ORIGINAL ARTICLE

MEMSurgery: An integrated test-bed for vascular surgery

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Abstract

Many surgical procedures require skillful manipulations of blood vessels, especially in conventional invasive or minimally invasive surgical procedures. Current surgical methods do not allow the surgeon to receive any real time feedback of the tissue properties when operating on the vessel. As a result, the unintentional application of excessive force may damage the blood vessel. To minimize such trauma, and to study the interaction of surgical instruments with the vessel structure, we have developed an integrated surgical test-bed called MEMSurgery (Microelectromechanical Sensory augmented Surgery). The test-bed integrates four elements: a) force sensors mounted on surgical appliances, b) a feedback control mechanism utilizing the intrinsic mechanical properties of the blood vessel, c) feedback of the force applied on the tissue back to the surgeon through a haptic feedback device, and d) visual feedback by a graphical computer model of the vessel. Finally, we evaluate the performance of MEMSurgery by testing the hypothesis that the combination of haptic feedback, feedback control based on vascular mechanical properties, and real-time visual representation of the vessel will help the surgeon decrease the probability of applying excess force while occluding the blood vessel. To this end, we designed a rodent experimental model to obtain the ideal minimum occlusion force (MOF). After a series of human performance studies, and subsequent comparison to direct application of force on the forceps (without feedback), the results show that the probability of applying reasonable MOF increases from 35.5% to 80%. After a brief training period, the probability increases to 90%.

Keywords: Haptic feedback, visual feedback, Minimum Occlusion Force(MOF), surgical augmentation, vascular surgery

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INTRODUCTION

Coronary artery bypass graft (CABG) procedures involve many skillful manipulations. This widely practiced procedure consists of vessels from the leg being harvested and grafted onto the heart to bypass the blocked or atherosclerotic artery [1, 2]. Such bypass procedures involve extensive handling of the vessel and its soft tissue. The surgical handling of the vessel that involves cutting or squeezing and compression is very likely to cause trauma to the vessel walls, and especially to the endothelium [3]. This problem is even more acute with the advent of a variety of minimally invasive coronary artery bypass procedures, where the surgeon's view and perception is blocked and thereby, the degree of difficulty in manipulating vessels increases. Because of the shortcomings of assisting technologies, excess force may unintentionally be applied when performing delicate operations such as the closure of these vessels or the exposure of the vascular endothelial surface (both necessary precursors to several procedures) [4]. This excess force may cause tissue damage, one of the major factors affecting the outcome of the surgery [5]. As a result, studies show that 10% of grafts become occluded within a month after surgery; this figure increases gradually to 50% ten years after the procedure. Approximately 60% of CABG patients require a second procedure within two years [6]. Estimates indicate that as much as 25 billion dollars are spent on treatment every year for this problem [7].

A recent study has shown that amplified force feedback improves performance during robot-assisted middle ear surgery [8–10]. Augmentation by haptic feedback with a control mechanism based on tissue mechanical properties in the sensing and manipulation of vessels, should extend the surgeons' perception during surgical procedures and help avoid excess force from unintentionally being applied. The tactile and mechanical properties of the tissue hold a wealth of information about its state and physiology. Surgeons rely a great deal on the intuition gained by the 'feel' of the tissue (a combination of these tactile and mechanical properties) while performing surgery and diagnosing disease. However, as the size of the operating area decreases, the feedback a surgeon receives from a tissue decreases. Tissue properties, heretofore obvious to the surgeon, enter a region of uncertainty when performing minimally invasive or robotic surgery. Thus, a device that provides amplified feedback to the user on the state of the tissue and the force applied to the tissue should allow the surgeon to operate with greater efficiency. Eventually, a robot manipulator and haptic interface can be combined into a single device to create a cooperative manipulator [11–13]. This would be particularly useful for surgeons in manipulating the tissue directly through the tool, while simultaneously feeling amplified forces.

These issues motivated us to develop a test-bed for microsurgery research - MEMSurgery that integrates surgical tools, computer interfaces and control devices, and feedback systems (such as visual and haptic). This paper describes the MEMSurgery test-bed, which integrates microsurgical tools, enhanced by incorporating sensors for measuring tissue properties. One particular example presented here is a pair of forceps mounted with strain gauges to measure forces applied to the blood vessel being operated on. The test-bed is interfaced to a computer-controlled driver for precisely compressing the blood vessel in a stepwise manner. The force applied and the vessel's reaction is returned to the surgeon via a haptic feedback device. This computer-controlled device can provide force feedback proportional to the forces perceived at the surgical tool. Moreover, to assist the surgeon with the visualization of the vessel compression procedure, we present a computer model of the vessel and its closure dynamics. This three-dimensional model presents not only a graphical view of vessel closure, but also correlates it with the tactile sensation of pulsation through haptic feedback.

