implement because inertia and friction compensation must be implemented to avoid relaying slave device dynamics to the user. Additionally, inaccuracies in robot modeling and force estimation make these force feedback systems vulnerable to instability. Instability may be exacerbated in the case of remote teleoperation, where significant time delay exists in the teleoperation loop. Finally, teleoperator system stability and transparency are directly conflicting design goals [15].

B. Sensory Substitution

To avoid some challenges of providing high-quality force feedback while preserving stability, sensory substitution methods have been used to convey force information via alternative senses. When using a force sensory substitute, we effectively open the force feedback control loop between the master and slave robot, thereby reducing the risk of instability.

1) Graphic and Auditory Sensory Substitution: Many researchers have used graphic or auditory feedback to convey force information. These methods provide the user of the teleoperated system with a graphic or auditory indicator of the forces they are applying on remote objects with the slave robot. One study found that when these methods are used to substitute for force feedback in a teleoperated suturing task, resulting suture tensions using both auditory and graphical feedback more closely matched suture tensions under ideal conditions (hand tied) than when using no feedback [6].

Although some studies have reported graphic and auditory feedback to be useful, others have reported negative effects. In another teleoperated suturing study, these methods had no positive effects and, in fact, resulted in a greater number of suture breaks among surgeons with robotic surgery experience [7]. A different study found that graphic sensory substitution actually slows task performance [8]. While Cutler *et al.* found that discrete auditory signals improved surgical precision in a simulated ophthalmic surgery [16], surgeons in [6] preferred continuous graphical feedback over discrete auditory feedback. Continuous auditory feedback could be used, but continuous sound could mask environment sounds that are important for task performance. In settings where human operators are already highly stimulated by visual and auditory cues, it may be prudent to convey force information through the sense of touch. Tactile sensory substitution has the added benefit of conveying the information through the sense for which it was originally intended.

2) Tactile Feedback: Tactile devices for force sensory substitution use a variety of actuation technologies. Vibration feedback, delivered via an eccentric rotating mass or a linear resonance actuator, is a common form of tactile feedback. Vibration feedback devices have been used in many form factors, including a vibrating cylindrical handle [9], and several vibrating glove configurations, such as [17] and [18].

Vibrotactile feedback has been used to substitute for force feedback in a variety of applications, including robot-assisted surgery [19], [20], where it was shown to increase sensitivity to tissue contact and improve users' ability to differentiate tissue softness in a simulated surgery task. Despite these advantages, a single vibrotactile actuator can only display force magnitude and frequency information, losing important directional signals. Multiple actuators can be used, but in addition to added costs and complexity, it can be difficult to differentiate between vibration actuators that are located in close proximity [21].

Researchers have used tactile cues to create compelling sensations. Kim and Colgate [22] developed a haptic device that used a tactor to display pressure, vibration, shear force, and temperature to the skin of an amputee, enhancing the subjects' prosthetic grip force control. Another tactile feedback method is cutaneous normal force [23], [24]. This method attempts to replace the full haptic interaction with a wearable device that provides normal force to the fingertip. Prattichizzo *et al.* [24] call this cutaneous normal force feedback "sensory subtraction" because it replicates the cutaneous normal force of an interaction, but the kinesthetic component of the interaction is subtracted out. This technique was successfully used in a virtual ring-and-peg task where it improved task performance when compared with

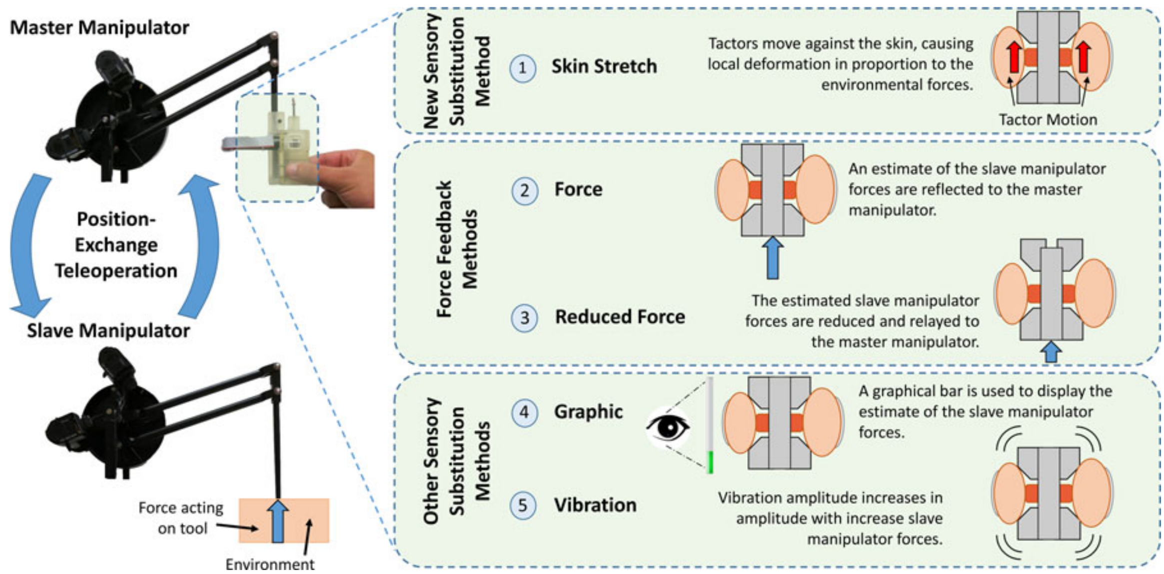


Fig. 1. Five feedback methods are tested for relaying force information to the user of a teleoperated system. In each case, the same forward control law is used, but a different feedback method is used to convey the forces from the slave environment.

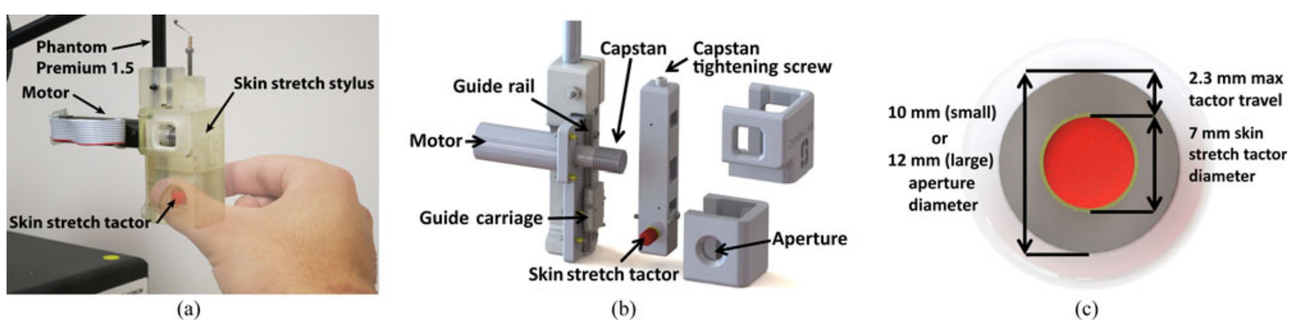


Fig. 2. (a) Participants gripped the skin-stretch stylus with their thumb and index fingers over the apertures. The tactor moved to convey force information during the skin-stretch feedback trials. For the other feedback methods, the tactor was actuated to maintain its position relative to the aperture. (b) The skin-stretch stylus: A DC motor actuates the rod containing the skin-stretch tactors. The tactors then move within the apertures. (c) Small and large apertures were used for participants with small and large fingers. The tactor origin was slightly below the center of the smaller aperture to allow the same 2.3 mm of travel.

graphic and auditory sensory substitution methods [25]. In that experiment, cutaneous normal force was applied to the fingertips based on the virtual slave device grip force. The TAKO-Pen tactile device is similar to cutaneous normal force feedback [26]. This device uses suction on the fingerpads to alter the user's perception of stiffness. During initial studies, increasing suction pressure increased users' perception of stiffness. However, many interactions contain cutaneous cues in nonnormal directions. We have chosen to investigate tangential deformation of the skin.

C. Skin-Stretch Feedback

Whereas sensory subtraction attempts to replicate the forces normal to the fingerpad that are present in natural interactions, skin-stretch feedback tangentially displaces the skin, creating a stimulation similar to the shear forces occurring during natural interaction. This method takes advantage of the finger's increased sensitivity to shear forces, compared to normal forces [27]. Skin stretch has been used successfully to communicate proprioceptive information [28], [29], stiffness [30], and friction [31]. Additionally, skin-stretch feedback provides the opportunity to convey directional information via actuation of the tactor in different directions [32]. Other devices have used a moving tactor with 2 DoFs to convey 5-DoF directional cues [33], and a 3-DoF device that combines skin stretch with normal indentation has successfully conveyed 3-DoF force information [34].

