A new approach to haptic rendering of guidewires for use in minimally invasive surgical simulation

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ABSTRACT

Guidewire insertion is an imperative task of minimally invasive medical procedures. During the procedure, surgeons need to steer long flexible thin wires through patient’s blood vessels to reach a clinical target. In this paper, we present a novel approach to model haptics of guidewire insertion process for training simulation. The algorithm also allows for the analysis of the insertion process through subtle physical behaviours of guidewires via force feedbacks. The method includes a 6-DoF dynamic coupling between a rigid body, i.e. the virtual tool and the deformation of the wire simulated as an elastic rod. Instead of using the frictional contact force or the acceleration of the guidewire tip for haptic feedbacks, we compute constrained forces by directly connecting the virtual tool to the end of the guidewire. Therefore, the coupling scheme transmits haptic interactions through constrained dynamics between the virtual tool and the guidewire. Both positional and rotational control modes are implemented and evaluated with respect to the dynamics of the guidewire, user inputs and feedback forces. Experiments highlight the usability of our algorithm for an insertion procedure simulation with complex blood vessel structures.

KEYWORDS
physics simulation; 1D soft bodies; haptic interaction; multimodal interactive virtual environment

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1. INTRODUCTION

Minimally invasive procedures into the vascular system begin with a needle puncture into, for example, the femoral artery. A guidewire is then inserted through the needle into the artery, the needle removed, and then the wire is navigated to the pathology. A catheter, which is used for treatment of the pathology, is slipped over the guidewire. A high level of expertise is required to perform such a procedure, thus presenting challenges to the safety and cost effectiveness of training in patients [1].

The continuous progress in multimodal interaction technologies over recent years has resulted in greater user experiences of virtual reality-based medical simulations. These tools can be used to train surgical skills and improve the accuracy of operations in controlled environments remote from patients [2–4].

Haptic interactions are of particular relevant in the context of guidewire insertion procedures, since dexterity, handling skills and imagination are needed from surgeons to steer long flexible thin wires to reach a clinic target that is significantly far from the insertion site [5].

Essential steering operations are pushing, pulling and turning the proximal end of guidewires as it is inserted. Pathological and medical conditions that may not be shown clearly on guided images, but might be perceivable through tactile feedbacks. As yet, forces returned by the dedicated medical haptic interface with which the physical guidewire is interacting are not derived from simulation dynamics itself, rather than from simplified frictional responses and/or positional derivative of the guidewire tip [5,6].

There are two impediments that make it difficult to simulate guidewire haptics: a physically based model is needed to allow the realistic real-time simulation of
guidewire dynamics that correspond to insertion operations correctly; the tactile perception subject to guidewire dynamics is minuscule and subtle, presenting difficulty in perceiving clear feedback forces using common commercial haptic devices.

To address these challenges, we introduce a novel algorithm for guidewire haptics by encapsulating coupled dynamics between the real-time deformation of the guidewire and a constrained rigid body that carries exact degrees of freedom (DoF) at any point on the wire. This algorithm enables the development of a software tool for training and the analysis of the insertion process. Both positional and rotational controls are implemented and evaluated.

Compared to the existing specialist medical haptic system, the one described in this paper offers a wide range of possible experiments with low costs. The dynamic coupling scheme can be easily used for haptic feedbacks with other types of applications involving simulations for 1D flexible objects, such as hair, ropes, tubes and cables, without the modification to models for rod simulations.

2. BACKGROUND

Guidewire haptics are underpinned by realistic real-time simulations of guidewire dynamic deformations that involve substantial bending, moderate twisting and minimum stretching.

In this section, we firstly review dynamic models for elastic rods and guidewires that have been developed for both computer animation and/or medical simulations, then highlight related haptic rendering algorithms with focus on medical simulations. We refer readers to Reference [7] for a detailed review of general haptic algorithms and References [4,8] for haptics applied in medical applications.

Models for strand, rod and guidewire: Various methods have been developed for simulating guidewire dynamics using articulated rigid body chains [9,10], dynamic splines controlled by fixed contact control points at predefined locations [11] and finite beam elements based on linear elasticity theory [12,13]. Some models handle bendings only [11,13] while others are more challenge for real-time simulations [14].

