Image-based Intraluminal Contact Force Monitoring in Robotic Vascular Navigation

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Abstract—Embolization, stroke, ischaemic lesion, and perforation remain significant concerns in endovascular interventions. Sensing catheter interaction inside the artery is advantageous to minimize such complications and enhances navigation safety. Intraluminal information is currently limited due to the lack of intravascular contact sensing technologies. We present monitoring of the intraluminal catheter interaction with the arterial wall using an image-based estimation approach within vascular robotic navigation. The proposed image-based method employs continuous finite element simulation of the catheter motion using imaging data to estimate multi point forces along catheter-vessel interaction. We implemented imaging algorithms to detect and track contacts, and compute catheter pose measurements. The catheter model is constructed based on the nonlinear beam element and flexural rigidity distribution. During remote cannulation of aortic arteries, intraluminal monitoring achieved tracking local contact forces, building contour map of force on the arterial wall and estimating structural stress of catheter. Shape estimation error was within 2% range. Results suggests that high risk intraluminal forces may happen even in low insertion forces. The presented online monitoring tool delivers insight into the intraluminal behavior of catheters and is well-suited for intraoperative visual guidance of clinician, robotic control vascular system and optimizing interventional devices design.

Index Terms—Robotic vascular intervention, intravascular sensing, contact force monitoring, finite element modeling, image-based estimation, vessel-instrument interaction, catheterization.

I. INTRODUCTION

ENDOVASCULAR interventions are leading treatments and diagnoses for cardiovascular disease. Despite improvements in tools and techniques, interprocedural risks of embolization, ischaemic brain lesions, and stroke in percutaneous procedures are still high, especially in carotid artery stenting [1], [2]. Studies reported that 50% of the cases after carotid stenting had a new ischaemic lesion on diffusion-weighted imaging (DWI) of the post-treatment scan due to embolism. The high number of microemboli, which has been reported during navigation of catheters and guidewires, highlights the importance of tool-tissue interaction [4], [5]. Studies have further pointed out other complications, perforation, thrombosis and dissection, as a result of catheter/guidewire interactions with arterial wall [6]. In the case of stenosis treatment, excess insertion force to cross occlusion could raise the likelihood of perforating [7]. Limited motion of under-actuated conventional catheter and guidewire can further add to the risk, especially in the case of a diseased and torturous vessel. These findings suggest that the intraluminal interaction contact force is one of the determinants of patient safety and procedure efficiency. Monitoring force information intraoperatively has the potential to enhance navigation safety and efficiency. Practical applications are threefold:

- Integrating CF data into intraoperative visual guidance for clinicians to bring safe catheter/guidewire manipulation. Intraluminal insight could minimize the risk of stroke, embolism, vessel perforation, or dissection. It could further limit human mishandling, especially in the case of novices, with implications to improve training for complex procedures.
- The development of an automated robotic surgery system that maintains CF in a safe range and improves smoothness using realtime intraluminal data. Vascular robotic technology demonstrated definite advantages in catheter controllability, stability, precision, and lower radiation for clinicians [8], [9], [10], [11], [12]. With the advent of artificial intelligence, a safe and autonomous form of robotic surgery can be introduced [13], [12].
- Intraluminal force information, along with catheter deflection data can also be used for research on design of interventional tools aimed to minimize vessel injuries and maximize maneuverability, torquability and deliverability. It would be beneficial to study the intraluminal performance of interventional devices and optimize their design.

Intraluminal information is limited due to the lack of force sensing elements in the endovascular tools or other remote sensing technologies. In electrophysiology, catheters’ tip force is essential due to the chance of over-burn in case of excessive CF or inferior lesion quality in weak contact [14]. Studies proposed ablation catheters with tip force sensors [15], [16], [17] and some are available commercially [18], [19]. However, extra cost, less maneuverability, and added weight are associated complications of sensorized catheters. Indirect force estimation of the tip on ablation catheter has worked based on modeling and shape analysis, e.g., kinematic model, cosserat rod, and piecewise planar elastica [20], [21], [22], [23], [24] where reporting promising results. On the other hand, proximal insertion force measurement and methods to
control it has been used in robotic catheterization [25], even though this force does not represent local vessel interaction forces. Quantitative analysis of the total contact force between instrument and cardiovascular phantoms has been established in robotic navigation [26] as well and used as a comparison metric [7], whereas the relation between total exerted force and local intraluminal CF is not entirely clear.

