Technical Section

A geometric approach to the modeling of the catheter–heart interaction for VR simulation of intra-cardiac intervention

Patricia Chiang\textsuperscript{a}, Yiyu Cai\textsuperscript{b}, Koon Hou Mak\textsuperscript{c}, Ei Mon Soe\textsuperscript{a}, Chee Kong Chui\textsuperscript{d}, Jianmin Zheng\textsuperscript{e},* 

\textsuperscript{a} School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore 639798, Singapore  
\textsuperscript{b} Institute for Media Innovation and School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore 639798, Singapore  
\textsuperscript{c} Mak Heart Clinic and School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore 639798, Singapore  
\textsuperscript{d} Department of Mechanical Engineering, National University Singapore, Singapore 119077, Singapore  
\textsuperscript{e} School of Computer Engineering, Nanyang Technological University, Singapore 639798, Singapore

\begin{abstract}
Cardiac intervention is a minimally invasive diagnostic and therapeutic procedure used to treat cardiac diseases. The mapping of heart geometry with minimal visual assistance presents a technical challenge for interventional cardiologists attempting catheter navigation. This paper presents a geometric approach to modeling the catheter–heart interaction for VR simulations of catheter navigation within a heart chamber. Three types of modeling are used to model the interaction between the catheter and the heart wall: non-slip, pseudo-slip and slip modeling. A two-step shape memory process that minimizes the bending of and strain on the catheter is designed for catheter deformation for non-slip or pseudo-slip contact, and a progressive group linkage bending process that constrains the catheter curvature and position within the volume enclosure is designed for catheter deformation for slip contact. The proposed model is consistent with the observations made during the experiment. The model is able to deform the catheter in any free-state shape within the volume enclosure and is independent of local motion increment. Thus, it presents advantages in terms of complexity and real-time requirements.
\end{abstract}

\section{Introduction}
Virtual reality (VR) simulators for cardiology and vascular surgery are increasingly being used in surgical education \cite{1}. For practical and ethical reasons, realistic VR simulators provide a good starting point compared to animal and patient trials \cite{2–4}. VR simulators allow the surgeon to master complex techniques using VR devices in vascular and intra-cardiac intervention procedures.

The early assessment of the severity of cardiac disease with associated heart wall damage caused by ischemia and myocardial infarction is critical to the survival of heart patients given that early treatments such as interventional procedures lead to higher chances of recovery. Unlike standard diagnostic imaging techniques, the electrophysiological mapping of the heart \cite{5} is a minimally invasive procedure in which a mapping catheter is inserted via the femoral artery, and is guided through to the aorta and finally into the heart chamber. By maneuvering the catheter within the heart via twisting, bending, inserting and retracting, the surgeon acquires contact points where the catheter tip touches the heart wall under stable conditions. The cardiac interventional device can detect the accurate location of ischemic regions with minimal risk, thus enabling cardiologists to assess the severity of heart wall damage in situ. In addition to electrophysiological mapping, a cardiac interventional device can also serve a therapeutic function by facilitating the transcatheter injection of stem cells to partially restore cardiac function in a diseased heart.

In the intra-cardiac intervention procedure, the captured positions and electrical signals are usually post-processed for display on a progressively reconstructed heart surface for analysis and the delineation of viable and non-viable myocardium \cite{6}. As in vascular intervention, precise hand–eye coordination is one of the main technical requirements involved in this process. However, unlike in vascular intervention, no fluoroscopic guidance is recommended during the cardiac mapping procedure. The operator must devise the mapping strategy based on the progressively reconstructed surface. Initial acquisition can be difficult without registered pre-operative 3D imaging modalities due to the poor geometry of the reconstructed surface. Efficient electromechanical mapping...
requires specific skills related to catheter navigation, suitable inclination at the contact surface, appropriate force for contact stability, and the control of the overall targeted mapping pattern. Thus, training on intra-cardiac intervention simulators is critical to mastering the skills required, especially for interventional cardiologists who are not familiar with the procedure.

One fundamental building block of such simulators is the modeling of the interaction between the catheter and the endocardium. Modeling catheter contact and deformation within the heart chamber makes it possible to simulate the heart mapping procedure. This paper discusses the modeling of the catheter for a VR simulation of cardiac intervention and focuses on the geometric deformation of the catheter when it interacts with the heart wall under static conditions.