Although skin stretch has been used extensively to communicate force information and guidance, it has yet to be applied in a teleoperation setting. In this study, we used a device that actuates the tactor in 1-DoF to test the ability of participants to use skin-stretch sensory substitution in a force-sensitive teleoperated palpation task.

III. Methods

A. Skin-Stretch Device Design

We designed a skin-stretch device in a stylus form factor [see Fig. 2(a)] to imitate interaction with an object surface via a hand-held tool. The stylus consists of two dome-shaped, 7-mm-diameter tactors (friction elements that displace the skin) actuated by a 16-mm geared DC Motor (Maxon RE16 with planetary gearhead GP16A). The DC motor drives a capstan which is connected by a steel cable to a vertical bar carrying the two tactors. A cable tensioning screw on the vertical bar adjusts the capstan cable tension. A carriage fixed to a linear guide rail constrains the bar to vertical motions.

Users place their fingertips on two conical apertures that are located on an outer shell of the device. Previously, we determined that a 12-mm-diameter aperture was less effective for participants with an index fingerpad width of less than 14 mm [30]. To mitigate the effect of participant finger size on perception, two apertures, depicted in Fig. 2(c), were used in this experiment. We used 10- and 12-mm apertures for participants with index fingerpad widths below and above 14 mm, respectively.

The two tactors of the skin-stretch stylus, which are Lenovo Trackpoint Classic tactors with a sandpaper-like finish, lie within the apertures. When the user holds the two apertures, friction between the tactors and the fingerpads generates shear forces that cause local skin deformation as the tactors move vertically in 1 DoF. A Maxon MR Type M encoder with 512 counts per revolution determines the location of the tactors within the apertures.

The skin-stretch device is similar in function to the device in [12], but includes several improvements. While the previous version used a crank arm with a nonlinear mapping between position and motor angle, the current version uses a capstan drive with a linear mapping between tactor position and motor angle of 4.75 mm/rad. The capstan drive also prevents a subtle clicking sound caused by the backlash in the previous version. Additionally, the new version is smaller than the previous version, making it possible to mount on a larger variety of master manipulators. The tactors must move quickly, because the desired tactor position must track fast changes in force produced by the slave manipulator. The tracking performance of the tactors was evaluated over the course of the experiment, and the total tactor position error based on backlash and controller tracking error was less than 10% of the tactor range of motion during 99% of the trial time of the experiments described in Section III-D.

B. Experimental Setup and Device Control

The skin-stretch stylus is attached to the end of a Phantom Premium 1.5 haptic device as shown in Fig. 2(a). The Phantom Premium with the stylus attached serves as the master manipulator to a second slave manipulator Phantom Premium 1.5, as depicted in Fig. 3. A video feed of the remote scene was displayed on a computer screen, and the manipulators were controlled using a custom application written in C/C++. When the stylus moves in space, the encoders of the master are used to determine the desired position and velocity for the slave robot. The position-exchange teleoperation controller is shown in Fig. 4. The haptic loop was controlled at 2000 ± 50 Hz. The force $\vec{F}$ commanded to the slave robot was

$$\vec{F} = k_\mathrm{p}(\vec{x}_\mathrm{m} - \vec{x}_\mathrm{s}) + k_\mathrm{d}(\dot{\vec{x}}_\mathrm{m} - \dot{\vec{x}}_\mathrm{s}) \tag{1}$$

where k_p and k_d are the proportional and derivative gains of 350 N/m and 1.2 N/mm/s, respectively, $\vec{x}_\mathrm{m}$ and $\vec{x}_\mathrm{s}$ are the positions of the master and slave manipulator's end effectors, and $\dot{\vec{x}}_\mathrm{m}$ and $\dot{\vec{x}}_\mathrm{s}$ are the estimates of the master and slave end effector velocities. The 350-N/m controller stiffness lies within the 300–400-N/m virtual stiffness range described by O'Malley and Goldfarm [35]. Below 300 N/m, perception capabilities were shown to improve with increasing virtual stiffness in a number of perception experiments. Participants reached a limit in their perception capabilities at stiffness levels of 300–400 N/m. The velocity estimates were calculated from backward differentiation followed by filtering with a second-order Butterworth low-pass filter with a cutoff frequency of 150 Hz, chosen during pilot testing to maintain haptic quality and stability. When performing a palpating motion in free space, similar to the motions performed within the experiment, the root-mean-square position error (RMSE) was 0.67 mm.

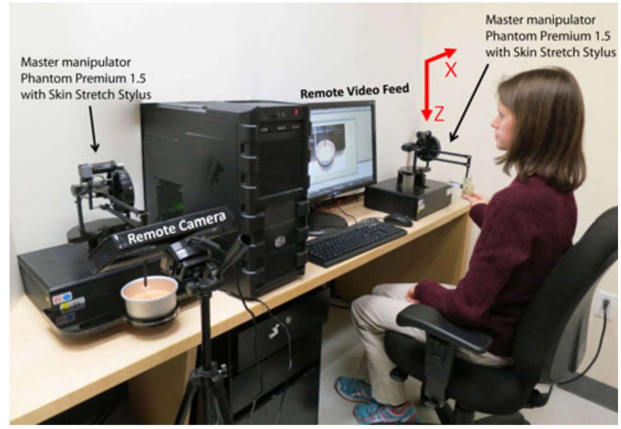


Fig. 3. Teleoperated palpation experimental setup. A user interacts with the master manipulator (Phantom Premium 1.5, SensAble Technologies) by moving the skin-stretch stylus that is attached to its distal link. The slave manipulator (another Phantom Premium 1.5) follows the motions and interacts with the tissue specimen. The video camera captures and displays the remote scene. During the actual experiment, a shroud blocked the slave robot from the participant's view.

The teleoperation system is programmed to provide one of five different kinds of force information feedback (see Fig. 1). Regardless of which feedback method was used, $\vec{F}$ was commanded to the motors of the slave manipulator. All feedback methods use $\vec{F}$ as an estimation of the environment interaction forces occurring at the slave robot. The relevant forces are listed in Table I. The participants were instructed to use vertical palpation motions because sliding forces across the tissue caused permanent damage to the artificial tissue surface. The palpation motions were predominantly vertical, and the workspace of the tissue container kept the end effector of the slave manipulator in a relatively vertical configuration. Thus, all feedback methods were implemented based on the vertical component of the force commanded to the slave manipulator, F_z. The following force information feedback methods were implemented with the parameters for each selected in pilot studies to maximize performance:

1) Skin Stretch: The vertical force F_z was transformed into a desired skin-stretch tactor displacement, z_ss:

$$z_\mathrm{ss} = \begin{cases} -r_\mathrm{ss} F_\mathrm{z}, & \text{if } F_\mathrm{z} \leq 3\mathrm{N} \\ -2.3\,\mathrm{mm}, & \text{if } F_\mathrm{z} > 3\mathrm{N} \end{cases} \tag{2}$$

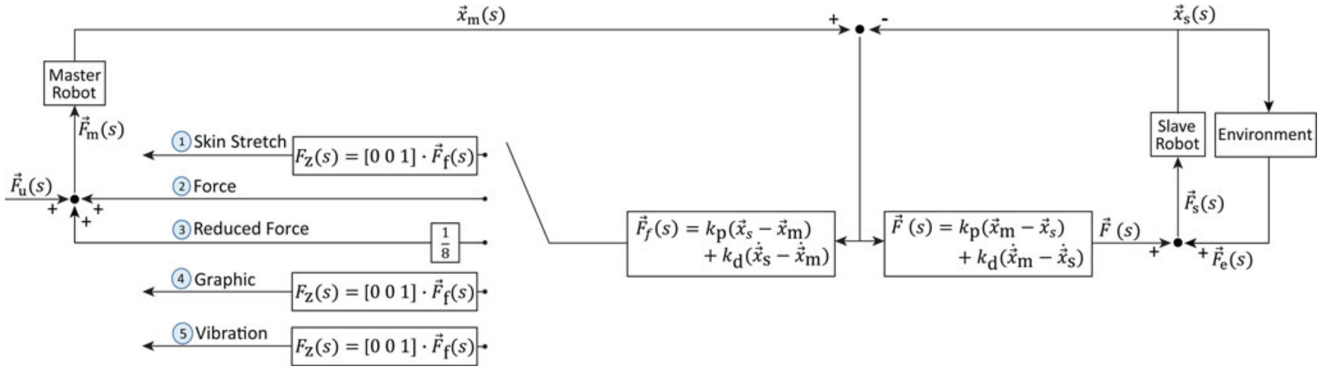


Fig. 4. Control diagram for the teleoperator system. The user applies force on the master manipulator, and a PD position controller regulates the movement of the slave robot. In each experimental trial, one of five different feedback methods is used. Numbers adjacent to the feedback methods represent their order of presentation in Fig. 1 and Section III-A.

