Force Controlled and Teleoperated Endoscopic Grasper for Minimally Invasive Surgery—Experimental Performance Evaluation

Jacob Rosen, Blake Hannaford,* *Member, IEEE*, Mark P. MacFarlane, and Mika N. Sinanan

***Abstract*—Minimally invasive surgery generates new user interfaces which create visual and haptic distortion when compared to traditional surgery. In order to regain the tactile and kinesthetic information that is lost, a computerized force feedback endoscopic surgical grasper (FREG) was developed with computer control and a haptic user interface. The system uses standard unmodified grasper shafts and tips. The FREG can control grasping forces either by surgeon teleoperation control, or under software control. The FREG performance was evaluated using an automated palpation function (programmed series of compressions) in which the grasper measures mechanical properties of the grasped materials. The material parameters obtained from measurements showed the ability of the FREG to discriminate between different types of normal soft tissues (small bowel, lung, spleen, liver, colon, and stomach) and different kinds of artificial soft tissue replication materials (latex/silicone) for simulation purposes. In addition, subjective tests of ranking stiffness of silicone materials using the FREG teleoperation mode showed significant improvement in the performance compared to the standard endoscopic grasper. Moreover, the FREG performance was closer to the performance of the human hand than the standard endoscopic grasper. The FREG as a tool incorporating the force feedback teleoperation technology may provide the basis for application in telesurgery, clinical endoscopic surgery, surgical training, and research.**

***Index Terms*— Endoscopy, force feedback, grasper, haptics, minimally invasive surgery (MIS), soft tissues, surgical simulation, teleoperation.**

I. Introduction

MINIMALLY invasive surgery (MIS) is a relatively new technique in which a surgeon operates with specially designed surgical tools through access ports requiring incisions of about 1 cm in size. This limits the surgical trauma to tissues, decreases the pain that the patients experience and results in a significant shortened recovery period. MIS technology generates two new user interfaces: 1) the monitor which gives the surgeon a two-dimensional (2-D) visual feedback of the internal anatomy and 2) the MIS surgical instruments which are inserted through the access ports. The instruments enable the surgeon to manipulate the internal organs by using finger loops outside the patient's body linked to the tool tip via a long tube/shaft including internal mechanism. Despite the benefits of MIS, this technique has some disadvantages due to the two new interfaces. The monitor, as a visual interface, reduces the surgeon's perception from a three-dimensional (3-D) to a 2-D view of the anatomy. Furthermore, the MIS instruments limit the surgeon's ability to gain the diagnostic information about the tissue being manipulated, as opposed to traditional surgery in which the surgeon examines the tissue by touching it directly with the hands. Moreover, due to the internal friction and backlash of the mechanism of MIS instruments the ability to perceive information by palpating tissues and organs is significantly reduced. Two components of the palpation information are tactile and kinesthetic information. Their combined use is referred to as haptic perception.

Previous work has explored manual driven material palpation with a regular instrumented endoscopic grasper [1], and automated palpation with an external robot [2]. Tactile sensors have been applied to endoscopic graspers which are coupled to tactile displays [3], [4]. These systems aim to enable the surgeon to discriminate textural or time-varying features of the tissue via endoscopic tools. An instrumented Babcock grasper, which measures forces and torques at the tool-tissue interface, has been reported, but does not measure or control grasping force [5]. In addition, teleoperation robots [6]–[8] and simulators implementing virtual reality with a force feedback haptic device have been developed [8], [9].

The importance of haptic feedback to safely perform surgery is unclear. Psychophysical experiments investigating the ability of humans to tactually discriminate the softness of objects showed that whereas tactile information was sufficient for rubber specimens both tactile and kinesthetic information was found necessary for spring cells [10]. Although color, texture, and visible aspects of tissue deformation in the surgical field convey important anatomic information, palpation may be critical in identifying otherwise obscure tissue planes, arterial pulsations, and regions of tissue thickening that may signify pathologies such as infection or cancer. Safe tissue handling requires tissue manipulation that is both secure and nondamaging. Much of the art of surgery and the implicit learning curve for traditional surgical technique depend on training to refine and educate the sense of touch. Training for endoscopic surgery is even more difficult because of the

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remote nature of videoendoscopic tissue manipulation. Recent literature emphasizes the importance of tactile feedback for accurate targeting of primary [11]–[13] and metastatic cancer [14]–[17] and identifying therapeutic margins for curative resection [12], [18], [19].

The loss of palpation for localization may seriously limit the efficacy and safety of minimally invasive treatment in some operative fields [17]. The present study aims to develop and characterize a grasper capable of restoring a degree of kinesthetic information to the surgeon about the tissue being grasped. The following goals were laid out: 1) improve the ability of the endoscopic surgeon to feel mechanical properties of tissues such as compliance, 2) make minimal changes to the form and function of existing surgical graspers to reduce cost, complexity, and certification difficulties (i.e., avoid adding sensors and wiring to the tool tip), and 3) take advantage of the declining cost of computer control.

This study is focused on two aspects of using the FREG: 1) objective aspect in which the FREG was used in an automatic mode for testing mechanical characteristics of soft tissue and viscoelastic material replications and 2) subjective aspect (psychophysic) in which the FREG was operated in bilateral force feedback mode examining the performance of test operators ranking materials according to their stiffness and comparing to the performance achieved by using a regular grasper and a gloved hand.

II. Methods

A. Force Feedback Endoscopic Grasper (FREG)

1) System Overview: The FREG (Fig. 1) incorporates teleoperation technology into an existing, reusable, endoscopic grasper (Fig. 2) for minimally invasive surgery [20]. The FREG system includes two subsystems. The master and the slave each consist of an actuator and a position encoder. The tool tip, pull/push rod and tube, is mounted on the slave subsystem that is inserted into the patient's body through an access port. The proximal end of the instrument tube is clamped to a supporting post of the slave. The pull/push rod operating the tool tip (jaws) is linked to the electromagnetic actuator via a ball and socket joint. The two finger loops (user interface) of the grasper are mounted on the master subsystem. The distal finger loop is connected to an actuator/encoder pair identical to those on the tool shaft enabling the surgeon to control the tool tip.