The MEMSurgery test-bed is used to carry out controlled experimental investigations on small vessels in rodents. The test-bed is also used to provide surgical training and to assess improvement in surgical skills with sensory augmentation. In an accompanying study [14], it is shown that this approach optimizes the minimum occlusion force (MOF) and provides training to surgeons.

SYSTEM DESIGN

MEMSurgery is a vascular surgery test-bed, presently configured to demonstrate its laboratory use in small animal (rodent) surgery, with a number of instruments integrated together. As depicted in Figure 1, the surgeon sits at the test-bed and operates a teleoperated device by means of a force feedback interface (also called a haptic interface). This device issues a position command to the computer, which is then relayed to a teleoperated probe. The teleoperated probe mechanically moves a surgical tool, such as a pair of forceps. The forceps are provided with sensors for measuring the compressive force that is applied at the tip. The computer also provides a visual display of the blood vessel. A computer model of the vessel presents visual information on the deformation of the vessel, and provides the surgeon with real-time, interactive feedback on the effect of vessel compression. Based on the measured forces, an amplified force is relayed to the haptic interface. This force feedback creates the perception of mechanical interaction with soft tissue when the surgical tool compresses the vessel, allowing the surgeon to experience the actual feel of the tissue held by the forceps. Other instrumentation modules are integrated into the test-bed to provide real-time and continuous information about vessel properties, such as blood flow (measured using a laser Doppler flow meter) and displacement (measured using a laser ranging system). Thus, this surgical test-bed serves as an excellent research setup to study vascular surgery procedures, to test the benefits of robotic assisted surgery, to evaluate surgeon's skills, and to provide surgical training.

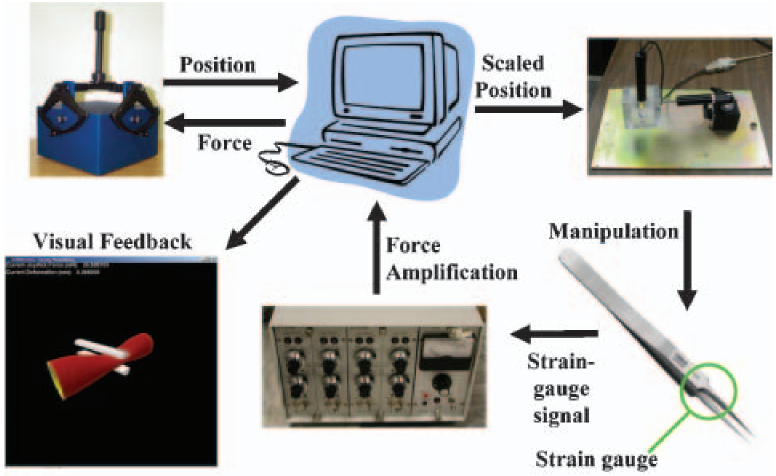

Figure 1 MEMSurgery framework.

Force-sensing surgical tools

Standard surgical tools follow designs that are fairly classical. However, such instruments can be enhanced by the addition of sensors that allow a system to provide feedback on the tissue properties. In our setup, we mounted several instruments with strain gauges in order to detect and measure the forces that are applied during surgery. The resistances of semiconductor strain gauges vary on the application of strain. We designed a pair of forceps (Figure 2) with measurement errors in the sub-milliNewton range. Our system also includes several mechanical actuators (M-231 and M-126, Physik Instrumente, Germany), which are interfaced with a standard motor control card using a PID (proportional-integral-derivative) controller (C-842, Physik Instrumente, Germany). These high-precision motors, which are accurate to within 0.01 micrometers, allow the computer to precisely manipulate the physical instruments. One of the motors closes and opens the forceps under computer control. The combination of the force sensitive tools, computer, haptic interface, and motors allow the user to control a surgical tool from a distance while still feeling the forces acting on that tool. In addition, utilizing previously collected experimental data, the same system (computer and haptic interface only, Figure 3) can be used for training purposes. The entire system operates in real-time, with some small latency imposed by the speed of the motors.

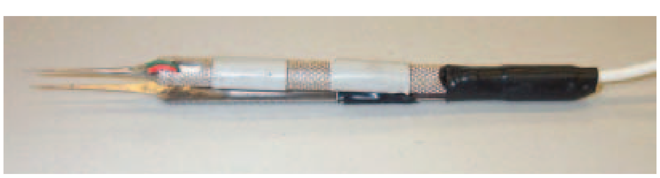

Figure 2 Forceps with strain gauges.

The computer uses the amplified strain gauge signals to reconstruct the forces between the forceps and the tissue. A customized program written in C++ allows the computer to both display the data on the monitor and control a haptic interface. The haptic device allows the user to feel some amount of force on their hand. Equivalently, in a practical application, this force should result in proportional augmentation of the force reported by the forceps (or other surgical tools). In surgical training and simulation, an experiment-based mechanical model of the blood vessel created on the computer can determine the force.

