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An Ergonomic Master Haptic Interface for the Robot-assisted Endovascular Catheterization System

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Abstract- **For most existing robot-assisted endovascular catheterization systems, the master interface takes the shape of a joystick or a haptic device, therefore potentially altering the natural behavior and motion patterns of physicians. Furthermore, existing studies on effective haptic feedback with robot-assisted endovascular catheterization system have been very limited. To address these issues, the paper proposes an MR fluids-based master haptic interface. The input catheter was utilized as the joystick, hence the physician can apply conventional motions (advance, retreat and rotate) on this interface. Meanwhile, the passive resistance force will be generated during the operating. The VCSEL (vertical-cavity surface-emitting laser) was used to measure the linear and rotational displacement of input catheter. The radial motion of the input catheter is detected by the hollow encoder. It was found that the mean accuracy of axial motion is 0.04 mm (precision, ±0.05 mm) for 100 mm length of advancing and retracting. And the considerable range of generated passive resistance force is from 28 mN to 1206 mN.**

Index Terms -MR fluids; Haptic Feedback; Catheter Interventional Surgery

I. INTRODUCTION

In recent two decades, the development of robot-assisted endovascular catheterization system was motivated by the desire to reduce fluoroscopy time, radiation dosage to surgeon and patient in addition to the reduction of surgeon fatigue, and improvement of position accuracy of the catheter. Unlike the conventional bedside technique, the robot-assisted endovascular catheterization system allows the operator to offer axial and radial motion by master robot placed in a remote location through the control console to guide the slave robot to advance, retract and rotate the catheter. Some commercial catheter navigation systems, all employed the master-slave control architecture, have demonstrated safety and efficacy in vascular and endovascular surgery, such as CorPath 200 and CorPath GRX vascular robotic systems (Corindus Vascular Robotics) [1]. Clinical studies implied that compared to conventional endovascular surgery, the robotic endovascular technology is effective in reducing procedure time, improving stability and precision, a shorter learning curve and decreasing the dosage of radiation exposure for both the patient and physician. Some research groups have developed remote catheter navigation systems are as follows. Thakur et al. developed a novel remote catheter navigation system by using an input catheter placed in a radiation-safe location to control a second patient catheter [2]. A compact telerobotics catheter navigation system was presented that had the accuracy of 0.1±0.1mm and 7±6 deg over 100mm of axial motion and 360 deg of radial motion [3]. A 4-degree-of-freedom master-slave catheter and guidewire driving system was developed by Cha et al. [4]. The system can enable the catheter and guidewire to be controlled independently in coaxial direction. In our previous studies, some novel remote-controlled vascular interventional robots were developed [5-12], which could operate the catheter in 2 DOFs and measure the proximal forces during the catheter insertion. Also, the VR-based haptic catheterization training system was presented [13], for new surgeons. The Fig. 1 shows the typical schematic diagram of the master-slave catheterization system.

The development of robotic platforms in recent years has aimed to reduce radiation exposure, increase precision and stability of motion, and add operator comfort. Currently, physicians overwhelmingly rely on 2-D visual feedback, as one of their dominant information sources, during robotic endovascular surgery [14]. However, lack of the sensation of touch or haptic feedback from catheter-tissues contact to the operator is a drawback in current robot-assisted endovascular catheterization systems. The medical professionals strongly rely on the sense of touch and their intuitive skills during endovascular surgery. However, the employment of the robot-assisted endovascular catheterization system removes the catheter from the physician's hands, also eliminates the direct contact between the clinician and patient. For most existing robotic solutions, the master interface takes the shape of a joystick or a haptic device, therefore potentially altering the natural behavior and motion patterns of experienced operators. Furthermore, existing studies on effective haptic feedback with robot-assisted endovascular catheterization system have been very limited. The physician can hardly evaluate the amount of force applied to the tissues that may increase the incidence of excessive forces and may rupture an aneurysm or the blood vessel, leading to fatalities. The ideal teleoperated surgery scenario is viewed as a physical extension of the human body. A high level of transparency is essential to physicians to make a right decision in robotic endovascular surgery. To this end, recreating effective haptic in master side becomes urgently in robot-assisted endovascular catheterization system.

