Real-Time Measurement of Blood Vessel Occlusion During Microsurgery

Christian M. Sauer, B.S., Damian H. Tomlin, M.S., Homayoun Mozaffari Naeini, M.D., Oleg Gerovichev, M.S., and Nitish V. Thakor, Ph.D.

Department of Biomedical Engineering, The Johns Hopkins School of Medicine, Baltimore, Maryland

ABSTRACT
Measurement and feedback of vascular properties during microsurgery is generally not available. We carried out real-time in vivo measurement and analysis of microsurgical occlusion of 1–2-mm diameter arteries and veins in rodents. A pair of forceps mounted with strain gauges was designed for applying and directly measuring the force on tissue. Forces between 0 and 450 mN were applied, with the device having a resolution of 0.5 mN. We performed in vivo experiments on the rat femoral (\( n = 5 \)) and abdominal (\( n = 8 \)) blood vessels to measure the elastic restoration force of the tissue in response to radial compression at different levels of force. On average, the minimum occlusion force was 57 mN for the rat artery. During steady application of force, the perturbations in the blood vessel due to heartbeat are visible in the force data. These force oscillations ranged between 1 and 3 mN around the mean steady-state force applied. It was determined that the magnitude of the Fourier spectral peak corresponding to heartbeat frequency can be used as a measure of the patency of the blood vessel, and can provide feedback to microsurgeons to avoid damage to the vessel by application of excess force. Comp Aid Surg 7:364–370 (2002). ©2003 Wiley-Liss, Inc.

Key words: force feedback; forceps; vascular compression; minimum occlusion force (MOF); pulsatile; surgical augmentation

OBJECTIVE
The tactile and mechanical properties of tissue hold a wealth of information about the physiology and health of that tissue. Surgeons rely a great deal on intuition gained by the “feel” of tissue (a combination of visual and tactile feedback) while performing operations and diagnosing disease.\(^1\)\(^2\) However, as the size of the operating area is reduced, the feedback a surgeon receives from a tissue decreases. During many microsurgical procedures, the blood vessels must be occluded to halt blood flow through the exposed area.\(^1\)\(^2\) Excess force may unintentionally be applied when performing delicate operations such as closure of these vessels or exposure of the vascular endothelial surface (both necessary precursors to several procedures), and this excess force may cause tissue damage, one of the major factors affecting the surgical outcome.\(^3\)\(^-\)\(^5\) Numerous earlier investigations reporting blood vessel damage due to excess force motivated us to develop a microsurgical assistant. In essence, it is a device that provides additional feedback about the blood vessel patency to the surgeon.

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Address correspondence/reprint requests to: Nitish V. Thakor, Johns Hopkins University, Traylor 701, 720 Rutland Avenue, Baltimore, MD 21205. E-mail: nthakor@bme.jhu.edu
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Previous experiments explored the mechanical properties of blood vessels.\textsuperscript{6-9} Other studies have used those results to determine the minimum occlusion force (MOF), the force at which the blood vessel is held closed (with no excess force being applied). This is of particular importance because microvascular clamp procedures can inflict considerable endothelial damage and create a concomitant threat of postoperative thrombosis at the lesion site, as shown by several scanning electron and light microscopy studies.\textsuperscript{6,10} These studies showed that the initial changes in the endothelium occurred in smaller radii of curvature, and that the possibility and extent of damage is directly related to the applied clamping force.\textsuperscript{3-6,10} However, the total view of the tissue \textit{in vivo} has been neglected, as have possible feedback mechanisms for informing the surgeon about the status of the vessel. Possible applications of intraoperative deformation monitoring include cardiovascular surgery, cerebral revascularization, and plastic/reconstructive procedures.

To be of any use to a surgeon, real-time calculation of the MOF or closure status of a vessel must be performed. To address this need, a force-sensitive instrument that has the potential for real-time user feedback was developed in this study and used to determine blood vessel properties.

\textbf{MATERIALS AND METHODS}

\textbf{Strain-Sensitive Forceps}

Semiconductor strain gauges are devices that vary in resistance as strain is applied to them. In our instrument, two strain gauges are installed on a pair of forceps, with one on either side of one leg of the forceps to measure the strain on that leg when either opening or closing the tool. We can treat the forceps leg as a cantilever beam configuration and use the appropriate mechanical model. The gauge on the lower surface is located precisely under the gauge on the upper surface; they measure bending strains that are of equal magnitudes but of opposite signs.

