Evaluation of the effect of catheter on the guidewire motion in a blood vessel model by physical and numerical simulations

Kazuto TAKASHIMA*, Atomu OIKE*, Kiyoshi YOSHINAKA**, Kaihong YU***, Makoto OHTA***, Koji MORI**** and Naoki TOMA*****

*Graduate School of Life Science and Systems Engineering, Kyushu Institute of Technology
2-4 Hibikino, Wakamatsu-ku, Kitakyushu, Fukuoka 808-0196, Japan
E-mail: ktakashima@life.kyutech.ac.jp
**National Institute of Advanced Industrial Science and Technology
1-2-1 Namiki, Tsukuba, Ibaraki 305-8564, Japan
***Institute of Fluid Science, Tohoku University
2-1-1 Katahira, Aoba-ku, Sendai, Miyagi 980-8577, Japan
****Graduate School of Sciences and Technology for Innovation, Yamaguchi University
2-16-1 Tokiwadai, Ube, Yamaguchi 755-8611, Japan
*****Graduate School of Medicine, Mie University
2-174 Edobashi, Tsu, Mie 514-8507, Japan

Abstract
Various catheter simulators using a computer or blood vessel biomodels have been developed for training of medical students and young physicians. Moreover, we have developed a system to simulate a guidewire in blood vessels that uses both numerical analysis and experimental observation for the quantitative analysis of treatment technique, surgical planning, intra-operative assistance, and to facilitate the design of new guidewires. However, not limited to our group, there is a lack of studies evaluating the motions of both the guidewire and catheter and their interaction in a blood vessel. Therefore, in the present study, we modified our computer-based system and experimental apparatus to evaluate the catheter motion and compared both results.

First, we added a mechanism to the experimental apparatus to move and evaluate the catheter in a poly (vinyl alcohol) hydrogel blood vessel model. Second, we added a new catheter model to the calculation. We subsequently evaluated the behaviors of the medical devices (the guidewire and the catheter) by measuring the three-dimensional position and the contact force between the medical devices and the vessel wall. Comparison of the calculation and experimental results showed that the trajectories and the contact forces of both the experimental and numerically analyzed medical devices had the same tendencies. By considering the flexibility of the catheter in both the experimental and numerical analysis methods, we could reproduce the phenomena seen in clinical situations, such as movement of the catheter during the insertion or removal of the guidewire, and movement of the guidewire during the insertion of the catheter.

Keywords: Endovascular treatment, Simulation, Guidewire, Catheter, Phantom, Numerical analysis

1. Introduction
Catheters and guidewires are used in the treatment of cerebrovascular diseases, such as ischemic strokes and intracranial aneurysms. However, the small diameter and tortuosity of blood vessels may make the procedure very difficult. Moreover, the length and flexibility of catheters and guidewires, as well as their limited number of degrees of freedom, may severely reduce the surgeon’s visual and tactile perception when manipulating these tools during the treatment. Therefore, various catheter simulators using a computer have been developed to improve safety in endovascular treatments (Otani et al., 2015; Chiang et al., 2013; Irie et al., 2012; Hirayama et al., 2011; Schafer et al., 2007; Cai et al., 2006; Lenoir et al., 2006; Bhat et al., 2005; Konings et al., 2003; Yamamura et al., 2003; Dawson et
force into a silicone aneurysm was less for the insertion at a constant speed by a motor than for the intermittent insertion was 0.05 s, and the velocity of both the guidewire and the catheter was ±2.1 mm/s. It was reported that the insertion was 23.4 mm and 30.7 mm, respectively. The maximum removal length of the guidewire was 25.5 mm. The acceleration and deceleration time was 0.05 s, and the velocity of both the guidewire and the catheter was ±2.1 mm/s. It was reported that the insertion force into a silicone aneurysm was less for the insertion at a constant speed by a motor than for the intermittent insertion.

In our previous study, we developed a computer-based system to simulate a guidewire in blood vessels using multibody dynamics method (Takashima et al., 2014, 2012, 2009, 2007, 2006). The system was able to predict the course of approach to lesions, present numerical results and animations for the quantitative analysis of treatment technique, treatment planning and intra-procedural assistance, and act as a communication tool between patients and the physician. These simulations are also expected to be useful for analyzing the structure of guidewires and may help to facilitate the design of new guidewires. In our previous study (Takashima et al., 2014), we also developed an experimental apparatus using a PVA-H blood vessel model that can be used in experiments under the same conditions as the numerical analyses to validate the results of the numerical analysis. This experimental apparatus can be used not only for validation of the numerical analyses but also as a simulation system. Comparison of the calculation and experimental results for validation of the simulation system showed that the trajectories and contact forces of the experimental and calculated guidewire tips had the same tendencies.