1.1. Related work

Previous studies have focused on modeling the interaction between the catheter and the blood vessel for vascular intervention rather than the interaction of the catheter and the heart wall within a chamber enclosure. For example, a multi-body system was used to simulate the interaction of the device with tissue [2]. The position and dynamics of the catheter were determined based on input contact as well as external and contrast injection forces. Cai et al. [3] developed a potential field representation of the vessel wall for modeling the deformation between the device and vessel. Lenoir et al. [7] modeled the catheter and guidewire as a set of beam elements in FEM with static non-linear characteristics. The combination of translation and rotation imposed a set of constraints on the composite catheter–guidewire FEM model for deformation. An optimization strategy [8] based on substructure decomposition was used for faster computation. Collision detection was made possible using a connectivity graph linking a set of connected segments. The accuracy of collision detection depends on the resolution. Bounding volume hierarchy [9] was used for collision detection in which the vascular model was represented by axis-aligned bounding boxes and the catheter/guidewire by spheres to balance collision detection accuracy with efficiency. Another physics-based approach proposed by Alderliesten et al. [10] modeled the propagation of the guidewire by minimizing the sum of the internal energy of the bending and torsion of the guidewire and the deformation of the vessel wall during interaction with guidewire, while considering the friction between the guidewire and vessel wall.

In all of the above methods, the catheter was assumed to move along the vessel wall when inserted or retracted. The 3D deformation of the catheter is also limited in space due to the small volume enclosed by the vessel segment relative to the heart chamber and the large number of points of contact between the catheter and the vessel wall. Therefore, these methods cannot be directly used in intra-cardiac intervention. The catheter kinematics involved in intra-cardiac navigation, as described by Ganji et al. [11], predict the catheter tip position based on external manipulation. However, they do not address catheter interaction with the heart wall within the heart chamber.

1.2. Our work

This paper presents a geometric approach to modeling the catheter–heart interaction in VR simulations. Note that in a simulator that models the heart-mapping procedure, tissue deformation plays a less significant role due to the less significant deformation of the heart wall as a result of direct catheter contact. The visualization of such tissue deformation is not as critical as visualizing the reconstructed surface in intra-cardiac intervention. Catheter deformation, on the other hand, is important in tracking the catheter’s movements within the heart chamber, its contact with the heart surface and to what degree it tends to slip along the heart surface. We propose to separate catheter deformation into three types: non-slip, pseudo-slip and slip. It is of note that non-slip deformation is important in intra-cardiac intervention but is not found in vascular intervention. Geometric methods are used to implement the deformations. Although our approach is geometrically based and heuristic, the modeling is based on observations from a workbench experiment that we used to characterize the interaction between the catheter and the heart tissue. Our experiment and the user study both indicate that our method can achieve real-time interaction using a common laptop platform and that its performance is consistent with user experience and expectations. To the best of our knowledge, this is the first study that has proposed a means of simulating the catheter–heart interaction in the context of intra-cardiac intervention.

This paper is organized as follows. Section 2 describes our scheme for modeling catheter and heart interaction. Section 3 presents the detailed algorithmic and implementation characteristics of catheter deformation. Section 4 presents validation experiments exploring catheter contact with heart tissue, with which the proposed interaction modeling is shown to be appropriately consistent. Section 5 discusses the results of our VR simulation, and Section 6 concludes the paper.

2. Modeling of the catheter–heart interaction

This section describes how we geometrically model the catheter and the heart and then presents our scheme for the catheter–heart interaction.

2.1. Catheter model

An electromechanical mapping catheter is a wire composed of three parts: the distal bending segment, which includes the catheter tip and is responsible for bending; the catheter body, which is in the shape of a straight line; and the handle, which contains the steering knob used to maneuver the catheter (see Fig. 1(left)). The catheter can be subjected to three different sets of movement during manipulation: insertion/retraction (or translation), rotation, and bending. Bending is effected on the bending segment as a bi-directional planar deflection. Using bending, rotation, and translation, the user can manipulate the catheter to reach different parts of the heart wall.

Based on the catheter structure, we propose to model the catheter as a series of N rigid links with N+1 data points \( P_i \) (\( i = 0, 1, \ldots, N \)) that indicate the location of the catheter in 3D space (see Fig. 1(right)). The catheter tip corresponds to the last point \( P_N \). We are particularly interested in the part of the catheter inserted into the lumen, which consists of the distal bending segment of \( K \) links and a short straight segment of \( N−K−1 \) links.

![Fig. 1. A physical catheter (left) and a geometric catheter (right) modeled as a series of links.](image-url)
Thus, \( P_{N-1} \) is the pivot point indicating the starting point for the bending, and link \( P_{1-1}P_{1} \) contains the point of insertion. The straight segment also serves as the rotation axis.

