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Steerable Catheter Microcoils for Interventional MRI: Reducing Resistive Heating

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Abstract

PURPOSE—To assess resistive heating of microwires used for remote catheter steering in interventional magnetic resonance imaging. To investigate the use of alumina to facilitate heat transfer to saline flowing in the catheter lumen.

MATERIALS AND METHODS—A microcoil was fabricated using a laser lathe onto polyimide-tipped or alumina-tipped endovascular catheters. *In vitro* testing was performed in a 1.5 T MR system using a vessel phantom, body RF coil, and steady state pulse sequence. Resistive heating was measured with water flowing over a polyimide tip catheter, or saline flowing through the lumen of an alumina-tip catheter. Preliminary *in vivo* testing in porcine common carotid arteries was conducted with normal blood flow or after arterial ligation when current was applied to an alumina-tip catheter for up to 5 minutes.

RESULTS—After application of up to 1 W of DC power, clinically significant temperature increases were noted with the polyimide-tip catheter: 23°C/W at zero flow, 13°C/W at 0.28 cc/s, and 7.9°C/W at 1 cc/s. Using the alumina-tip catheter, the effluent temperature rise using the lowest flow rate (0.12 cc/s) was 2.3°C/W. *In vivo* testing demonstrated no thermal injury to vessel walls at normal and zero arterial flow.

CONCLUSION—Resistive heating in current carrying wire pairs can be dissipated by saline coolant flowing within the lumen of a catheter tip composed of material that facilitates heat transfer.

Keywords

catheter; MRI; safety; heating

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Introduction

Current applied to microcoils fabricated on an endovascular catheter tip can be used to steer a catheter within a clinical magnetic resonance imaging (MRI) scanner (1,2). The current applied induces a magnetic moment at the catheter tip, which experiences a torque in the presence of the scanner main magnetic field, producing remotely controllable deflections that can be used for navigating vessel turns and branches or holding the catheter tip in a specific orientation. The electric currents required to generate desired magnetic moments, however, can generate enough heat through resistive dissipation to cause temperatures unsafe for blood or vascular walls. Unwanted heating could also occur in the wires running down the lumen of the catheter due to field coupling with radiofrequency (RF) magnetic fields used for MR imaging.

The FDA-recommended temperature rise due to specific absorption rate (SAR) should not exceed 1°C on or in the head and 2°C in the torso and extremities (3). Beyond the well-known case of metallic implants that can cause heating hazards (4), several authors have investigated the specific case of metallic wires in the setting of interventional MRI (5-8). Most studies have shown that the specific absorption rate measured in the presence of a metallic wire can exceed the SAR limitation of 2 W/kg (9-11). Heating of conductive wires in the presence of RF magnetic fields can arise through electromagnetic field coupling to a long wire (e.g., a guide wire), known as the antenna effect (12,13). The use of microcoils for remote control of catheter tip navigation, however, requires application of direct current which results in resistive heating. Resistive heating generated by these microcoils at the catheter tip has been shown to be higher than that produced by RF heating in previous experiments (14).

In this study, we investigated heating due to Ohmic resistance in catheter deflection microcoils supplying current to the microcoils. Separate experiments were conducted to investigate RF heating (please see co-submitted manuscript). Two sets of *in vitro* resistive experiments were conducted. The first used a microcoil fabricated on a polyimide tube suspended in the center of polyflow tube so that water could flow over the coil. This configuration simulates a microcoil cooled directly by blood flow in an artery.

In the second set of experiments, a microcoil is fabricated on an alumina tube attached at the distal end of a catheter. Insulation is provided by heat shrinkable tubing that surrounds the tip and the adjacent 2 mm of the catheter. This catheter tip design insulates the microcoil from the surrounding fluid, and instead transfers heat generated by the microcoils to saline coolant flowing through the catheter lumen. This design is based on the simple theoretical calculation that 1 W of heating, sufficient to provide the deflection of the catheter required for catheter guidance in an MRI setting, causes a flow of 0.2 cc/s to rise in temperature by 1.2°C. A local increase in the temperature of blood by 1.2°C will not cause coagulation, hemolysis, or other adverse effect (15,16).