Monitoring intraluminal contact forces through the length of catheter-vessel interactions is a gap. In previous work [27], we presented a sensor-less approach to estimate multi-point load forces at the side or tip of interventional tools. This study develops the approach to an intraluminal contact force monitoring system for the test case of Carotid and Subclavian arteries robotic cannulation. A two-degree freedom (DOF) robotic insertion unit was fabricated for remote catheter navigation in an anthropomorphic aortic arch phantom. The 6 DOF force/torque sensor is coupled to the phantom to study total force exerted, i.e., to compare the resultant of contact forces with intraluminal forces. Flexural rigidity distribution of catheter/guidewire is measured through a sequential bending test over its length. A real-time image-based pose measurement algorithm is developed. A nonlinear finite element model is set to simulate the catheter using image-based measurement data. Solving the inverse FEM model estimates the contact forces and catheter’s deflected shape. The quality of the model is assessed by analyzing errors in shape estimation. We present the contour map of intraluminal CF on vessel boundaries and the tracking of local CF. We also performed a quantitative analysis of CF compared with resultant forces. Furthermore, we established the stress in the catheter body from the FEM model.

II. METHODS AND ALGORITHM

A. Contact force estimation

The accuracy of the proposed sensor-less force estimation concept was shown in the previous work [27] through a direct force measurement setup, i.e., single contact point phantoms mounted on a 6-DOF F/T sensor. Estimation accuracy over 87% was achieved in agreement with force sensor measurements [27]. The proposed method is based on continuous simulation of catheter deflection as a beam using data from real-time imaging. Image processing segments and tracks deflected catheter shape and locates interaction with the internal wall of the vasculature. Using image processing data and prior knowledge about the intrinsic shape of the catheter, an FEM model is built and solved. Deflections induced by vessel walls at the contact point are the boundary conditions of the model, and reaction forces at the same boundary conditions return contact forces interaction. Deflections are fed to the model continuously to simulate catheter manipulation. The FEM model is based on the Timoshenko beam element, considering nonlinear large deformation. The model assumes the catheter structure is a uniform round beam with an equivalent bending modulus.

Let us describe a schematic view of a catheter-vessel interaction as shown in Fig. 1. A part of the catheter begins at any desired point ending to the tip is being modeled as a cantilever, which has all contact points included (window shown in Fig. 1). The model is created in the local coordinate of the catheter, having the x axis tangent to its base. This is to compute deflection from the catheter base perspective and isolate it from the global coordinate. Cantilever model has intrinsic catheter body with the length equal to the deflected shape in the window, $L_{cath} = L_{tip}$. The model length varies as catheter advances or retracts during navigation. Initial positions of contacts on the model are the ones having equal lengths as on the deflected catheter. For instance, position of contact point $i$ is $CP_i$, which has length $L_i$ along the deflected shape, and the corresponding point with the same length on cantilever model would be the boundary condition $BC_i$, i.e., it is the contact point before being deflected. Contact deflection, $\vec{d}_i$, is the displacement between deflected contact point position, $CP_i$, and its rest position on the model, $BC_i$ (see (1) and (2)). Deflections, $\psi(d) = \{\vec{d}_i\}$, are applied to boundary conditions $BC = \{BC_i\}$ respectively. By solving the inverse FEM model, the catheter is being deflected as it is interacting with the vessel, which simulates the current state. The FEM model computes the reaction forces, $CF = \{F_i\}$, at boundary conditions as estimates of contact forces on the vasculature. Even though the magnitude of contact forces matters for safety, CF can be broken into components of normal ($f_n$) and tangential/frictional forces ($f_t$) if such information is needed. As the catheter navigates through the vasculature, its deflected shape, the position of contact points, and the amount of deflections change. The cantilever FEM model parameters and computed contact forces are being updated in real-time accordingly.

$$\vec{d} = \vec{u}_x + \vec{u}_y$$  \hspace{1cm} (1)

$$\psi(d) = \begin{bmatrix} \vec{d}_1^x \\ \vdots \\ \vec{d}_i^x \\ \vdots \\ \vec{d}_n^x \\ \vec{d}_1^y \\ \vdots \\ \vec{d}_i^y \\ \vdots \\ \vec{d}_n^y \end{bmatrix} = \begin{bmatrix} \vec{x}_{CP_1} - \vec{x}_{BC_1} & \vec{y}_{CP_1} - \vec{y}_{BC_1} \\ \vdots & \vdots \\ \vec{x}_{CP_n} - \vec{x}_{BC_n} & \vec{y}_{CP_n} - \vec{y}_{BC_n} \end{bmatrix}$$  \hspace{1cm} (2)

Near-end part of the catheter can be simplified as a planar beam, where the effect of minor out-plane deflection is negligible compare to large planar deflections.

