Design and Analysis of a Prototype Haptic Device for Cardiovascular Interventions

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Abstract—In this paper, we present the details of the mechanical design for a haptic device intended to be used during minimally invasive cardiac surgeries. In addition, we develop the kinematics and dynamics equation of motion of the end effector of the device using the Euler-Lagrange equations. The viability of the design is demonstrated by integrating a virtual representation of the left ventricle to define unilateral constraints and finding a safe corridor for haptic guidance and control of the surgical robot.

I. MECHANICAL DESIGN OF THE HAPTIC DEVICE

The objective of the present work is to design a device to assist the surgeon by feeding back the forces acting on the end effector of the catheter or a medical tool during the operation while maintaining the necessary translational and rotational movements. In a conventional operation, the surgeons apply translational force for translational movement of the surgical tool with their wrist and/or arm movement. They may also apply rotational movement with their fingertips, which is necessary to rotate the tool tip inside the vessel, and thus perform a good navigation through the junctions and curves of the vessel or the heart. The surgeons may also apply both translational and rotational movement at the same time. In order to maintain an effective force feedback, the designed haptic device should provide all the rotational and translational forces to the surgeon’s hands and/or fingertips; yet, it must be a modular system, so that in case of a complication, which requires an open surgery, the surgeon can move freely. Considering these constraints, we propose a design method to avoid using any device, electronics or sensors attached to the surgeon’s hands or body. The design maintains all the necessary force feedback and movements using a separate device through which the surgical tool passes through.

The device is designed to operate in two modes: direct operation and teleoperation. The operational mode is decided based upon the reliability from the surgeon’s viewpoint and the health provider’s comfort. In the direct operation mode, the guide wire (or medical tool) passes through this single device, which is located near the patient and the surgeon operates the device as if she/he conducts a conventional operation. Namely, the surgeon operates the device by pulling, pushing or rotating the medical tool which passes through the device while it provides the necessary force feedback to the operator. In the teleoperated mode, two devices are used. The first one is stationed near the patient and the second one may be located at a distant teleoperation location that a surgeon has access. The tool passes through the first device to perform the operation. The second one is actuated by the surgeon and is used to feed back the forces acting on the first medical tool’s end effector. Fig. 1 illustrates the components of the designed device.

The working principle of the device is described next. As shown in Fig. 1 and Fig. 2, the device consists of three DC motors. DC motors-1 and 2 are in contact with the medical tool and used to actuate it in translational direction, where DC motor-0 is used to rotate the inner disc mechanism along with the medical tool. The dc motors-1 and 2 are placed symmetrically inside the inner disc mechanism to prevent any unbalanced mass during the rotational motion of the medical tool. At the tip of the translational motors, a soft material with high friction coefficient is used to prevent slipping of the medical tool while giving it force feedback.

Using the measurements from the end effector position of the tool, the motors can hold the tool or simulate a virtual environment as if the tool was moving in a highly viscous environment. This is necessary in order to prevent the operator from moving the end effector to an unsafe operating region, which can be identified in real-time using MR images. The viscous effects are zero in the safe zone, and gradually increase corresponding to an increase in the penalty coefficient which depends on the divergence of the
end effector from its pre-defined path or workspace.

II. KINEMATICS AND DYNAMICS OF THE HAPTIC DEVICE

A. Kinematics of the Haptic Device

The end effector of the catheter (or medical tool) is assumed to be eccentric. This type of tip structure allows the operator to navigate through the junction of the veins by simply rotating the guidewire/medical tool around its longitudinal axis. For a fixed eccentric shape, simultaneous rotation and translation of the tool defines a workspace that lies on the surface of a cylinder. Fig. 3 shows the structure of the end effector in a cartesian space.

\[
x = r_e \sin \theta_0 = r_e \sin \left(\frac{r_0}{r_{01}} \theta_0\right) \tag{1a}
\]
\[
y = r_e \cos \theta_0 = r_e \cos \left(\frac{r_0}{r_{01}} \theta_0\right) \tag{1b}
\]
\[
z = \frac{r_{1,2} \pi}{180} \theta_{1,2} \tag{1c}
\]

where \(\theta_0\) and \(\theta_{01}\) are the rotations of the dc motor-0 shaft and the inner disc shaft, respectively, \(r_0\) and \(r_{01}\) are the radii of the contact disc attached to the dc motor-0 shaft and the inner disc mechanism contact shaft, respectively. \(r_e\) is the eccentric distance of the end effector, \(r_{1,2}\) and \(\theta_{1,2}\) are the contact discs’ radii which are attached to the dc motor-1 and dc motor-2 shafts, and the rotation of the dc motors-1 and 2 shafts, respectively.