Of the three motions of the catheter, two (translation and rotation) can be easily achieved using an affine transformation on all of the link points. The difficulty of the catheter model lies in how it represents bending. The bending segment includes a pair of parameters \((k, h)\) where \( k \in \{N-K, \ldots, N\} \) is the index of the link points, \( h \in [-H, -1, 0, 1, \ldots, H_{1}] \) is the bending unit and \( H, H_{1} \) are the maximum bending units in the clockwise and counter-clockwise directions, respectively. Now we present a simple method of computing the coordinates of the link points using a local coordinate system defined on the bending plane. The bending segment is assumed to be a straight line when there is no bending. The segment at the maximum bending is required as an input. Referring to Fig. 2, the pivot point is chosen as the origin in the two directions. In summary, we need the following information to define a catheter:

- integers \( N, K, I, H_{1} \) and \( H_{1} \);
- \( N+1 \) points \( P_{i} \);
- a local coordinate system defined in the bending plane;
- two sets of angles \( \theta_{i}(H_{1}) \) and \( \theta_{i}(H_{1}) \) for maximum bending in the two directions.

### 2.2 Heart model

For catheter navigation within the heart chamber, a hybrid representation is used to describe the heart. In the hybrid representation, the heart is modeled using a tetrahedron mesh, and the volume that encompasses the mesh model is discretized into a set of voxels. Each voxel in the heart lumen or the heart wall, close to the heart surface, is specified by voxel type (surface, heart, non-heart) and is further described by the signed distance to the nearest heart surface. The associated plane equations of the tetrahedra are also computed to facilitate rapid interaction detection and response.

To derive the inner surface representation of the heart from the tetrahedral mesh, first, each voxel is tested against tetrahedra in the neighborhood and classified as either a heart or non-heart voxel. Then, a 3D operator scans the entire volume to identify the surface voxels and associated surface triangular facets. The distances of each voxel from various identified surface triangular facets in a neighborhood are computed, and the minimum distance is recorded as the distance of the voxel from the nearest heart surface.

Using this hybrid representation, the catheter movement can be easily tracked based on the distance of the catheter points to the nearest heart surface: for example, how close the catheter is to the heart surface or whether the catheter has moved out of the heart lumen into or outside the heart wall. If the catheter has moved into the heart wall, a collision event has been detected and a corresponding interaction response follows.

### 2.3 Catheter–heart interaction scheme

In the intra-cardiac intervention procedure, the catheter is navigated within the heart chamber by external means such as twisting, bending, insertion and retraction. The point of insertion of the catheter into the heart chamber is assumed to be fixed and held into the position by the aortic valves and aortic arch, whereas the catheter tip is initially free. A collision occurs when the catheter touches or crosses the heart wall. When a collision occurs, the catheter is deformed. The contact of the catheter tip with the heart wall is deemed to be stable if the tip touches at least the lower boundary and if the point of contact is maintained during the cardiac cycle with the catheter becoming deformed during diastole. If the point of contact slips due to instability, the catheter will bend and move to a new position along the heart wall. Therefore, we propose three types of catheter–heart interaction responses when a collision is detected between the heart wall and the catheter due to the input movement of the catheter: (i) no-slip contact, in which the catheter tip may be affixed to the heart wall at the initial contact while the catheter body undergoes deformation between the tip and the aortic insertion point; (ii) pseudo-slip contact, in which the catheter tip glides along the heart surface and the contact persists but the position is different; and (iii) slip contact, in which the initial contact is completely lost and the catheter is deformed along the heart wall. Non-slip and slip are confirmed during the experiment to be dependent on the contact angle between the catheter and tissue.

Based on the above analysis, we model the catheter–heart interaction as follows (see Fig. 3). The catheter can be in either an idle or a touch state depending on whether there is contact with the heart wall. The input movement of the catheter can trigger a transition from an idle to a touch state or maintain the idle state through a series of processes. Upon catheter motion, collision detection determines if the catheter has collided with the heart tissue. If no collision has occurred, the catheter returns to the idle state. If a collision has occurred, the first point of contact of the catheter with the heart wall is determined. Based on the first contact, the slip condition is determined. If no slip occurs, the catheter has undergone non-slip deformation. If a pseudo-slip occurs, a new contact point will be found, and the deformation is similar to that under the non-slip condition. If a slip occurs, the catheter will undergo slip deformation. The contact is analyzed after each instance of deformation to determine the validity of the...
contact. The catheter transitions to the touch state if the contact remains and returns to the idle state if contact is lost.

The slip conditions are determined by bending energy, strain energy, and the angle between the catheter and the tissue. Bending energy, $R_b$, is estimated based on the difference between the current deformed catheter state and the new undeformed catheter state. Strain energy $R_s$ is estimated using the difference between the current deformed catheter state and the current undeformed catheter state. The catheter–tissue angle $\phi$ is computed as the angle between the last link in the current deformed catheter and the nearest heart surface plane. Based on the experimental observations in Section 4, we identify the non-slip, pseudo-slip and slip deformations using $R_b, R_s$ and $\phi$. When the catheter–tissue angle $\phi$ is larger than a certain threshold (say 60°) and the bending and strain energies are smaller than particular thresholds, non-slip deformation is detected. Otherwise, slip deformation is identified. The phrase pseudo-slip is used to refer to deformations that are between slip and non-slip deformations.