![Fig. 1: Force estimation concept showing deflected catheter and constructed cantilever model. $CP_i$s are the contact points with length of $L_i$ on deflected shape and $BC_i$s are associated boundary condition on model.](image-url)
B. Image segmentation and contact tracking

Image segmenting obtains data required for the model, including lengths from catheter base to every contact point and tip, positions and deflections of contact points in the local coordinates of the catheter (see Fig. 1). The OpenCV (Open Source Computer Vision) library is used for image processing by programming in C++. Images are continuously taken through an RGB camera with 1920*1080 pixels and 30Hz frequency. The RGB image is converted to grayscale to be similar to X-ray fluoroscopic images. Medical imaging and visualization systems (X-ray fluoroscopy, CT and MRI) enable detection and tracking of the shape of the catheter and vessel boundaries in cardiovascular interventions. In this study, we detect the vessel using the contrast between the vascular phantom and the background, similar to injecting contrast agent to visualize the lumen in X-ray fluoroscopy. We assume there is no extreme movement or change in vessels during the procedure; however, an online calibration technique can be used to match the vessel image in case of any movement like breathing in a clinical case.

Image segmentation is divided into two main phases: extracting mask images of catheter and vessel boundaries, followed by search and tracking (Fig 2). A Gaussian filter is applied to filter out noise and then, a canny-edge detector algorithm is used to extract vessel boundaries [28], [29]. To extract catheter pixels, a thresholding operation is implemented on the grayscale frame. Then, dilation followed by erosion are applied to fill possible gaps and connect lines. The imaging algorithm is tested on the cannulation of aortic arteries by a guidewire (experimental setup is explained in later sections). Fig 2(b) and (d) display images of the sample of phantom boundary mask and a guidewire mask, respectively.

An algorithm with a moving search window normal to the guidewire curve is designed to sweep the entire length of the guidewire, as depicted in Fig 2. The algorithm moves a rectangle search window along the tangent to the guidewire θ(s) = dy/dx, where s is curve length. The starting point of the search is determined with an initial search step over the base frame. The search window finds guidewire pixels and computes their centroid $C_G$. In each window, a parallel logic operation is made on masked vessel image to find the point of contact with boundaries (VB) where the distance of the guidewire centerline and VB, $C_V$, is less than a predefined contact distance. The search window position is updated along θ(s) once each centroid point is found and moved toward the tip. The tip is detected where the number of contiguous pixels is less than a predefined amount. Algorithm 1 presents the pseudo-code of extracting and moving search window method in segmentation and tracking. The proposed moving window search is robust to possible gaps or missing points.

Fig. 3 shows step by step segmentation results for guidewire interaction with the aortic arch phantom. Fig. 3(g) and (f) depicts segmented guidewire-vessel interaction overlaid on original frames, which verifies the effectiveness of the segmentation and search algorithm.

Last step converts pixel values to metric using camera calibration parameters. Afterward, the length of any desired point along the guidewire can be computed by numerical integration. Derived parameters are: tip pose $P_t = \{x_t, y_t, l_t\}$, contact point pose $\{P_{CPi}\} = \{x_{CPi}, y_{CPi}, l_{CPi}\}$ and GW center-line pose $\{C_{GW}\}$.

Algorithm 1 Segmentation and tracking

```
Function image processing
video.read(frame)
2: frame ← RGBtoGry(frame) ▷ filter out noise
frame ← Gaussianfilter(frame) ▷ extract vessel boundaries
4: BVS ← canny(frame) ▷ extract guidewire
while video.read(frame) do ▷ tracking
5: GW ← threshold(frame) ▷ extract guidewire
GW ← dilate and then erode GW
6: while Moving search window is true do ▷ search
for search pixel of window normal to catheter
7: if pixel value is 0 in GW then ▷ compute center of pixels
compute center of pixels
end if
end for
10: C_{GW} ← \{centroid(x), centroid(y)\}
if C_{GW} - BVS < predefined then ▷ contact detection
14: P_{CPi} ← \{centroid(x), centroid(y), L_{CPi}\}
end if
16: if nPixel ≤ minNPixel then ▷ tip detection
P_t ← \{centroid(x), centroid(y), l_{tip}\}
else
20: update the tangent to the catheter
\{P_{MovingWindow}\} ← update and move search window
end if
22: end if
24: return C_{GW}, BVS, P_{CPi}, P_t
```