The forceps are used in conjunction with two sensors. 1) A laser displacement sensor (Philtec, Inc., Annapolis, MD, USA) records the displacement (closure) of the vessel with a resolution of 10 microns. This position-adjustable instrument is connected to a computer with a data acquisition card. The voltage produced is sampled at 500 Hz. 2) A flow meter (Periflux 4001, Perimed, Inc., North Royalton, OH, USA) measures the flow within the vessel and provides a marker for the point at which the vessel is completely occluded.

Haptic feedback system

The forces involved during microsurgery are in the milliNewton (mN) range. These are too subtle to be fully sensed and appreciated by the surgeon's tactile senses. The lack of adequate sensory feedback is more acute in teleoperated or minimally invasive surgery, where the surgeon may be operating the surgical tool at a distance or under limited visibility

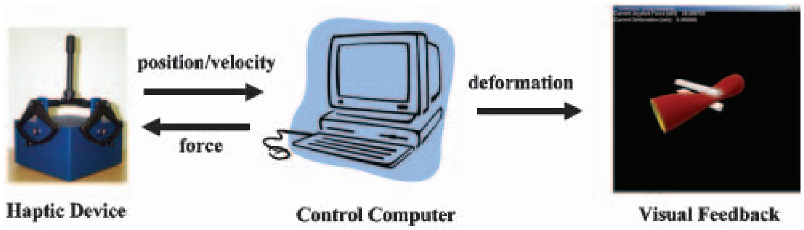

Figure 3 Training framework.

constraints, with no direct tactile feedback available at all. Hence it is very desirable to provide feedback of the amplified forces being applied on the surgical tool. Force feedback is accomplished using a haptic interface, the IE2000 (Immersion Corp., Mountain View, CA, USA).

Control mechanism based on tissue mechanical properties

Although haptic and visual feedback are helpful for the surgeon to operate on a blood vessel, the achievement of the minimal occlusion force is still dependent on the surgeon's experience and the adjustment they make. Therefore, introduction of a control mechanism based on intrinsic mechanical properties of the blood vessel is very important to safely and reliably achieve the minimal occlusion force. Using extensive experimental data, a statistical estimate of the mechanical properties of the blood vessel can be obtained. For example, Figure 5, Figure 6, and Figure 7 show the relationship between the force applied and the deformation of vessel. There are two obvious characteristics of tissue mechanics that could be used as feedback inputs. The first is the abrupt jump in force feedback from the previous level of compression (Figure 5), which happens only when the vessel is completely occluded. The second is the change in mean curvature of the graph relating force and deformation, with increased compression of the forceps (Figure 7). The mean curvature from the beginning of occlusion to full occlusion is larger than that corresponding to any other deformation position prior to that point. These characterizations are incorporated into the control and data collection program (using LabView) written to control the motor movement in real time, so that the system acquires data pertaining to MOF automatically and reliably.

Visual feedback system

The surgeon operating on the vessel does not have a high-resolution view of the vessel being operated. Moreover, minimally invasive surgery procedures make this task even more difficult since the visualization of the surgical field is greatly obstructed. Hence an amplified visual display of the surgical area would be extremely helpful to the surgeon. Data required for such viewing of the surgical field can be obtained with the aid of endoscopic cameras or other optical imaging techniques. In addition, the surgeon can be provided with artificial visual feedback of the vessel on the computer screen. In our MEMSurgery test-bed, a three dimensional model of the vessel along with the surgical tool is presented to the surgeon on the computer screen. The computer model recreates not only the vessel deformation but also integrates the interaction of the surgical tool with the vessel.

A computer model of a blood vessel and the tool was created by using the OpenGL programming environment. This model, presented to the surgeon on a computer console from different, rotated and scaled viewpoints, makes it convenient to see the vessel compression. Finally, vessel dynamics are simulated. *In vivo*, the vessels pulsate at the heart rate. The pulsations can provide valuable tactile information to the surgeon, providing a perception of tissue mechanical properties such as compliance and patency. Thus, visual display of the vessel spanning different orientations along with dynamics such as pulsations help augment the surgeon's perceptions of the vessel surgery and is anticipated to help achieve improved surgical outcomes.

To construct a computer graphical model of the vessel soft tissue and its deformation under externally applied force, we model the vessel by a NURBS (Non-Uniform Rational B-Splines) surface approximation. Figure 4 shows the layout of control points. There are 8 control cross-sections, with each section having 9 control points around the perimeter. The first and last control point overlap, so as to get a closed surface. The following is the mathematical model of NUBRS surface (for details, please see reference [15]).