The strain gauges were interfaced to a Wheatstone bridge circuit. Any strain in the forceps causes the bridge to become unbalanced. The voltage output of the bridge is amplified and sampled for real-time data acquisition and analysis. Resistance changes of the same sign produced by the axial loads are canceled because the two active gauges are in the adjacent arms of the Wheatstone bridge. Likewise, any resistance changes of thermal origin will be negated when both gauges and the specimen experience the same changes in temperature. Figure 1 shows the finished forceps with attached strain gauges.

The calibration of the instrument was accomplished in two phases. The first phase involved measuring the response of the tool to a known force. Known weights were suspended from the leg of the forceps mounted with strain gauges, and a calibration curve was obtained (Fig. 2a). From this curve, which is essentially linear, a coefficient was determined to allow the output voltage to be converted into force applied.

The second phase of the calibration was the study of an artificial vessel. The forceps were used to apply force to artificial vessels while fluid was being pumped through them. The fluid was circulated with a pulsed-flow pump (VWR Scientific Products, West Chester, PA), simulating the pulsations that would occur in real arteries. This testing was necessary to determine the proper placements for other instruments in relation to the forceps. It also gave some insight into the situation that would occur during an actual experiment. The procedure developed with these artificial vessels was identical to that later used for the \textit{in vivo} measurements. The resulting data were used to determine the performance of the forceps \textit{in vivo} with real blood vessels.
Surgical Procedure

As *in vivo* experimental models, we used 13 Wistar rats weighing 250–350 g. Animal care and experimental and surgical procedures followed the approved guidelines of the Institutional Animal Care and Use Committee (IACUC) at the Johns Hopkins University. Anesthesia was given as an intraperitoneal bolus injection of natrium pentobarbital (0.25–0.35 ml) and heparin i.v. (5000 IU/kg) was injected. Body temperature was maintained at 37 ± 0.5°C. The rats were randomly divided among two equal groups of experiments.

In the first group (Group I), 1 cm of the femoral artery and vein was identified and isolated by making a small incision and retracting the muscles on the left femoral area of the rats after shaving and disinfecting the area. In the second group (Group II), the abdominal aorta and inferior vena cava were identified and isolated a few mm below the renal arteries by opening the abdominal wall after shaving and disinfecting the area.

In all cases, the forceps were positioned perpendicularly on the vessel walls. The mechanical stress on the vessel walls was minimal, but isolation of the vessels was unavoidable and slight damaging effects must be taken into account. It was expected that the internal endothelium would not be damaged during closure because the acting forces were less than or equal to the MOF.

Other Tools

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

**In Vivo Measurement Technique**

The forceps were positioned on the vessel to be tested. The two remaining tools were then positioned; the laser directly above the tip of the forceps and vessel, and the flow meter proximal or distal to the forceps but directly on the vessel. The forceps were then closed in small increments, with each force level sustained for 10 s (to allow the vessel to settle). The increments were controlled using a computer-controlled linear positioning stage (Physik Instrumente GmbH & Co., Waldbronn, Germany). Figure 2b shows the setup: the forceps are labeled A, the flow sensor is labeled B, and the robotic stage is labeled C. Flat forceps were used to avoid vessel scission. Measurements at several steps above the MOF of the vessel were taken to ensure that proper data were collected. The force was then reduced in the same steps. The results were reproducible. The statistics on the vessel properties and the MOF are shown in Table 1.

**In Vivo Arterial Pulse Measurement**

Arterial walls pulsate with each heartbeat. The vessel wall thus exerts a small force on devices, such as clamps or forceps, that are in contact with it. This force varies directly with the degree of occlusion of the vessel, increasing at first as force is applied to the vessel, then decreasing as the vessel...
is slowly occluded. By monitoring these force perturbations, we can determine the exact point at which the blood vessel closes. This forms a critical link between the data collected in the study and clinical applications.