The inverse kinematics of the haptic device (shown in (2a) and (2b)) will be used to determine the desired dc motor shaft angles for a given reference trajectory in cartesian space. To design and analyze feedback control algorithms, we need to derive the dynamic equations of motion of the system.

\[
\theta_0 = \frac{r_{01}}{r_0} \arcsin \left(\frac{x}{r_e}\right) = \arccos \left(\frac{y}{r_e}\right) \tag{2a}
\]
\[
\theta_{1,2} = \frac{180 \theta_{1,2}}{r_{1,2} \pi} \tag{2b}
\]

B. Dynamics of the Haptic Device

The Euler-Lagrange equations are employed to derive the dynamics of the system, which describes the behavior of the mechanical system subject to holonomic constrains. In order to derive the Euler-Lagrange equations of motion, we must first form the Lagrangian of the system, which is the difference between the total kinetic energy and the total potential energy [1].

The Lagrangian is defined as

\[
\mathcal{L} = \mathcal{K} - \mathcal{P} \tag{3}
\]

where \(\mathcal{K}\) is the total kinetic energy and \(\mathcal{P}\) is the total potential energy of the system. After deriving the Lagrangian, the equations of motion can be determined using the Euler-Lagrange equation as

\[
\frac{d}{dt} \frac{\partial \mathcal{L}}{\partial \dot{q}_i} - \frac{\partial \mathcal{L}}{\partial q_i} = \tau_i \quad i = 1, \ldots, n \tag{4}
\]

where \(q_i (i = 1, \ldots, n)\) are the joint variables and \(\tau_i\) is the related torque term.

The total kinetic energy of the haptic robot is defined as the summation of the rotational \((\mathcal{K}_{rot})\) and translational \((\mathcal{K}_{trans})\) kinetic energies as

\[
\mathcal{K} = \mathcal{K}_{rot} + \mathcal{K}_{trans} \tag{5}
\]

where the rotational kinetic energy is defined by the rotation of the inner disc mechanism including the attached dc motors and the rod (simulating the medical tool), and the translational kinetic energy is defined by the translation of the rod as follows,

\[
\mathcal{K}_{rot} = \frac{1}{2} J_{m0} \dot{\theta}_0^2 + \frac{1}{2} J_{id} \dot{\theta}_{01}^2 + \frac{1}{2} J_i \dot{\theta}_{1,2}^2 + 2 \left(\frac{1}{2} J_{m1,2} \dot{\theta}_{1,2}^2\right) \tag{6a}
\]

\[
\mathcal{K}_{trans} = \frac{1}{2} m v_t^2 \tag{6b}
\]

where \(J_{m0}\) and \(J_{m1,2}\) are the inertias of the DC motor-0 and DC motor-1 and 2 shafts respectively (with contact cylinders attached to the shafts), \(J_{id}\) is the inertia of the inner disc (with dc motors attached) and \(J_i\) is the inertia of the medical tool (with a mass attached to the eccentric end-effector). \(m\) represents the overall mass of the medical tool and \(v_t\) represents the linear velocity of the center of mass of
the medical tool. Hence, using (6) the total kinetic energy (5) can be expressed as

\[ K = \frac{1}{2} \left( \frac{r_0^2}{r_0^2} J_m + J_{id} + J_t \right) \dot{\theta}_{01}^2 + J_{m1,2} \dot{\theta}_{1,2}^2 + \frac{1}{2} m r_{1,2}^2 \dot{\theta}_{1,2}^2 \] (7)

where \( v_t = r_{1,2} \dot{\theta}_{1,2} \).

The potential energy term is found from the rotation of the rod with a mass attached to the eccentric tip as

\[ P = m g h_t \cos \theta_{01} \] (8)

where \( m \) represents the overall mass of the medical tool at the center of mass and \( h_t \) represents the distance from the inner disc rotation axis to the medical tool center of mass.

The Lagrangian (3) for the haptic device is formed as

\[ L = \frac{1}{2} \left( \frac{r_0^2}{r_0^2} J_m + J_{id} + J_t \right) \dot{\theta}_{01}^2 + J_{m1,2} \dot{\theta}_{1,2}^2 + \frac{1}{2} m r_{1,2}^2 \dot{\theta}_{1,2}^2 - m g h_t \cos \theta_{01} \]

Using (4) and defining the joint variables as \( q_1 = \theta_{01} \) and \( q_2 = \theta_{1,2} \) we derive the torque equation for inner disc shaft and dc motors-1 and 2 shafts as follows.

\[ \tau_{01} = \left( \frac{r_0^2}{r_0^2} J_m + J_{id} + J_t \right) \ddot{\theta}_{01} - m g h_t \sin \theta_{01} \] (9a)

\[ \tau_{1,2} = (2 J_{m1,2} + m r_{1,2}^2) \ddot{\theta}_{1,2} \] (9b)

Reflecting the related damping terms to the total dc motor-0 torque yields

\[ \tau_{01} = \tau_{m0} - (B_0 \dot{\theta}_{01} + B_{id} \dot{\theta}_{01}) = \tau_{m0} - (B_i \frac{r_{m1,2}}{r_0} + B_{id}) \dot{\theta}_{01} \] (10)

where \( \tau_{m0} \) is the dc motor-0 torque, \( B_0 \) is the dc motor-0 damping, and \( B_{id} \) is the damping associated with the inner disc.