In our previous study (Takashima et al., 2014), we did not evaluate the catheter motion using numerical analysis and experimental observation. The proximal part of the guidewire in that study was constrained by a fixed needle mimicking a catheter to determine the initial position of the wire. During actual endovascular treatments, however, after insertion of the guidewire, a catheter with a hole is slipped over the guidewire until it reaches the lesion and is then used to perform various procedures (e.g., coil embolization, balloon angioplasty, and stent deployment). It is difficult to enhance the treatment efficiency if the medical devices are normally delivered to the lesions. For example, the position and the shape of the delivered catheter affect the catheter stability required for adequate coil embolization. In clinical situations, the position and the shape are empirically selected through a trial and error process. Therefore, it is very important to evaluate the motions of both the guidewire and catheter. Note that, not limited to our group, there is a lack of studies evaluating the motions of both the guidewire and catheter and their interaction in a blood vessel experimentally and using numerical analysis. In the present study, we added a flexible catheter to the computer-based system and the experimental apparatus, and evaluated the motions of the catheter and guidewire using numerical calculations and experimentation. We compared the calculated and experimental results from the perspectives of the contact forces and trajectories of the tips of the medical devices.

2. Methods
2.1 Experiment

The new experimental apparatus including the mechanism to move the catheter is shown in Fig. 1. The three-dimensional coordinates are also shown in this figure. We shifted the mechanism to move the guidewire using a two-axis automatic stage (Sigma Koki Co., Ltd., SGSP20-85(X), SGSP-40YAW) upward (negative y-axis direction), and added a mechanism to move the catheter under it. The proximal part of the catheter was fixed on another automatic stage (Sigma Koki Co., Ltd., SGSP20-85(X)). To fix the catheter, we used a fiber holder (Suruga Seiki Ltd., F260-2L) and attached two pipes at both ends of the fiber holder to reduce the deflection of the medical devices during the insertion experiment (left panel of Fig. 1). The outer diameter, inner diameter and total length of the upper pipe were 6 mm, 4 mm and 140 mm, respectively. The outer diameter, inner diameter and total length of the lower pipe were 2.5 mm, 0.7 mm and 100 mm, respectively. The motions of the proximal ends of the guidewire and the catheter along the y-axis are shown in Fig. 2. The maximum insertion lengths of the guidewire and the catheter were 23.4 mm and 30.7 mm, respectively. The maximum removal length of the guidewire was 25.5 mm. The acceleration and deceleration time was 0.05 s, and the velocity of both the guidewire and the catheter was ±2.1 mm/s. It was reported that the insertion force into a silicone aneurysm was less for the insertion at a constant speed by a motor than for the intermittent
insertion by the manual operation of the doctor (Nagano et al., 2010). Therefore, we will evaluate the effects of the intermittent motion in future studies.

The PVA-H blood vessel model is also shown in Fig. 1. Similar to our previous study (Takashima et al, 2014), the PVA solution was cast into an acrylic box containing a torus-shaped mold, and then PVA crystallization was promoted by freezing. After gelatinization, the mold material was removed by the lost-wax technique. The centerline of the blood vessel model port for inserting the medical devices was located on the y-axis. We used a torus-shaped vessel model with an inner diameter of 4 mm. This shape is the same as those used in our previous studies (Takashima et al, 2014, 2012, 2007, 2006). We defined the centerline by connecting a half circle of radius 4 mm. During the experiment, the
PVA-H model was filled with working fluid to decrease the reflection and refraction on the inner wall of the blood vessel model. The working fluid was a glycerin aqueous solution created to match the refractive index of the blood vessel model (refractive index: 1.447). A white acrylic resin plate was set behind the model as a white background.