The third set of experiments consisted of preliminary testing of catheter constructs *in vivo*. A single axis solenoidal alumina-tipped catheter with luminal saline drip was navigated under x-ray guidance to the common carotid artery (CCA) of a pig via transfemoral percutaneous access. The pig was moved to a 1.5T clinical MR scanner and imaging was performed with a steady state free precession (SSFP) sequence. Continuous direct current was then applied to the catheter microcoils for zero to 5 minutes at various points in the CCA. The catheter was then navigated to the contralateral CCA, the CCA was ligated proximal to the catheter tip (in order to achieve zero flow conditions in the artery) and the experiments were repeated. Postmortem histologic analyses of the pig CCAs were performed to assess potential thermal or mechanical damage to the arterial walls.

Methods

Microcoil fabrication

Microcoils were fabricated with lithographic technique called Laser Lathe (17), which allows non-planar surfaces such as cylinders to be patterned with feature sizes as small as 5 μm . Solenoidal microcoils for the present experiments were 5 mm long and made of copper electroplated through a mask of photoresist to form a 50 μm wide trace approximately 15 μm thick with 50 μm spaces between turns (Figure 1). The solenoid contained 50 turns. The microcoil substrates were 1 mm outer diameter (O.D.) polyimide tubes and 99.8% pure 1.2 mm O.D. alumina tubes (Ortech Advanced Ceramics, Sacramento, CA, USA). Both are chemically inert to the processing and applications conditions.

In Vitro Experimental design

Figure 2 shows the experimental set-up for testing of microcoils on polyimide. The microcoil assembly is suspended inside a 1/4" polyethylene flow tube (inner diameter, ID = 4 mm). The microcoil diameter is 1 mm and a thermocouple is attached to the coil through an electrically insulating but thermally conducting film. The exposed area of the thermocouple is covered with a thermally insulating layer to ensure that it measures the temperature of the coil, and not that of the coolant.

The experiment differs in two respects from the situation that would be encountered by a catheter tip in an artery. First, the polyethylene tube is insulating whereas a blood vessel wall would be able to remove heat. Second, water and blood have slightly different thermal conductivity (0.6 W/m-K vs. 0.5 W/m-K, respectively).

In an effort to transfer heat generated by the microcoils to saline coolant flowing in the catheter and to allow minimal heat to flow radially outwards into the surrounding fluid, microcoils were then fabricated onto alumina (instead of polyimide). Alumina has 150 times higher thermal conductivity than polyimide, which allows heat to be transferred in ~10 milliseconds to the entire thermal mass of the alumina despite the fact that the wall thickness of alumina tube is several times greater than that of polyimide (18). Transfer to the coolant occurs by diffusion at the inner wall of the tube and by convection in the bulk of the coolant. Finally, since heat is being removed by the coolant, an insulating layer can be interposed between the coil on outer surface of the alumina and the blood. This structure is shown in Figure 3. Aluminum oxide (alumina) has a low magnetic susceptibility (X_m) of $-3.7 \times 10^{-5} \text{ m}^3/\text{mol}$, and thus is also an appropriate material for use in the interventional MRI environment.

A microcoil was fabricated on a 1.3 mm O.D., 0.41 mm I.D. 99.8% alumina tube and a thermocouple was placed on the coil as in the previous experiment. Heat shrinkable tubing was slipped over the tube/thermocouple and an abutted 2.8 Fr catheter and then shrunk in place by heating. The heat shrink tubing shrinks in diameter as much as a factor of two so the alumina and the catheter do not have to have the same outer diameter. The final O.D. of the catheter tip was measured at just under 2.0 mm including heat shrink tubing. A second thermocouple was placed 1 mm beyond the face of the alumina tube to measure the temperature of the effluent. Saline flow was provided by a commercial saline intravenous drip bag elevated 1.3 meters above the microcoil-tipped catheter.

Microcoil-catheter construction

Commercially available vascular infusion microcatheters (Tracker-18 and Renegade-18, Boston Scientific, Natick, MA, USA), were used as substrates for the microcoil-catheter technology. The catheters had no ferrous components and ranged in size from 2.5 F to 3.0 F

at the catheter tip, which are standard sizes used for cerebrovascular and peripheral vascular interventional procedures. All catheters were 150 cm in length.