3. Catheter deformation

We now describe the catheter deformations in detail. The basic approach is first to compute the undeformed catheter state based on the motion of the catheter and then to compute the deformation if a collision occurs. The deformation is assigned to the non-slip, pseudo-slip, or slip contact group. In this section, we use $R_i(t)$ and $P_i(t)$ to denote the link points with index $i$ on the undeformed catheter and deformed catheter at time $t$, respectively. $R_i(t)$ is obtained using the method described in Section 2.1.

3.1. Non-slip contact

Typical stable contact may be described as catheter deformation with two fixed ends: the catheter point nearest to the point of insertion and the tip that makes contact with the heart wall. The deformation is modeled to ensure shape continuity during deformation and to minimize the strain and bending of the catheter with respect to its initial shape. For this purpose, we propose a two-step heuristic shape memory algorithm for deformation.

First, the initial shape deformation is computed as the weighted force displacement from the previous deformed catheter state with boundary constraints. The force displacement is the vector difference between the current and new undeformed catheter states. We let $P(t)$ denote the points on the initial deformed catheter at time $t$. The boundary constraints are $P(t) = P_0(t-1)$, which keeps the catheter tip $P_t$ stationary, and $P_i(t) = R_i(t)$, which maintains the location and orientation of $P_{i-1}P_t$ (which is the link containing the point of insertion after motion). The initial shape deformation is thus modeled as follows:

$$P_i(t) = P_i(t-1) + w_i(R_i(t) - R_i(t-1)), \quad i = 1 + 1, \ldots, N-1$$

where $w_i$ is a weight distribution with a larger value near the point of insertion and a smaller value near the catheter tip. We set $w_i = 1 - ((i-1)/(N-1))^2$. If necessary, the weight $w_i$ is further adjusted to ensure that the deformed shape $P_i(t)$ is within the volume enclosure.

Next, the initial shape deformation obtained is refined by adjusting the link points toward or away from the straight line connecting the fixed catheter point near the point of insertion and the catheter tip. We let $Q_i$ denote the projection of $P_i$ onto the straight line $P_0P_t$. Then, the adjustment is made using $P_i = Q_i + \mu(P_i - Q_i)$ where $\mu$ is a global coefficient. The global coefficient is computed by minimizing the following energy function:

$$E = \lambda \sum_{i=1}^{N} ||P_i - R_i|| + (1-\lambda) \sum_{i=1}^{N-1} ||P_{i+1} - P_i - L_{i+1}||$$

where $\lambda$ is a tradeoff factor (we choose $\lambda = 0.3$ in our implementation) and $L_{i+1} = |R_i - R_{i+1}|$ is the link length of the undeformed catheter. The first term represents the bending of the catheter from the pre-defined shape in a free-energy state, and the second term represents the catheter strain based on the original data for link lengths. Fig. 4 shows the two-step process through which non-slip catheter deformation occurs.

3.2. Pseudo-slip contact

Pseudo-slip contact represents the transition between stable and non-stable contact in which the catheter transitions from the non-slip to the slip condition. Because the heart surface has a network of trabeculae, the catheter tip will meet obstacles that slow down the slipping. Pseudo-slip contact is classified as a sub-class of the non-slip contact interaction in which the catheter tip is allowed to deviate slightly from the contact point when the external force applied to the catheter is insufficiently counteracted by the frictional wall forces.

The initial slip vector $S^0$ of the catheter tip is computed as the component of motion vector $R(t) - R(t-1)$ along the projection...
For some vector \( \mathbf{P}_\text{tan} \) of the catheter onto the tangent plane of the heart surface at \( \mathbf{P}_N(t-1) \):

\[
\mathbf{S}^0 = \left( (\mathbf{R}_N(t) - \mathbf{R}_N(t-1)) \cdot \mathbf{P}_\text{tan} \right) \mathbf{P}_\text{tan} / ||\mathbf{P}_\text{tan}||^2
\]

If the unit normal vector of the surface at \( \mathbf{P}_N(t-1) \) is \( \mathbf{n} \), then \( \mathbf{P}_\text{tan} = (\mathbf{P}_N(t-1) - \mathbf{P}_N(t-1)) - (\mathbf{P}_N(t-1) - \mathbf{P}_N(t-1)) \cdot \mathbf{n} \mathbf{n} \). The new catheter tip position is obtained by moving \( \mathbf{P}_N(t-1) + \mathbf{S}^0 \) perpendicularly towards the heart surface:

\[
\mathbf{P}_N(t) = \mathbf{P}_N(t-1) + \mathbf{S}^0 + \mathbf{n} \mathbf{n}
\]

for some \( \mathbf{S}^0 \) such that \( \mathbf{P}_N(t) \) is on the heart surface, and therefore, the final slip vector becomes \( \mathbf{S} = \mathbf{P}_N(t) - \mathbf{P}_N(t-1) \).