The heartbeat rate in rats was observed to be 300–450 beats/min. Thus, the perturbations in the arterial walls occur at frequencies between 5 and 7 Hz (in our study, 6 Hz was found to be most common). In the experiment, we looked for these force perturbations by performing a spectral analysis on the force data. By filtering the resulting frequency information and isolating the 5–10 Hz band, the peaks due to heartbeat pulsations were determined. The relative magnitude of the spectral peak in this band was correlated with the vessel closure.

RESULTS

The applied force as recorded by the forceps is given in Figure 3. The force was applied in levels, with each force value being held for approximately 10 s. The large jump in force seen at approximately 100 s corresponds with the vessel being fully occluded. Figure 4 shows a magnification of certain levels of force from Figure 3 (10 and 15 mN represent the forceps just barely touching the vessel, while 85 mN occurs when the vessel is nearly occluded. Also note that these levels have been normalized around their average value to show only the oscillations). These force oscillations display a large variation from one level to another. These variations are caused by the heartbeat pulsations of the arteries and have been noted in other force studies. Figures 5a (artery) and 5b (vein) show displacements that resulted from the force applied to the vessels. These graphs show the average forces and displacements over the 10 s that each force level was held. Note that displacement is negative. It was measured from a base point, defined as the point where no force was applied and the vessel is fully open. As the vessel was closed, it moved away from the origin, giving a negative displacement. The curve shows two parts: the force applied to compress and fully close the vessel

<table>
<thead>
<tr>
<th></th>
<th>MOF (mN)</th>
<th>Diameter (mm)</th>
<th>Trials (animals)</th>
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<tr>
<td></td>
<td>Mean</td>
<td>Std. dev.</td>
<td>Mean</td>
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<tr>
<td>Femoral artery (Group I)</td>
<td>57.3</td>
<td>12.8</td>
<td>.61</td>
</tr>
<tr>
<td>Abdominal aorta (Group II)</td>
<td>70.6</td>
<td>9.9</td>
<td>1.13</td>
</tr>
<tr>
<td>Inferior vena cava (Group II)</td>
<td>49.3</td>
<td>6.0</td>
<td>1.34</td>
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Fig. 3. Temporal force profile as applied by forceps to the rat abdominal aorta. Force is increased until the vessel is occluded (at approximately 100 s) then decreased. Force levels are held for approximately 10 s each.

Fig. 4. Oscillations at several levels of force in the rat abdominal aorta. The graph is the magnification of several levels of applied force from the temporal profile. The oscillations are shown with the static background force removed.
(loading, A curve) and the force as the vessel is slowly released to open again (unloading, B curve).

Each force level was sustained for 10 s to allow enough data to be gathered to perform a Discrete Fourier Transform (DFT) on each level. It was found that a 2048-point DFT gave the optimal result. A DFT is a transformation that reveals the composite frequencies of signals and their relative magnitudes. In short, it reveals the frequency composition of the signal. As a frequency becomes more dominant in a signal, the magnitude of its peak (the power of that frequency) will increase. This transform was performed using the FFT algorithm implementation in MATLAB (Release 11). With a sampling rate of 500 Hz, the minimum time to gather enough data on each level was slightly over 4 s. Each level was sustained for a longer time to obtain a “steady-state” set of points. The DFT of each level shows much noise on the entire spectrum. When the range of interest (the frequency of rat pulse, from 5 to 10 Hz in the DFT plot) is isolated, only one major peak exists (Fig. 6). This spike indicates the presence of heartbeat oscillations, with its magnitude signifying the power of the perturbations.

The resulting force vs. perturbation power graph is slightly noisy. Taking several trials on a single blood vessel and averaging the results eliminated most of this noise. Figure 7a shows the change in displacement and power as the force is increased and the vessel is closed. Figure 7b shows the same data as the vessel is released and returns to open status. As the force increases, the DFT magnitude increases at first and then drops off. As force decreases, the reverse happens. As expected, the two graphs look very similar: Group I (experiments involving the femoral artery) and Group II (experiments involving the abdominal aorta) show similar behaviors. However, in the case of Group

Fig. 5. Displacement–force dependency for rat blood vessels: three trials on one animal (A = loading curve with force increasing to the right; B = unloading curve with force decreasing to the left). Displacement is measured from the initial resting state of the vessel, hence the negative values. (a) Displacement–force dependency for rat femoral artery. This graph shows very little hysteresis, revealing the elastic nature of the artery. (b) Displacement–force dependency for rat femoral vein. A large degree of hysteresis is shown.