C. Finite element model

Consider a planar two-node beam element of length l, where each node has three degree of freedom, as defined in Fig. 4. Nodal displacement vector contains longitudinal ($u_x$), transverse ($u_y$) displacement and rotation ($\varphi$) for both nodes.

$$\mathbf{e} = \begin{bmatrix} u_{x1} & u_{y1} & \varphi_1 & u_{x2} & u_{y2} & \varphi_2 \end{bmatrix}^T$$  \hspace{1cm} (3)

Global elements coordinate vector is

$$\{\mathbf{e}\} \equiv \begin{bmatrix} u_{x1} & u_{y1} & \varphi_1 & \ldots & u_{x(N+1)} & u_{y(N+1)} & \varphi(N+1) \end{bmatrix}^T \in \mathbb{R}^n$$  \hspace{1cm} (4)

where $N$ is the number of element and $n = 3(N + 1)$ is the number of degrees of freedom.

In the previous study, we presented results based on Euler-Bernoulli (EB) beam theory. EB assumes thin beams where angular distortion due to shear deformation is considered negligible, and cross-section is perpendicular to the bending line (neutral axis). Timoshenko beam theory is generic compared to EB theory. Its employed formulation accounts for large axial, bending, and shear deformations as well as large translation
and rotation of the beam structure. The governing equations of Timoshenko beam theory are the following:

\[ \frac{d^2}{dx^2} (EI \frac{d\varphi}{dx}) = q(x) \]  

(5)

\[ \frac{d\psi}{dx} = \varphi - \kappa A \gamma \frac{d}{dx} (EI \frac{d\varphi}{dx}) \]  

(6)

where \( E \) and \( G \) are the elastic modulus and shear modulus. \( L, I \) and \( A \) are length of the beam, second moment of area and cross section area, respectively. \( \kappa \) is the Timoshenko shear coefficient and \( q(x) \) is the load. Fig. 4 compares Timoshenko beam deformation to EB, where the cross-section of a beam element remains plan but not necessarily perpendicular to the beam axis, i.e., \( \varphi(x) \neq \frac{dw}{dx} \) and \( \gamma_{xy} \) is shear. Timoshenko can be used for thick as well as slender beams, which is the more accurate choice for modeling of catheters and guidewires, especially where the beam cross-sectional dimensions is small compared to typical distances along its axis.

The equilibrium of a finite element beam model at the time \( t \) is

\[ M \ddot{\mathbf{u}} + C \dot{\mathbf{u}} = \mathbf{R} - \mathbf{F} \]  

(7)

where \( M \in R^{n \times n} \) is the mass matrix and \( C = \alpha M \) is the damping matrix proportional to mass with coefficient \( \alpha \). \( R \in R^n \) is the external load vector and \( F \in R^n \) is the internal force vector in global coordinates. \( \mathbf{u} \) and \( \dot{\mathbf{u}} \) are acceleration and velocity vectors, respectively. The velocities and acceleration vectors are small during navigation because of continuous contact with the vessel. Further, considering the low mass of the catheter/guidewire, the dynamic force is small compared to large external forces caused by large deflections. Thus, we are implementing a quasi-static solution to minimize the computational cost and reach a real-time execution.

Nodal force is formed based on individual vectors consid-
Nlgeom with active associated beam model is solved using ABAQUS, B21 beam next section. The FEM modeling is coded in C++ and the to its associated boundary condition. Each element has an vector having an equal length to contact points L conditions located on the model as of the points (with L L in Algorithm 2. The part is made based on intrinsic shape characteristics is needed. A geometrical computation is essential to create the cantilever model from imaging data, as given

\[ u \]_t vectors an inverse finite element since we have nodal displacement \( K \) and \( \Delta t \) parts [34]. Strain-displacement equations contain nonlinear terms that must be considered in the nonlinear part of the stiffness matrix. \( K_{NL} \) is achieved by applying Castigliano’s theorem to the strain energy which counts the interaction between axial load and lateral deformation [35].