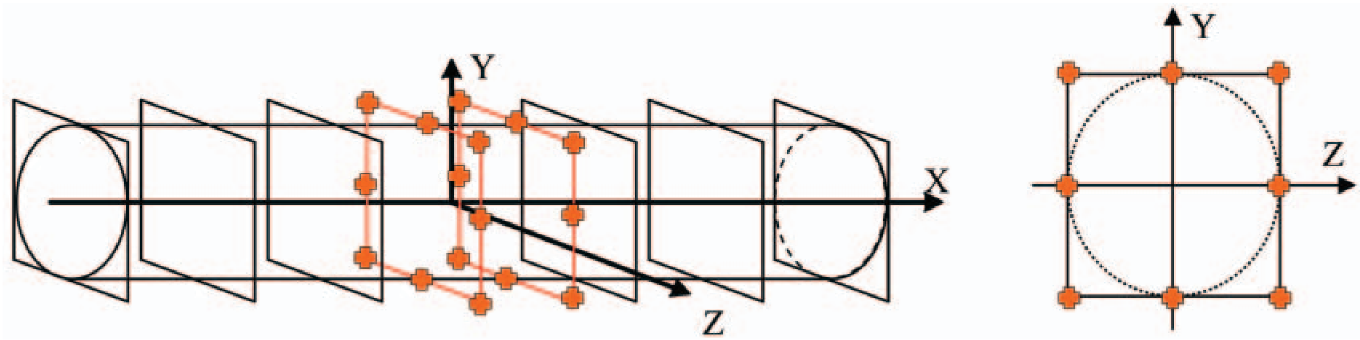

Figure 4 Control points distribution.

$$S(u,v)=\sum_{i=0}^{n}\sum_{j=0}^{m}R_{i,j}(u,v)P_{i,j}\,;$$

$$R_{i,j}(u,v)=\frac{N_{i,p}(u)N_{j,q}(v)w_{i,j}}{\sum_{k=0}^{n}\sum_{l=0}^{m}N_{k,p}(u)N_{l,q}(v)w_{k,l}}$$

$S(u,v)$ and $R_{i,j}(u,v)$ are vector-valued functions of the parameter dimensions u and v, which are normalized to [0,1]. $P_{i,j}$ are control points; $R_{i,j}(u,v)$ are piecewise rational basis functions; $N_{i,p}(u)$ and $N_{j,q}(v)$ are B-spline Basis Functions; p, q respectively stand for the number of degrees along u and v parameter dimensions; $w_{i,j}$ are the weights; m+1, n+1 are the numbers of control points along u and v dimensions.

In order to integrate pulsation and deformation into our model, the relative control points frequently need to be repositioned. To simulate deformation due to compression of the forceps under the applied force, the distance between two orange sections is equal to the width of the forceps, as shown in Figure 4. 12 control points (6 points on the top and 6 on the bottom of the two sections) move towards X-Z plane, and another 12 control points (6 points in the front and 6 points behind of the two sections) move away from X-Y plane. The deformation of the vessel produced by the forceps' compression needs to be translated into a movement

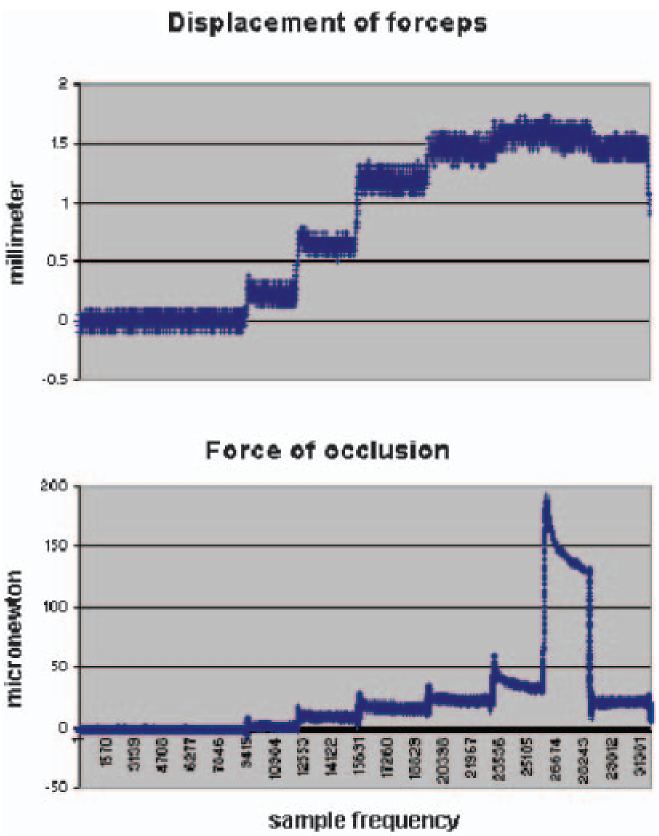

Figure 5 Force profile applied to the rat abdominal aorta by the forceps. Force was gradually increased until the aorta was completely occluded, then decreased. Each position level was held constant for about 6 seconds. Relaxation phenomenon of the soft tissue is observed at each position level. The thick trace is the result of displacement occurring during pulsatile movement of the vessel (which is palpable).