Catheters were then assembled by running the two 0.127 mm (0.005 in.) wires through the entire length of the microcatheter inner lumen¹. Then, 0.0508 mm (0.002 in.) wire was soldered to the 0.127 mm wire at the catheter tip, and was fed through holes that had been drilled in the polyimide tube near the coil connection pads. The 0.0508 mm wires were then soldered to the coil connection pads. The microcoil assembly was next glued to the tip of the microcatheter and secured with heatshrinkable tubing (Raychem, North Spring, TX, USA). In the case of the microcoil on alumina the heat shrink tubing covered all but the distal end of the microcoil assembly while in the case of the microcoil on polyimide it covered only the proximal end of the microcoil assembly leaving the microcoil in direct contact the surrounding fluid. The 0.127 mm copper wires at the microcatheter hub were connected to a 5-pin or 8-pin electrical connector (a male telephone or Ethernet jack). The jack was plugged into a female connector on an electrical switch box (used to change the electrical polarity of the DC currents), which was connected to up to 3 DC power supplies.

Testing was performed in the above-described phantoms using a 1.5 Tesla clinical scanner (Achieva, Philips Medical Systems, Best, The Netherlands) without imaging.

Catheter Preparation for In Vivo Testing

A 2.7F Tracker-18 microcatheter (Boston Scientific, Fremont, CA) served as the substrate. A solenoid of 75 turns was created from 0.0015 inch copper wire on the outer surface of a 1.3 mm diameter alumina tube, with wire wound into thermal epoxy for adherence. Final layers of epoxy and heat shrink were applied over the solenoid, making the final outer diameter of the catheter approximately 2.0 mm. Two 0.005 inch copper lead H-poly insulated wires were pulled through the 150 cm catheter. Lead wires were attached to microcoil leads (0.0015 inch diameter copper wire) proximal to the catheter tip. Coil lead wires were 2.5 inches and 3.25 inches long, separating the electrical connection point and keeping the central catheter lumen patent. The leads were pulled through a modified Thuoy-Borst Y-adaptor at the microcatheter hub and were subsequently attached to the catheter. Power leads from the phone jack power adaptor center two leads were brought through the center bore of the Y-adaptor, allowing saline to be infused through the side port. The total resistance of the assembly was 9 ohms.

In Vivo Testing at 1.5 Tesla

Under a protocol approved by the UCSF institutional committee on animal care and research, a 30 kg farm pig served as a test subject to investigate potential thermal damage from resistive heating (or RF heating, see accompanying manuscript) related to use of the microcoil-tipped catheter. After fasting for 12 hours, the pig was sedated with acepromazine (1.1mg/kg) followed 30 min later by ketamine (20-30mg/kg), intubated and anesthetized with 1.5%–2.5% isoflurane and 2%–3% nitrous oxide, 1% oxygen, and for both surgery and imaging. Respiration was assisted with an artificial tidal volume of 15 ml/kg and a frequency of 15 breaths/minute. All monitoring and anesthesia equipment was MR-compatible, thus allowing continuous care of the animal in both the MR and X-ray components of the multimodality imaging suite.

¹To make this system useful for MR-guided interventions such as coil embolization and particle delivery, an alternative to the above-described intraluminal copper wire design has been tested. Although we utilized non-braided, non-ferrous microcatheters in the experiments reported here, we have used microcatheters with copper wires embedded in the catheter wall (Modified Polymer Compoents, Sunnyvale, CA) while still maintaining the small microcatheter caliber. Implementation of this open-lumen design is also key in allowing saline flush to be used as a coolant, as well as permitting coils, particles, drugs, and other materials to be delivered through the catheters.

Under aseptic conditions, a 9F sheath was inserted into the femoral artery percutaneously followed by 50 IU/kg heparin. A 9F guide catheter was navigated over a guidewire to the proximal left CCA under x-ray guidance. The coil-tipped microcatheter construct was then inserted through the guide catheter and advanced to the distal CCA. The animal was then transported to the MRI scanner. Because UCSF has a combined X-ray/MR suite, animals can transition from the X-ray angiographic suite to the MR imaging suite on a single sliding table. Imaging and catheter heating experiments were then performed in the left CCA as outlined below.

The pig was returned to the x-ray angiography suite and the proximal right CCA was catheterized with the 9F guide catheter. The coil-tipped microcatheter construct was then inserted through the guide catheter and advanced to the distal right CCA. The right CCA at the tip of the 9F catheter but well proximal to the microcoil-tipped catheter tip was then ligated with a silk suture in order to obtain zero flow in the CCA. The animal was then transported to the MRI scanner. Imaging and catheter heating experiments were then performed in the left CCA as outlined below.