To increase sensing resolution, the encoder wheels are connected to the actuation axes via pulleys and a Kevlar drive belt having a multiplication ratio of 1 : 3.6. As a consequence, both master and slave position sensors have 1400 quadrature position counts over the full 0.6 rad (34.4°) motion range. The FREG actuators are flat coil actuators modified from hard disk drive head positioning actuators. Hard disk drive head actuators have many advantages for precision robotics and force feedback devices [21]. Actuators taken directly from 5.25 in (133 mm) hard drives with a maximum torque of 0.1 Nm at 2.0 A (based on steady-state coil temperature of 93 °C) did not produce convincing subjective grasping sensations. The actuator magnets were replaced with custom-made Nd–Fe–B magnets having approximately triple the energy product of the Al–Ni–Co magnets used in the disk drive actuator [22]. The coil and bearing assembly was retained. To realize the full flux increase from the new magnets, new frames were built from high permeability iron to prevent backing iron saturation. The new actuator magnets and frames increased the torque output to 0.3 Nm, but preserved the desirable qualities of low torque ripple, low friction, and low back-driving inertia.

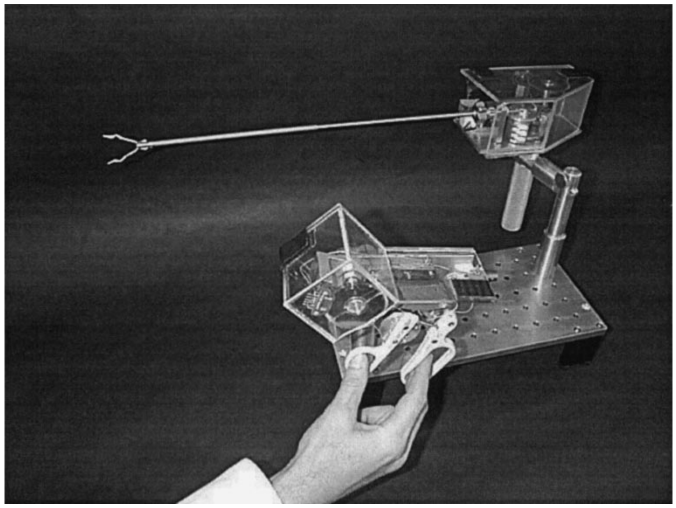
The laparoscopic instrument used in these experiments is a stainless steel Babcock grasper (Carl Storz Inc., model No. 30 420 BL) with a square jaw grasping surface area measuring 9.4×8.5 mm (Fig. 2). The tool shaft is 5 mm in diameter and 38-cm long from the proximal attachment to the instrument tip. The shaft and mount allow 360° rotation of the tool about its long axis. This system allows easy change of shaft length, diameter, and tool tip conformations. Laparoscopic tools compatible with the mounting system are readily available from various manufacturers.

2) Control: The control system supports two modes of operation: 1) bilateral force feedback-teleoperation and 2) programmed automatic grasping (palpation) operation for tissue characterization (Fig. 3). Proportional-derivative (PD) controllers were designed for both the master and slave using a linear dynamic model of the device and conventional control techniques [24]. Integral feedback is not desirable in position error based force feedback control because it creates a time-varying force feedback under conditions of steady-state contact. The force feedback controller is based on the well-known bilateral, position error based, teleoperation system [25]. In this design, the measured position of each side serves as the reference position input for the other.

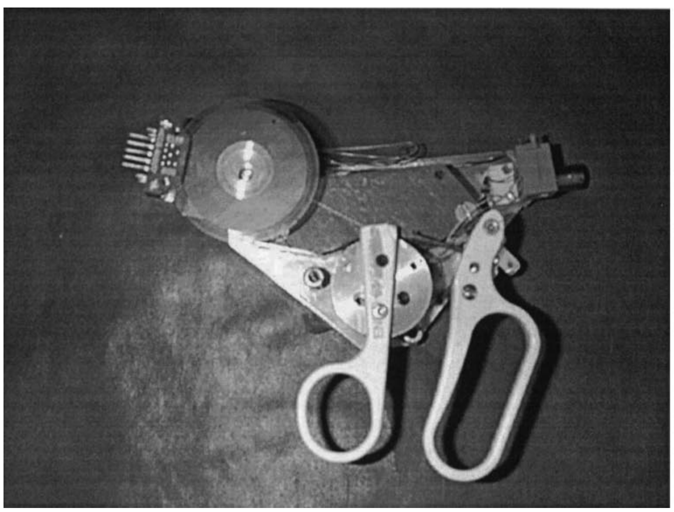
A desirable quality of a force feedback system is a high effective stiffness between master and slave sides. In the position error based architecture, this requirement can be translated into the need for a high value of the proportional feedback gain, K_p [25]. An additional controller design constraint is introduced from the actuator limit of 0.3 Nm maximum torque. Experience shows that users feel a subjective loss of contact sensations when a force feedback device saturates at its maximum force output. There is, thus, a tradeoff between K_p and the deflection at which saturation occurs. For high values of K_p, the user will feel high effective stiffness, but saturation will occur at relatively smaller position errors. The K_p value was selected such that the position error corresponding to the actuator saturation point is set at one quarter of the motion travel range (1)

$$K_p = \frac{I_{\max}}{0.16} = 12.6. \tag{1}$$

The remaining parameter K_d was determined by placing the dominant closed loop pole for an 8-ms settling time constant and a damping ratio of 0.5. For the slave, this design method resulted in an unstable controller, possibly because of backlash in its mechanism. An acceptable controller was recomputed with a lower initial K_p value. The resulting gains are given in Table I.