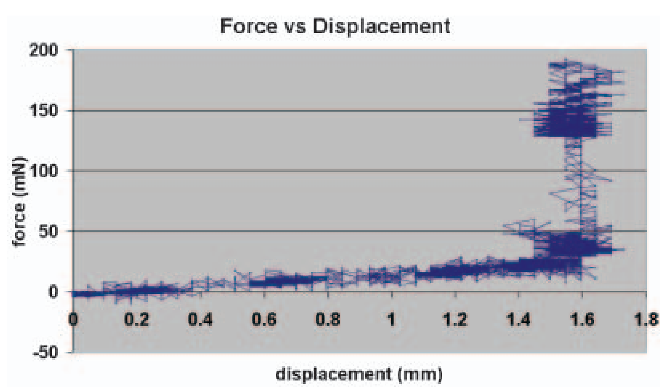

Figure 6 Relation between experimentally obtained force and displacement recorded during vessel occlusion. An abrupt jump in the force applied occurs when the vessel is fully occluded. The jagged lines were caused due to pulsation and tissue relaxation of the aorta, and the noise of the system.

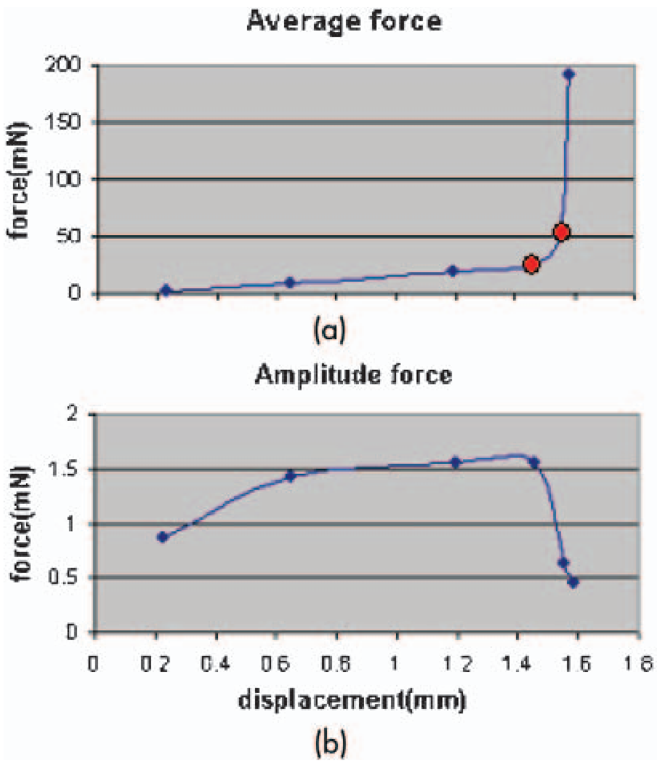

Figure 7 Relation between F_{avg}, F_{amp} and displacement by interpolation. The top diagram (a) shows mean curvatures before started occlusion (left red point) and after full occlusion (right red point) are very close to 0, hence corresponding to two linear mechanical properties. There are abrupt changes of curvature around the two red dots, positive curvatures exist between the dots. This characteristics is used in providing feedback. The bottom diagram (b) shows amplitude force is very small during occlusion.

of these control points. If we assume that a point on the vessel gets deformed by an amount d along the radial direction V (unit vector) due to the movement of a most influential control point, $P_{k,l}$, which translates by a distance α in the direction V from $P_{k,l}$ to $\bar{P}_{k,l}$, and α is determined as follows:

$$\bar{P}_{k,l} = P_{k,l} + \alpha V; \alpha = \frac{d}{|V| R_{k,l}(\bar{u}, \bar{v})}$$

Where, $R_{k,l}(\bar{u},\bar{v})$ is a piecewise rational basis function; $\bar{u},\bar{v}$ denote parameter coordinates for the deformed point on the surface. $P_{k,l}$ is replaced by $\bar{P}_{k,l}$ in the control point array, and then the model is updated, resulting in the deformed vessel model. This technique extends easily to the modification of multiple control points [15, 16]. To simulate pulsation, 9 control points in a section are repositioned at one time and then these control point deformations are passed from one section to the next according to the pulsation frequency. The model is thereby continuously updated by the new control point array.