TABLE I
LIST OF FORCES USED IN THE CONTROL DIAGRAM

Symbol	Definition
$\vec{F}_u(s)$	Force applied to master manipulator by user
$\vec{F}_m(s)$	Total force acting on master manipulator
$\vec{F}_e(s)$	Force applied to slave manipulator by environment
$\vec{F}(s)$	Force applied to slave manipulator by controller
$\vec{F}_s(s)$	Total force acting on slave manipulator

where r_{ss} is the skin-stretch ratio, mapping force to factor displacement. A negative tactor displacement, z_{ss}, causes skin stretch in the opposite direction of the applied force—emulating the tactile sensation when interacting with the environment. The desired tactor position was filtered with a second-order Butterworth low-pass filter with a cutoff frequency of 5 Hz to prevent a transient vibration-type feeling from the tactor resulting from fluctuations in the slave robot position around the master position when tracking dynamic motions. This cutoff frequency is acceptable because the frequency of human hand movements in various medical contexts [36], and the palpation movements in our experiment, is well below 5 Hz. We chose a skin-stretch ratio of 0.75 mm/N based on prior work and on a pilot experiment. In prior work, we tested three skin-stretch ratios for which stiffness discrimination ability was similar to that of kinesthetic force feedback [12]. During pilot studies, several values within the range of the previous study were tested, and 0.75 mm/N was chosen based on participant preference and the ability of participants to discern artificial tissue. Tactor displacement was limited to 2.3 mm due to the physical constraints of the apertures [see Fig. 2(c)]. This tactor displacement corresponded with 3 N of force occurring at the slave manipulator.

The exact deformation of the skin is unknown and may be smaller than tactor displacement due to potential slip of the tactor relative to the skin. Participants were coached to hold the device tightly enough such that the tactor did not slip against their fingerpad, but rather caused local deformation. We did not monitor the extent of skin stretch, and instead focused on the participants' use of the tactor displacement stimulus.

2) Force: The kinesthetic force F_{f} that is commanded to the motors of the master manipulator is

$$\vec{F}_{\mathrm{f}} = -\vec{F}. \tag{3}$$

This relation directly mirrored the commanded slave forces and displayed them back to the user.

3) Reduced Force: The reduced gain kinesthetic force F_{rf} commanded to the motors of the master manipulator was

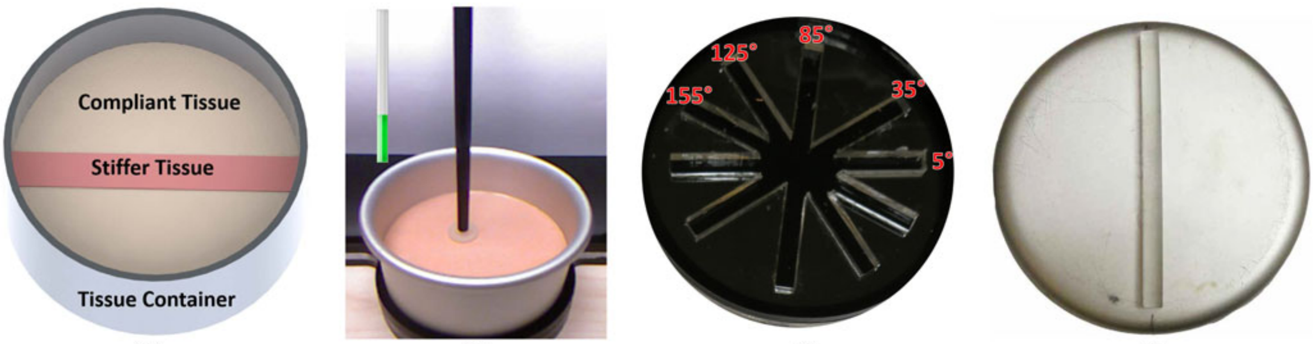
$$\vec{F}_{\mathrm{rf}} = \frac{1}{8}\vec{F}_{\mathrm{f}}. \tag{4}$$

Reduced force feedback may be used in situations where stability constraints limit the maximal gains of the force feedback, such as in the case of time delay. The $\frac{1}{8}$ gain reduction was chosen experimentally such that the system was barely stable with a one-way time delay of 50 ms, which is on the order of the delay experienced during intercontinental teleoperation [37]. This time delay was not implemented in the current study, but the reduced gain force feedback serves as a benchmark for performance in situations where force sensory substitutes might be used.

4) Graphic: A vertical bar, as seen in Fig. 5(b), was displayed graphically on the teleoperator video feed next to the region of interaction. The percentage fill of the graphical bar, h_g, was proportional to F_{z} and reached maximum value at 100% when the magnitude of F_{z} was equal to 4 N:

$$h_{\mathrm{g}} = \begin{cases} 0\%, & \text{if } F_{\mathrm{z}} \leq 0.16\,\mathrm{N} \\ r_{\mathrm{g}} F_{\mathrm{z}} - 4\%, & \text{if } 0.16 \leq F_{\mathrm{z}} \leq 4.16\,\mathrm{N} \\ 100\%, & \text{if } F_z > 4.16\,\mathrm{N} \end{cases} \tag{5}$$

where r_{g} is the graphical bar percentage-to-estimated force ratio. During pilot studies, a 0.16-N dead-band was chosen to prevent the forces of moving the slave manipulator in free space from generating graphical feedback. Participants rarely palpated with greater than 5 N of force. However, participants tended to briefly interact with the environment in a matter that saturated the graphical feedback, no matter how high those forces were. Consistent palpation forces of greater than 4.5 N caused permanent damage to the artificial tissues. We chose $r_{\mathrm{g}} = 25\%/\mathrm{N}$, which resulted in a usable force range of 4 N



(a) (b) (c) (d)

Fig. 5. (a) Rendering of the artificial tissue and container. Color differences were created in this diagram to easily identify the shape and location of the stiff region; the actual tissues used for the experiment had a uniform appearance. (b) Example of the visual display that was presented to participants for all feedback cases. The graphic bar used to convey force information only during the graphical feedback case can be seen in the upper left. The percentage fill of the bar, h_g, is an indication of the vertical forces occurring at the slave manipulator. During other feedback trials, the graphic bar remained empty. (c) Custom-designed fixture that was used for registering the artificial tissue container to the slave Phantom Premium in a known orientation. This allowed for the consistent orientation of the stiff region across trials. (d) Underside of the artificial tissue container had a rigid member that slots into the registration fixture.

and caused graphical feedback saturation at 4.16 N, keeping the majority of interaction forces below that value. This kept the artificial tissues from being damaged and also prevented overheating of the slave robot's motors.

5) Vibration: This feedback was implemented by commanding a high-frequency sinusoidal force to the master manipulator motors. The participants felt the resulting vibration through their contact with the stylus. We used amplitude modulation to communicate force magnitude due to its efficacy over other methods of vibration modulation [38]. The range for peak vibration sensitivity of the fingertip varies between 150 and 550 Hz depending on the source [21], [39]. We chose 200 Hz based on participant preference during pilot testing. The vibration forces were commanded to the master manipulator along the x-direction (see Fig. 3) to avoid confounding the vertical input to the master manipulator, although users perceived the vibration as a magnitude without directional information [40]. These forces were

$$F_{\mathrm{v}} = \begin{cases} 0, & \text{if } F_z \leq 0.4\,\mathrm{N} \\ r_{\mathrm{v}} F_{\mathrm{z}} \sin(200 \times 2\pi t), & \text{if } F_z > 0.4\,\mathrm{N} \end{cases} \quad (6)$$

where $r_{\mathrm{v}} = 0.3$ is the vibration feedback gain. This gain was allowed the vibration to be perceptible at small forces of interaction, but did not cause discomfort. The 0.4-N dead-band prevented occasional small vibrations during manipulation in free space that could confuse participants. An accelerometer was used to confirm that the stylus vibrated at 200 Hz.