(a)



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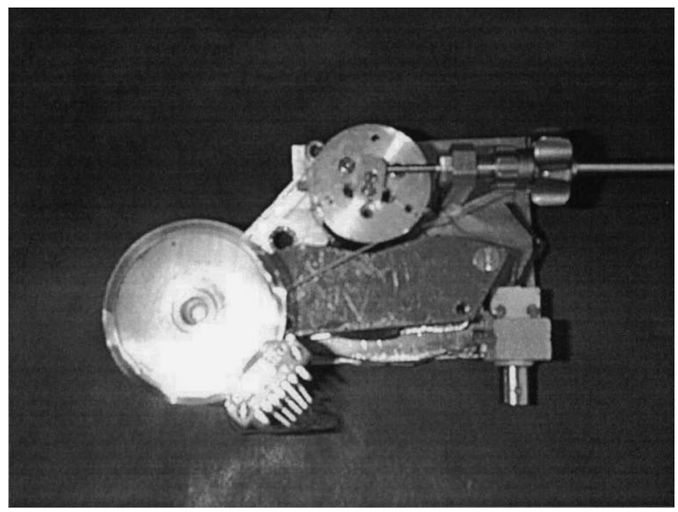

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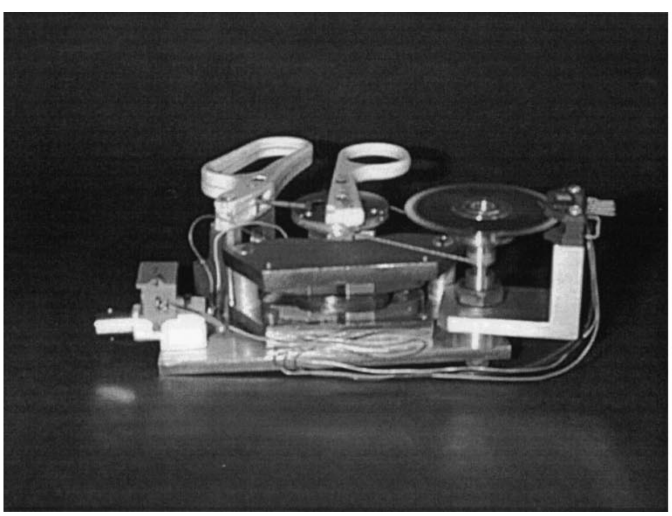

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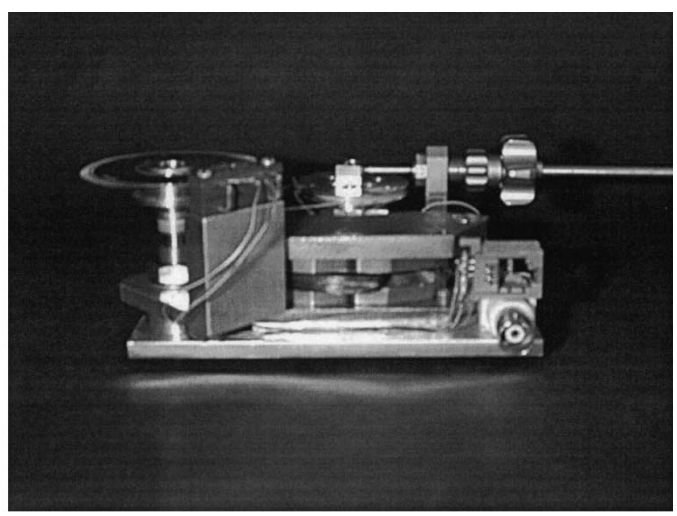


(e)

Fig. 1. FREG. (a) System overview, (b) master—front view, (c) master—top view, (d) slave—front view, and (e) slave—top view.

3) Mechanism Analysis: The endoscopic grasper has a unique mechanism to transfer the position and moments applied by the surgeon on the finger loops to the tool at the tip which is grasping the tissue. The following static analysis takes into account only the grasper geometry and not the system dynamics and its friction. Fig. 4 shows a scheme

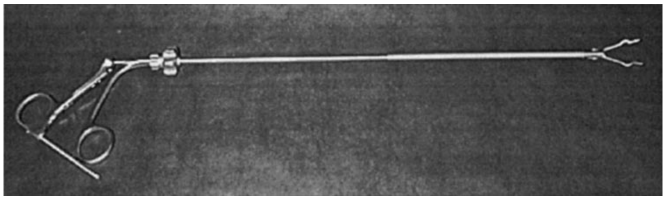


Fig. 2. Endoscopic grasper with a Babcock tip—Carl Storz model No. 30 420 BL.

of the endoscopic grasper internal mechanism in two typical positions: 1) tool tip jaws closed (reference position) and 2) tool tip jaws in an intermediate position. Given the geometry defined in Fig. 4, the finger loop angle (θ) as a function of the tool tip jaws angle (φ) is defined by (2). The transfer function between the moment applied on the finger loops relative to their joint (T_H), and the moment developed at the tool tip jaws relative to their joint (T_T) is defined by (3). Using the numerical geometry dimensions of the Babcock grasper (Carl Storz Inc., model No. 30 420 BL) the transfer functions (2), (3) are plotted in Fig. 5(a) and (b)

$$\theta = \tan^{-1}\left\{\frac{1}{R}\left[L_0 - L_1\sqrt{1-\left(\frac{L_2}{L_1}\sin\left(\beta_0+\frac{\varphi}{2}\right)\right)} - L_2\cos\left(\beta_0+\frac{\varphi}{2}\right)\right]\right\} \tag{2}$$

$$\frac{T_L}{T_H} = \frac{L_2\cos\theta}{2R}\left[\sin\left(\beta_0+\frac{\varphi}{2}\right) + \frac{\frac{L_2}{2L_1}\sin\left(\beta_0+\frac{\varphi}{2}\right)}{\sqrt{1-\frac{L_2}{L_1}\sin\left(\beta_0+\frac{\varphi}{2}\right)}}\right]. \tag{3}$$

An ideal transfer function of the endoscopic grasper would be a linear transfer with a gain of one for the hand to tool tip position [Fig. 5(a)], and a constant value of the tool tip moment to handle moment ratio as a function of the handle position [Fig. 5(b)]. However, using the geometry dimension of the grasper under study shows a gain of 1.2 between the handle position and the tool tip position [Fig. 5(a)] and nonlinear moment ratio between the handle and the tool tip as a function of the handle angle [Fig. 5(b)]. The grasper mechanism moment/position transfer function might be another reason for the kinesthetic distortion that exists in endoscopic tools.

