Novel, Robust, and Efficient Guidewire Modeling for PCI Surgery Simulator based on Heterogeneous and Integrated Chain-Mails

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Abstract

Despite the long R&D history of interactive minimally-invasive surgery and therapy simulations, the guidewire/catheter behavior modeling remains challenging in Percutaneous Coronary Intervention (PCI) surgery simulators. This is primarily due to the heterogeneous heart physiological structures and complex intravascular inter-dynamic procedures. To ameliorate, this paper advocates a novel, robust, and efficient guidewire/catheter modeling method based on heterogeneous and integrated chain-mails, that can afford medical practitioners and trainees the unique opportunity to experience the entire guidewire-dominant PCI procedures in virtual environments as our model aims to mimic what occurs in clinical settings. Our approach’s originality is primarily founded upon this new method’s unconditional stability, real-time performance, flexibility, and high-fidelity realism for guidewire/catheter simulation. Considering the front end of the guidewire has different stiffness with its conjunctive slender body and the guidewire length is adaptive to the surrounding environment, we propose to model the spatially-varying six-degree of freedom behaviors by solely resorting to the generalized 3D chain-mails. Meanwhile, to effectively accommodate the motion constraints caused by the beating vessels and flowing blood, we integrate heterogeneous volumetric chain-mails to streamline guidewire modeling and its interaction with surrounding substances. By dynamically coupling guidewire chain-mails with the surrounding media via virtual links, we are capable of efficiently simulating the collision-involved inter-dynamic behaviors of the guidewire. Finally, we showcase a PCI prototype simulator equipped with haptic feedback for mimicking the guidewire intervention therapy, including pushing, pulling, and twisting operations, where the built-in high-fidelity, real-time efficiency, and stableness show great promise for its practical applications in clinical training and surgery rehearsal fields.

Keywords:
Guidewire Simulation, Heterogeneous Chain-Mails, Guidewire-Vessel Interaction, Guidewire-Blood Interaction, PCI Simulator, Haptic Feedback

1. Introduction and Motivation

During the past 15 years, virtual reality (VR) based simulation has been gaining momentum in the medical community, because it plays more and more significant role in the medical education, training, the rehearsing of higher risk clinical situations, certification and the safe adoption of new technologies. And the Food and Drug Administration of USA has advocated VR simulation training as a part of the approval of new procedures/devices. Specifically, as the well recognized effective treatment of heart diseases, Percutaneous Coronary Intervention (PCI) has a wide range of complex intravascular procedures, which requires cardiologists to gain or maintain proficiency via experiencing more situations to avoid the potential life-threatening complications. Although the American Board of Internal Medicine has also adopted simulation training as the required recertification process for Interventional Cardiology Board certification, the current coronary intervention simulation can only accommodate summative portrayal and attentive-level operator training. Although many methods, ranging from simple spring-mass structure [1], moderately complex kirchoff elastic rod [2] to complex Cosserat-theory-based models [3], have been employed to simulate the guidewire behaviors, suffering from the heterogeneous heart physiological structures and complex intravascular inter-dynamic procedures, the guidewire/catheter behavior modeling is still challenging in PCI surgery simulation due to lacking magic tradeoffs among seemingly competing criteria of unconditional stability, real-time
performance, adequate flexibility, and high-fidelity realism. In particular, key technical challenges are highlighted as follows.

First, since heart has tortuous 3D blood flow pathways with multiple inlets and outlets, PCI simulation involves complex guidewire-dominant procedures, which requires the simulated guidewire should have six degrees of freedom. And [4] states that real-time efficiency is more important than accurate physical simulation in response to simulated interventions, to improve efficiency while retaining certain simulation accuracy simultaneously, it is nontrivial to design a robust and efficient geometric/physical model to depict and control the guidewire’s behaviors with high fidelity.

Second, to facilitate the feeding direction tuning, the guidewire usually has a small segment of relatively-soft hook in the front end, whose stiffness is different from its connecting guidewire body. In addition, the discrete sampling of the guidewire should be adaptive to the surrounding environment. This makes the desired spatially-varying guidewire behavior model even more unimaginable.