C. Artificial Tissue Models

Artificial tissue models were molded using Ecoflex 0030 platinum cure silicone rubber, similar to [5]. We adjusted the compliance of the tissues by using silicone thinner and added pigment to achieve a skin-like color. Each tissue was created in a three-step process that resulted in a circular tissue with a bisecting stiffer region as shown in Fig. 5(a). First, a wooden block was placed into a circular container bisecting the circle and a compliant mixture was poured into the container to form the first layer. After curing, the wooden block was removed and the stiffer mixture was poured into the resulting cavity. Finally, a 2-mm layer of compliant mixture was poured over both parts to give it a uniform appearance. The Ecoflex part A, part B, thinner, and Silc Pig pigment were mixed in a ratio of 12:12:24:1 in the compliant region, and 12:12:0:1 in the stiff region. The diameter of the container was 98 mm, and the total thickness of the tissues was 20 mm. The stiffer tissue region was 14.5 mm wide.

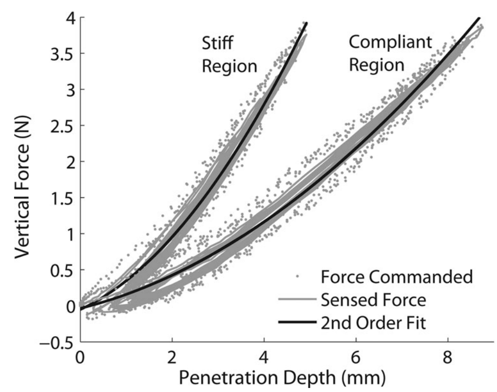


Fig. 6. Force–displacement relationships for a typical artificial tissue under several palpations. The stiff and compliant tissue can be distinguished using both the commanded and sensed force. A second-order polynomial was fit to the soft and stiff region ($R^2 = 0.99$ for both) before and after each experiment participant. The second-order fits from before and after each participant were statistically compared to ensure that tissue properties did not significantly change over the course of the experiment.

A new tissue model was used for each participant to avoid potential damage. The tissue model mechanical properties were monitored by performing a force–displacement analysis before and after each participant completed the experiment using the Phantom Premium probe with an attached ATI Nano17 force/torque sensor. An example set of tissue force–displacement curves is shown in Fig. 6. The average linearized stiffnesses of the stiffer and more compliant regions of tissue were 683 and 433 N/m, respectively. When combining the tissue stiffness with the proportional teleoperation controller stiffness, the resulting stiffnesses for stiffer and more compliant regions were 231 and 194 N/m, respectively, which result in a 19.4% difference between the regions. This difference is greater than the 14.1% Weber Fraction for kinesthetic stiffness discrimination determined by a previous experiment using the same skin-stretch device attached to a Phantom Premium rendering virtual stiffness [12].

The force–displacement curves shown in Fig. 6 indicate that the commanded slave force, $\vec{F}$, was similar to the force reading obtained by the ATI force sensor. Additionally, due to the sterilizability, durability, and cost concerns associated with using a force sensor, the commercial da Vinci surgical robot does not have force sensors on its tools. Therefore, we used $\vec{F}$ to estimate the forces acting on the slave robot, more closely replicating the force information that is available in the da Vinci surgical robot, which does not have direct force sensing. We performed several manual palpations and found an RMSE between $\vec{F}$ and the sensed force of 0.15 and 0.14 N for the stiff and compliant regions, respectively.

The artificial tissues were registered to the location of the Phantom Premium through a custom-made fixture with linear cutouts at angles 5°, 35°, 85°, 125°, and 155° from horizontal. These angles spanned the possible range of orientations, and also avoided the common angles of 0°, 45°, and 90° in order to avoid guessing. A beam was attached to the bottom of the artificial tissue container and slotted into the linear cutouts. The fixture and the beam on the artificial tissue container are shown in Fig. 5(c) and (d). This allowed the stiffer region of the artificial tissue to be oriented according to the described angles.

D. Experiment Procedures

Ten participants (three females) between the ages of 20 and 28 participated after giving informed consent, in a protocol approved by the Stanford University Institutional Review Board. Eight participants were self-reported right-hand dominant. The mean age was 25.2 with a standard deviation of 2.1 years. All participants were students and seven had previous experience with haptic devices.

In a teleoperated palpation task, participants were asked to find the location of a stiffer region bisecting the circular artificial tissue at an unknown orientation. They were told to complete the task as accurately and quickly as possible, and to focus on accuracy. Prior to the study, participants were told that they would be using five distinct feedback methods—skin stretch, force, reduced force, graphic, and vibration. They were shown how to hold the skin-stretch stylus [see Fig. 2(a)]. This grasp remained consistent across all feedback modalities. Participants moved the master manipulator with the attached skin-stretch stylus and palpated an artificial tissue at the remote side using the teleoperated slave manipulator. For each sensory substitution feedback method, it was explained that amplitude of the feedback was based on the vertical forces occurring between the slave robot and the environment.

Each experiment trial consisted of two phases. First, the participants freely palpated the artificial tissue and attempted to determine the orientation of the stiff region. Once the participants determined their estimated orientation of the stiffer tissue region, they pressed a key and then traced the orientation of the stiff region with the slave robot end effector while hovering slightly above the tissue. A linear fit to these data was used to determine the participants' estimated angle orientation of the stiffer region.

The experiment was performed in two consecutive days for each participant. On the first day, the participant completed a 15-trial training session followed by a test session of two blocks of 12 trials; each block was performed using a different feedback method. On the second day, the participant completed the remaining three 12-trial blocks. The training session consisted of 15 trials: three consecutive trials using each of the five feedback methods. The order of presentation of the feedback methods was determined through a balanced Latin Squares design in order to minimize the effects of the feedback presentation order. During the first of the three consecutive trials of each training block, the participants were told the orientation of the stiff region in advance. In the second trial, the participants were told the orientation of the stiff region only after they recorded their answer. In the third trial, the participants found the orientation of the stiff region, but were not provided with any orientation feedback. During the main experiment sessions, the participants were not given any orientation feedback.

After completing the training session, participants moved on to the first two trial blocks. The experiment blocks for each feedback method were ordered according to the same balanced Latin Squares method as the training. The first two trials of each 12-trial block were not analyzed. These two trials allowed the participant to refamiliarize themselves with using each feedback method. The remaining ten trials of each block consisted of two repetitions of each of the five possible tissue orientation angles arranged in pseudorandom order. After the participants recorded the estimated angle for the tissue orientation, they proceeded to the next trial of the block. There was at least a 1-min break between blocks. After the skin stretch and vibration blocks, breaks of 5 min were taken in order to prevent desensitization effects on the fingers for the next trial block [41]. Due to the loud noise produced by the manipulator motors when displaying vibration feedback, participants wore noise canceling headphones playing white noise during vibration trials to keep their focus on the haptic feedback. On the second day, the participants were given a 30-s free exploration time with each feedback method to refamiliarize themselves before proceeding with the remaining three feedback blocks.

After completing the experiment, the participants completed a postexperiment survey in which their personal details were collected including finger size and any finger abnormalities. They were given the opportunity to provide subjective feedback about each of the feedback methods.

E. Data Analysis

Time, position, commanded forces, filtered velocity of both the master and slave robot, and the displacement of the skin-stretch tactor were recorded at 100 Hz. We used the

TABLE II
FRIEDMAN TEST RESULTS FOR MEAN ABSOLUTE ANGLE ERROR, PEAK PALPATION FORCE, TRIAL COMPLETION TIME, AND PALPATION DISTANCE DIFFERENCE

Factor	χ^2 or p	Angle Error [degrees]	Palpation Force [N]	Completion Time [s]	Palpation Distance Difference [mm]
Feedback Method	χ^2	17.93	247.60	7.29	14.28
	p	0.0013	< 0.0001	0.1214	0.0065

MATLAB `findpeaks` function on the slave robot position data to identify the location of the local maxima of each penetration into the tissue and used the position as an indicative location for each palpation movement in further analysis. Additionally, by looking at the corresponding forces in these local maxima instances, the peak force of each palpation was determined.

To assess the efficacy of the feedback methods for the teleoperated palpation task, we examined the following metrics.