B. Materials

Two types of materials were used in order to evaluate the FREG performance: 1) soft tissue (pig internal organs—small bowel, spleen, colon, stomach, liver, and lung) and 2) latex and silicone materials. The soft tissue, latex, and silicone material were used for evaluation of their mechanical characteristics using the FREG in automatic palpation mode (objective protocol—Section II-C1). In addition the silicone materials were also used to test the FREG performance in teleoperation mode compared to other tools in ranking the materials stiffness by test operators (subjective protocol—Section II-C2). Table II summarizes the materials, tool usage and mode of operation in each type of experimental protocol.

The latex and the silicone materials were manufactured by Simulab Corporation. For the purpose of further discussion, latex materials were designated L1, L2, L3, L4, and L5. All the latex materials were shaped in the same cylindrical form with a diameter of 13 mm and a length of 45 mm. The above designation referred to each material based on its stiffness, where L1 is the softest material and L5 is the stiffest material. L1 to L4 can be considered viscoelastic materials representing artificial replication of soft tissues, while L5 can be defined as a solid which exhibits the upper limit of physiological stiffness, and can simulate bone.

The silicone materials' compliance characteristics were controlled by the percentage (weight) of catalyst used during manufacturing. Thus, a set of eight materials were obtained (0%, 5%, 10%, 15%, 20%, 25%, 30%, and 35% of catalyst). All the silicone materials were shaped as a cylinder with a diameter of 14.7 mm and a length of 150 mm with the same color and texture.

C. Experimental Protocol and Data Analysis

Two types of protocols were defined and tested in order to evaluate the performance of the FREG system in its two modes of operation. The objective experiment focused on the biomechanical characteristics of soft tissue and viscoelastic material replication using the FREG in an automatic mode. In the subjective experiment, test operators used the FREG in a bilateral mode using a protocol which examined the ability to rank material stiffness, relative to the performance achieved by a standard endoscopic grasper tool and the human hand (Table II).

1) Objective Experiment: The FREG is controlled by software running in real-time mode on a PC. This feature allows automated grasping and palpation functions to be implemented in software. Using this function, the deflections and forces being applied on the tissues are measured in order to extract information about tissue mechanical properties. The aim of the protocol was to measure mechanical characteristics of biological and artificial soft tissue materials with the endoscopic grasper tool.

A material can be defined by its constitutive equation, however those equations can only be determined by experimental methods [26]. Previous studies analyzing soft tissue were focused mainly on testing tissues under uniaxial tensile conditions using the quasi-linear viscoelasticity theory (QLV) [26]. Selected tissues, which are inherently under physiological compression conditions (e.g., bone, cartilage, and intervertebral discs), have been studied under uniaxial compression using the biphasic viscoelastic theory [27].

A full characterization of soft tissue and latex as viscoelastic materials requires an extensive experimental database which includes the material time domain response (creep and relaxation) and its frequency response (complex modulus). Fung [26] in his QLV theory suggested that if a step increase in elongation is imposed on the specimen, the stress developed will be a function of time (t) as well as of the material

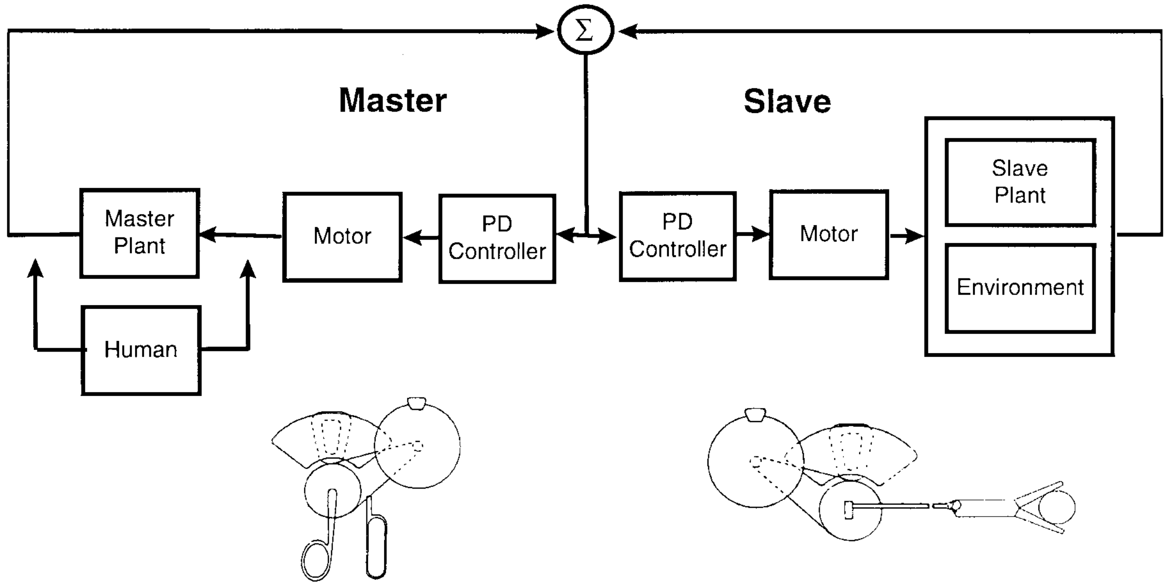


Fig. 3. FREG bilateral force feedback (teleoperation) control scheme.