Reflecting the related damping terms to the combined total dc motors-1 and 2 torques yields

\[ \tau_{1,2} = 2 \tau_{m1} - (2 B_{1,2} + B_i r_{1,2}) \dot{\theta}_{1,2} \] (11)

where \( \tau_{m1} \) is the dc motor-1 and 2 torques, \( B_{1,2} \) is the damping term for dc motors 1 and 2 and \( B_i \) is the damping term of the rod (the medical tool) translational motion.

Putting all together, using (9) we derive the dynamic equations of motions for inner disc rotation and medical tool translation as:

\[ \begin{align*}
\tau_{m0} &= \left( \frac{r_0^2}{r_0^2} J_m + J_{id} + J_t \right) \ddot{\theta}_{01} + \left( B_0 \frac{r_{m1,2}}{r_0} + B_{id} \right) \dot{\theta}_{01} - m g h_t \sin \theta_{01} \\
2 \tau_{m1} &= (2 J_{m1,2} + m r_{1,2}^2) \ddot{\theta}_{1,2} + (2 B_{1,2} + B_i r_{1,2}) \dot{\theta}_{1,2}.
\end{align*} \]

### III. Prototyping and Experimental Implementation

#### A. Prototyping

Having the components of the device modeled in Computer Aided Design (CAD) software with exact dimensions, the device is designed to be compact to assure the maximum user comfort and mobility. Thus, the overall dimensions of the device are 17 cm in length, 12 cm in width and 12.8 cm in overall height (including the inner disc mechanism). Fig. 4 shows the assembly of the prototype and Fig. 5 shows the components of the prototype in detail. Notice that the device dynamically changing boundaries of the safe operating region.

Fig. 4. Assembled prototype of the haptic device for minimally invasive cardiac intervention.

Fig. 5. Prototype components of the haptic device.

#### B. Hardware and Software Implementation

The device is printed using ABS plastic material. For controlling the device, a dspace® 1103 real-time data acquisition system is used, and the control algorithm is developed using Matlab/Simulink®. DC motor drivers, which are commanded from the real-time data acquisition system, are used to supply the necessary currents to the motors shown in Fig. 6.

The virtual representation of the heart is used as a virtual environment that provides contact force feedback. The dynamically changing boundaries of the safe operating region
is fed to the control algorithm as constraints. Inside these boundaries, the operator is able to position the end effector without feeling any force feedback from the system. In fact, the friction coefficients of the haptic system are compensated by introducing a virtual environment with negative friction. Using this frictionless environment within safe boundaries allows the operator to control the haptic device with minimum effort. When the operator moves the end effector to the safe boundaries less than a margin $\epsilon$, the control algorithm applies a force which is opposite to the force that the operator is already applying.

The virtual representation of the left ventricle of the heart is defined as the model of the left ventricle of the heart. This model and the safe boundary extracting algorithm is developed in [6]. The model is programmed in C\# programming language, and the safe boundary data from the C\# code is integrated by the dSpace real-time data acquisition and control system. This way, the necessary measurements (i.e., the dynamically changing width and depth of the safe boundary) taken from the virtual environment is fed to the control algorithm in real time. While actuating the prototype, the operator follows the end effector position of the rod (medical tool) inside the dynamically changing left ventricle in real time. We integrated the C\# code with dSpace by calling the Matlab engine from the C\# code. We also developed a Matlab function to read from and write to the dSpace using the MLIB library that the dSpace company provided.

C. Virtual Representation of the Left Ventricle

The virtual representation of the left ventricle is taken from Yeniarias et al. [6]. The authors achieved a realistic modeling by using MRI images of the heart taken from healthy volunteers. They used a standard CINE pulse sequence that collected 24 frames of eight short axes and three long axes planes [6]. An access corridor was generated based on the criterion that it should not contact the endocardium (see Fig. 8), which is the innermost layer of tissue that lines the chambers of the heart, for all slices. The implemented algorithm determined a “base” surface ($S_{\text{base}}$) common to all slices and frames that satisfied the above criterion.

1) The left ventricle was segmented and the endocardial boundary was extracted in all frames of all the short and long axes slices with manual tracing and Insight Segmentation and Registration Toolkit (ITK) functions. ITK is an open-source, cross-platform system that provides developers with an extensive suite of software tools for image analysis.

2) For each time frame ($I = 1$ to 7 slices), the common volumetric access corridor was determined by superimposing the access area for all seven slices.
3) The process in step 2 was then repeated for all time frames (i.e. for $J = 1$ to 24), to find the common access areas for all times sampled with the CINE sequence. The result from steps 2 and 3 was the surface $S_{\text{base}}$. In practice, the algorithm in steps 2 and 3 is equivalent to superimposing the segmented left ventricles for all short axis slices and for all time frames.

4) The 3D access corridor was then generated by extending the $S_{\text{base}}$ from the apical to the basal short axis slices.

As a result, the workspace of the robot (shown in Fig. 8 in shade) is determined using the four-step procedure described above, as shown in Fig. 9. The figure shows the segmented left ventricle contraction at three different time instants. The shaded vertical region shows the safe operating region for the haptic system end effector. We used the dynamically changing safe boundaries as the input to the control algorithm developed for providing the haptic effects to the operator.

REFERENCES