Finally, the data from our experimental studies on arteries are incorporated into the model, so that users can get haptic feedback according to our specific experimental data. At each level of vessel deformation, the average recovery force F_{avg} and amplitude force F_{amp} are calculated by the following formulations:

$$F_{avg} = \frac{\sum_{i=1}^{n} f_i}{n}; \; F_{amp} = \frac{\sum_{i=1}^{n} \left| f_i - F_{avg} \right|}{n}$$

Where, f_i denotes the i^{th} sample force at one level, and n denotes the number of sample at one level. Of course, F_{avg} and F_{amp} at the starting and ending part of every level are different due to relaxation of soft tissue, especially when the vessel is occluded (Figure 5). For simplicity and without loss of accuracy, F_{avg} and F_{amp} are calculated according to the data at the starting part of every level, such as the sampled data during the 1st second. The relationships between F_{avg}, F_{amp} and deformation are approximated by Piecewise Cubic Hermite Interpolating Polynomial (PCHIP) [17, 18] and implemented by haptic API (Application Programming Interface) [19], which allows users to incorporate their own functions of force and deformation.

EXPERIMENTS AND RESULTS

General surgical preparation

The experimental studies were carried out on adult Wistar rats (300 g). All experimental procedures in this protocol followed the guidelines presented in the Rodent Survival Surgery manual, approved by the Animal Care and Use Committee of Johns Hopkins University [20]. Rats did not receive any treatment prior to anesthesia. The actual surgery was under complete anesthesia. All surgical tools and the surgical site were sterile. The experiment started with intra-peritoneal injection of Xylazine/Ketamine 0.05–0.10 ml/100 g IP [mix. 8.75 ml of Ketamine (100 mg/ml), and 1.25 ml of Xylazine (100 mg/ml)] and then given as needed during the experiment (~every 30 min) to keep the rat under the desired anesthesia level. After disinfecting the skin of the rat's abdominal area, a long midline incision was made through the skin and then through the abdominal muscles. The intestine was moved to expose the abdominal aorta and vena cava. With careful blunt dissection, the fat layer surrounding the vessels was removed and the vessels were separated from each other along their length for about 1 inch (2–3 cm). It is very important to highlight the fact that the walls of the vessels were touched as little as possible to prevent unintentional damage or contraction reflex. We successfully tried to prevent any injury to the main vessels and their collaterals. Some of the small-attached blood vessels were however ligated to prevent bleeding. After the experiment, the rats were euthanized by an overdose of Nembutal.

Characterizing the profile of applied force

We setup our experiment, calibrated the laser displacement sensor, and the force-sensing forceps, and collected data as described earlier [4]. The gradual displacement of the forceps and the resulting force were recorded and shown in Figure 5. Figure 6 shows the relation between displacement and force. F_{avg}, F_{amp} are calculated from the experimental data according to the formula described earlier. Figure 7 represents approximately the relationship between F_{avg}, F_{amp} and the deformation (in particular, Matlab PCHIP was used to interpolate the relation curves in our implementation). The force applied to the forceps is a sum of F_{avg} and a dynamic force with an amplitude of F_{amp}. F_{avg} is a static force induced by the arterial wall and blood. F_{amp} is the amplitude of a dynamic force induced by the arterial wall pulsation with each heartbeat. In the experimental investigations (Figure 7), F_{avg} first increases linearly until the vessel starts to get occluded. A steep non-linear increase in the slope is observed as the vessel becomes further occluded. Finally F_{avg} increases linearly again with a steeper slope after the vessel is fully occluded. F_{amp} increases and reaches its maximum when the vessel just gets occluded and then steeply decreases with further occlusion. These results are easily incorporated into the visual model of the blood vessel using a haptic interface to provide an augmented force feedback. The results also show that it is possible to explore the blood vessel occlusion by monitoring these force perturbations.

System performance and evaluation

In order to evaluate and test the performance of the MEMSurgery test-bed, we designed the following experimental procedure: 1) In order to test MOF, volunteer subjects were asked to occlude the vessel by applying a minimum force on the forceps according to their judgment. 2) The profile of the applied force versus compression that we acquired beforehand during animal surgery was incorporated into our MEMSurgery system. The subjects were subsequently asked to occlude the vessel by applying MOF using their judgment of visual and force feedback. 3) We showed them properties of the vessel, particularly some characteristics when the vessel was occluded, so that they had some knowledge about how to reach MOF, and then asked them to use our training simulator. 4) After gaining knowledge of MOF, the subjects were asked to occlude the vessel as described in step 2. 5) System performance was evaluated by a comparison of the MOFs that were acquired in steps 1, 2 and 4. The particular experimental protocol is as follow:

Experiments were conducted with 30 randomly chosen volunteer subjects aged 21–35. The experiment consisted of 3 steps. In every step, the subjects were supervised by two other collaborators to increase the precision of measurements and decrease the probability of undesirable operation. In all the steps, the subject sat comfortably and held the forceps for step 1 and held the joystick for step 2, 3 and 4 in their dominant hand. Before starting, the procedures and goals of our study were explained to the subjects (for example, a general explanation of where and how to apply the force was necessary for subjects without a surgical background), but the subjects were not told how much force was necessary to occlude the vessel. The procedure was repeated twice. The subject had a few minutes to relax between each trial. If the subject felt they had an undesirable movement or was not totally comfortable with the procedure, we eliminated that segment and the corresponding data from our study. The subject was asked to repeat that step again when they were ready. We carried out four experimental steps for each subject.