In computer animation, Cosserat model for flexible rods is considered for static surgical thread [15] and for dynamics of flexible [16]. In Reference [17], spatially adaptive dynamic Cosserat model has been applied for interactive knot tying in ropes. An articulated rigid body system was also proposed to simulate knot tying [18] and a specialised solver that treats torsion as a state variable has been used to simulate thread [19]. More recently, Bergou et al. [20] developed a discrete geometry treatment for modelling elastic rod.

Considering comprehensive haptic interactions requiring 1 kHz frame rates, we aim at a real-time algorithm for guidewire simulation, which handles not only bending but also torsion effects, and at the same time with minimum stretching of the wire.

With respect to preserving length constraints for the guidewire during the simulation, although generalised coordinate systems such as articulated rigid body chains and other higher order finite element models like Cosserat schemes conserve length and allow twist to be modelled [12,17,18,21,22], these models make computations for collisions and constraints more difficult. Moreover, their relatively expensive per element computation leads to difficulties in haptic rendering. The super helix model [21], for example, scales quadratically as the number of elements of the rod are increased and even with the improved recursive scheme [22], the computation is still expensive for our simulation purpose. In particular, guidewires are long thin and flexible rods with length approximately 170 cm per wire.

Based on Reference [20] and extending our work [23], our guidewire simulation model (§3.3) has a minimal number of DoF (3-DoF for bending and 1-DoF for twisting) with a reduced coordinate formulation for dynamic computation. Furthermore, the discrete representation for the centerline of the guidewire handles collisions easily. To enforce length preservation, we apply an augmented Lagrangian formulation by introducing additional variables (i.e. Lagrange multipliers) to constraint the stretch of each edge of the centerline.

Although algorithms for 1D object simulations have been increasingly addressed in both computer graphics and medical simulation communities as discussed above, as yet, such 1D structures have not been exhaustively considered for haptic rendering models.

Haptic rendering: Various haptic rendering algorithms and approaches exist to cover many areas of applications, including medical simulations [7]. We consider the guidewire insertion task as tool-based manipulations (versus hand-based interactions), operating on 6-DoF (position and orientation) object manipulations and propose to apply constraint-based coupling of rigid body dynamics with rod dynamics.

In Reference [24], constrained rigid body dynamics was simulated as a quadratic programming problem on contact forces. Later, the author [25] extended the algorithm developed in Reference [26] for solving frictionless linear complementarity problem to the friction case, and achieved linear-time performance in practice. There is an increasing research development in addressing the rendering of interactions with solid deformable objects with particular attention devoted to surfaces and volumes [27–29]. Haptic interactions with 1D objects, however, have received comparably little attention. Recently, Bonanni et al. [30] presented a force rendering method for Cosserat elastic rods by adding an additional torque force to rod dynamics as the haptic input.

Haptic rendering was used in training for high-risk operations, including minimally invasive procedures. Several surgical procedures are performed with 4-DoF laparoscopy tools, which are valid only in a limited number
of situations and cannot capture full 6-DoF object manipulations [31]. Zorcolo et al. proposed a catheter insertion simulation with combined visual and haptic feedbacks based on a volumetric description of the simulation environment. A 3-DoF control mode was implemented and the feedback force was reconstructed from the movement trajectory of the tip of the PHANToM stylus [6].

The fundamental difference of our approach from the work discussed above is that the virtual tool is directly connected to the end of the guidewire and we compute haptic feedback forces based on the dynamic coupling between the virtual tool and the deformation of the guidewire, instead of only based on the frictional contacts or the velocity change of the guidewire tip. A related work was presented in Reference [32], in which a virtual tool was attached to the tip of a thread modelled as a mass-spring connection for guiding the suture thread through a soft tissue that was also base a mass-spring network. The haptic feedback returned by the environment with which the thread interacts was not derived from the rod dynamics itself, which, in contrast, our methods computes haptic feedbacks from dynamic deformations of the rod.