1) *Absolute angular error (degrees)*: Calculated for each trial from the difference between participants' estimated and actual orientation of the stiff region in the artificial tissue.
2) *Average peak palpation force (N)*: Calculated for each trial by averaging the forces that occurred at each local maxima of penetration into the tissue.
3) *Completion time (s)*: Calculated as the time in seconds that elapsed between participants starting free exploration and indicating that they were ready to provide their answer as to the stiff region orientation.
4) *Palpation proximity (mm)*: Calculated by evaluating the position of each palpation and determining the distance to the center line of the stiff tissue region during different parts of the trial.

After evaluating the normality of the data for each performance metric, we found that the individual subject and grouped subject data did not follow a parametric distribution. Thus, we performed the Friedman test on the effect of feedback modality on the various performance metrics. Statistical significance was determined at the $\alpha = 0.05$ level. Individual comparisons were performed using the Wilcoxon signed-rank test with Bonferroni correction for family-wise error due to multiple comparisons to determine the statistical significance of the differences in performance between the different feedback methods. For this reason, statistical significance for Bonferroni-corrected comparisons was determined at the $\alpha_{\mathrm{adj}} = 0.005$ level, and the associated p-values are shown as p_{B}. Statistical analysis was performed using custom-written MATLAB code.

To investigate the exploratory strategies of the participants, we examined the locations of the local penetration maxima. We used the locations to calculate the spatial distribution of the palpations at specific time windows, throughout the course of exploration during the trial. We calculated the distance of each palpation from the known position of the stiff region within the tissue by calculating their distance from the nearest point of the stiff region. We determined the proximity of the participants' interaction to the stiff region over the course of each trial and characterized how the different feedback methods affected the exploration strategies of the users.

For each individual trial, the stiffer region of the artificial tissue was set to one of the five possible orientations. Although

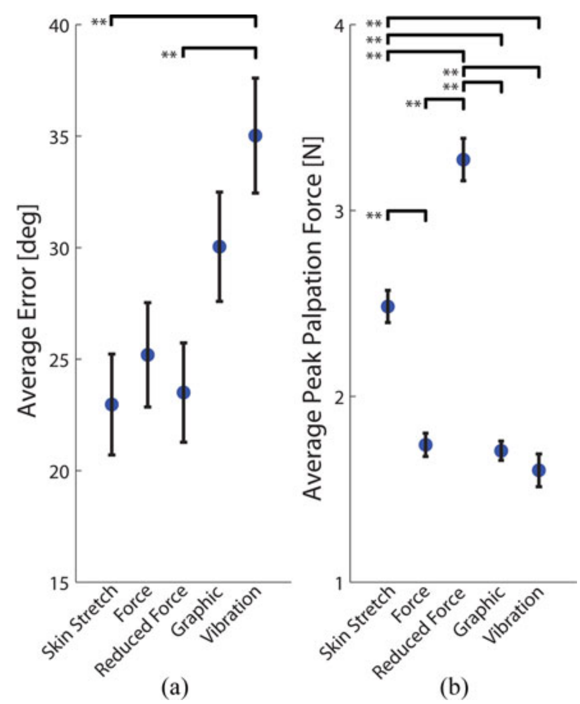


Fig. 7. (a) Absolute angle error was determined by taking the absolute value of the difference between the participants' indicated tissue orientation angle and the actual orientation angle. (b) Palpation force was calculated by recording the peak force that occurred during each palpation. The error bars indicate the standard error for each feedback method. Bonferroni-corrected post-hoc t-test comparisons are indicated where there was statistical significance at the 0.005 (*) and 0.001 (**) level.

angle error data could be calculated from raw data, in order to see a visual trend from the palpations, we rotated the palpation position data such that the stiff region was oriented in a horizontal manner across all trials. When observing a spatial distribution of the palpations, this allows us to see a visual trend over the course of the trials as the participants palpated in closer proximity to the stiff region.

IV. RESULTS

Statistical significance was determined at the $\alpha = 0.05$ level, and Friedman test results are shown in Table II. Individual, Bonferroni-corrected comparison significance was determined at the $\alpha_{\mathrm{adj}} = 0.005$ level, and associated p-values are indicated as p_{B}.

Participants were as successful using skin-stretch feedback to determine the orientation of the stiff region as when they used either force or reduced force feedback and were statistically significantly more accurate than when using vibration feedback. The Friedman test results indicate that there was a statistically significant effect of the feedback method on average angular

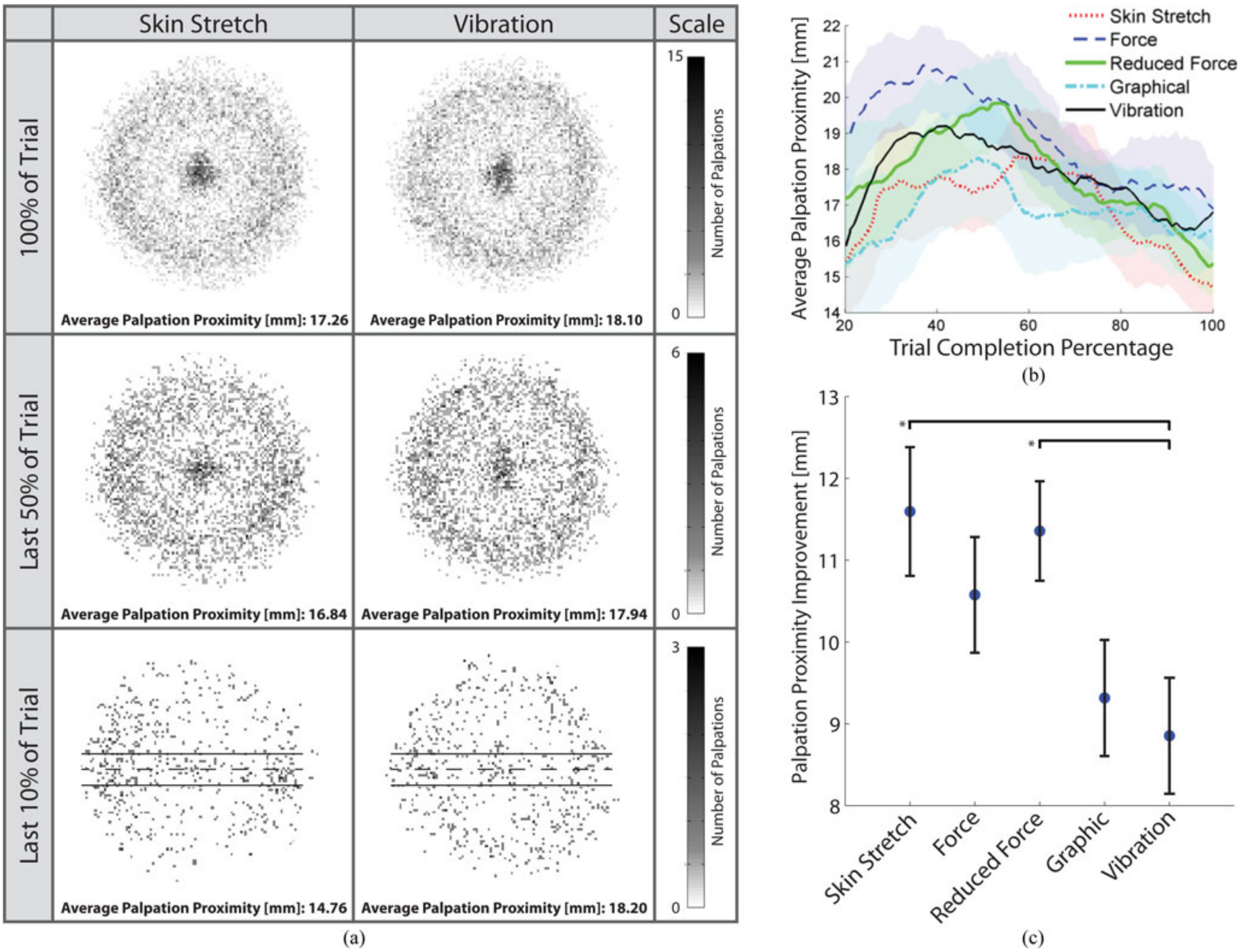


Fig. 8. (a) Spatial distribution of palpation locations for all participants. Data from each trial were visually rotated to account for the differing orientation of the stiff region from trial to trial. Also included is the average palpation proximity, which indicates the average distance of each palpation from the center line (dashed line) of the stiff tissue region. The region boundary is shown by the solid lines in the bottom palpation panes. (b) Average distance of each palpation from the stiff region plotted against trial completion percentage. This percentage was evaluated over a 20% sliding window of the total trial time. Average trial time was about 50 s for each feedback method. (c) Average improvement from furthest palpation proximity to end of trial for each feedback method. Bonferroni-corrected significance is indicated at the 0.005 level (*).

error ($\chi^2_{(4,N=100)} = 17.93; p = 0.0013$). Fig. 7(a) shows that participant angular error using skin stretch was not statistically different from force ($Z = 1.05; p_{\mathrm{B}} = 0.2943$), reduced force ($Z = 0.50; p_{\mathrm{B}} = 0.6157$), or graphical feedback ($Z = 2.18; p_{\mathrm{B}} = 0.0290$), but was statistically smaller than when using vibration feedback ($Z = 3.62; p_{\mathrm{B}} = 0.0003$). Participant angular errors using reduced force feedback were also statistically smaller than when using vibration feedback ($Z = 2.65; p_{\mathrm{B}} = 0.0009$). While participants using skin stretch did not perform statistically significantly better than when using graphical feedback, angular error using skin stretch was statistically significantly lower than when using vibration, while performance using graphical feedback was not better.