Third, the heart is a four-chambered muscular organ containing an involuntary conduction system, which pumps blood throughout the blood vessels by repeated, rhythmic contractions. It is hard to simulate the dynamic coupling among guidewire, blood, and blood vessels. Thus, even if fully physics-based models can mimic such complex mechanics, the inevitable computation burden can not satisfy the interactive requirement of PCI surgery operations. Meanwhile, the stability is also a nontrivial problem that is commonly occurred in fully physics-based systems.

To tackle the aforementioned challenges, we focus on a brand-new, integrated, and robust guidewire modeling method towards its high-performance simulation and interactive response to the beating vessels and flowing blood in the PCI surgery simulator. Our approach is a generalized 3D chain-mail model with heterogeneous dynamics, enabling tight coupling of guidewire and its surrounding media in a unified solution. This is the first research attempt in adapting 3D chain-mail models to PCI simulation in VR. Fig. 1 intuitively illustrates the architecture of our method. The trainee inserts guidewire into the vessel using haptic device, when the guidewire is moving in the virtual environment, its strong coupling with blood flow is realized by building virtual chain-mail links and its interaction with vessel wall is achieved by collision detection and response, then the new state of the guidewire is obtained by numerically integrating the chain-mails. Compared with other existing methods, our system can achieve unconditional stability, full automation, real-time response, and physically plausible interaction with participating media involved in minimally-invasive surgery and therapy procedures. Specifically, the salient contributions can be summarized as follows:

- We propose a spatially-varying six-DOFs (degrees of freedom) guidewire behavior model based on the generalized 3D chain-mails, which allows to handle the soft guidewire tip and its more rigid body in a simple and consistent way, and at the same time, can accommodate the overall guidewire length and its discrete representation to be adaptively modified according to the specific PCI intervention and the surrounding intravascular environment.

- We synchronously incorporate the surrounding-media interactions into the 3D chain-mails based guidewire model by streamlining the representation of vessel and blood with heterogeneous volumetric chain-mails and coupling their motion constraints based on the dynamically constructed virtual chain-mail links, which enables stable and real-time integrated behavior simulation among guidewire, blood flow, and vessel wall.

- We augment our guidewire simulation by integrating it with a PCI-specific force feedback device, together with penalty force based haptic feedback.

Figure 1: The flowchart of our new guidewire simulation method.
calculation, FEM-based tissue deformation, ray-casting based X-ray simulation, and 3D realistic rendering, which collectively give rise to the concerted efforts towards physical, visual, haptic, and procedural realism of PCI simulator, and also contribute to many other VR-based interactive simulations.

2. Related Work

Physics-based Guidewire Modeling and Simulation. A popular method to simulate guidewire-like deformable objects is the mass-spring model [5]. Typically, linear spring forces try to preserve the spring length, and angular springs aim to preserve the rest angles between adjacent springs [6]. Luboz et al. [7] employed the hybrid mass-spring particle system for the guidewire simulation, however, to facilitate correct force propagation between the soft tip and rigid body, guidewire tip had to be specially handled. Mi et al. [8] proposed a multi-body mass-spring model by discretizing the guidewire into tip, link, and body. Link is defined as a rigid body, which connects the tip and body that are represented by mass-spring model. Generally speaking, mass-spring model is intuitive and easy to understand, however, it is hard to handle material torsion, while a guidewire being inserted into the blood vessel must undergo rotation. As for more complex physical models, FEM-based method is also very popular in guidewire simulation. Wei et al. [9] used 3D FEM-based beam element to simulate the guidewire, where each beam consists of two six-DOF nodes (three degrees for translation and the other three for rotation). The FEM-based method is more physically realistic, but suffers from being computationally expensive [10]. Besides, the Cosserat-based approaches consider the material frames when formulating the strain-stress relations of guidewire. For example, Tang et al. [11] proposed a real-time and realistic elastic rod model to simulate minimally invasive vascular interventions by limiting the guidewire to move along the centerline of the vessel. Sueda et al. [12] proposed a general framework to handle the large-scale deformation of highly constrained strands but without torsion. Recently, Mao et al. [3] designed a Cosserat-based guidewire model, which affords the guidewire to freely move (forward or backward), bend, or twist subject to collision contact, however, they ignored the guidewire-blood interaction. Although Cosserat-based method can obtain high physical accuracy, it is not stable, and needs to consume large amount of time to handle the continuous interaction with the surrounding environment.