Similar to our previous study (Takashima et al., 2014), two web cameras (Cam1 and Cam2, 15 frames per second, 1920 × 1080 screen resolution (Logicoool)) were positioned at the sides of the vessel model and were used to acquire images during the experiment. To measure the total force applied by the guidewire along the y-axis, three load cells (Kyowa Electronic Instruments Co., Ltd.) were also located under the blood vessel model. We used the sum of the outputs of three load cells as the total contact force along the y-axis. The tip of the catheter was located at \((x, y, z) = (1, 0, 0)\) as the initial position.

The catheter tip was straight and the guidewire tip was bent in shape (right panel of Fig. 1). The bending angle of the guidewire tip was 45°. The distance between the tip and the fixed position of the guidewire (total length) was 348 mm. The fixed position of the guidewire was initially located at \((x, y, z) = (1, -341, 0)\). The distance between the tip and the fixed position of the catheter (total length) was 25 mm. The specifications of the guidewire and catheter are summarized in Table 1.

| Guidewire (RG-GA1218s: Terumo) | Diameter: 0.30 mm |
|--------------------------------|-------------------|
| Bending angle of tip: 45°      |                   |
| Catheter (Excelsior SL-10: Stryker) | Outer diameter: 0.60 mm |
|                                  | Inner diameter: 0.42 mm |
|                                  | Tip shape: straight |

We attempted to use thinning processing to track the catheter tip as we did for the guidewire tip in our previous study (Takashima et al., 2014), but the small difference in color tone between the guidewire and catheter resulted in the two devices being extracted as one object. Therefore, we painted the tip of the catheter using a yellow oil-based paint marker, and carried out the tip tracking of the catheter using the range extraction center-of-gravity method (Fig. 3). We adjusted the color tone, chromaticity and brightness of the acquired images to make tracking the guidewire and catheter tips easier. Similar to our previous study (Takashima et al., 2014), the acquired images were assessed by binary transformation and thinning processing, and the three-dimensional position of the guidewire tip was tracked. The three-dimensional motion coordinates of the tips of both medical devices were subsequently measured using the direct linear transformation (DLT) method. We used WriggleTracker and Move-tr/3D (Library Co., Ltd.) for thinning and tracking processing, and the DLT method, respectively.

![Fig. 3 Tracking the catheter tip.](image)

**2.2 Numerical analysis**
Similar to the experiment, in the numerical simulation, we also inserted the guidewire and catheter models into the blood vessel model and calculated the contact forces and the trajectories of the tips of the medical devices (Fig. 4). The guidewire and blood vessel models, and the simulation method are similar to those used in our previous studies (Takashima et al., 2014, 2012, 2007, 2006). The vessel is a circular elastic tube, whose shape is defined by the centerline and the radii. We calculated the contact force vector between the guidewire and vessel, as well as the induced moment of each segment, and then calculated the effects of these forces and moments on other segments. The guidewire model is composed of viscoelastic springs (three degrees of freedom) and segments. \( T \) is defined as the sum of the viscoelastic forces of a mobile joint as follows:

\[
T = -K_g(q - q_o) - D_g\dot{q} \tag{1}
\]

where \( q \) is the joint displacement vector, \( q_o \) is the joint displacement vector when no loads are applied, and \( K_g \) and \( D_g \) are elastic and damping coefficients, respectively. The viscosity of the medical devices would be negligibly small comparing to that of vessel wall. For example, most parts of the guidewire consist of elastic metal alloy. But, we modelled the guidewire and catheter as a viscoelastic material in order to consider the damping effect by the vessel wall and the working fluids. We changed the damping coefficients \( D_g \) of the guidewire joint and evaluated the effects of these parameters on the contact force because this parameter was not measured and it is difficult to set an appropriate value. Similar to our previous study (Takashima et al., 2014), we neglected gravity, blood flow and the friction between the guidewire and blood vessel because the same tendency was seen in the trajectories and the contact force of both the experimental and simulated guidewire tips previously. We will consider the effects of these parameters in future studies.

In this study, similar to the guidewire model, the catheter model is composed of viscoelastic springs (three degrees of freedom) and segments, and calculated the contact force against the vessel. Moreover, we also calculated the contact force between the guidewire with an outer diameter of \( 2R_g \) and the catheter with an inner diameter of \( 2R_c \) according to the following processes. Namely, assuming the guidewire joints and the catheter segments as spheres and cylinders, respectively, we considered the contacts of the spheres inside the cylinders. The joints of the guidewire can be located in the space of the catheter using these contact forces.