In Vivo Imaging and Heating Experiments

Testing was performed using a 1.5 Tesla clinical MR scanner. Microcatheters were tested at six positions in each CCA, with the most distal location being the first test point, and each subsequent test point separated by a 2 cm manual pull back of the catheter to ensure an adequate margin between test points in case thermal damage occurred at any given test point. As outlined in Table 1, The initial 3 test points in each artery were directed at resistive heating caused by running current (300 mA for 30 seconds, 1 minute, or 5 minutes) through the catheter microcoils during imaging and the subsequent 3 test points in each artery were directed at RF heating in catheters receiving no current by just being imaged (0 mA for 30 seconds, 1 minute, or 5 minutes). Imaging was performed with a SSFP sequence (TR = 5.5 ms, TE = 1.6 ms, flip angle = 30°, 128 × 128 matrix, 5-6 mm slice thickness, SAR = 3.7 W/kg), with real-time imaging at 3 - 5 frames per second. Left CCA experiments were at physiologic arterial flow; right CCA experiments were at arterial stasis. For all experiments, room temperature normal saline solution was perfused through the central lumen of the microcatheter at a drip rate of approximately 0.1 cc per second, based on the results of *in vitro* experiments.

Euthanasia and Postmortem Gross and Histologic Analyses

At the conclusion of the surgery and imaging, the guide catheter and microcoil-tipped catheter were removed. The animal was maintained under anesthesia for an additional two hours in order to allow time for histologic changes from potential thermal damage to mature. The animal was then euthanized by the injection of pentobarbital (200 mg/kg IV), saturated KCl (2ml/kg IV), and bilateral thoracotomy to collapse both lungs, consistent with the recommendations of the Panel on Euthanasia of the American Veterinary Medical Association. The common carotid arteries were then excised en bloc, opened longitudinally with a scalpel for gross examination, fixed in formaldehyde solution, embedded in paraffin embedded in paraffin, sectioned (4 µm thick) and stained with hematoxylin and eosin and Masson trichrome.

Results

The results of the first set of experiments using microcoils on polyimide are shown in Figure 4. The temperature rise of the coil at three flow rates, 1 cc/s and 0.28 cc/s and 0 cc/s, is plotted against power dissipated in the microcoil. At zero flow, the temperature of the coil rises 23°C/W; at 0.28 cc/s the temperature rises at 13°C/W; and at 1 cc/s the temperature

risks at 7.9°C/W. Each data point represents the long-time asymptote of a temperature vs. time curve as the system equilibrates at each new power level. Recalling that we want to use on the order of 1 W of power for tip deflection and that 2°C is the maximum temperature rise to which blood and tissue should be exposed, it is clear that blood or tissue should not be in direct contact with the microcoil on polyimide.

To facilitate heat transfer into fluid flowing through the tube, polyimide was replaced by alumina as the microcoil substrate. The high thermal conductivity of alumina allows passage of heat through the tube wall with only a small thermal gradient. The results of the flow experiments using an alumina catheter tip are summarized in Figure 5. Three flow rates were used: 0.12 cc/s, 0.25cc/s, and 0.48 cc/s. At 0.12 cc/s and 0.96 W, the effluent temperature rise is 2.3°C, about as expected while the microcoil and the ceramic outer surface of the catheter rose about 7.6 °C. The heat shrinkable tubing surrounding the microcoil vastly reduces heat flow into the surrounding fluid compared with the case of direct contact between that fluid and the copper microcoil. Higher flow rates reduced the effluent and outer surface temperatures proportionately. Even at the highest flow rates in the polyimide experiment, 1 W power dissipation would expose blood to a surface 8°C above body temperature, while at the lowest flow rate in the experiment using alumina, 1 W power dissipation would only expose blood to coolant effluent at 2.3°C above body temperature.

Under the conditions tested, no thermal or mechanical damage to the catheterized porcine common carotid arteries was detected by gross examination or histologic analysis. Under conditions of zero arterial flow, platelet and fibrin coagulum was detected adherent to the endothelium of the ligated artery or detached in the arterial lumen in 3 of 6 samples (Table 1). Under conditions of normal arterial flow, however, no histologic damage or coagulum was identified.