Fig. 6. Calculated DFT of the 25 mN force level. A shows the entire 250-Hz spectrum of data collected. B is a magnification of the 5–10 Hz range. The vertical scale has been preserved between the graphs.
II. the oscillations in force were of a much greater magnitude. This is to be expected, considering both the larger size of the abdominal aorta and its much greater proximity to the heart.

DISCUSSION

The delicate nature of the endothelium predisposes it to damage from force-applying agents such as clamps and forceps. Vessel trauma caused by vascular clamps ranges from nonnuduing ultrastructural changes to endothelial cells to gross injury resulting in both endothelial cell and smooth muscle necrosis. In small arteries (≤1 mm in diameter), vascular clamping has been shown to result in endothelial cell denudation and damage to the underlying smooth muscle cells of the vessel matrix.12

The application of force to a blood vessel is unavoidable and necessary during any involved surgical procedure. In most cases, this force is applied to decrease or stop the amount of blood flow in a vessel. Theoretical calculation of the MOF in these cases is of little benefit to the surgeon. The MOF of a vessel varies according to several vessel properties (diameter, age, blood pressure, wall thickness, shape, blade contact area, elasticity, and previous and current pathology). Thus, simply determining a value or formula model for the force needed to occlude a vessel has limited practical application without a detailed study of the chosen vessel. The average MOF for the femoral artery was approximately 57 mN (see Table 1).

Such small forces mean that surgeon will be unlikely to hold a single force value to keep the vessel closed during the procedure. It is more likely that the surgeon will apply a much greater force than needed, even when aware of the MOF. However, if a real-time feedback system is used that notifies the surgeon when that point has been reached, it becomes possible for the surgeon to stay within a small range of applied force. This allows the surgeon to keep blood flow in the vessel to a minimum and not damage the tissue.

The mechanical properties of blood vessels show that they are inelastic when taken as a whole object. It can be seen from the applied force graph that the vessels exhibit stress relaxation. That is, when held at a constant displacement, the force on the tissue slowly decreases (this can be seen on the highest level of force in Fig. 3). This is an important mechanical property that occurs in all inelastic materials. The second indicator comes from the vessels’ force–displacement relationship graph (Fig. 5a and b). The vessel follows two different curves depending on the loading or unloading of the vessel. This is hysteresis and is an important characteristic of the tissue. It is also important to note the difference in the graphs for the artery and vein: besides requiring more force to close, the artery is much more elastic. The two force–displacement curves of the artery are very close, indicating only a small amount of hysteresis, while the vein demonstrates this property to a much greater degree. The two paths are very disparate. The arteries are expected to be much more elastic, as they have thicker walls and are required to handle a greater load of pressure, while the veins are much more compliant with thinner walls and show this in the two separate paths taken during the procedure.

As described earlier, the surgeon could benefit greatly by knowing the force applied to the vessel and the closure status in real time, allowing a corrective manipulation of the instrument. Future work will focus on finding ways of delivering this
information to the user. An important consideration is the presentation of the deformation of the vessel to the surgeon in the most effective manner. The most obvious method is visual display, which requires the surgeon to look away from the operating site constantly to receive real-time information. A more complex system would supply haptic force feedback to the surgeon, augmenting the tactile feel by amplifying milliscale forces applied to the vessel. Integration of force feedback at the user input stage would provide real-time precision control and compensation for the surgeon during microsurgery, ultimately benefitting the patient.

CONCLUSION

While during many surgeries it is necessary for a surgeon to close a blood vessel, with the help of innovative instruments it may be possible to limit the amount of damage caused during this procedure. We have created an instrument that gives real-time measurement of some of the properties of blood vessels in vivo. With this instrument, we have measured the MOF of several blood vessels in rodents. In addition, we have found a method of determining the occlusion status of vessels. This method requires only one measurement during surgery—the force the surgeon is applying to the vessel. This measurement can then be delivered to the surgeon to increase his awareness of the operating field. Further research must be done to determine the most beneficial method of delivering the real-time data gathered by our instrument to the surgeon. Although this article only studied the vessel properties in a research setting, we see great potential for this instrument in the clinical field.

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REFERENCES