After transferring finite element matrix of the local principal axis of the elements to global cartesian co-ordinate and performing element assemblage process, the incremental equilibrium equation of quasi-static analysis is:

\[ (t^L_K + t^L_{K_{NL}}) \Delta t u = t^+ \Delta t R - t^L F \] (10)

where \( t^L_K \) and \( t^L_{K_{NL}} \) are stiffness matrices at time \( t \), \( t^L F \) is nodal point force of time \( t \) and \( t^+ \Delta t R \) is externally applied nodal load at time \( t + \Delta t \) [36]. The case of our problem is an inverse finite element since we have nodal displacement vectors \( u \) from imaging data and calculating \( R \) as interaction contact forces. \( R \) is the resultant of all external forces applied to catheter from vessel, including normal and friction forces.

\[ t^L R = t^L F + t^L R_n \] (11)

\( R \) is computed regardless of type, condition, or material properties of the external environment. Viscoelastic effect of tissue is already seen in nodal displacement \( u \). Consequently, no information about contact condition and tissue characteristics is needed. A geometrical computation is essential to create the cantilever model from imaging data, as given in Algorithm 2. The part is made based on intrinsic shape information and total length of GW, \( L_{tip} \). Position of boundary conditions located on the model as of the points (with \( L_{BC} \)) having an equal length to contact points \( L_{CP} \). Displacement vector \( i \) is between \( BC_i \) and \( CP_i \) (see 1) which is assigned to its associated boundary condition. Each element has an individual mechanical property \( EI \), which is elaborated in the next section. The FEM modeling is coded in C++ and the associated beam model is solved using ABAQUS, B21 beam with active Nlgeom.

**Algorithm 2 Model preparation**

1. **Input** Data from image-based pose measurement
2. \( L_{beam} \leftarrow L_{tip} \)
3. Create cantilever beam from intrinsic shape and \( L_{beam} \)
4. for \( i < N_{CP} \) do
   5. \( L_{BC_i} \leftarrow L_{CP} \)
   6. Locate position of \( P_{BC_i} \) on cantilever beam based on \( L_{BC_i} \)
   7. Apply \( d_i \) to \( BC_i \)
   8. for \( n < N_{elements} \) do
      9. Compute element length from tip \( S_e \)
      10. \( EI_{element} \leftarrow EI(S_e) \) (assign mechanical property

**Fig. 5:** Three-point bending test: schematic concept (a), Electroforce dynamic testing machine (b), deflected guidewire under test (c).

**Fig. 6:** Flexural rigidity distribution of the guidewire based on distance from distal tip. Average value and deviation are shown.

### D. Flexural rigidity distribution

Cardiovascular devices have different mechanical properties, designs, and geometries to achieve specific tasks. The local flexural rigidity of guidewires varies along the length. Clinical performance, i.e., steerability, torquability, penetrability, deliverability, and safety, depends on mechanical property profile over the length [37]. It needs lower flexural rigidity at the distal tip while higher rigidity is desired beyond the tip (proximal end). Flexural rigidity distribution is measured by sequential three-point bending tests along the length.

Three points bend test shown in Fig 5(a) behaves as described in (12), where \( F \) is applied load, \( w \) is downward deflection at the middle of support span, \( l \) is support span, \( E \) is bending modulus and \( I \) is second moment of inertia. For circular beams (rod or tube), i.e. guidewire, catheter or guidewire in support catheter, \( I \) can be expresses as of \( t = \pi r^4 \).

\[ F = \frac{48wEI}{l^3} \] (12)

Bending modulus is proportional to the fourth power of the radius, so a slight change in radius can greatly affect the accuracy of bending modulus value, consequently force estimation. Hence, we aim to determine and utilize \( EI \). Fig 6 shows flexural rigidity distribution of Zipwire™ Stiff guidewire (Boston Scientific, USA) measured as a function of length from tip, \( EI(s) \). To do so, sequential three-point bending
tests were performed over 0-500 mm distal end at increments of 10 mm (over 0 - 200mm), 20 mm (over 200 - 400mm) and 50mm (over 400 - 500mm). We used a Bose® Electroforce 3200 dynamic testing machine (Bose Corp., Massachusetts, US). The support span length was 30 mm, and loading speed of the machine was fixed at 20 mm/min to eliminate any potential dynamic effect. Each test repeated three times, and the average value and standard deviation were computed. Rigidity changes sharply in 15 cm of the distal tip and stays consistent beyond it. In the model creation, the algorithm measures element lengths from the tip and assigns individual mechanical properties to the elements based on the rigidity graph (see algorithm lines 9 - 11).