3. DYNAMIC COUPLING BETWEEN HAPTICS AND GUIDEWIRE

3.1. System Overview

Figure 1(A) depicts an overview of the dynamic coupling scheme, in which the force $F_h$ and the torque $T_h$ are haptic inputs set by the device to the virtual tool for the guidewire insertion simulation. As indicated in Figure 1(B), the constrained point $P_c$ at the end of the guidewire connects to the virtual tool. To enable basic pushing, pulling and twisting operations, the force $F_{push}$ and the turning angle $\Delta \theta$ are computed from the control inputs of the virtual tool.

During an insertion simulation, we also compute Axis Aligned Bounding Box collision detections alongside dynamic computation on bending and twisting forces for each node of the centerline of the guidewire. The dynamic response of the guidewire to the insertion operations as well as to contact collisions with the blood vessel’s inner wall generates the deformation force $F_g$ of the guidewire at the constrained node. As a result of the dynamic coupling between the rigid body and the flexible wire, haptic feedbacks $F_c$ and the torque $T_c$ are computed. We set the pushing force at the end of the guidewire $F_{push} = F_h$, the haptic input force.

3.2. Constrained Rigid Body Dynamic Coupling

To solve the haptic feedbacks $F_c$ and the torque $T_c$, the virtual tool has the centre of mass $M_i$ at the position $P_i$ and a vector $r$ from the centre of mass to the constrained node $P_c$ [Figure 1(B)]. The dynamics equations for the virtual tool are solved for the linear and angular acceleration terms of the rigid body:

$$ \ddot{P}_c = M_i^{-1} F_c + M_i^{-1} F_{ex} + \alpha_t \times r + \omega_t \times (\omega_t \times r), $$

$$ \alpha_t = I_i^{-1} (r \times F_c) + I_i^{-1} \tau_{ex} - I_i^{-1} (\omega_t \times L_t) $$

$$ L_t = r \times P_c $$

The subscript $t$ indicates the virtual tool, $F_{ex}$ is the external force on the body, which include gravity and damping forces and, $\alpha_t$ is the angular acceleration. The inertia tensor $I_i$ and the external torque on the body $\tau_{ex}$ can be found by standard rigid body dynamics and the angular velocity $\omega_t$. We have following constrained equations:

$$ P_t - r - P_c = 0 $$

The subscript $c$ indicates the constrained point $P_c$ at the end of the guidewire. Substituting (2) into (1) and rearrange for a system of linear equations $Ax = b$, we have:

$$ \ddot{P}_c = [M_i^{-1} - \bar{r} t_i^{-1} \bar{r}] F_c + b_t $$

$$ b_t = M_i^{-1} F_{ex} + \alpha_t \times (\omega_t \times r) + [t_i^{-1} \tau_{ex} - t_i^{-1} (\omega_t \times L_t)] \times r $$

$$ T_c = F_{ct} \times R $$

In Equation (5), $\bar{r}$ is the skew symmetric matrix of $r$. In Equation (7), $F_{ct}^0$ is the force vector that is normal to the tangential force vector to the centreline of the guidewire at $P_c$. 

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3.3. Guidewire Dynamics

The guidewire is simulated as a discrete elastic rod [20], where the total potential energy of the wire due to deformation is computed as:

\[ E(G) = E_{\text{bend}} + E_{\text{twist}} + E_{\text{stretch}} \]  

(8)

\[ E_{\text{bend}} = \frac{1}{2} \int \mu \kappa^2 ds \]  

(9)

\[ E_{\text{twist}} = \frac{1}{2} \int \beta m^2 ds \]  

(10)

\[ E_{\text{stretch}} = \frac{1}{2} \sum_{j=0}^{n-1} k (e_j - e_j)^2 \]  

(11)

To compute bending energy, we find the centreline’s curvature vector \( \kappa \) is equal to the vector that is normal to the tangential vector of an arc length parameterised curve \( s \) in \( \mathbb{R}^3 \). The twist energy is found by computing the twist quantity expressed in terms of the pointwise twist \( m_i \) about the centreline at each node \( i \), which is found by a quasi-static update on the elastic rod dynamics. We simulate the guidewire as a naturally straight isotropic rod. Given appropriate boundary conditions, the assumption leads to a simplified computation of twist \( m_i = l_i(\Delta \theta)/E \), where \( E \) is the twice of the initial guidewire length, the boundary condition \( \Delta \theta \approx -\mathbf{(T}_h \mathbf{R}^{-1})/dr \), where \( \mathbf{R} \) is the cross section radius of the rigid body at \( \mathbf{p}_c \).