The initial catheter displacement is then adjusted by the amount of the slip. The adjustment adds a second force displacement term to compensate for the slip and to interpolate the shape deformation near the catheter tip. The second force displacement is the normal component of the motion vector along the final slip vector weighted using the weight distribution \( w_i \), increasing in a quadratic fashion. That is,

\[
P_i^0(t) = P_i(t-1) + w_i(R_i(t) - R_i(t-1))
\]

\[
+ w_i \left( \frac{(R_i(t) - R_i(t-1)) \cdot (R_{ref}(t) - R_{ref}(t-1)) \cdot \mathbf{S}_i}{||\mathbf{S}_i||} \right)
\]

and

\[
w_i = \begin{cases} 
0, & \text{if } i < N + \frac{l}{2} \\
\frac{2i - N - l}{N - l}, & \text{if } i \geq N + \frac{l}{2}
\end{cases}
\]

Finally, the refinement is computed based on the non-slip contact.

### 3.3. Slip contact

Non-stable contact occurs when there is catheter deformation involving a fixed end (the catheter point nearest the point of insertion) and a moving end (the catheter tip). The non-stable contact occurs when the catheter deformation is large and the frictional forces of the contact are unable to hold the catheter tip in position.

For this mode of interaction, our approach is to adjust the catheter points iteratively, progressing from the fixed catheter point near the aortic valve to the moving catheter tip until all catheter points outside the lumen fit within the volume enclosure. First, the intersection between the heart wall and the catheter is computed. Then, the new position of the intersecting catheter link is determined. Using the proximal data point as reference, the distal point is manipulated so that it touches the tangent plane of the surface. To simulate more realistic deformation and model catheter stiffness, a group of catheter links are bent collectively instead of as an individual link. Next, the bending is extended from the intersecting link to the catheter tip to preserve the curvature of the catheter. This interaction process is repeated for the next catheter link. During this iterative adjustment process, the catheter bends and deforms from the pivot to the moving tip.

Beginning with the fixed end of the catheter, each catheter point is assessed to determine if it has penetrated the heart wall. If a catheter point \( R_{cur} \) has moved within the heart wall, the new position of \( R_{cur} \) must be determined, which should be on the heart wall in that region. Suppose that the previous catheter point proximal to the current catheter point \( R_{cur} \) is \( R_{ref} \); then, link \( R_{ref} R_{cur} \) intersects the heart wall surface at some point. We denote the normal of the heart surface at the intersection point as \( \mathbf{n} \). Based on the principle of minimal energy for deformation, the new contact position must lie within the plane, denoted by \( \Pi \), that contains \( R_{ref} R_{cur} \) and has a normal vector orthogonal to \( \mathbf{n} \). Thus, the normal of plane \( \Pi \) is computed using \( \mathbf{n}_b = (R_{cur} - R_{ref}) \times \mathbf{n} \). In addition, the length of the catheter link \( R_{ref} R_{cur} \) should remain constant. Thus, the problem of finding the new position of \( R_{cur} \) can be described geometrically as the intersection of a sphere with center \( R_{ref} \) and radius \( ||R_{cur} - R_{ref}|| \), plane \( \Pi \) and the heart wall surface. Note that the intersection of the sphere and plane \( \Pi \) is a circle (see Fig. 5). If we let \( \mathbf{T} = \mathbf{n}_b \times (R_{cur} - R_{ref}) / ||\mathbf{n}_b|| \), then the circle in 3D has the following parametric representation:

\[
P(x) = R_{ref} + \cos(x)(R_{cur} - R_{ref}) + \sin(x)\mathbf{T}, \quad x \geq 0.
\]

The new position of \( R_{cur} \) can be obtained by computing the intersection of this parametric curve and the heart wall surface.