Discussion

Microcoils fabricated on polyimide, as described above, are effectively thermally isolated from any coolant that might be inside the tube. For a 1 mm diameter polyimide tube, wall thickness is typically 100 microns and its thermal conductivity is ~ 0.15 W/m-°C. Since the wall thickness is very small compared to the tube diameter, and the length of the tube is large compared to the diameter, the heat flow across the tube wall is the same as the heat flow across a flat sheet of the same length as the tube, width equal to the circumference of the tube and thickness equal to the tube wall thickness, i.e.

$$J = -\kappa \cdot \text{grad}T \quad \text{becomes} \quad J = 1/\pi d = \kappa \Delta T / x$$

where the heat flux, J (W/m²), is 1 W divided by the surface area of the microcoil, πd , and $\text{grad}T$ is just the temperature difference between the outer (hotter) and inner (cooler) tube surfaces, ΔT , divided by the tube wall thickness, x . ΔT is the only unknown in this equation.

In the absence of heat flow into the fluid outside the tube, the temperature difference that would be required to sustain 1 W of heat flow across the tube wall in our coil geometry is ~ 45 °C. In other words, the coil itself would have to be 45 °C hotter than the coolant inside the tube. Thus, the fluid outside the tube in direct contact with the microcoil must dissipate the heat produced by Ohmic losses in the coil. But as the results above show, the microcoil surface will nevertheless be too hot.

The situation improves greatly with the use of alumina tubing (thermal conductivity ~ 22 W/m-°C). The wall thickness is not small compared to the tube diameter but since thermal power crossing the inner boundary of the tube has to be the same as that coming from the

coil at the outer boundary, the heat flux and the temperature difference between these surfaces depends only on the logarithm of the ratio of the outer to the inner radius. Now the result of dissipating 1 W at the outer surface of the alumina tube is a temperature drop of $\sim 1.5^{\circ}\text{C}$ at the inner surface. This temperature difference is biologically tolerable.

The temperature difference between inner and outer surfaces of the microcoil tube can be reduced further if desired. First, alumina tubes can be fabricated with thinner walls. Second and more significantly, alumina can be replaced by aluminum nitride. This material has a thermal conductivity in excess of $140\text{ W/m}\cdot^{\circ}\text{C}$ and, like alumina, is an excellent electrical insulator. Thus, another factor of 5-10 reduction in temperature drop across the tube wall is possible.

The heat transfer experiments reported here used solenoidal microcoils in which currents circulate around the catheter axis and the resulting magnetic moment points along the catheter long axis. In a practical MR-guided catheter, independent coils with magnetic moments on two or three orthogonal axes are required. For example, the microcoil in Figure 6 is a helical coil with magnetic moment perpendicular to the catheter axis. In order to limit the length of the rigid catheter tip, it is desirable to stack coils radially rather than fabricate them side by side along the length of the catheter. For good heat transfer it is desirable to provide a low thermal resistance interleaved dielectric between stacked coils.

The virtue of the alumina design presented here is that the outermost coil is insulated from the blood by the outer layer of polymer tubing. Thus, the outermost coil could be at elevated temperature, sufficient to sustain the required heat flow through the interlayer dielectric, without significantly raising the temperature at the outside of the shrink tubing. That is to say, the thermal impedance of the shrink tubing (the product of the thermal resistivity and the thickness of the tubing wall) can be large compared with the sum of the thermal resistances of the two interlayer dielectrics and that of the ceramic tubing substrate. A typical interlayer dielectric might be $25\text{ }\mu\text{m}$ thick. In the worst case, the thermal resistivity of that layer might be comparable to polyimide, i.e. the layer is a polymer. Then the condition that thermal impedance for heatflow inward be small compared to impedance for heatflow outward into the blood amounts to the requirement that the thickness of the shrink tubing be greater than $50\text{ }\mu\text{m}$. This condition is easily satisfied without unduly increasing the outer diameter of the catheter tip.