Non-physics-based Guidewire Modeling and Simulation. Non-physics-based methods are oftentimes highly related to geometry processing. Spline-based models, also known as continuous deformation models, can discretize a rod into nodes to serve as a set of control points of a spline curve. In this model, the differential quantities, such as curvature and (geometric) torsion, can be accurately computed in closed form without resorting to numerical approximation. Kaldoret et al. [13] addressed the simulation of knitted cloth at the yarn level, wherein the yarn is simulated with a single spline curve. Theetten et al. [14] proposed a spline based model to handle the physical simulation of deformable curve-shaped objects. Although these models can achieve good performance, they are hard to handle material torsion. Ganji et al. [15] employed the forward kinematics approach to predict the catheter’s tip position by assuming the catheter bends with zero torsion and constant curvature. Rungjiratananon et al. [16] simulated complex hairstyle by employing an overlapped chain-shape-matching method and neglecting the torsion deformation. Non-physical models have advantages in efficiency and stabilities for the guidewire simulation in complex environment, but they can only achieve physically-plausible effects.

Inter-Dynamic Behavior Coupling. Considering the influence on guidewire behaviors from flowing blood, we mainly review the coupling between the blood flow field and guidewire, which belongs to solid-fluid coupling issue. Akinci et al. [17] proposed a novel, versatile method for the two-way coupling of SPH fluids and rigid bodies by using boundary particles to sample the surface of rigid objects. Another two-way coupling technique is proposed to manipulate the forces between fluids and hair by Lin [18], wherein the motion of hair and fluids is simulated by evaluating the hydrodynamic forces with SPH-based boundary handling techniques. How to effectively handle collision for long curve-shaped deformable models is also nontrivial. Servin et al. [19] described an approach for interactive simulation of wires contacting with rigid bodies using massless contact nodes. But their model can not handle self-contacts in high-tension regions. Durville [20] proposed a contact-friction model to simulate knot tightening, which can detect various contact configurations between elastic beams.

3. Guidewire Modeling based on 3D Chain-Mails

3.1. Overview on Generalized Chain-Mail Model

3D Chain-mail model is designed to deform large data sets at interactive rates, which permit to directly ma-
The black arrow means the moving vector of the “sponsoring” element, represented by the blue-filled rectangle. When the distance of adjacent neighbors is less than the minimal compression or larger than the maximal stretch, elements adjust to satisfy the constraint, indicated with the red circle.

Li et al. [22] proposed an extension to the original chain-mail algorithm allowing non-uniform rectilinear meshes to be modeled. In the generalized chain-mail model, each element can be arbitrary positioned, and is linked to any number of neighbors; and it can be extended to any dimension. As shown in Fig. 3(a), we illustrate the generalized chain-mail model in 2D. Element A has four neighbors: B, C, D, and E. When A moves to a new position, its neighbors update their positions according to certain updating rules. Fig 3(b) shows the initial positions of A and C, Fig 3(c) shows the potential valid region of C, controlled by the compressing (through $\text{minDx}$), stretching (through $\text{maxDx}$) and shearing (through $\text{maxShearDy}$) constraints.

Medical guidewire can be treated as elastic object that tends to bend or twist rather than stretch. By manipulating the end of the guidewire, the body should immediately propagate the translation or rotation to the tip. A high-fidelity simulation should reflect the subtle dynamics of the guidewire tip when the guidewire is being steered around a bifurcation.