In Equation (11), \( \mathbf{e} \) and \( \mathbf{r} \) are initial and current edge length, respectively. In Equations (9)–(11), \( \mu \), \( \beta \) and \( k \) are material constants.

Despite the computation efficiency, the algorithm for guidewire simulation requires a method to maintain minimum stretching of the wire. We employ an augmented Lagrangian formulation by introducing Lagrange multipliers to enforce length constraints during the time integration step of the algorithm. At each node of the centerline, there are four DoF, i.e. position \( \mathbf{x}_i \), and the twist \( \theta_j, i = 0, \ldots, n \), where \( n \) is the number of nodes. Therefore, the constrained Lagrangian equations for the guidewire is:

\[ \frac{d}{dt} \left( \frac{\partial \mathbf{K}}{\partial \mathbf{x}_i} \right) - \frac{\partial \mathbf{K}}{\partial \mathbf{x}_i} = \mathbf{F}_i - \sum_{j=0}^{n-1} \frac{\partial 
E(G)}{\partial \mathbf{x}_j} \frac{\partial \theta_j}{\partial \mathbf{x}_i} - \Sigma \lambda_j \mathbf{C}_j \]  

(12)

where \( \mathbf{K} \) is the kinetic energy and \( \mathbf{x}_i \) and \( \theta_j \) are the DoF at node \( i \). The external force acting on each node \( \mathbf{F}_i \) includes collision and frictional forces and \( 
E(G) \) is found by equation (8). \( \mathbf{C}_j \lambda_j \) is the stretch constrain force on each node, where \( j \) is the number of edges of the centerline.

Equation (12) is solved using a system of linear equations:

\[
\begin{bmatrix}
M & 0 & -C^T_l
0 & M & -C^T_l
C_l & C_l & C_l & 0
\end{bmatrix}
\begin{bmatrix}
\dot{P}_x
\dot{P}_y
\dot{P}_z
\lambda
\end{bmatrix} =
\begin{bmatrix}
B_x
B_y
B_z
C \dot{P}^{-1}
\end{bmatrix}
\]  

(13)

Solving Equation (13), we obtain the acceleration \( \ddot{P}_c \) at the constrained point. Rearranging Equation (5) for the haptic feedback force \( \mathbf{F}_c \), we have:

\[ \mathbf{F}_c = [\mathbf{M}_c^{-1} - \mathbf{r}_c^{-1}] \mathbf{r}_c^{-1} (\mathbf{b}_c - \dot{\mathbf{P}}_c) \]  

(14)

\( \mathbf{F}_c \) is found by solving Equation (14).

4. HAPTIC CONTROL AND FEEDBACKS

The haptic control of the virtual tool to the end of the guidewire produces deformation distortions of the wire, which in turn generates feedback forces through the constrained dynamic coupling. We evaluate the effect of the guidewire deformation on the feedback force by fixing the tip of the guidewire and applying positional and rotational controls to the guidewire.

4.1. Positional Control

When only applying a constant velocity to the end of the wire, bending deformation starts at the fixed tip of the wire and is gradually transmitted across the guidewire. Thus, the haptic feedback force corresponding to guidewire deformations is solely generated by the bending distortions. Our haptic algorithm for generating the feedback force is sensitive to the bending of the wire, but more so when the deformation is near to the constrained point connecting the virtual tool. Figure 2 shows a good agreement of the haptic feedback force \( \mathbf{F}_c \) (the green curve) as a result of bending deformations of the wire with the bending force \( \mathbf{F}_c \) at the constrained point on the guidewire (the blue curve).