TABLE I
CONTROLLER PARAMETERS

	K_p [Nm/rad]	K_d [Nm sec/rad]
Master	12.6	0.05
Slave	9.6	0.04

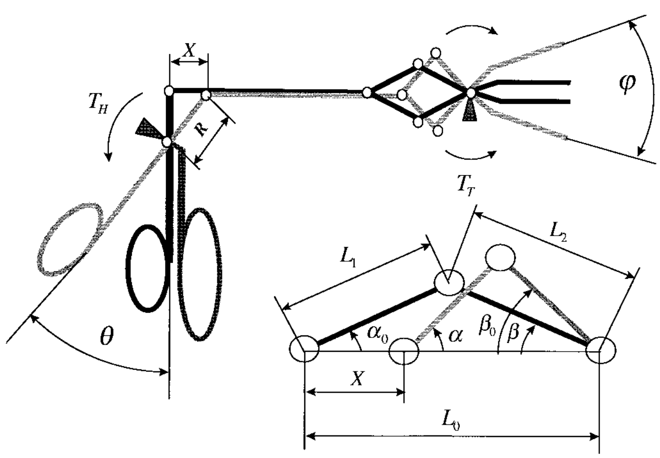


Fig. 4. Endoscopic grasper—schematic diagram of the mechanism in two typical positions.

stretch ratio (λ). The history of the stress response, called the relaxation function ($K(\lambda, t)$), is assumed to be of the form

$$K(\lambda, t) = G(t) * T(\lambda) \tag{4}$$

in which $G(t)$ is a normalized function of time, called the reduced relaxation function and $T(\lambda)$ is a function of the stretch ratio alone, called the elastic response. Fung [26] proposed (5a) for defining the elastic response of the material under tension conditions.

In our study the integration constant (C) in (5a) was determined by using the initial condition in a natural state in which, by definition, $T = 0$ when $\lambda = 1$. Moreover, (5a) was modified to define the elastic response in compression condition (5b) by multiplying the right-hand side of (5a) by -1. We define the stress in compression condition with a negative sign compared to tension, and we substitute the definition of tensile strain $\varepsilon_T = (L - L_0)/L_0 = \lambda - 1$ with the definition of compressive strain $\varepsilon_C = (L_0 - L)L_0 = 1 - \lambda$

$$T_T(\lambda) = Ce^{\alpha\lambda} - \beta \tag{5a}$$

$$T_C(\lambda) = \beta\left(1 - e^{\alpha(1-\lambda)}\right) \tag{5b}$$

$$T = \frac{F}{A_0} \tag{6}$$

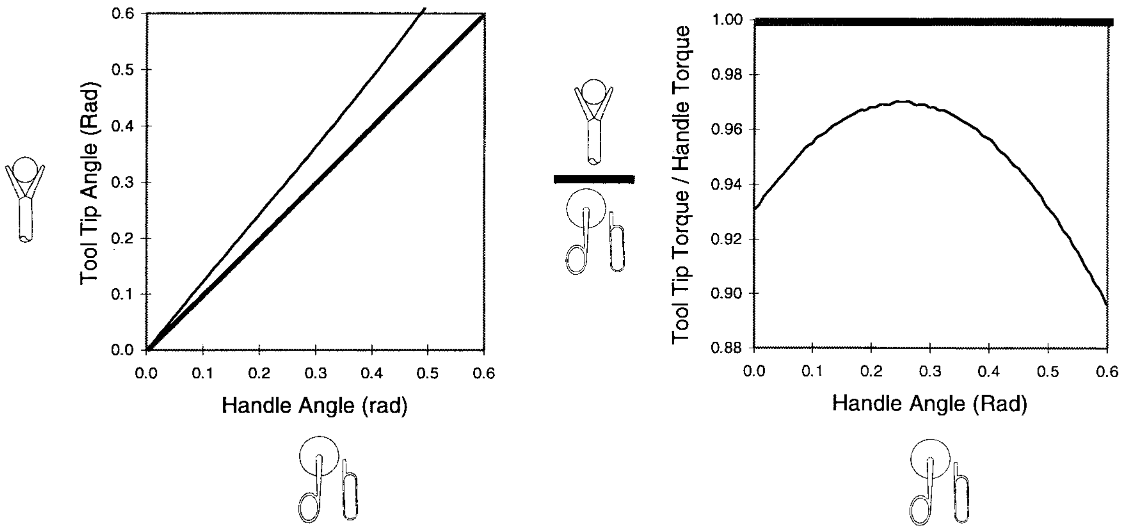
$$\lambda = \frac{L}{L_0} \tag{7}$$

$$K(\lambda) = \frac{dT}{d\lambda} = \alpha\beta e^{\alpha(1-\lambda)} \tag{8}$$

where α and β are parameters, λ is the compression–length ratio, T_T is the uniaxial tension stress, T_C is the uniaxial compression stress, F is the force applied on the specimen by the grasper, A_0 is the grasper contact cross-sectional area, L is the length of the material compressed by the load, L_0 is the length of the material at zero load, and K is the material stiffness.

To perform the automated palpation function, the slave position controller was driven by a sinusoidal displacement command while recording position, position error, and its motor torque. Three cycles of a 1-Hz sinusoidal displacement were applied as the command position input to the slave controller. The amplitude of the sinusoid corresponded to full opening and closing of the jaws (0.6 rad). During the automated palpation the slave received its control commands directly from the computer and the master was disconnected. The PC recorded the slave torque and angular displacement for 3 s at a sampling frequency of 1 KHz.

Torque versus displacement data were first isolated in time to the segment involving initial contact and compressive



(a) (b)

Fig. 5. Endoscopic grasper transfer functions (Thick line—linear transfer function; Thin line—actual transfer function). (a) Tool tip angle as a function of handle angle. (b) Tool tip moment—handle moment ratio as a function of handle angle.