Step 1: In vivo, the subjects directly applied the force to the forceps with their hands, which they thought was necessary to occlude the vessel. After a few minutes of relaxation they repeated the procedure. We recorded the force and the variation of the force applied on the vessel on a computer. We did not provide any information or feedback to the subjects during the experiment.

Step 2: The subjects used MEMSurgery to identify MOF of the vessel. They applied force on the vessel gradually, via haptic interface (i.e. joystick), until they thought that they had fully occluded the vessel. In the process, they were guided by the visual and haptic force feedback. After a few minutes of relaxation, the subjects were asked to repeat the same procedure.

Step 3: We told the subjects about the mechanical characteristics of the vessel at the time it was occluded, and the subjects were trained by MEMSurgery using the diagram of mechanical properties as indicated (Figure 8).

Step 4: After training, the subjects applied force that they thought was necessary to occlude the same vessel without feedback from the diagram of mechanical properties, but using the knowledge gained from the training using visual and force feedback. After a few minutes of relaxation, the subjects were asked to repeat the procedure again. The data were recorded separately.

According to our in vivo experimental data (Figure 5 and Figure 6), we know that there exists a jump from 50 mN to 130 mN, while the forceps produce a very small compression corresponding to the jump. There is a similar slope in Figure 6 and Figure 7 after 130 mN. Which means that the vessel is almost fully occluded by a force of about 50 mN, this value is however specific to the vessel we worked on. It is reasonable for us to infer that its MOF should be between 50 mN and 130 mN. Based on this observation of MOF, we analyzed the above performance test data. The results (Figure 9) showed that 35.5% of the subjects (11 subjects in 30) without visual and force feedback applied reasonable MOF to occlude the vessel. With the help of MEMSurgery, 80% of the subjects (24 subjects in 30) applied reasonable MOFs, even if they did not know the mechanical properties of the vessel they were occluding. After training and getting knowledge of the mechanical properties of the vessel, 90% of the subjects (27 subjects in 30) applied reasonable MOFs. In addition to that, with the help of MEMSurgery, the distribution of MOFs that were applied by subjects fell in a narrower range. After training, there is a 10% increase in the number of subjects who used reasonable MOF, and 3.3% of the subjects applied more than 130 mN (around 180 mN). In fact, this is still acceptable. According to our previous preliminary experiments and other estimates, the MOF of the rat's abdominal aorta could reach a high of 200 mN without damaging the endothelial cells, the intima or the vessel wall.

DISCUSSION AND CONCLUSIONS

The use and performance of the MEMSurgery test-bed in vascular surgery research and surgical training are described. The test-bed integrates a number of technologies - the surgical tool, control system for occluding blood vessels, force sensing and haptic feedback, and a graphical computer model. This test-bed has been successfully used to provide training to non-skilled individuals to achieve desirable minimum occlusion force (MOF) [14]. The test-bed has yielded useful data on tissue properties, distinguishing the mechanical properties of veins and arteries [4]. Together, this information should lead to improvements in surgical procedures,

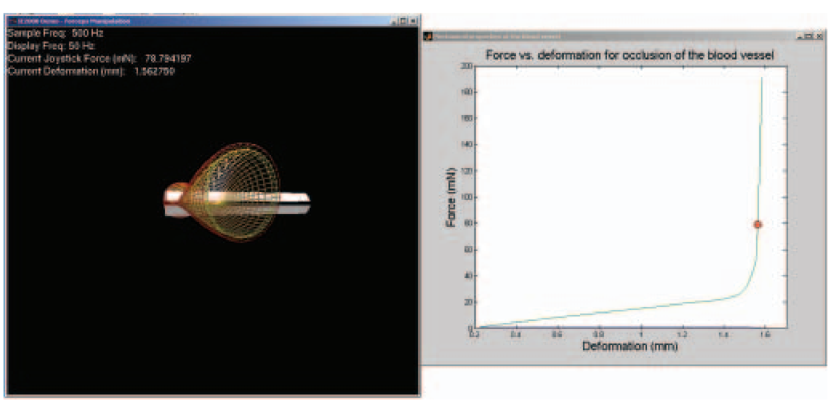

Figure 8 Training simulator with indication of mechanical properties, the red point shows in real-time the position of force and displacement when users apply force to occlude the blood vessel. Therefore, in addition to visual and haptic feedback, the relative position of red point in mechanical properties provides users deep insights to occluding the blood vessel.