We construct the guidewire representation model based on the generalized 3D chain-mails. We discretize the guidewire as a chain of small segments, and each segment denotes a small rigid rod, which is neither compressible nor bendable. And the hinge point connecting two segments controls the allowable local bending. As shown in the Fig. 4(a), the red edges represent small rigid rod (chain-mail), the points on the lines are the joints. Our guidewire model consists of two main parts: rigid slender body and soft tip. As shown in Fig. 4, the softer tip is more likely to bend. By specifying spatially-varying control parameters for the body and tip, our chain-mail model can represent heterogeneous guidewire behaviors.
Pushing. As shown in Fig. 4(a), when the guidewire is pushed, the translation direction of the first chain-mail in the guidewire body is calculated by normalizing vector \( v \), which is obtained by subtracting the position of the first tip chain-mail from that of the last tip chain-mail, respectively denoted as \( x_0 \) and \( x_3 \). Then the first body chain-mail will translate along the direction \( v \) according to the speed control parameter \( t \). Taking the first body chain-mail as the sponsoring element, the translation is then sequentially propagated to other chain-mails under the compressing and stretching constraints (and shearing constraints).

Pulling. As shown in Fig. 4(b), when the guidewire is pulled, the last body chain-mail is moved along the normalized body direction \( v \), which is calculated by subtracting the position of the last two chain-mails of the guidewire body. Analogously, taking the last body chain-mail as the sponsoring element, the positions of other chain-mails can be updated under the compressing and stretching constraints (and shearing constraints).

Rotation. As shown in Fig. 4(c), when the guidewire is rotated, we can get the rotation axis by normalizing the vector obtained from the first two chain-mails of the guidewire body. Then the first tip chain-mail will be rotated with an angle \( \theta \), which will further lead other chain-mails to move under the shearing constraints (and compressing and stretching constraints).

3.3. Numerical Computation of Guidewire Dynamics

Based on the constructed chain-mail model, we can conduct the guidewire deformation via two steps. In the geometric deformation step, when arbitrary chain-mail moves, the motion is sequentially transmitted to its neighboring chain-mails guided by the spatially-varying compressing, stretching, and shearing constraints. In the physical optimization step, each chain-mail’s position will be locally adjusted to keep the previous shape as rigid as possible, so that the system can arrive at the minimum energy state as quickly as possible.

Geometric Deformation. In our guidewire model, each chain-mail has two neighbors except the two terminal ones. Supposing the chain-mail \( P(x_1, y_1, z_1) \) is the sponsoring element, and the chain-mail \( Q(x_2, y_2, z_2) \) is one neighbor of \( P \), then we have:

\[
\begin{align*}
\Delta x &= |x_1 - x_2| \\
\Delta y &= |y_1 - y_2| \\
\Delta z &= |z_1 - z_2|
\end{align*}
\] (1)

When \( P \) moves to the new position \( P^*(x_1^*, y_1^*) \), we can calculate its displacement vector \( \beta \) as:

\[
\beta_1 = P^* - P_1^*; \beta_2 = P^* - P_2^*.
\] (6)

Figure 5: Illustration of physical optimization. From top to bottom: initial shape, deformed shape, optimized shape of the three guidewire chain-mails.

If \( x_2 \leq x_1, y_2 \geq y_1 \), and \( z_2 \geq z_1 \), we can get

\[
\begin{align*}
x_{\min} &= x_1^* + (\alpha_{\max} \Delta x - \beta \Delta y + \Delta z) \\
x_{\max} &= x_1^* + (\alpha_{\max} \Delta x + \beta \Delta y + \Delta z) \\
y_{\min} &= y_1^* + (\alpha_{\max} \Delta y - \beta \Delta z + \Delta x) \\
y_{\max} &= y_1^* + (\alpha_{\max} \Delta y + \beta \Delta z + \Delta x) \\
z_{\min} &= z_1^* + (\alpha_{\min} \Delta z - \beta (\Delta x + \Delta y)) \\
z_{\max} &= z_1^* + (\alpha_{\min} \Delta z + \beta (\Delta x + \Delta y))
\end{align*}
\] (3)