III. EXPERIMENTAL PLATFORM AND STUDY PROTOCOL

A. Experimental setup

The setup has a transparent, realistic, anthropomorphic training phantom representing aortic arch, and extended carotid structure with normal configuration (Fig. 7). A camera is mounted on top of the phantom to provide image feedback for force estimation and also operator visual guidance. A robotic platform is designed to remotely navigate the guidewire (Fig. 7 (b)). The robotic driver (slave) is controlled by the operator (master) using a keyboard/joystick to command manipulation procedure. Zipwire™ Stiff guidewire with angled tip shape is used with no support catheter. Operator performs navigation under image-guidance using live grayscale image which is simulated as the 2D fluoroscopy and projected on a monitor. It is a general case study of any non-steerable catheter or guidewire navigation through cardiovascular anatomy.

Part of the aim of this paper is to study and compare the total force exerted on the vasculature, i.e., insertion force in our setup, with intraluminal interaction contact forces. To do so, the phantom is mounted on a six-axis force-torque (F/T) sensor as shown in Fig 7 (d) (Mini40; ATI Industrial Automation, Inc, Apex, NC). It provides force measurement in X, Y and Z directions. The phantom is placed on its gravity center to hinder tilting or vibrating during the procedure. The force sensor platform and phantom are isolated from any other support or contact, so that force measurement is corresponding to the resultant catheter-vessel force vector. The F/T sensor reading are obtained through a Data Acquisition system (National Instrument DAQ) using a software which records data at 60 HZ and computes resultant magnitudes of the 3D force measurements. Measurements are zeroed at the very beginning of each test to omit weight of the phantom or any undesired loads.

Robotic Navigation System: A robotic driver system is designed based on the methods in conventional manual navigation of non-steerable catheters/guidewire, i.e., techniques similar to the push, pull, and twist. The design (Fig. 7 (c)) has two degrees of freedom as of insertion and rotation similar to the designs in [38]. It consists of a translational driver unit mounted on a slip-ring gantry, which allows the rotational motion of the catheter simultaneous to insertion motion. Simultaneous rotation-insertion motion is a acquired skill in manual manipulation that is featured in the proposed robotic system. The design includes two servomotors (Dynamixel XH430 series, ROBOTIS, CA, US) under velocity control law based on a PID controller. Catheter translation motion is achieved by a fractional drive wheel and a secondary spring-loaded idler roller coupled opposite-side of the drive wheel to guarantee sufficient frictional force. The second motor rotates the whole translation unit on a housing gantry equipped with a slip-ring. It allows unlimited rotation of catheter to facilitate maneuverability.
B. Study protocol and data analysis

One operator performed remote robotic navigation of the guidewire through supra-aortic vessels, including the right subclavian artery (RSA), the right common carotid artery (RCCA), the left common carotid artery (LCCA) and the left subclavian artery (LSA), see Fig. 2 (e). Cannulations were repeated five times for each targeted artery. The position and length of contact interactions are taken from image segmentation along with the CF estimation results from FEM model to visualize CF monitoring on vessel boundaries. Resultant insertion forces on the phantom are recorded with F/T sensor. Quantitative analysis is performed on the parameters of intraluminal CF and resultant forces, including: average of peak force, standard deviation (STD) of peak force, average of mean force, force-time impact (force integral over time).

FEM estimation’s error is analyzed based on GW deformed shape estimation, i.e., plane distance between actual GW and simulated results. If \( (h_i, q_i) \) represent coordinates of each points on the actual shape and \( (H_i, Q_i) \) as the coordinates of points from FEM modeling, then the estimation error \( (E_P) \) is expressed by

\[
E_P = \sqrt{(h_i - H_i)^2 + (q_i - Q_i)^2}
\]

The structural stress of GW is obtained from the FEM model. It results in an intraluminal stress study during navigation.