At both ends of the segment of the catheter, we arranged \( n+1 \) nodes. The nodes are successively numbered from 0 to \( n \) from the proximal end of the catheter. The nearest and the second end point from a distal end point from a guidewire segment (index \( i \) were successively numbered from the proximal segment) are \( X_{c1} \) and \( X_{c2} \), respectively (Fig. 5). \( A_i \) and \( B_i \) are defined as follows.

\[
A_i = X_{gi} - X_{c1} \tag{2}
\]

\[
B_i = X_{c2} - X_{c1} \tag{3}
\]

Using \( \theta_i \) (the angle between \( A_i \) and \( B_i \)), the following equation can be obtained.
\[ \mathbf{A}_i \cdot \mathbf{B}_i = |\mathbf{A}_i||\mathbf{B}_i|\cos\theta_i \]  

(4)

Using these vectors, we judged whether \( X_{gi} \) is inside or outside the catheter segment between the nodes \( X_{c1} \) and \( X_{c2} \). First, we calculated \(|\mathbf{A}_i|\cos\theta_i\) and the following three cases were considered according to the position in relation to the centerline.

(a) \( 0 \leq |\mathbf{A}_i|\cos\theta_i \leq |\mathbf{B}_i| \)

The node \( X_{gi} \) is judged as inside the catheter segment between \( X_{c1} \) and \( X_{c2} \).

(b) \( -0.5 < |\mathbf{A}_i|\cos\theta_i < 0 \)

\( X_{gi} \) is judged as inside the catheter with some condition. The contact force was multiplied by the coefficient \( u \) to prevent the sudden state transition when the guidewire was moved into or out of the catheter.

(c) \( |\mathbf{A}_i|\cos\theta_i \leq -0.5 \)

\( X_{gi} \) is judged as outside the catheter.

In the case of (a) or (b), the distance between \( X_{gi} \) and the nearest point on the catheter segment \( (l_{ci}) \) is expressed as follows.

\[ l_{ci} = |\mathbf{A}_i| - (s|\mathbf{A}_i|\cos\theta_i)|\mathbf{B}_i| / |\mathbf{B}_i| \]  

(5)

where \( s \) is the coefficient to prevent a rapid change in the direction when two segments of the catheter model are bent. We then calculate the contact force \( f_{ci} \) when \( |l_{ci}| - R_c + R_g \) is positive as follows.

\[ |f_{ci}| \propto (|l_{ci}| + R_g - R_c)^{3/2} \]  

(6)

In this numerical study, penalty method is adopted as contact analysis. As \( f_{ci} \) increases according to the penetration depth, the joint of guidewire model which penetrates the inner wall of the catheter model is pushed to near the surface of the inner wall. When the calculation at each time step is finished, the penetration is enough small.

The motions of the proximal ends of the guidewire and catheter were the same as those of the experiment. The parameters of the guidewire and catheter model for simulation are indicated in Table 2. The mechanical properties of the guidewire are same as those in our previous study (Takashima et al., 2014). We also measured the mechanical properties and the shape of the commercial catheter used in the experiment, and used the results for the simulation. We calculated the bending stiffness of the catheter using the bending rigidity measured through the 3-point bending test. Assuming isotropic material and a Poisson’s ratio of 0.3, the torsional stiffness is approximated as 0.77 times as large as the bending one.

Based on the above assumptions, we calculated the Newton–Euler equations of motion using the contact force at every finite time step using numerical differentiation formulas. The time steps were not fixed in order to make the relative errors smaller than permissible values at each time step. When contact occurred within a time step, we divided the time step at the contact point. This simulator was developed in a MATLAB/Simulink (The MathWorks, Inc.)
environment. We also used SimMechanics (The MathWorks, Inc.), which provides a multibody simulation environment for three-dimensional mechanical systems.

Table 2 Parameters of the guidewire and catheter model for the simulation.

|                     | Guidewire             | Catheter             |
|---------------------|-----------------------|----------------------|
| Bending stiffness (N·mm/rad) | 0.144\textsuperscript{a} | 0.425                |
| Torsional stiffness (N·mm/rad) | 0.111\textsuperscript{b} | 0.327                |
| Damping coefficient (N·m·s/rad) | 0.005, 0.003 (=D\textsubscript{g}) | 0.0025               |
| Bending angle of tip (degrees) | 45                    | 0                    |
| Number of segments | 18                    | 6                    |
| Length of segment (mm) | Distal tip: 1.25      | Both ends: 1.25      |
|                     | Proximal part: 77.625, 155.25, 78.875 | Other: 2.5           |
|                     | Other: 2.5            |                      |

\textsuperscript{a} Bending stiffness of two joints of the proximal part is 0.00232 N·mm/rad.