TABLE II
SUMMARY OF TOOLS AND MATERIALS USED IN THE EXPERIMENAL PROTOCOL

Protocol	Type of Material	No. of Specimens	Tool	Operation Mode
Objective	Soft Tissue	6	FREG	Automatic
	Latex	5	FREG	Automatic
	Silicone	8	FREG	Automatic
			Indentor	Automatic
Subjective	Silicone	6	FREG	Teleoperation
			Grasper	Manuel
			Hand	Manuel

displacement. Compared to other types of surgical grasping instruments, the geometry of the Babcock tool suggests that it creates a relatively uniform stress distribution under the contact sites. The torque-displacement data measured at the handle were transformed to the uniaxial compression stress-length ratio using the geometry of the slave. Then, the stress–length ratio data were fitted, using the least squares method, with (5b).

The device performance was evaluated with *in vivo* palpation of pig internal organs (small bowel, spleen, colon, stomach, liver, and lung). Tissues were inserted in the jaws with the tool handheld. Three experimental sessions were performed compressing the tissue in three different locations. Each experimental session included three sinusoidal compression cycles in the same locations. Thus, a total of nine stress–length ratio data sets were examined for each material. Protocols for anesthetic management, euthanasia, and survival procedures were reviewed and approved by the Animal Care Committee of the University of Washington (Seattle, WA) and the Animal Use Review Division of the U.S. Army Veterinary Corps. In addition to the soft tissue the same experimental protocol was used to test the latex and silicone materials.

In order to independently test the materials with a method which is less dependent on the testing tool type, the stress compression–length ratios of the silicone materials were measured by a parallel surface compression test. In this testing method the silicone materials were placed on a flat surface plate and a metal cylinder indentor with a diameter of 7 mm and a flat contact area which was parallel to plate was penetrated to the materials. The indentor was moved against the plate along the silicone material diameter applying compression conditions while measuring the force and deflection along the line of action. No additional boundary conditions except the flat surface and the indentor restricted the materials from expanding freely in all other directions.

2) Subjective Experiment: An objective method to evaluate the FREG performance in discriminating between the stress–length ratio characteristics of soft tissue and viscoelastic materials was described in the previous section. However, the endoscopic graspers are usually manipulated by a human. This situation raises the question regarding the psychophysical aspects of the FREG performance. The experimental protocol was designed to examine the operator ability to rank a group of materials according to their stiffness using three tools: 1) the FREG in a bilateral force feedback mode; 2) a regular endoscopic grasper, commonly used in MIS; and 3) touching the materials with the hand (latex gloved) as traditionally done in open surgery. The same Babcock tool was used by the FREG and the regular grasper during the subjective experiment. Moreover, the same type of tool was also used during the objective experiment performed by the FREG.

Ten test operators divided into two groups (surgeon group and control group) were asked to rank eight materials according to their stiffness using the three tools. The surgeon group included experienced general surgeons specialized in minimally invasive surgery. The control group included engineers without any medical background. The two groups included subjects with the same gender (male) and similar average age. In order to avoid the fatigue and learning effects, each test operator performed the experiment using the three tools in different order. A typical experiment, using one of the tools mentioned above, began with a 2–5 min learning period in which the materials were presented to the test operators in the correct order of increasing or decreasing stiffness. The test operators were allowed to learn by using the current tool, the differences in stiffness between the silicone samples. During the testing session, the materials were presented to the test operators in random order (keeping the same random order for all the test operators). The test operators were then instructed to rank the materials in regard to stiffness four times (two times in increasing stiffness order and two times in decreasing stiffness order). The experiment was performed in such a way that the test operators could not see either their hand or the tool tip, in order to avoid visual feedback of material deformation. Using this method, the test operator's material ranking was based solely on their haptics sense (touch). The test operators wore a single pair of surgical latex gloves during the experiment to simulate operating room conditions. In the case of ranking the material stiffness with the hand, the test operators were instructed to apply only compression forces along the material's diameter (cylinder shape) rather then shear forces. The ranking methodology that the test operators were instructed to follow was to build a subgroup, by excluding out of the whole group 2–4 materials with the highest or lowest stiffness and then to rank the materials in this group. If the test operator was unsure regarding the stiffness of the last material in the subgroup, it was possible to return it to the initial group. However, once the materials in the subgroup were ranked and excluded by the test operator, he was not allowed to compare the stiffness of the materials that were left against the stiffness of the materials in the excluded subgroup.

The statistical null hypothesis (H_0) was that all the tools (standard endoscopic grasper, FREG, and hand) are equal in their performance for ranking material stiffness. The index of performances was calculated as the mean squared error (MSE) of the estimated ranking (ER) relative to the correct ranking (CR)—(9)

$$\mathrm{MSE} = \frac{\sum_{i=1}^{n} (\mathrm{ER} - \mathrm{CR})^2}{n}. \tag{9}$$

For the six materials [$n = 6$ in (9)] used in the subjective experiment, the MSE value has a range of $\langle 0, 11.66 \rangle$. When a test operator ranks the materials exactly according to their correct stiffness order the MSE has a value of zero. The upper boundary of MSE value of 11.66 can be achieved when the test operator ranks the material in the opposite manner.

Two-way analysis of variance (2-D ANOVA) was used to analyze the differences between instruments within group-3 levels (regular endoscopic grasper, FREG, and hand) and between groups-2 levels (surgeon group and control group).

III. Results

A. Objective Experiment

The stress–length ratio curves for the compression phase of each material grasping (pig internal organ soft tissues, latex, and silicone materials) were fitted with the exponential function (5b) using the least squares method. Typical measured data and the corresponding curve fit are plotted in Fig. 6 for each material. Note, that for presentation purposes, minus signs indicating compression stresses were omitted. Since the software generated three close/open cycles, there were three compression tests recorded 1 s apart for each grasp. Tissues typically got stiffer in the second and third compression of each sequence. The quality of the numerical fit was verified by using the correlation ratio factor R^2. The computed R^2 values were typically very close to one ($R^2 > 0.999$), indicating very high quality of fit between (5b) and the experimental data.