IV. RESULTS AND DISCUSSION

A. Intraluminal contact forces

Fig. 8 depicts an example of guidewire FEM model and simulated results at a given time during cannulation of RCCA. Subfigure (a) presents FEM model structure: guidewire beam model (meshed in red), deflection vector \( d \) which is applied to boundary condition \( BC \), simulated guidewire in blue contour highlighting reaction forces, direction and magnitudes of four local contact force vectors and the contact positions on the vessel wall. In this example, contact forces are: \( |\vec{f}_1| = 0.17 N, |\vec{f}_2| = 0.26 N, |\vec{f}_3| = 0.09 N, |\vec{f}_4| = 0.03 N \). Contact forces closer to distal end (\( f_3 \) and \( f_4 \)) are significantly lower than \( f_1 \) and \( f_2 \) which is because of lower GW flexural rigidity in that region. Fig. 8(b) compares modeling results represented as green dots to the actual shape from image segmentation in blue line. A visual comparison suggest model accuracy to mimic GW behavior.

Fig. 9 displays the graph of tracking local CFs over navigation of RCCA. At the beginning of the procedure, GW has only one contact, and the number of contacts is increasing to four as GW is advancing. Maximal range of \( CF_3 \) and \( CF_4 \) are 0.04 N - 0.11 N whereas the range of \( CF_1 \) and \( CF_2 \) are about 0.205 N - 0.26 N. The ascending trend is seen for all CF values, but \( CF_1 \) and \( CF_2 \) forces are increasing to greater amounts. This is because of the fact that \( CF_1 \) and \( CF_2 \) positions are moving away from the tip over insertion which gradually shift contact to the stiffer regions, whereas \( CF_3 \) and \( CF_4 \) are just happening at the softer regions close to the tip. Magnitude of CFs are fluctuating because of the GW slip-stick motion on the phantom wall. Maximal intraluminal CF, as the main safety performance metric in interventional procedures, is highlighted on a gray shadow in Fig. 9. Maximal CF was on \( CF_1 \) and turns to \( CF_2 \) after guidewire gets in contact with more points and is being more deflected at the middle region (see Fig. 9 where \( CF_3 \) starts). It was observed that every procedure has a unique CF monitoring trend; however, maximal contact forces were always happening on contact points far from tip rather than near. Map of CF on vessel interaction boundaries over the time of procedure is presented in Fig. 10 (results are for the same trial presented in Fig. 2). CFs magnitude are color-coded on their coordinate of interaction. The guidewire continuously interacts with the inner lumen arterial wall in which some parts of the wall got in contact several times. Maximum CF is visualized in case of having several interactions on a part of the lumen. The path of interactions starts from both side of the descending aorta, gets to the aortic arch bifurcations and moves into RCCA where it mainly interacts with the right side of RCCA. Maximal contact forces are localized at the bifurcation edge of RCCA in the aortic arch, about 0.26 N, and downward in the right side of the descending aorta, about 0.2 N. It is because of the large deflection induced to GW at those regions and also the guidewire’s high rigidity modulus.

Fig. 11 depicts CF contour map of RSA, LCCA, and LSA
Fig. 10: Guidewire-vessel interactions and intraluminal contact forces map computed during navigation of RCCA.

In these examples maximum CFs are 0.31N, 0.08N and 0.11N for RSA, LCCA and LSA, respectively. Observing the CF contours suggest that maximal CF in the all procedures mainly happens at the bifurcation edges, where the guidewire path changes, and it is being forced to bend by the edge wall while advancing. In other words, the bifurcation edge wall acts as a support for the GW. Additionally, guidewire sticking to the phantom wall leads to more friction forces and consequently, more push force and CF. RSA contours looks similar to RCCA in terms of magnitudes and maximal CF locations on the vessels. This is because of common navigation procedure up to RCCA-RSA bifurcation, which consequently results in similar GW motion and interactions map.

Table I shows intraluminal contact force metrics in all cannulation experiments. Cannulation of RCCA and RSA are associated with higher contact forces compared with LCCA and LSA, since these procedures are more dexterous and engaging a longer portion of GW with larger deflections and larger friction forces. The highest CF peak up to 0.37 N was observed in RSA cannulation. Average of peak CFs were 0.24 N, 0.305 N, 0.092 N and 0.114 N at RCCA, RSA, LCCA and LSA respectively and average of mean CF were 0.14 N, 0.172 N, 0.041 N and 0.046 N. The standard deviation of peak CF was from 0.021 N to 0.053 showing a big variation over trials. It infers that not all procedures have small CF, but some may cause damage to arterial cells. Force time integration or force impact is a performance metric for quality of the procedure; a lower value indicates faster procedures with lower CF, i.e., the best desired scenario. Force impact was from 1.17 N.s at LSA up to 9.85 N.s at RSA. The higher value of CF and longer time of cannulation to reach the targeted point in RSA and RCCA resulted in significantly greater F-T integration.