To simulate catheter stiffness and resistance to bending, the above interaction process is used with a group linkage bending...
The above four constraints are used to determine the length of the group linkage and the regional bending behavior. Given an input M linkage associated with catheter inertia, the group linkage is tested to determine that it satisfies all constraints in a process referred to as the group linkage search. If the conditions are not satisfied, M is decremented, and the process is repeated until the conditions are satisfied. The corresponding group linkage bending angle \( \angle R_j R_k R_{M-k} \) is computed using the interaction process described earlier in Fig. 5 with \( R_k = R_{ref} \) and \( R_{M-k} = R_{ref} \). The group linkage bending angle indicates the collective deviation of M links from the current catheter shape. The group bending angle is distributed across M links using linear scaling to compute the unit bending angle \( 2(k-j+M)/(M(M+1)) \), for link \( R_j \) that makes the proximal end link near the point of insertion bend the least and the distal end link near \( R_k \) bend the most. Then, starting from point \( k-M+1 \), each point \( i \) in the group linkage consisting of points \( \{k-M+1,k-M+2,\ldots,k\} \) is adjusted by rotating its unit bending angle approximately \( R_{k-1} \) on the plane formed by \( R_{k-M}, R_k \) and \( R_{k-1} \). After each adjustment, the remaining length of the catheter composed of the points beyond the adjusted point towards the free end are readjusted. The chain adjustment is performed with respect to the curvature (or the 2nd order derivative) of the undeformed catheter:

\[
R_i^\prime = 2R_{i-1} - R_{i-2} + (R_i - 2R_{i-1} + R_{i-2}).
\]

Because the group linkage bending involves the collective adjustment of individual links, there may be some resolution differences that result in \( R_k \)'s moving into the heart wall. Hence, multiple attempts may be required to move the catheter back to the heart lumen.

4. Validation

A workbench experiment was designed for use in analyzing the interaction between the catheter and the heart tissue. The aim was to provide quantitative measurements for haptic devices in the context of an intra-cardiac intervention simulator. Force-displacement characteristics were investigated for different deflection angles between the catheter and the heart tissue.

4.1. Materials and method

A biomechanical apparatus specially designed for use in measuring small amounts of force against displacement in tissue was used for the experiment. The distance sensor measures a range between 60 mm and 260 mm with a linearity of 360 \( \mu \)m and a resolution of 100 \( \mu \)m. The force sensor measures a range of \( \pm 1500 \)g with a maximum sensitivity of \( 1.43 \times 10^{-3} \)g. The 3D positioning with a stepper motor and the recording of sensor data are controlled using the LabView software. An electrophysiology mapping catheter from St. Judes Medical was used. The catheter has a deflectable tip controlled by rotating the top part of the catheter shaft. A holder for mounting the catheter to the force sensor is designed and fabricated with polymethylmethacrylate. The holder should not generate torque, which might affect the force measurements, but should nevertheless maintain full length to preserve deflection control. Fresh porcine hearts were purchased from a butcher.

The procedure included the following steps once the apparatus setup was in place. Tissue samples from the septal walls were incised from the left ventricle of the fresh porcine hearts. The tissue was mounted onto an elevated platform with nylon thread. Care was taken to ensure that the mounting was parallel to the platform. The system was first calibrated with force–distance measurements for the sensor only and the sensor and catheter-holder only.
The catheter was inserted into the holder and held securely by set screws. The catheter tip was extended 95 mm beyond the holder. The force–displacement measurements were taken for a range of 40 mm. The tip of the catheter was moved precisely 3 mm above the desired position of contact. It was necessary to maneuver the positioning system because the catheter’s nearest contact with the heart surface varied with each change in deflection. The catheter was then programmed to move a vertical distance of 40 mm towards the tissue at a speed of 2.8 mm/s. The corresponding force and vertical displacement measurements were recorded. The catheter deflection was changed using a calibrated knob on the catheter shaft. The steps from tip movement to catheter deflection were repeated for catheter–tissue angles ranging from 90° to 15° at 15° intervals.

4.2. Results

The force–displacement graph for the experimental setup is presented in Fig. 7. The data have been supplemented with calibration measurements. It was observed during the experiment that continuous slip occurs at the catheter–tissue angle of < 60° and that stable contact of the catheter with the heart wall can be established at a catheter–tissue angle of > 60°. The catheter–heart interaction is highly dependent on the local tissue characteristics, and care was taken to ensure that the catheter tip touched nearly the same spot in different catheter–tissue angle setups. Referring to the 90° graph, we can see four phases of catheter–heart interaction. During phase A, the catheter moves towards the heart tissue, and the average force is constant with the displacement. When the catheter touches the tissue at a relative displacement level of 3 mm, initial contact is established with a slight depression in the heart wall. Further displacement causes the catheter to deform in phase B. As the heart wall becomes more rigid and resists catheter insertion, the catheter deformation exerts upward force on the sensor, causing a decrease in the measured amount of force. The measured amount of force decreases linearly with the amount of displacement. At a point near relative displacement 30–33 mm, referred to as the sudden slip point, the contact becomes unstable, and the catheter enters phase C. The contact between the catheter and tissue is lost, and the catheter slips suddenly in the direction of least obstruction, releasing the stored energy. The force detected increases sharply as the resistance is removed suddenly. Either the catheter tip hits another contact point or the catheter wire glides along the heart wall, creating constant force displacement in phase D. The 75° graph similarly characterizes the catheter–heart interaction. However, the 60° graph depicts two small sudden slips. For angles smaller than 60°, the catheter slips directly along the heart surface and exhibits no initial firm contact. The force increases non-linearly upwards to a small degree as the displacement increases due to gentle downslope towards the apex (Fig. 8b).