If \( x_2 < x_1 \), then the definitions of \( x_{\min} \) and \( x_{\max} \) change to

\[
\begin{align*}
x_{\min} &= x_1^* - (\alpha_{\max} \Delta x - \beta (\Delta y + \Delta z)) \\
x_{\max} &= x_1^* - (\alpha_{\max} \Delta x + \beta (\Delta y + \Delta z))
\end{align*}
\] (4)

Similar cases will occur when \( y_1 < y_2 \) or \( z_1 < z_2 \). And different settings for the parameters \( \alpha_{\min}, \alpha_{\max} \) and \( \beta \) make the guidewire heterogeneous in slender body and soft tip.

Physical Optimization. As shown in Fig. 5, we further refine the chain-mail positions by minimizing the deformation energy of the entire guidewire. Given the initial position of chain-mail \( P(x, y, z) \) and its two neighbors \( P_1(x_1, x_1, x_1), P_2(x_2, x_2, x_2) \), we can obtain their relative-position vectors \( \delta_i \) as

\[
\delta_1 = P - P_1; \delta_2 = P - P_2.
\] (5)

After deformation, \( P \) moves to the new position, we can calculate its displacement vector \( \beta_i \) as:

\[
\beta_1 = P^* - P_1^*; \beta_2 = P^* - P_2^*.
\] (6)
Therefore, according to the Hooke’s Law, we can formulate the deformation energy as:

$$E = \sum_{i=1}^{2} \| \beta_i - \delta_i \|_2^2, \quad (7)$$

Then we need to calculate the new position of $P^*$ to minimize the energy. By defining:

$$Q_1 = P_1 + \delta_1; \quad Q_2 = P_2 + \delta_2; \quad (8)$$

we can rewrite the energy formula with $Q$ as

$$E = \sum_{i=1}^{2} \| P^* - Q_i \|_2^2. \quad (9)$$

Minimize the energy, we can get the optimized position of $P^*$ as the centroid of the $Q_1Q_2$.

$$P^* = \frac{Q_1 + Q_2}{2}. \quad (10)$$

Algorithm 1 documents the numerical computation procedure for guidewire dynamics, wherein lines 3 to 18 depict the geometric deformation, and lines 19 to 22 describe the physical optimization procedure. Here sponsoring-element denotes the motion-triggering chain-mail. As each chain-mail is processed, its position is compared to the sponsoring one, and if it is outside the potential valid region, the chain-mail will be moved into the valid region with the minimum distance cost. If so, it in turn becomes a new sponsoring element, and its neighboring chain-mails are added to the candidate lists waiting-list to wait for further processing. The flag processing-state records whether the chain-mail has been updated or not, which can guarantee that each element is processed at most once.

As described above, different materials can be simulated by adjusting the compressing, stretching and shearing parameters respectively: $\alpha_{\text{min}}, \alpha_{\text{max}}$ and $\beta$. In Fig. 6(a), one end of the guidewire is fixed to the wall, because of the gravity, the guidewire falls and collides with the ground spontaneously. Different shearing parameters lead to different deformation degrees. It shows that the larger the shearing parameter is, the greater the bending degree will be, which means the object is softer. Correspondingly, being similar to elastic ropes, a large proportion of the guidewire contacts with the ground. On the contrary, the small shearing parameter gives rise to stiff guidewire simulation. When the shearing parameter is zero, the guidewire will be totally rigid and can not fall down under the influence of gravity. And Fig. 6(b) shows more complex guidewire deformation results obtained by respectively manipulating its soft tip with pulling, pushing, and twisting operations.