The statistical analysis also shows a significant effect of feedback method on the average peak palpation force ($\chi^2_{(4,N=100)} = 247.60; p < 0.0001$). Fig. 7(b) shows that palpation forces using skin stretch were statistically significantly lower than when using reduced force feedback ($Z = 4.92; p_{\mathrm{B}} < 0.0001$). Force, vibration, and graphical feedback methods all had statistically significantly lower peak palpation force than reduced force ($Z = 8.59; p_{\mathrm{B}} < 0.0001, Z = 8.67; p_{\mathrm{B}} < 0.0001,$ and $Z = 8.35; p_{\mathrm{B}} < 0.0001,$ respectively) and skin-stretch feedback ($Z = 5.36; p_{\mathrm{B}} < 0.0001, Z = 7.72; p_{\mathrm{B}} < 0.0001,$ and $Z = 7.77; p_{\mathrm{B}} < 0.0001,$ respectively). From this, we conclude that skin stretch was very effective in terms of angular error, but also caused larger interaction forces when compared to force feedback.

Statistical analysis did not reveal an effect of feedback method on trial completion time ($\chi^2_{(4,N=100)} = 7.29; p = 0.1214$); skin stretch is similar to force feedback in terms of task completion time. Average trial completion time was between 46 and 53 s for all feedback methods.

We also examined the spatial distribution of the palpations occurring over various subdivisions of trials. Fig. 8(a) displays examples of these distributions for skin stretch and vibration feedback, for 100% of the trial, last 50% of the trial, and last 10% of the trial. Fig. 8(b) uses these distributed palpation data and shows the average distance of each palpation from the stiff region of the artificial tissue as a function of the percentage of trial completed. The average distance was evaluated over a sliding 20% window of the total trial time. The Friedman test indicated that feedback method had a significant effect on the difference between peak and final distance of palpation from the stiff tissue region ($\chi^2_{(4,N=100)} = 14.28; p = 0.0065$). The Wilcoxon signed-rank test shows that participants using skin-stretch feedback and reduced gain force feedback exhibited a larger difference between peak and final distance of palpation than participants using vibration feedback ($Z = 2.81; p_{\mathrm{B}} < 0.0050$ and $Z = 3.05; p_{\mathrm{B}} = 0.0023$, respectively). The difference between the peak and final distance of palpations from the stiff region can be seen in Fig. 8(c). Skin-stretch feedback exhibits the largest difference between peak and final palpation distance out of all the feedback methods, while vibration feedback exhibits the smallest difference between peak and final palpation distance.

To ensure that the artificial tissue properties did not deteriorate during the experiment, we fit a second-order polynomial curve to the force–displacement data for all of the artificial tissues. The R^2 values averaged 0.99 for the soft and stiff regions, before and after the experiment. The polynomial fit coefficients from before and after each participant's experiment were statistically compared using a paired t-test. There was no statistically significant difference in the second-order, first-order, or constant values for the compliant region or for the stiff region.

V. Discussion

A. Detection Angular Error

When performing a teleoperated palpation task, participants were able to determine the orientation of a stiff region in surrounding tissue using skin stretch as well as when using force feedback and more accurately than when using vibration feedback, a common force sensory substitute. Additionally, while the current experiment provided 1-DoF skin-stretch feedback, skin stretch provides the ability to convey tactile force information in multiple directions [32], [42], a capability frequently lacking in sensory substituted vibration systems [20], [43]. Furthermore, tactile skin stretch conveys the environment force information through the sense of touch, the sense for which it was originally intended. Meli *et al.* [25] found that a tactile "sensory subtraction" device used to communicate grip force information improved performance over both graphic and auditory feedback in a virtual ring-and-peg task.

While participant angular error using skin stretch in our experiment was not statistically better than when using graphical feedback, in many teleoperated scenarios, such as surgical teleoperation, the users are overly stimulated with visual information from the environment. Ungrounded tactile feedback methods, such as skin stretch and sensory subtraction, appear to be promising alternatives to other sensory substitution methods for replacing force feedback in scenarios where stability concerns arise.

The properties of the artificial tissues were varied during pilot testing in order to adjust the difficulty of the task and show differences in capability when using the various feedback methods. The orientation estimation is not, in itself, a surgical task. Rather, it provided a single metric of performance. Additionally, the stiff tissue region was 14.5 mm wide. This means that it is possible for the participant to have an error of 9.5° and still be palpating over the tissue region at the edge of the container. For these reasons, the relative angular error performance is of more importance than the absolute error.

B. Palpation Force

Despite similar angular error and trial completion time, there was an increase in palpation force when using skin stretch and reduced force feedback in place of standard force feedback. However, although skin stretch increased interaction forces when compared with force feedback, skin stretch displayed a decrease in palpation forces when compared to reduced force feedback. This finding is consistent with previously reported results indicating that interaction forces increased as force feedback gains were lowered in a teleoperated blunt dissection task [4]. Part of this can be attributed to the decrease or absence of a grounded kinesthetic force that naturally pushes the master robot out of the tissue when using reduced force feedback or other tactile feedback methods. Indeed, high-gain force feedback would be ideal in many teleoperation scenarios, but there exist many applications, such as robot-assisted minimally invasive surgery [44], for which the risk of instability associated with high-quality force feedback [15] prevents its implementation. In these scenarios, tactile feedback, such as skin stretch, has the potential to reduce interaction forces, and possibly tissue trauma, when compared to lower quality force feedback, such as the reduced force feedback in this experiment.

It may be possible to further reduce the interaction forces exhibited using skin stretch by appropriately adjusting the skin-stretch ratio. A larger skin-stretch displacement per unit force could create a stronger perception of stiffness [30] as well as tactor saturation at lower forces and possibly encourage a lighter touch. Future studies are needed to examine the effect of such adjustment of the skin-stretch gain on the overall task performance as well as the magnitude of interaction forces.

While vibration feedback resulted in lower palpation force than skin stretch in this experiment, participants using vibration feedback were not able to accurately locate the stiff region. Therefore, while palpation forces were reduced, so was the ability to obtain information about the stiffness characteristics of the environment. According to postexperiment surveys, many participants reported the vibration feedback a nuisance and actually kept their interaction forces limited in order to avoid experiencing strong vibration feedback. The virtual bar presented in the graphical feedback case also limited interaction forces. Since the bar had a displayed maximum level, participants were inclined to keep their interaction within the presented range. This resulted in fewer interactions with especially high force. It is possible that decreasing the range of the graphical feedback may cause participants to use even less palpation force, but the decrease in the perceivable force range may have made it more difficult for participants do distinguish the differences between the tissue regions.

Interestingly, despite the clear difference in angular error and peak palpation force, there was no statistically significant difference in task completion time between feedback methods, even though the participants were instructed to prioritize accuracy over completion time.