Fig. 8 shows two snapshots of the catheter–heart interaction experiment. The catheter with a pre-set bending and tip displacement
of 3 mm above the mounted sepal tissue moved vertically downwards a distance of 40 mm. The catheter came into contact with the tissue and was deformed initially (in the non-slip condition), with the catheter tip fixed at a contact point (Fig. 8a). Further vertical movement resulted in the sudden deflection of the catheter tip away from the initial contact point. The catheter was subsequently deformed (in the slip condition), with the catheter tip gliding along the surface of the heart tissue (Fig. 8b).

Our proposed interaction modeling for the catheter and the heart wall is consistent with the data obtained from the catheter–heart interaction experiment. There is an explicit difference between the non-slip and slip scenarios. Given a larger catheter–tissue angle, a transition occurs from the non-slip to the slip condition. The initial catheter deformation under non-slip contact (Fig. 8a) may be modeled, and the transition to deformation under slip contact is supported by the increasing catheter bending or deviation. The pseudo-slip state has not been identified in this experimental dataset, but it may be difficult to observe in force–displacement graphs because the transition time is small. At smaller catheter–tissue angles, there is continuous slipping, and catheter deformation under slip contact (Fig. 8b) may be modeled. Preliminary validation has also shown that the path of the slipping is dependent on local tissue structure, the initial shape of the catheter when the contact is stable, and the geometry of the hard surface when the contact is unstable.

We also seek to validate the correlation between the simulation data and the experimental data. The bending and strain energies of the catheter during external manipulation can be tuned to the relative force from the experiment to indicate the scalable resistive force as haptic feedback in the simulator. We define the resistive force \( F_R \) as the weighted square sum of the bending and strain energies:

\[
F_R = k_0 R^2_b + k_s R^2_s
\]

where \( k_0 \) and \( k_s \) are constants that match the force range of the experiment \((-13.38; k_s = -43.62)\). The simulated measurement for the catheter tip approaching the lateral wall of the left ventricle at a 2/5 distance from apex to base is shown in Fig. 9. The angle that the tip makes with the heart wall surface during the initial contact is close to \( >80^\circ \). There are four stages that occur as the catheter is continuously inserted from the aortic valve until a displacement of 32 mm is achieved. Initially, there is no contact between the catheter and the heart wall. The first contact occurs at 5.8 mm, and the catheter undergoes non-slip deformation in which the magnitude of the simulated bending and strain force increases rapidly. Subsequent insertion results in pseudo-slip deformation with a similar increase in the magnitude of the simulated force until a point at 21.8 mm, at which the tip of the catheter gives way, the catheter undergoes slip deformation and the simulated force magnitude decreases sharply. In slip deformation, the force magnitude increases with increased displacement. The trend in the force–displacement graph of the simulation is generally consistent with the results of the catheter–heart tissue experiment.

5. Experiments

The modeling of the catheter–heart interaction is performed on a laptop with an Intel Core2 Duo CPU P8600 at 2.40 GHz, 4 G RAM and Windows Vista Home Premium SP1. The heart surface of the left ventricle is pre-computed from a cardiac dataset \([12]\) composed of 148,516 vertices and 728,321 tetrahedra. The left ventricle is represented with dimensions \( 94 \times 109 \times 112 \text{ units}^3 \) and is digitized at a resolution of \( 0.4 \times 0.4 \times 0.4 \text{ units}^3 \). The deflectable portion of the catheter is represented by 16 data points with 4 units per link and up-sampled \((8 \times)\) via linear interpolation.