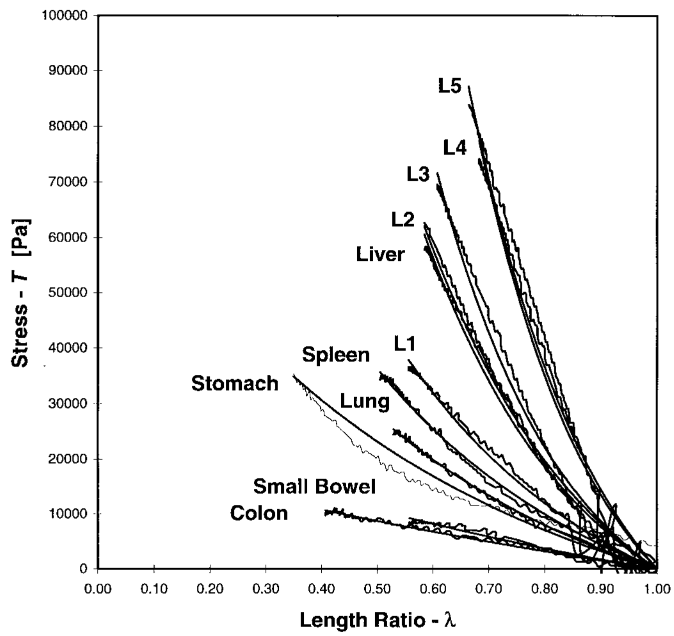


Fig. 6. Typical uniaxial compression stress as a function of length ratio and the corresponding exponential curve fit for different pig organs and latex materials.

The resulting parameters of the exponential fit α and β as computed from the FREG's measurements were displayed as a scatter plot for the pig soft tissues and for the latex and silicone materials (Fig. 7). Generally speaking, stiff materials have high values of α and β, and vice versa. In some cases, as the stiffness of the material increases, β increases whereas α may decrease (as happened for the silicone materials tested using the indentor technique—Fig. 8). However, the product $\alpha \cdot \beta$ always increases as the material stiffness increases, indicated by (8). Data points of α and β formed into clusters for each material that was tested. Each cluster consists of nine data points. Rectangles, defined by the standard deviations computed from the organ data clusters, did not overlap except for lung and spleen. These variances are partly due to the variation in stiffening of tissues under repeated compression as described above.

In addition to the measurements performed by the FREG, the eight silicone materials were tested using the indentor experimental protocol. Stress–length plots of the eight silicone materials indicated that from the mechanical point of view three of the eight materials (20%, 25% and 30%) had the same stress-strain curves, so that it was impossible to distinguish between their stiffness with the tools used in the subjective experiment. For the purpose of analyzing the data, the three materials with the similar mechanical characteristics were lumped into one. The exponential fit parameters α and β obtained by indentor testing method have different values

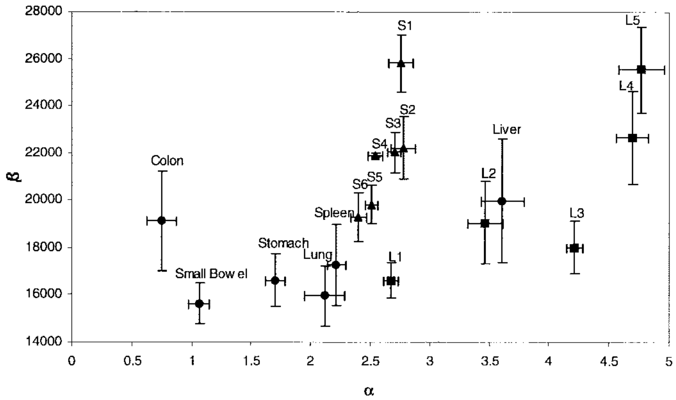


Fig. 7. Scatter plot of the biomechanical parameters α and β for different pig soft tissues, latex (L1–L5) and silicone (S1–S6) materials.

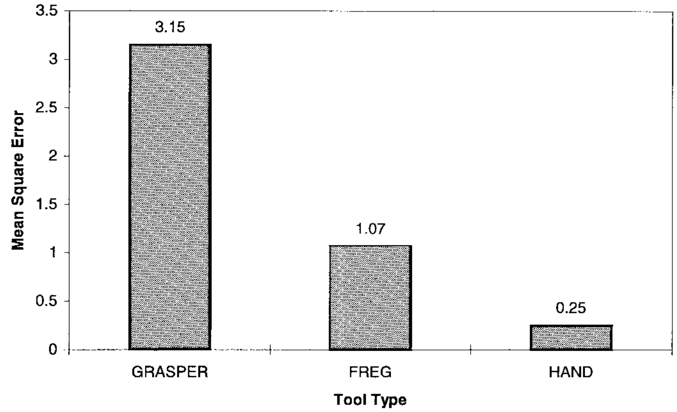


Fig. 9. The MSE of ranking the stiffness of six materials with three different tools.