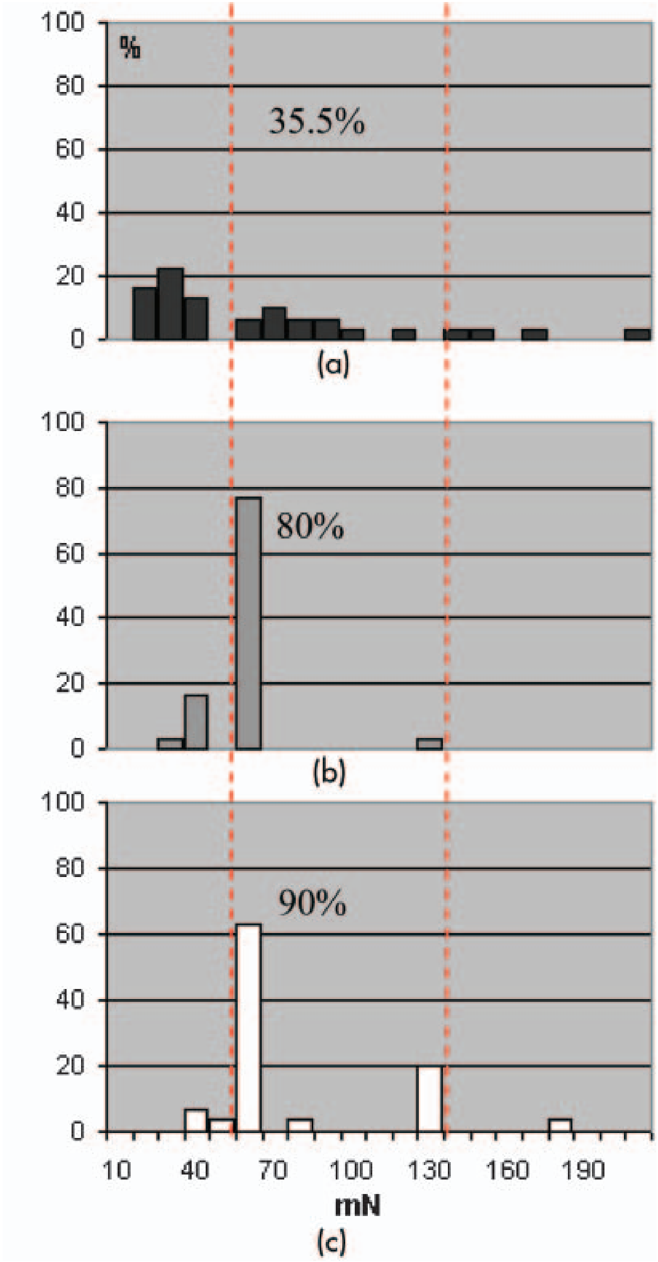

Figure 9 Histograms of MOF: a) The top figure shows directly applied force to the forceps; b) The middle figure shows force applied with visual and force feedback via MEMSurgery, the user has no knowledge about the characteristics of MOF; c) The bottom figure shows force applied with visual and force feedback via MEMSurgery and knowledge about the characteristics of MOF. The two red dashed lines indicate the range of MOF according to our experimental data. The percentage of MOF that lie in this range are 35.5%, 80%, and 90% respectively.

especially minimally invasive or robot assisted approaches that would minimize surgical trauma. However, a great deal of additional technological development, basic research on vascular properties, and translation of this basic research to clinical devices and clinical practice remains to be done.

Future developments will mainly focus on: 1) utilizing real-time medical imaging (e.g. high resolution ultrasound, optical coherence tomography etc.) and experimental mechanical methods to obtain more accurately the mechanical properties of blood vessels in vivo; 2) modeling the mechanical characteristics of the blood vessel and the interaction between the vessel and the surgical tools; 3) optimizing the shapes of the surgical tools, such that the vessel experiences less strain or stress at sensitive spots under the same deformation. In addition, these tools will be enhanced by the addition of new sensors and telemetry that allow them to report applied force without limits imposed by wires; and 4) integrating a specialized haptic device for displaying the ''squeezing forces'' between the two fingers to a surgeon [21]. This system could as well be used for interaction with a virtual environment (such as a model of a blood vessel obtained from *in vivo* data) or to display amplified forces during a teleoperated or a cooperative robot-assisted procedure. Successfully developing all of the above technologies will improve the MEMSurgery test-bed and set the stage for using MEMSurgery for future scientific investigation of the entire bypass procedure in chronic animal studies. Extensive histological examination of the vessel properties (such as damage to the endothelium, intima, and reocclusion) and the success of the bypass vessel in restoring blood flow in long-term bypass studies, will be needed to fully demonstrate the benefits of MEMSurgery test-bed. Subsequently, these instruments and surgical practices will have to be translated for clinical practice. Such translational research is always quite challenging as it involves not only technological hurdles, but also regulatory and ethical considerations. Nevertheless, the MEMSurgery technology remains promising and it has uses in vascular surgery and number of other microsurgical application, such as nerves, spine, and other delicate organs requiring precision surgery.

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