Using a clinically achievable saline coolant flow rate based on *in vitro* results preliminary *in vivo* testing demonstrated no thermal injury to vessel walls at normal and zero arterial flow. Although tissue temperature was not measured directly, the most important parameter in ultimate clinical application, is the likelihood of thermal injury to tissues by the catheter either from resistive heating while current is applied for catheter navigation or RF heating while the catheter is being imaged in the MR scanner (see accompanying manuscript for further discussion). The lack of apparent thermal injury is encouraging, and will serve as the basis for further *in vivo* testing under different scenarios of catheter navigation.

It is important to note that in actual clinical use, it is likely that the catheter tip coils will only be energized for a few seconds to achieve catheter tip deflection, followed by manually advancing the catheter or guidewire into a target vessel. With currents in the 200-400 mA range, catheter tip deflection angles between 20 and 40 degrees have been achieved at 1.5 Tesla with the catheter configurations described. This tip deflection occurs virtually instantaneously with current activation (within 1 second) and persists as long as the current is active. The experiments outlined in this manuscript are designed to simulate the maximum amount of heating if the catheter were activated for a long period of time, in order to determine how long currents can safely be activated.

Conclusion

These experiments are instrumental in the development of a catheter tip design that protects the body from exposure to surfaces heated by the electrical currents required to deflect the tip as it is steered through the vascular system. The use of saline coolant and alumina to facilitate heat transfer are feasible options not only for this microcoil-catheter construct but potentially for other interventional MRI coil-catheter designs that use microcoils for imaging or tracking.

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Figure 1. Image of microcoil solenoid (50 turns) fabricated with lithographic technique called Laser Lathe, 5 mm long, and made of copper electroplated through a mask of photoresist to form a 50 μm wide solenoidal trace approximately 15 μm thick with 50 μm spaces between turns, on a 1 mm O.D. polyimide tube.

Figure 2.

Schematic diagram depicting experimental set-up for testing microcoils on polyimide. The microcoil assembly is suspended inside a 1/4" polyethylene flow tube (ID = 4 mm). The microcoil diameter is 1 mm and a thermocouple is attached to the coil through an electrically insulating but thermally conducting film. The exposed area of the thermocouple is covered with a thermally insulating layer to ensure that it measures the temperature of the coil.

Figure 3. Schematic depicting microcoil solenoid wound on alumina substrate, attached to catheter tip using Raychem shrink tubing (which also acts as an insulator shielding blood from the heated coil). A second thermocouple 1 mm downstream of the end of the alumina tube is not shown.

Figure 4.

Temperature vs. power dissipated in the microcoil on a polyimide substrate, at four flow rates: 1.26 cc/s, 1 cc/s and 0.28 cc/s and 0 cc/s. Each data point represents the long-time asymptote of a temperature vs. time curve as the system equilibrates at each new power level.

Figure 5. Temperature vs power dissipated in the microcoil on an alumina substrate, at three flow rates: 0.12 cc/s, 0.25cc/s, and 0.48 cc/s. Temperatures at the outer surface of the ceramic are given, as well as that of the effluent 1 mm beyond the end of the alumina. Flow = effluent temperature. Ceramic = temperature of alumina.

Figure 6.
A Helmholtz or “saddle” shaped microcoil wound on the outer wall of a catheter, which creates a magnetic moment perpendicular to the catheter’s long axis.

Table 1

Gross and histologic findings in porcine common carotid arteries after testing of alumina-tipped microcatheters with various durations of bFFE imaging and microcoil current application

Current to Microcoils (mA)	Duration of Current to Microcoils (sec)	Duration of bFFE Imaging (sec)	Gross and Histologic Findings
Left CCA: Normal Arterial Flow			
300	30	30	No thrombus, no damage to arterial wall
300	60	60	No thrombus, no damage to arterial wall
300	300	300	No thrombus, no damage to arterial wall
0	0	30	No thrombus, no damage to arterial wall
0	0	60	No thrombus, minimal medial vacuolization likely within normal limits
0	0	300	No thrombus, poorly preserved specimen limits histologic evaluation
Right CCA: Zero Arterial Flow			
300	30	30	No thrombus, no damage to arterial wall
300	60	60	No thrombus, no damage to arterial wall
300	300	300	No thrombus, no damage to arterial wall
0	0	30	Platelet and fibrin thrombus adjacent to endothelium, no damage to arterial wall
0	0	60	Platelet and fibrin thrombus adjacent to endothelium, no damage to arterial wall
0	0	300	Platelet and fibrin thrombus detached from endothelium, no damage to arterial wall