\textsuperscript{b} Torsional stiffness of two joints of the proximal part is 0.00178 N·mm/rad.

3. Results and Discussion

3.1 Experiment

The tracking results of the guidewire and catheter tips in the three experiments are shown in Fig. 6. The trajectories of the tips of the medical devices at every 1 s are also shown using points. The photographs of the blood vessel model are simultaneously shown under the trajectories to indicate the actual inner wall of the model with some manufacturing errors. Figure 6 (a), (b) and (c) show the tracking results of the tips of both medical devices during insertion of the guidewire, insertion of the catheter and removal of the guidewire, respectively. As we could not track the guidewire tip hidden inside the catheter during removal of the guidewire, the tracking results of the guidewire tip is not shown in Fig. 6 (c). These results indicate that the positions of the guidewire and catheter tips can be tracked successfully. In these figures, movement of the medical device tips, which are fixed at the proximal parts, is clearly evident. Namely, the movement of the catheter tip during insertion of the guidewire (Fig. 6 (a)), the movement of the guidewire tip during insertion of the catheter (Fig. 6 (b)), and the swinging of the catheter tip during removal of the guidewire (Fig. 6 (c)) can be seen. These phenomena are also seen in clinical situations (Sutou et al., 2004). The first contact point between the guidewire tip and inner wall of the blood vessel model was observed at \(x \approx -7.5\) mm (the arrow in Fig. 6), which shifted largely in the negative \(x\)-axis direction, in comparison with our previous study using a metallic needle mimicking a catheter (Takashima et al, 2014). The catheter used in the present study deformed more easily than the needle used in our previous study. Therefore, the guidewire tip could be inserted deeply (negative \(x\)-axis direction), and first contacted the blood vessel wall. The guidewire tip moved from one vessel wall to the other along the \(z\)-axis (right graph in Fig. 6 (a)). Compared with the movement of the guidewire tip, that of the catheter tip along the \(z\)-axis was smaller during the insertion of the catheter (right graph in Fig. 6 (b)). This is because the movement of the catheter is restricted by the internal guidewire.

The results of the transition of the contact force between the medical devices and the blood vessel wall along the \(y\)-axis in the three experiments are shown in Fig. 7. As shown in this figure, the contact force gradually increased during insertion of the guidewire, and then rapidly increased as the catheter was inserted. As these medical devices were inserted into the curved portion of the blood vessel model, these devices were gradually bent. Therefore, the contact force would gradually increase according to the bending deformation. The contact force rapidly decreased as the guidewire was removed. As the proximal part of the guidewire is not fixed rigidly, the difference of the contact forces between the three experiments is large during the insertion of the guidewire, in comparison with our previous study (Takashima et al, 2014). Two large contact forces in this figure are marked A and B for comparison with the numerical simulation results presented in the next section. The similar tendencies and small differences in the results of the tip tracking and the contact force in the three experiments indicate good repeatability.
3.2 Numerical analysis