Algorithm 1 Guidewire Dynamics

1: **Input:** initial shape of guidewire $G$.  
2: **Output:** new position of guidewire $G^*$.  
3: $(x, y, z) \leftarrow (x^*, y^*, z^*)$, update position of $P$.  
4: add the neighbors of $P$ to waiting-list with $P$ as sponsoring-element.  
5: **while** waiting-list is non-empty do  
6: for every $q \in$ waiting-list, check the processing-state  
7: if flag is TRUE then  
8: remove $q$ from list.  
9: else check $q$ against the valid region $R$ with $P$ is sponsoring-element.  
10: if $q$ lies outside $R$ then  
11: move $q$ to the nearest point $q^*$ in $R$  
12: add its neighbors to the waiting-list  
13: $q$ is sponsoring-element  
14: $G(q) \leftarrow q^*$  
15: processing-state $\leftarrow$ TRUE.  
16: end if  
17: end if  
18: end if  
19: **end while**  
20: for $i \leftarrow 1$ to N do  
21: $n_i \leftarrow$ number of neighbors of $p_i$.  
22: $P_i^* = \sum_{j=1}^{n} \frac{p_j + p_i - p_j}{n}$  
23: $G(p_i) \leftarrow P_i^*$.  
24: **end for**

4. Interactions with the Surrounding Environment

4.1. Model Coupling based on Virtual Links

To couple the behaviors among guidewire, blood, and vessels, we uniformly construct volumetric chain-mails for vessel and blood flow field. Similar to the guidewire model, for each volumetric chain-mail, we connect it to its neighbors by real links, and set the compressing, stretching and shearing constraints according to
their heterogeneous properties respectively. As shown in Fig. 7, each vascular chain-mail is connected to its one-ring neighbors. For each chain-mail in the blood flow field, its neighboring chain-mails are determined by a sphere with certain radius.

So far, we have got three chain-mail models: 3D guidewire chain-mail model $G$, volumetric blood-field chain-mail model $B$, and volumetric vascular chain-mail model $V$. The links in such three models are called real links, when some chain-mails update, the change is spread via those real links within the corresponding independent model. To simulate cross-model interactions, we resort to virtual links, which are responsible to spread the motion from one chain-mail model to another. We dynamically build the virtual links between $G$ and $B$ to simulate the flowing-blood influence on guidewire behavior, and dynamically build the virtual links between $V$ and $B$ to simulate the vessel motion influence on guidewire behavior.

In our PCI surgery simulator, the vessels deform with heart beating. The blood flow field, which is represented with millions of cells, should flow under the influence of vascular model via virtual links. Therefore, we should dynamically construct virtual vessel-blood links, because both of vessel and blood are moving. We respectively set the compressing, stretching and shearing parameters for the virtual vessel-blood links to $(1.0, 1.0, 0.0)$ to achieve close coupling. Suppose the sponsoring vascular chain-mail $p$ moves to new position $p^*$ as

$$q^* = q + (p^* - p). \quad (11)$$

When the guidewire moves in the vessel, blood flow should affect the guidewire motion. Analogously, we also set the compressing, stretching and shearing parameters for the virtual guidewire-blood links to $(1.0, 1.0, 0.0)$. Suppose the current position of the guidewire chain-mail is $q(x, y, z)$, one of its neighboring virtually-linked blood chain-mail is $p_i(x, y, z, v)$, and $v$ is the velocity. The guidewire motion induced by the virtual links can be calculated as

$$\begin{align*}
q^* &= q + \sum_{i \in S} w_i \cdot \Delta p_i \\
\Delta p_i &= \frac{w_i}{\Delta t}, \\
\omega_i &= \nabla \cdot \Delta t
\end{align*} \quad (12)$$

where $S$ is the neighboring blood chain-mail set, $w_i$ is the weight that is inversely proportional to the distance $d_i$ between guidewire chain-mail and its neighbors, and $\Delta t$ is the time step.

$$q^* = q + (p^* - p). \quad (11)$$

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\end{align*} \quad (12)$$

where $S$ is the neighboring blood chain-mail set, $w_i$ is the weight that is inversely proportional to the distance $d_i$ between guidewire chain-mail and its neighbors, and $\Delta t$ is the time step.