C. Palpation Strategies

Our analysis of the average distance of palpations from the stiff region of the artificial tissue gives insights about palpation strategies using different feedback modalities. Since participants were told before the experiment that the stiff tissue region bisected the circular tissue, most chose to start their trials by palpating the center of the circle, a region in which they were sure they would feel the stiffer tissue. We can see that this region is very dense with palpations in Fig. 8(a). Then, participants generally moved their exploration away from the center in an arbitrary direction to make comparisons with the surrounding softer tissue and determine the orientation of the stiff region. During the postexperiment survey, many participants indicated that this was, in fact, the basis of their exploration strategy. From Fig. 8(b), exploration with each feedback method started with a relatively low palpation distance from the stiff region as participants were still located near the center of the circular tissue. This distance then increased for all feedback methods as participants explored the surrounding area. As the trial concludes, participants' palpations begin to move back toward the stiff region as they refine their estimation of the stiff region orientation. Additionally, participant palpations were closer to the stiff region using skin-stretch feedback than any other feedback method at the end of the trial.

Since the initial exploration out from the center of the circle was generally arbitrary, we chose to compare the peak and final palpation distance from the stiff region for each feedback method to determine the progress towards the stiff region made over the course of the trial. By analyzing the difference between the peak and final proximities of palpation from the stiff region, we determined that skin stretch exhibited the most improvement in determining the location of the region. Graphic and vibration feedback methods exhibited the smallest difference between peak and minimum palpation distance. Over the course of the trial, participants using these feedback methods palpated in a manner that trended less toward the stiff region of interest. This could be an indication that participants were less deliberate in their exploration using these methods and, therefore, used a less direct and more randomly distributed approach.

D. Sensory Augmentation

This study demonstrates the versatility of the skin-stretch device for users with various finger sizes. Based on a prior study that indicated a reduced effect of skin-stretch feedback on participants with index fingers smaller than 14 mm wide when using the 12-mm aperture [30], a second 10-mm aperture was used for participants with reduced finger size. Two participants used the smaller aperture, and both had performance characteristics typical of the other participants. All participants were able to discern skin-stretch cues for determining the orientation of the stiff tissue region. We found some interparticipant variability, which is consistent with prior research on skin stiffness [45], tactile sensitivity [46], and mechanical stimulus neurological threshold [47]. The average angular error for individual participants ranged from 10.6° to 30.0°, with a standard deviation of 7.0°. This variation between participants was less than that when using force feedback, which resulted in a standard deviation of 11.3°. However, the standard deviation for trial completion time using skin-stretch feedback was 20.9 s, higher than the 14.6 s when using force feedback. The additional variance in trial completion time when using skin-stretch feedback may have resulted from the different rates at which participants adapted to using skin-stretch feedback for force sensory substitution.

One possible extension for skin-stretch feedback in teleoperation is augmentation with force feedback. While it may not be possible to stably render high gain force feedback in some situations, reducing the feedback gains can improve stability while sacrificing the quality of the force interaction. In this study, skin-stretch feedback reduced interaction forces when compared with reduced force feedback. Integrating the two feedback methods may allow decreased interaction forces while retaining some of the natural boundary created by kinesthetic force feedback. In [30], it was shown that the perceived stiffness of a virtual environment can be increased by rendering skin-stretch feedback in conjunction with force feedback. In a teleoperated situation, this augmentation could be used to improve the fidelity of low gain force feedback.

VI. Conclusion

We have compared various forms of force information feedback in a teleoperated palpation task. Skin-stretch feedback, which has the ability to convey force information through cutaneous tactile cues, was compared to force feedback and other traditional sensory substitutes such as graphic and vibration feedback.

We found that skin-stretch feedback may be a suitable alternative for vibration or reduced gain force feedback when performing teleoperated stiffness discrimination. First, angular error was statistically significantly less when using skin-stretch feedback than when using vibration feedback. Also, peak palpation force was significantly less when using skin-stretch feedback than when using reduced gain force feedback, which had similar angular error performance. Finally, trial completion time was not affected by the feedback method.

Skin-stretch feedback may also be a suitable replacement for force feedback or graphical feedback when peak interaction force is not critical. The average participant angular error was similar when using skin-stretch feedback and force feedback, which we generally consider the "gold standard" for stiffness discrimination. However, peak palpation force was statistically significantly increased when using skin-stretch feedback rather than force feedback or graphical feedback. Despite that disadvantage, skin-stretch feedback does not require obscuring part of the user's view, which may be important in visually demanding scenarios. Further study is needed to determine what specific tasks are most suitable for this alternative feedback method, and how skin-stretch feedback gains might be modified to lower peak interaction forces.

Providing skin-stretch feedback is straightforward, because a skin-stretch feedback device can be attached to the master manipulator of existing commercial teleoperation systems. While 1-DoF skin-stretch feedback was sufficient for our simple palpation task, more complicated tasks may require feedback with higher degrees of freedom. Multi-DoF skin-stretch devices [34] may be used to convey force information in these situations. Applications such as surgical teleoperation will be investigated in order to determine the effectiveness of 3-DoF cutaneous skin-stretch feedback in more complex and potentially clinically relevant scenarios.

References

[1] S. J. Lederman and R. L. Klatzky, "Sensing and displaying spatially distributed fingertip forces in haptic interfaces for teleoperator and virtual environment systems," *Presence: Teleoperator Virtual Environ.*, vol. 8, no. 1, pp. 86–103, 1999.

[2] C. R. Wagner, R. D. Howe, and N. Stylopoulos, "The role of force feedback in surgery: Analysis of blunt dissection," in *Proc. 10th Symp. Haptic Interfaces Virtual Environ. Teleoperator Syst.*, 2002, pp. 68–74.

[3] G. Tholey, J. P. Desai, and A. E. Castellanos, "Force feedback plays a significant role in minimally invasive surgery," *Ann. Surg.*, vol. 241, no. 1, pp. 102–109, 2005.

[4] C. Wagner, N. Stylopoulos, P. G. Jackson, and R. Howe, "The benefit of force feedback in surgery: Examination of blunt dissection," *Presence: Teleoperators Virtual Environ.*, vol. 16, no. 3, pp. 252–2620, 2007.

[5] J. C. Gwilliam, M. Mahvash, B. Vagvolgyi, A. Vacharat, D. D. Yuh, and A. M. Okamura, "Effects of haptic and graphical force feedback on teleoperated palpation," in *Proc. IEEE Int. Conf. Robot. Autom.*, 2009, pp. 677–682.

[6] M. Kitagawa, D. Dokko, A. M. Okamura, and D. D. Yuh, "Effect of sensory substitution on suture-manipulation forces for robotic surgical systems," *J. Thoracic Cardiovascular Surg.*, vol. 129, no. 1, pp. 151–158, 2005.

[7] C. E. Reiley, T. Akinbiyi, D. Burschka, D. C. Chang, A. M. Okamura, and D. D. Yuh, "Effects of visual force feedback on robot-assisted surgical task performance," *J. Thoracic Cardiovascular Surg.*, vol. 135, no. 1, pp. 196–202, 2008.

[8] M. J. Massimino, "Improved force perception through sensory substitution," *Control Eng. Practice*, vol. 3, pp. 215–222, 1995.

[9] T. Debus and T. Jang, "Multi-channel vibrotactile display for teleoperated assembly," *Int. J. Control, Autom., Syst.*, vol 2, pp. 390–397, 2004.

[10] M. A. Srinivasan and R. H. LaMotte, "Tactual discrimination of softness," *J. Neurophysiol.*, vol. 73, no. 1, pp. 88–101, 1995.

[11] W. M. B. Tiest and A. M. L. Kappers, "Cues for haptic perception of compliance," *IEEE Trans. Haptics*, vol. 2, no. 4, pp. 189–199, Oct.–Dec. 2009.

[12] S. B. Schorr, Z. F. Quek, R. Y. Romano, I. Nisky, W. R. Provancher, and A. M. Okamura, "Sensory substitution via cutaneous skin stretch feedback," in *Proc. IEEE Int. Conf. Robot. Autom.*, May 2013, pp. 2341–2346.

[13] G. Tholey, J. P. Desai, and A. E. Castellanos, "Evaluation of a laparoscopic grasper with force feedback," *Surgical Endoscopy*, vol. 18, no. 5, pp. 863–867, 2004.

[14] P. Puangmali, K. Althoefer, L. D. Seneviratne, D. Murphy, and P. Dasgupta, "State-of-the-art in force and tactile sensing for minimally invasive surgery," *IEEE Sensors J.*, vol. 8, no. 4, pp. 371–381, Apr. 2008.

[15] D. Lawrence, "Stability and transparency in bilateral teleoperation," *IEEE Trans. Robot. Autom.*, vol. 9, no. 5, pp. 624–637, Oct. 1993.