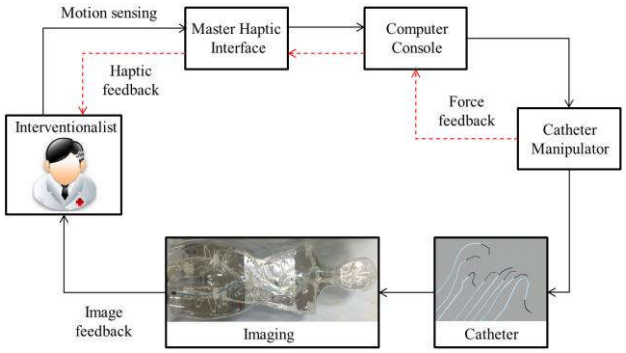

Fig. 1 Schematic diagram of the master-slave catheterization system

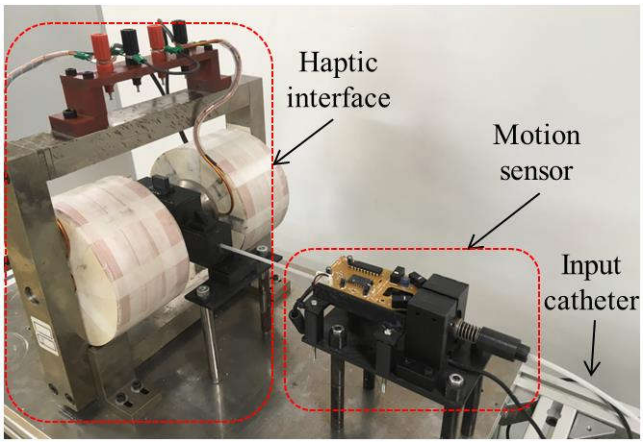

Fig. 2 The master haptic interface

To address these issues, the proposed master haptic interface which can provide the operator with the ability to use their conventional bedside skills in the robot-assisted endovascular catheterization procedure. Meanwhile, the haptic feedback can be provided to the operator during the operation.

The remainder of this paper is organized as follows. The MR fluids-based master haptic interface is presented in section II. In section III, the experiment is presented to evaluate the proposed haptic interface. The experimental results are presented in section IV. Section V concludes this paper.

II. MR Fluids-based Master Haptic Interface

The uniqueness of the proposed master haptic interface is that the operators can apply the natural motions (push, pull or twist) on the input catheter and perceive the haptic feedback through the input catheter during the manipulation. The structure of the master haptic interface, shown in Fig.2, was designed to consist of an input catheter motion sensing part (measuring the axial and radial motions of an input catheter), and an MR fluids-based haptic interface (generating haptic feedback).

A. Motion Sensor Design

The prototype of axial motion sensor, previously described in [15], its schematic diagram is shown in Fig. 3, is a laser-based navigation sensor that measures the axial motion of the input catheter. The VCSEL (vertical-cavity surface-emitting laser) emits the laser light to the surface of the catheter, and then the optical sensor will receive the image of the detection

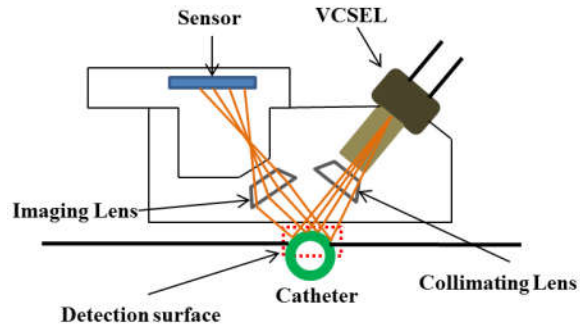

Fig. 3 Working principle of the axial motion sensor

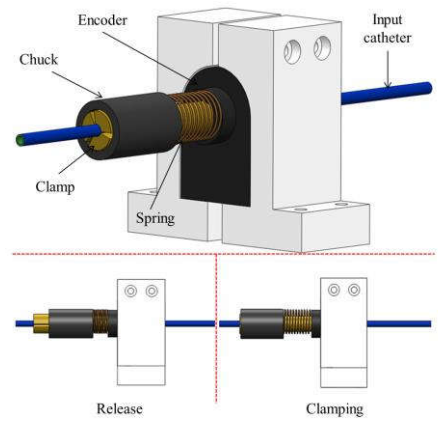

Fig. 4 Working principle of the radial motion sensor

surface. Hence, when the input catheter is pulled or pushed, the changes of two adjacent detection surface images can be detected. The coherent nature of the collimated laser light, along with the specular optics, allowed the sensor to track on more surfaces compared to the LED-based mouse sensor. This non-contact detection technique of measuring the linear displacement of the catheter was not only provided the high accuracy (1mm in axial motion over 100mm) and sensitivity (3000 Hz) but also increased the authenticity of the operation.

The radial motion of the input catheter is detected by the hollow encoder (UN-2000, MUTOH, Japan). The input catheter goes through the clamp, which is mounted on the hollow shaft of the encoder. A cylinder chuck is coaxial with the hollow shaft and placed over the clamp. The spring, over the clamp, is located between the chuck and the encoder in the axial direction. The generated grip force of designed clamp is big enough to prevent the sliding of input catheter. The rotational signal of input catheter will be directly measured by the encoder when it is gripped. The operators can apply pull motion to the chuck, and pressure the spring to release the input catheter. The schematic diagram is shown in Fig. 4.