4.2. Collision Handling

We employ sign distance field (SDF) method [23] to conduct collision detection. The distance field function of surface $S$ together with its sign function are respectively defined as

$$D(p) = \min(\|p - q\|, \forall p \in R^3, q \in S), \quad (13)$$

$$\text{sgn}(p) = \begin{cases} 
-1, & \text{if } \langle p - q | n \rangle < 0 \\
1, & \text{if } \langle p - q | n \rangle \geq 0 \end{cases} \quad (14)$$

Here $\langle \cdot | \cdot \rangle$ denotes the inner product, $q$ is $p$'s closest point in $S$, and $n$ is the surface normal.
The calculation of SDF is shown in Fig. 8(a). By extending the vascular triangles along the face normal-Box, we can get the prisms, and calculate the axis-aligned bounding boxes(AABB) for the prisms. Then we can compute the signed distance for each point in bounding box. As shown in Fig. 8(b), the blue region represents parts of the negative distance field of the heart. To avoid excessive bending, we improve the collision detection by adding dynamic ray test. When the guidewire is pushed, we first update the position of the first tip chain-mail \( p_b \) and check if there is collision. If collision happens, collision response works to modify the chain-mail \( p_b \) to the legal position, and then we update the guidewire with the sponsoring chain-mail \( p_p \). To reduce the searching time and simplify the algorithm, we implement the collision detection based on CUDA.

As for collision response handling, we also calculate the signed distance \( d \) for each guidewire chain-mail \( p \) via the trilinear interpolation of its 8 corner points of the corresponding grid cell, and get the normal \( n \) by normalizing the analytic gradient of the trilinear interpolation. If the a guidewire chain-mail is detected to be closer to the vessel than the given threshold \( \epsilon \) or the chain-mail is outside the vessel, we modify its position via

\[
p' = p + |\epsilon - d| \cdot n.
\]

(15)

\( \Delta p = p' - p \), the displacements along the normal direction \( \Delta p_n \), and the displacements along the tangential direction \( \Delta p_t \), can be calculated as

\[
\Delta p_n = n \cdot (\Delta p \cdot n) \\
\Delta p_t = \Delta p - \Delta p_n
\]

(16)

where \( n \) denotes the normal of the closest surface to the chain-mail \( p \).

To reflect the friction behavior, we use Coulomb’s model for friction computation, which considers both static and kinetic frictions. We can formulate the friction force as

\[
F_{\text{friction}} = -C_f \cdot \Delta p_t, \quad 0 \leq C_f \leq 1.
\]

(17)

Specially, we set \( \beta = -\max\left(\frac{\Delta p_t \cdot C_t}{|\Delta p_t|}, 0\right) \) as a factor to indicate tangential movement, wherein \( \beta = 0 \) represents no movement and \( \beta = 1 \) means no friction. Therefore, we can rewrite the friction force formula as

\[
F_{\text{friction}} = (\beta - 1)\Delta p_t.
\]

(18)

4.3. Coupling Results

As shown in Fig. 9, through virtual links, we can easily realize uniform cross-model coupling. When the influence of the blood flow is imposed, the guidewire tends to move along the direction of the blood flow. As shown in Fig 10, the guidewire will stop until its tip collides with the vessel and the friction force between the guidewire and the vessel is large enough.

![Figure 9: Demonstration of the guidewire-vessel coupling. When dragging a vessel chain-mail, under the influence of virtual links, the motions of the blood flow and guidewire are affected. The circles in (a) show the coupled deformation results, the corresponding circles in (b) show the deformation-reverted effects.](image)

![Figure 10: Demonstration of the guidewire-blood coupling. From left to right: the guidewire motion without flow field, blood flow field, and the guidewire motion with flow field. The color-encoded lines in middle picture shows the flow velocities, wherein the colors indicates the speed while the direction of the line means the direction of the velocity.](image)