The feedback force \( \mathbf{F}_c \) (B in Figure 2) increases to 0.48 N when both large and small bending deformations occur across the wire [the bottom image (B) of Figure 2 shows the corresponding deformation], whereas a relatively straight section of the guidewire near to the virtual tool [the bottom image (C) of Figure 2] decreases the feedback force [(C) in Figure 2] to 0.03 N. The feedback force due to bending is then increased to 1.12 N [(D) in Figure 2] by continuously moving the end of the wire closer to the fixed tip via a constant displacement 0.2 mm per frame, thus, bending the wire even further [the bottom image (D) of Figure 2].

4.2. Rotational Control

To test haptic feedback forces produced by twisting effects of the wire, we fix both ends of the wire and start with a default natural straight configuration.
When only applying a rotational control to the virtual tool by a constant turning angle about the centreline, the wire transmits the twist across the wire from the end to the tip. As a result of a constant turning, the feedback force $F_c$ (the green curve) in response to twisting deformations is evaluated in terms of the total of turning angles with respect to the force $F_c$ at the constrained point (the blue curve). The feedback force increases sharply to 3.35 N at the beginning [(B) in Figure 3] due to the twist of 63.2 radians and decreases to 0.02 N when the twisting energy is dissipated across the entire guidewire [(C) in Figure 3]. The insert images correspond to the twist deformations at points B–C.

5. EXPERIMENT

To demonstrate how the encapsulated feedback force responses to an insertion process, we place a number of simulated plaques at predefined locations inside a vessel.

Intuitively, during an operation with constant linear and angular velocities, when the tip of the guidewire collides with a clot, as shown in the simulation snap shots (A) and (C) in the bottom images of Figure 4, the guidewire bends from the tip, leading to an expected increase in the haptic feedback force until the clot is broken by a stronger force applied from the guidewire tip. Also consequently, the feedback force is expected to decrease immediately once the blockage is cleared, as shown in the simulation snap shots (B) and (D) in the bottom images of Figure 4.

As in the chart of Figure 4, the deformation force $F_g$ at the constrained point is almost constant at about 0.16 N and haptic feedback force $F_c$ is also a constant at about 0.14 N before clot collisions. When the tip of the wire collides with blockages, the force $F_g$ at the constrained point $P_c$ increases to 0.28 and 0.24 N, respectively at points A and C, whilst, in response to the deformation, the haptic feedback force $F_c$ increases to 0.24 N and 0.31 N, respectively at points A and C. The deformation force $F_g$ decrease to 0.01 N at point B and 0.06 N at point D as soon as the clot blockages are cleared, as shown in the snapshot B and D of the bottom images in Figure 4. The feedback force $F_c$ is decreased to 0.11 N at point B and to 0.12 N at point D.
As stated in Reference [7], the force update rate has effects on stability of haptic rendering. In different insertion cases, instabilities may arise in the simulation of collision contacts due to the rapid change of the bending of the guidewire. In these cases, haptic devices must react with large changes in force to small changes in the position. As shown in the above tests, there are places where the feedback force oscillates. The oscillations, however, are not caused by the force update rate since our algorithm is efficient for real-time computations at 90 fps on the Intel core 2 Duo Processor P7350 CPU at 2.0 GHz, but largely because of the imperfect preservation on edge lengths. The issue will be addressed in future work.

6. CONCLUSION

The new haptic algorithm for guidewire insertion allows us to simulate guidewire haptics with 6-DoF inputs to manipulate the medical instrument using a commercial haptic device. The algorithm is also capable of encapsulating haptic feedback forces accurately in response to bending and twisting of the guidewire deformations. The validation tests illustrate that feedback forces produced by the dynamic coupling between a virtual tool and dynamic deformations of the instrument are sensitive and accurate. We are currently considering the effect of frictional contacts on the guidewire and its subsequent effects on haptic rendering.

The use of the Lagrange multipliers leads to additional variables to be solved during time integration step. We would like to explore alternative methods for the length constraint enforcement without energy being dissipated during the constraint enforcement step.

Finally, a user study is an important aspect to be included in our future development. Another area of future work is to compare the results with force measurements on real guidewires. In addition, an interesting related topic would be in simulating an isotropic properties of guidewires to ensure higher order of accuracy in modelling guidewire dynamics, thus accurate haptics.

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