C. Resultant exerted forces

Some studies proposed the measurement of total catheter/GW insertion forces or total contact forces applied on a phantom for either contact force evaluation or insertion force control [39], [26]. Total catheter insertion force does not represent the local CF applied on the inner arterial wall, while it is likely to be the resultant of intraluminal CFs plus unknown friction forces from introducer sheet, access point, etc. Here, we are comparing maximal intraluminal CF with resultant exerted force (RF) on the phantom recorded by the F/T sensor platform. Fig. 12 shows samples of force comparisons within arteries. There is no consistent trend; maximal intraluminal CF is higher or lower than RF; however, in the majority of times, intraluminal CF is larger. RF may fall through retractions while CF remains large. It also suggests that RF is not a promising indication of complications since it is not as high as the CF, and their peaks do not match. Table I includes intraluminal CF and RF metrics. Max RF values may be higher compared to max CF, whereas mean RF is always showing lower values. This means that low catheter/GW insertion force could increase deflection gradually, which may lead to a high intraluminal CF with a risk of arterial wall injury. Monitoring and controlling RF, or insertion force, might smoothen the procedure, but it does not necessarily prevent excessive CF and subsequent complications.
D. Stress

In the proposed method, we can extract stress and strain from the GW/catheter FEM model at any instant of time. Fig 13 presents samples of GW stress contour in arteries. Using intraluminal information, GW structure can be optimized to maximize navigation performance while minimizing CF and stress.

E. Discussion

The observed errors in estimation could be due to measurement error in GW bending modulus over its length. The sharp change in section may not be appropriately seen in sequential experiments. However, manufacturers have bending information based on their design, which can be used if such information is provided. Another error source is that the GW model is initially considered as a straight cantilever, whereas small deflection of the structure is observed for high initial lengths. The accuracy of force monitoring varies depending on the correctness of image segmentation as well, but is not sensitive to it. Camera calibration could cause such errors in this study, which may not be an issue in commercially available clinical imaging systems. Proposed force estimation does not require information on arterial wall material properties and tissue deflections, which is considered an advantage; however, having that information would help to estimate local pressure profile on contact. It is almost certain that a distributed contact may happen but it results in wider contact area, which has lower pressure and risk. In this study, the amount of forces, deflection, stress, and other parameters are reported for navigation of a GW in a training phantom, therefore, they are not necessarily the same magnitude in clinical practice, and such conclusion should not be made.

V. Conclusion

Catheter-vasculature interaction was reported to be the leading cause of complications and major concern, such as embolization, stroke, ischaemic lesion, and perforation, in endovascular interventions. In this study, the case of catheter contact point forces is of interest, as it is the extreme condition with a high risk of over-pressure on the vasculature. We implemented an image-based intraluminal catheter-vessel interaction monitoring tool based on imaging data and numerical computation. The image segmentation algorithms successfully detect and track contacts, vasculature boundaries, catheter and compute pose measurements needed. FEM model effectively simulated manipulation and predicted contact force...
and structural stress. Remote cannulation of the aortics was performed using a robotic unit and intraluminal contact force monitoring was achieved by tracking local CFs and building a contour map of CF on the arterial wall. Results suggest that RCCA and RSA associated with higher CF where maximal CF happens at the bifurcation edge of the aortic arch. Estimation error was low, showing the fidelity of the model. Contact forces can be visualized intraoperatively for clinicians to prevent injuries and reduce the learning curve for novices. CF could be used by the robotic system as a feedback for restrained force control. Resultant forces exerted on phantom was directly measured compared with local CF trend, where small RF and catheter insertion forces have seen while CF is large on the vessel. Model predicted structural stress of GW besides CF data, which helps to optimize the design of interventional catheter/guidewires. Nearly all cardiovascular procedures are under 2D real-time imaging. Physicians keep imaging plan normal to catheter motion, which we have simulated experimentally in this study. Extending the proposed methodology to 3D is an intuitive step that requires an upgrade simulated experimentally in this study. Extending the proposed imaging plan normal to catheter motion, which we have

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