Temporal configurations of guidewire and catheter are shown in Fig. 8. As shown in this figure, the guidewire and the catheter models were inserted along the blood vessel model. The tracking results for the guidewire and catheter tips from the numerical simulation are shown in Fig. 9. The trajectories of both tips at every 1 s are also shown in Fig. 9.
this figure, the effect of damping coefficient $D_g$ (0.005, 0.003 Nms/rad) on the trajectory is also shown. The same tendency was seen in the trajectories of the tips of both the experimental and simulated devices. Similar to the experimental results, the simulated catheter tip was inserted along the outside wall of the blood vessel. Moreover, the movement of the catheter tip during insertion of the guidewire (Fig. 9 (a)), the movement of the guidewire tip during insertion of the catheter (Fig. 9 (b)), and the swinging of the catheter tip during removal of the guidewire (Fig. 9 (c)) are also shown in this figure. The average final positions of the catheter tips in Figs. 6 (c) and 9 (c) were ($-11.8, 7.5, \ldots$).
1.6) and (−11.8, 5.3, 0), respectively, and the difference was 2.8 mm. For realistic visualization in a training simulator, an accuracy of 10% of the lumen diameter seems acceptable to produce artificial, ‘fake’ X-ray images simulating the fluoroscopy images of a real endovascular intervention (Konings et al., 2003). Therefore, this difference may be large and can attribute to the lack of the constraint of the catheter tip in this region. Moreover, there were several differences between the calculation and experimental results. Unlike in the experiment as shown in Fig. 6 (a), at the beginning of the guidewire insertion, the guidewire tip moved smoothly in parallel along the y-axis (blue dotted circle area in Fig. 9). One reason for the difference may be that we did not consider the plastic deformation of the guidewire and the friction between the blood vessel wall and medical devices in the numerical simulation. For example, in the experiment, the plastic deformation of the guidewire tip occurred because of the large volume of the polymer coating on the distal part of the shape-memory alloy wire. In future studies, we will adopt a friction model based on our previous models (Takashima et al, 2012, 2007, 2006).

The transition of the contact force between the medical devices and blood vessel model along the y-axis from the numerical simulation is shown in Fig. 10. Similar to the results of our previous study (Takashima et al, 2014), the contact force in the simulation was larger than that in the experiment and one reason is that the damping term is large and not accurate as described in section 2.2. However, the same tendency of the contact force was seen. For example, the contact force gradually increased during insertion of the guidewire, rapidly increased during insertion of the catheter, and rapidly decreased during removal of the guidewire in both the numerical simulation and experiment. Moreover, there were two large contact forces, as shown in Fig. 10 (marked A and B in this figure). A similar phenomenon is visible in Fig. 7. However, there was a difference in the time when the two large contact forces occur between the calculation and experimental results. In Fig. 11, the appearance of the medical devices at two positions A and B are shown. As shown in Fig. 11, the contact forces in the experiment increased when many parts of the medical devices contacted with the blood vessel wall. On the other hand, in the calculation, the maximum forces occurred as soon as the medical device tips contacted the blood vessel wall. The difference in the time between the calculation and experimental results may be due to the difference of the trajectories of the medical device tips between the two methods. In Fig. 10, the force when $D_g = 0.003 \text{ Nm/ rad}$ was slightly smaller than that when $D_g = 0.005 \text{ Nm/ rad}$, although we were not able to change the time when the large contact forces occurred. However, the damping coefficient would greatly affect the contact force. Moreover, the reason why the catheter insertion did not produce an increase in the contact force would be that the friction between the catheter and the guidewire was ignored.
4. Conclusions

We constructed a new experimental apparatus that included a mechanism to move a catheter, and were able to reproduce the catheterization that is commonly seen in clinical situations. We also added a new flexible catheter model to the calculation. Furthermore, we measured the three-dimensional position of the guidewire and catheter tips and the force applied by the medical devices on a blood vessel model during insertion and removal. We also compared the numerical simulation results with the experimental results.

Similar to actual clinical situations, the movement of the catheter tip during insertion of the guidewire, the movement of the guidewire tip during insertion of the catheter, and the swinging of the catheter tip during removal of the guidewire are visible because we added a mechanism to move and evaluate the flexible catheter to the experimental apparatus and numerical simulation model. The same tendencies were seen in the trajectories of both the experimental and simulated guidewire tips. Therefore, it is necessary to consider the catheter motion to reproduce the actual clinical situation.

The same tendency of the contact force was also seen in both the experimental and calculation results. For example, the contact force gradually increased during insertion of the guidewire, rapidly increased during insertion of the catheter, and rapidly decreased during removal of the guidewire in both the numerical simulation and experiment. The contact force in the experiment increased when many parts of the medical devices contacted with the blood vessel wall. On the other hand, in the numerical simulation, the maximum force occurred as soon as the medical device tips contacted the blood vessel wall. The contact force decreased when the damping coefficient of the guidewire decreased. However, we were not able to simulate the time when the maximum contact force occurs. Namely, the damping coefficient would greatly affect the contact force. In future iterations of the numerical analysis and simulations, we will modify the simulation parameters to minimize the differences in the guidewire and catheter motions during the experiment and simulation using several types of medical devices and blood vessel models.

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