Besides, to verify the accuracy and stableness of our method, we respectively compare our inter-dynamic guidewire simulation result with those from position based dynamics (PBD) method and Cosserat-based method. As shown in Fig. 11, for the complex heart and blood vessel models, our method is as stable as the well-recognized PBD method, and our method outperforms it in physical realism. Meanwhile, although Cosserat-based method is fully based on physics and can achieve high fidelity, our method can also achieve comparable simulation result. However, Cosserat-based method is not only very unstable but also very inefficient when handling drastic force, frequent multi-point collision and long-time guidewire manipulation. For more vivid comparison demonstrations, please refer to...
5. Application In Virtual PCI Surgery

We integrate our guidewire simulation into a PCI surgery simulator. Our PCI surgery simulator consists of six modules: guidewire simulation, blood flow simulation, penalty force based haptic feedback calculation, PCI-specific haptic instrument, ray-casting based X-ray simulation, and realistic rendering system. Guidewire is one of the most important instruments. Fig. 13 shows the workflow of our surgical simulator. The trainee controls the virtual instrument through the haptic device. And the haptic device is applied to connect the real guidewire to the virtual guidewire. When the trainee inserts the guidewire from aortic arch, and feeds it to the target location along the vessel via pushing and rotating operations, the force feedback is transmitted to the haptic device at the same time, wherein the force feedback is calculated by summing the internal force and friction force resulted from the virtual guidewire as

\[ F_{\text{feedback}} = F_{\text{friction}} + F_{\text{internal}}. \]  

(19)

When the guidewire deform, the internal force can be calculated according to the displacement of the first two chain-mails with respect to its previous position. According to Hooke’s law, we have

\[ F_{\text{internal}} = (\text{Length}(P_1^t - P_2^t) - \text{Length}(P_1^{t-1} - P_2^{t-1})) \cdot k, \]  

(20)

where the \( P_1^t \) and \( P_2^t \) are the current positions of the first two chain-mails, \( P_1^{t-1} \) and \( P_2^{t-1} \) are their previous positions, and \( k \) is the constant coefficient.

Meanwhile, when inserting the guidewire in PCI simulator, the guidewire length should be determined by the actual situation. To this end, we build the Axis-Aligned Bounding Box (AABB) for the whole vascular model. If the number of guidewire chain-mails outside the AABB exceeds the specified threshold, we delete the particle from the tail end, and the guidewire becomes shorter. Similarly, if the number is less than another number threshold, which is not necessary the same with the former one, we add a new chain-mail \( Q(x, y, z) \) to the end of guidewire. The guidewire simulation result in our PCI simulator is shown in Fig. 14. Our method can achieve physically-plausible inter-dynamic guidewire simulation with unconditional-stableness and real-time efficiency, which can fully satisfy the actual applicative requirements of PCI surgery simulation. For more vivid simulation results, please refer to our supplementary video.

6. Conclusions

We have detailed a novel guidewire simulation method enabling inter-dynamics of the guidewire geometry, physical deformation, and their coupling with vessel wall and blood flow. Our new approach is based on the streamline of modeling, behavior, and interaction.
using the generalized, heterogeneous chain-mails. As a result, our method allows the guidewire to freely move (forward or backward), bend, or twist subject to the surrounding flowing blood and beating vessel constraints as well as collision contact. Meanwhile, we have integrated our guidewire simulation into a prototype PCI surgery simulator equipped with haptic feedback, wherein the guidewire length is allowed to be adaptively varying in order to enhance its flexibility and reduce the overall computational expense. The observed stableness, real-time efficiency, flexibility, and high-fidelity realism collectively validate the effectiveness of our method and show great promise for its practical applications in clinical training fields.

However, currently we only consider the one-way coupling that comes from vessel and blood to guidewire, but ignore the guidewire’s influence on the motion of blood and vessel. We plan to continue to improve our system’s performance via two-way coupling in order to make our PCI simulator more realistic and practical for medical practitioners.

References

[18] W.-C. Lin, Coupling hair with smoothed particle hydrodynamics fluids.