5.1. Real-time performance

The catheter–heart interaction can be modeled in real-time in intra-cardiac intervention simulations. The deformation and time statistics for 10k small catheter movements are represented in Table 1. The datasets shown are selected because at least 33% of the catheter movements exhibited valid deformation. The catheter is first initialized to a common position close to the surface of the left ventricle, and the catheter is then moved, with equal probability of the different catheter motion types. The step motions of the catheter are \(15^\circ\) per rotation, 8 units per translation \((\sim 8\% \text{ of left ventricle length dimension})\) and 4 units per bending \((0.5\% \text{ maximum deflection})\). Larger step motions \((20^\circ, 16 \text{ translation units}, 8 \text{ bend units})\) are used in the last five datasets. It is observed that each catheter movement can be computed within a range of 2.2–5.8 ms. Assuming that catheter deformation consumes more time than collision
detection, the average time assumed per deformation is 5.4–10.5 ms. Under normal circumstances, the catheter is carefully navigated within the heart chamber, and high acceleration or large catheter movements are unlikely. Any hard contact of the catheter with the heart wall should be minimized during the cardiac intervention procedure to prevent defibrillation or injury to the heart wall. The large step movements and 33% deformations per catheter movement used in the statistics tests are intended to provide a worst case simulation. With an average of 8 ms per deformation, it is possible for the haptic device to sample catheter movements every 10 ms, thus providing smooth catheter control and utilizing 26.4% of the CPU capacity. The performance of catheter deformation in this manner has exhibited advantages in terms of the degree of complexity and real-time requirements involved.

5.2. Visual effects

Fig. 10 illustrates various catheter–heart interactions with the catheter bent away from (a), rotated away from (b), and forward of (c) the initial catheter position shown in the left column. The left ventricle is represented by the mesh with the aortic and mitral valves. The catheter is represented by the blue-striped rod in its energy-free state and by the red-striped rod in its deformed state. The shape and position of the catheter are determined initially by the insertion point at the aortic valve and tilt angles. Subsequently, the deformation of the catheter via a set of bending, rotating, and pull–push motions is determined using catheter–heart interaction modeling. Progressive incremental catheter motion may displace part of the catheter outside the heart lumen. If the catheter tip has penetrated the left ventricle wall (indicated by the blue stripe), the catheter will be deformed using the interaction algorithm (indicated by the red stripe). In non-slip contact, the catheter is deformed with the catheter tip fixed at the initial contact position. In slip contact, the catheter is progressively deformed, beginning with the fixed catheter point near the insertion point and ending with the catheter tip.

Fig. 11 shows the effect of global linkage bending on the catheter–heart interaction. Without group linkage bending, a kink may result from the simple interaction process. If care is taken in controlling catheter stiffness via group linkage bending, the deformation of the catheter is much smoother, and the results of the deformation are much more acceptable.

5.3. User study

Due to the complexity of intra-cardic intervention and the dearth of efficient acquisition devices, it is very difficult to perform an effective evaluation of our work. Therefore, a user study of an intra-cardiac simulator simulating catheter deformation is conducted instead. The participants are 10 doctors with an average of 2 years of experience in general medicine and surgery. The participants are first briefed on the procedure of cardiac intervention and the features of the simulator. They are then required to perform two mappings of the left ventricle of the heart using a virtual reality catheter unit. There is a graphical display indicating their progress during the catheter navigation.
process and showing the contact of the catheter with the invisible heart wall and with the reconstructed surface mesh. The performance of the intra-cardiac simulator is assessed via participant feedback. The results of the assessment of the catheter interaction module are given in Table 2. A consensus emerges that the catheter–interaction model is in real-time and that the simulated catheter navigation process is acceptably consistent with user experience and expectations.

6. Conclusions

This paper has presented a geometric approach to addressing the catheter–heart interaction in a VR simulation of intra-cardiac intervention. We represent the catheter using a series of rigid links and the heart using a hybrid representation consisting of mesh and voxels. The interaction between the catheter and the heart wall is classified as non-slip, pseudo-slip or slip deformation under external twisting, bending, insertion and retraction of the catheter. Catheter deformation under non-slip or pseudo-slip contact is achieved via recursive shape deformation that minimizes the bending and strain of the catheter. Catheter deformation under slip contact is achieved using a progressive geometric deformation that constrains the catheter curvature and position within the volume enclosure. The key advantage of the proposed approach is that it can be used to deform the catheter into any free-state shape within the volume enclosure and is independent of local motion increment. The deformation can be achieved in a single motion step, thus ensuring real-time interaction. The modeling is consistent with the observations culled from our experimental procedure.

It should be pointed out that this is the first attempt to model the catheter–heart interaction within a VR simulation. The current modeling is not very accurate, although it provides a reasonable simulation for training purposes. A great deal of research will be needed to produce a more accurate simulation. In particular, to produce a simulation based on physics should be a goal of future research. In addition, the proposed method is limited to interactions under static conditions. With the availability of heart motion reconstruction, catheter–heart interaction modeling could be extended to model catheter deformation under dynamic conditions.

Acknowledgments

This work is supported by ARC 9/09 Grant (MOE2008-T2-1-075) from Singapore. The authors would like to thank Mr. Yang Liangjing for his help with the catheter–heart interaction experiment and Mr. Chia Yak Khoong for his help in designing and constructing a holder for mounting the catheter to the force sensor.

References