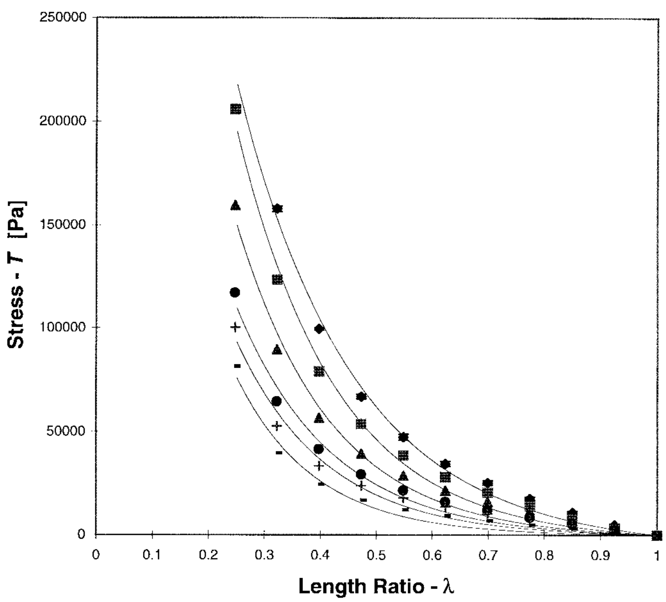


Fig. 8. Uniaxial compression stress (standard method) as a function of length ratio and the corresponding exponential curve fit for the six silicone materials used for the subjective test.

($\alpha\langle 4.76, 5.57, 5.85, 5.89, 6.14, 7.07\rangle$ and $\beta\langle 6300, 3034, 1830, 1379, 941, 380\rangle$ for (S1–S6) compared to those measured by the FREG (Figs. 7 and 8). This phenomena can be explained by the FREG's inherent stiffness due to its structure and internal mechanism. The FREG stiffness most probably was measured as part of the material's stiffness. However, the relationship between the material parameters multiplication $\alpha \cdot \beta$ and the material stiffness remained the same for the two testing methods. Moreover, since the six silicone materials are evenly graded as measured by stress–length ratio characteristics (Fig. 8), this makes them ideal for the purpose of the subjective testing experiment.

B. Subjective Experiment

The results of the subjective experiment are summarized in Fig. 9. The 2–D ANOVA statistical test showed a significant difference between the performance obtained by the three tools ($p = 1.7 \cdot 10^{-6}$). The best performance in ranking the materials according to their stiffness was achieved by using the hand (MSE $= 0.25$), whereas the worst performance was obtained by the standard endoscopic grasper (MSE $= 3.15$). The performance of the FREG (MSE $= 1.07$) was between the previous two, and closer to the performance of the hand than the grasper. The analysis suggests that there is no significant difference between the two operator groups (surgeons and control) that were tested ($p = 0.065$). From the results above, the null hypothesis (H_0) may be rejected (Section II-B1). There is a significant difference between the performances of the three tools in ranking the materials according to their stiffness.

IV. Conclusions

Part of the haptic information that is lost when a surgeon manipulates a soft tissue using an endoscopic tool/grasper may be regained by using the bilateral force feedback technology implemented in an endoscopic instrument. The FREG is capable of controlling the force or displacement of the jaw (tool tip) with interchangeable tools. To minimize cost and complexity, the system works with existing interchangeable reusable tools. The FREG controller was designed to maximize position control gain while preserving stability under unloaded conditions. Separating the human interface (finger loops) from the endoscopic tool allows one to generate a new human–machine interface (transfer function) in a way that enhances performance by overcoming the distortion that exists in the current mechanical endoscopic grasper setup.

The first phase in this two-phase study was the objective experiment. The scatter plot (Fig. 7) shows that the FREG in automatic mode is capable of discriminating between different soft tissues and latex/silicone simulated tissue, and that the material's intrinsic biomechanical parameters can be identified for compression conditions. Moreover, a correlation from the mechanical characteristic perspective between the latex material and the soft tissue was found. For example, the data indicate that the latex material L2 might simulate the liver. The α and β parameter values may be used to design this material for simulation purposes. The α parameter values for hollow organs (e.g., colon, small bowel, and stomach) tend to be lower than the α parameter values for solid organs (e.g., spleen and liver). Although hollow organs have greater macroscopic variability (mucosa, muscular layers, and serosa) than solid organs, they also contain lumenal air which could explain their softer characteristics.

Different values of α and β parameters for the six silicone materials were obtained using the FREG compared to a testing method in which the contact areas were parallel. This result suggests that the measurements performed by the FREG are tool-dependent. However, the two methods show the same trends. For example, the product $\alpha \cdot \beta$ may be an indicator for the material stiffness. The current values of the α and β parameters may be used for designing materials as tissue replication for training usage. However, since the material parameters measured by the FREG were found to be tool-dependent, the usage of the α and β parameter values should be restricted to developing phantom materials for the purpose of training in MIS applications.

The second phase of this study focused on the psychophysics aspects of operating the FREG. The statistical analysis (2-D ANOVA) of performance (MSE) measured for the ten test operators ranking six materials according to their stiffness suggests significant improvement in the performance of the FREG relative to a standard endoscopic grasper. The FREG performance was closer to the human hand, in rating material stiffness, which defines the upper performance limit, than the standard endoscopic grasper, which defines the lower limit. Even in the hand-in-glove conditions, the test operators were not capable of ranking the material stiffness correctly in all the cases. This fact may raise the need for more advanced instruments like the FREG capable of increasing the haptic sensation beyond the capability of an unaided hand.

The approach outlined in this study might be replicated in future studies with different endoscopic tools and control algorithms. This future research may study the soft tissue biomechanics under shear-compression conditions, and from another perspective, the correlation between the control algorithms and their parameters and the FREG performance. These studies may expand the current knowledge of the tissue/tool interface and on the human/machine interface in MIS.

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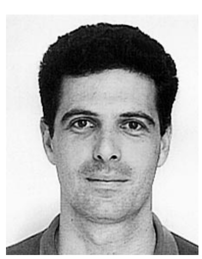

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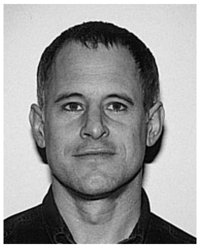

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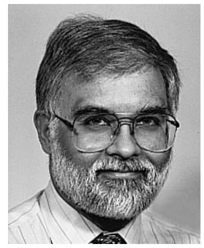

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