[16] N. Cutler, M. Balicki, M. Finkelstein, J. Wang, P. Gehlbach, J. McGready, I. Iordachita, R. Taylor, and J. T. Handa, "Auditory force feedback substitution improves surgical precision during simulated ophthalmic surgery," *Investigative Ophthalmol. Visual Sci.*, vol. 54, no. 2, pp. 1316–1324, 2013.

[17] P. Galambos, "Vibrotactile feedback for haptics and telemanipulation: Survey, concept and experiment," *Acta Polytechnica Hungarica*, vol. 9, pp. 41–65, 2012.

[18] Y. Muramatsu, M. Niitsuma, and T. Thomessen, "Perception of tactile sensation using vibrotactile glove interface," in *Proc. IEEE Int. Conf. Cognitive Infocommun.*, 2012, pp. 621–626.

[19] R. Schoonmaker and C. Cao, "Vibrotactile force feedback system for minimally invasive surgical procedures," in *Proc. IEEE Int. Conf. Syst., Man Cybern.*, 2006, vol. 3, pp. 2464–2469.

[20] W. McMahan, J. Gewirtz, D. Standish, P. Martin, J. Kunkel, M. Lilavois, A. Wedmid, D. I. Lee, and K. J. Kuchenbecker, "Tool contact acceleration feedback for telerobotic surgery," *IEEE Trans. Haptics*, vol. 4, no. 3, pp. 210–220, May/Jun. 2011.

[21] L. Jones and N. B. Sarter, "Tactile displays: Guidance for their design and application," *Human Factors: J. Human Factors Ergonomics Soc.*, vol. 50, no. 1, pp. 90–111, 2008.

[22] K. Kim and J. Colgate, "Haptic feedback enhances grip force control of sEMG-controlled prosthetic hands in targeted reinnervation amputees," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 6, pp. 798–805, Nov. 2012.

[23] K. Minamizawa, D. Prattichizzo, and S. Tachi, "Simplified design of haptic display by extending one-point kinesthetic feedback to multipoint tactile feedback," in *Proc. IEEE Haptics Symp.*, 2010, pp. 257–260.

[24] D. Prattichizzo, C. Pacchierotti, and G. Rosati, "Cutaneous force feedback as a sensory subtraction technique in haptics," *IEEE Trans. Haptics*, vol. 5, no. 4, pp. 289–300, Fourth Quarter 2012.

[25] L. Meli, C. Pacchierotti, and D. Prattichizzo, "Sensory subtraction in robot-assisted surgery: Fingertip skin deformation feedback to ensure safety and improve transparency in bimanual haptic interaction," vol. 61, no. 4, pp. 1318–1327, Apr. 2014.

[26] L. Ben Porquis, D. Maemori, N. Nagaya, M. Konyo, and S. Tadokoro, "Presenting virtual stiffness by modulating the perceived force profile with suction pressure," in *Proc. IEEE Haptics Symp.*, Feb. 2014, pp. 289–294.

[27] J. Biggs and M. Srinivasan,, "Tangential versus normal displacements of skin: relative effectiveness for producing tactile sensations," in *Proc. 10th Symp. Haptic Interfaces Virtual Environ. Teleoperator Syst.*, 2002, pp. 121–128.

[28] K. Bark, J. Wheeler, P. Shull, J. Savall, and M. Cutkosky, "Rotational skin stretch feedback: A wearable haptic display for motion," *IEEE Trans. Haptics*, vol. 3, no. 3, pp. 166–176, Jul.–Sep. 2010.

[29] J. Wheeler, K. Bark, J. Savall, and M. Cutkosky, "Investigation of rotational skin stretch for proprioceptive feedback with application to myoelectric systems," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 18, no. 1, pp. 58–66, Feb. 2010.

[30] Z. F. Quek, S. B. Schorr, I. Nisky, A. M. Okamura, and W. R. Provancher, "Sensory augmentation of stiffness using fingerpad skin stretch," in *Proc. IEEE World Haptics Conf.*, 2013, pp. 467–472.

[31] W. R. Provancher and N. D. Sylvester, "Fingerpad skin stretch increases the perception of virtual friction," *IEEE Trans. Haptics*, vol. 2, no. 4, pp. 212–223, Oct.–Dec. 2009.

[32] B. Gleeson, S. Horschel, and W. Provancher, "Perception of direction for applied tangential skin displacement: Effects of speed, displacement, and repetition," *IEEE Trans. Haptics*, vol. 3, no. 3, pp. 177–188, Jul.–Sep. 2010.

[33] A. L. Guinan, N. C. Hornbaker, M. N. Montandon, A. J. Doxon, and W. R. Provancher, "Back-to-back skin stretch feedback for communicating five degree-of-freedom direction cues," in *Proc. IEEE World Haptics Conf.*, 2013, pp. 13–18.

[34] Z. F. Quek, S. B. Schorr, I. Nisky, A. M. Okamura, and W. R. Provancher, "Sensory substitution using 3-degree-of-freedom tangential and normal skin deformation feedback," in *Proc. IEEE Haptics Symp.*, 2014, pp. 27–33.

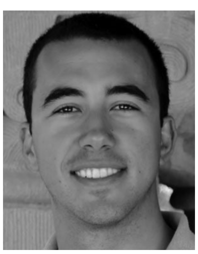
[35] M. O'Malley and M. Goldfarb, "The effect of virtual surface stiffness on the haptic perception of detail," *IEEE/ASME Trans. Mechatronics*, vol. 9, no. 2, pp. 448–454, Jun. 2004.

[36] J. D. Brown, J. Rosen, L. Chang, M. N. Sinanan, and B. Hannaford, "Quantifying surgeon grasping mechanics in laparoscopy using the blue dragon system," *Stud. Health Technol. Informat.*, vol. 98, pp. 34–36, 2004.

[37] J. Marescaux, J. Leroy, M. Gagner, F. Rubino, D. Mutter, M. Vix, S. E. Butner, and M. K. Smith, "Transcontinental robot-assisted remote telesurgery: Feasibility and potential applications," *Ann. Surg.*, vol. 235, no. 4, pp. 487–492, 2002.

[38] C. E. Stepp and Y. Matsuoka, "Vibrotactile sensory substitution for object manipulation: Amplitude versus pulse train frequency modulation," *IEEE Trans. Neural Syst. Rehabil. Eng.*, vol. 20, no. 1, pp. 31–37, Jan. 2012.

[39] J. Bell, S. Bolanowski, and M. H. Holmes, "The structure and function of pacinian corpuscles: A review," *Prog. Neurobiol.*, vol. 42, no. 1, pp. 79–128, 1994.
[40] N. Landin, J. Romano, W. McMahan, and K. Kuchenbecker, "Dimensional reduction of high-frequency accelerations for haptic rendering," in *Proc. EuroHaptics*, Jul. 2010, pp. 79–86.
[41] N. E. Hochreiter, M. J. Jewell, L. Barber, and P. Brown, "Effect of vibration on tactile sensitivity," *J. Am. Phys. Therapy Assoc.*, vol. 63, no. 6, pp. 934–937, 1983.
[42] Z. Quek, S. Schorr, I. Nisky, W. Provancher, and A. M. Okamura, "Sensory substitution using 3-degree-of-freedom tangential and normal skin deformation feedback," in *Proc. Haptic Symp.*, 2014, pp. 27–33.
[43] J. T. Dennerlein, P. A. Millman, and R. D. Howe, "Vibrotactile feedback for industrial telemanipulators," in *Proc. ASME Dynam. Syst. Control Division*, vol. 61, 1997, pp. 189–195.
[44] A. M. Okamura, "Haptic feedback in robot-assisted minimally invasive surgery," *Current Opinion Urology*, vol. 19, no. 1, pp. 102–107, 2009.
[45] Q. Wang and V. Hayward, "In vivo biomechanics of the fingerpad skin under local tangential traction," *J. Biomechanics*, vol. 6, pp. 27–32, 1983.
[46] D. Loesch and N. Martin, "Finger ridge patterns and tactile sensitivity," *Ann. Human Biol.*, vol. 40, pp. 851–860, 2007.
[47] R. Johansson and A. Vallbo, "Tactile sensory coding in the glabrous skin of the human hand," *Trends Neurosci.*, vol. 11, no. 2, pp. 113–124, 1984.



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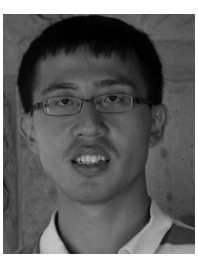

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