B. MR fluids-based Haptic Interface

The MR fluids is the suspension of micrometer-sized ferromagnetic particles in a carrier fluid. When exposed to an applied magnetic field, the particles form a kind of chain-like structures aligned parallel to the direction of the applied magnetic field. Consequently, the aligned parallel structure can act to resist shearing or flow of the fluid [16]. These chains have a significant effect to change the apparent viscosity of the

fluid. Its rheological behavior can be changed reversibly within in several milliseconds in response to the external magnetic field. These features have motived the design of haptic interfaces and rehabilitation devices based on MR fluids in medical applications. MRFs-based actuators were used to develop the 2 DOFs haptic interface for medical applications, the devices showed great potential for stability and transparency on master-slave teleoperation [17]. MR fluids actuated miniature tunable stiffness haptic interface has been designed and fabricated [18]. The Bingham plasticity model is effective in describing the essential field dependent fluid characteristic. In our applications, we utilized Bingham plastic model to describe MR fluids field-dependent behavior. In this model, the total shear stress is given by

$$\tau = \tau_0 \operatorname{sgn}(\gamma) + \eta\gamma \tag{1}$$

where τ_0 is the yield stress caused by the applied magnetic field. The second term in this equation is the shear stress related to the motion, η is the viscosity independent of the applied field, γ is the shear rate. The first term of equation describes the dynamic yield stress acting on the contact surface covered by the MR fluids. The second term is uncontrollable shear stress, which is the source of friction force when the device is in off-state. The shear rate depends on the relative speed of the two moving surfaces and the thickness of the MR fluid gap between the surfaces.

There are three different operating modes of MR fluids, shear mode, valve mode and squeeze mode, as shown in Fig. 5. In the shear mode, the MR fluids is constrained between two parallel surfaces which are in relative motion with the magnetic field that is perpendicular to the surface direction. The force is applied to one of the surfaces, making it move laterally relative to the other, directly shearing the fluid layer. Such mode is mainly used in the rotary application (brakes and clutches). This mode is used in the haptic interface described in this paper. The valve mode, also called flow mode, exploits the fluids between two fixed walls, the magnetic field is applied perpendicular to the flow direction. The resistance to the fluid flow can be controlled by varying the intensity of the applied magnetic field. This mode is typical for damper and shock absorber applications. Squeeze mode is used mainly for bearing and impact damper applications; the fluids is constrained between two parallel surfaces while an external magnetic field is applied perpendicular to the surfaces. The force is applied perpendicularly to one of the surfaces, making it move towards the other, squeezing the fluid layer. For a given applied force, the displacement of the moving surface is controlled by the intensity of the applied magnetic field. The small displacement is achieved with high resistance force.

In our previous study [15], the range of controllable force is around 125 mN to 500 mN. This force is generated when the master side received the feedback warning signal from slave side. As because of the small force range, only the interactions between the virtual catheter tip and the vascular system can be reflected. In real endovascular surgery, effective haptic feedback that allows the physician to feel and interact during catheter manipulation is important for procedure success. In order to provide the effective haptic feedback to physicians, the bigger controllable force must be generated by the new designed haptic interface.

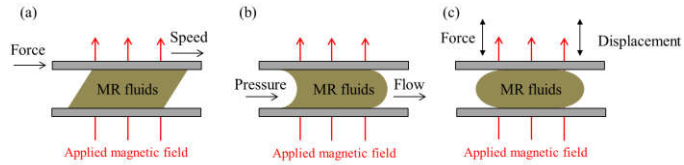

Fig. 5 Three different operating modes of MR fluids, (a) shear mode, (b) valve mode, (c) squeeze mode

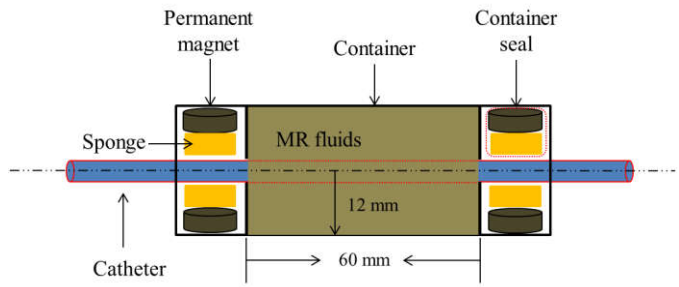

Fig. 6 Schematic diagram of the MR fluids container

There are two main parts, MR fluids container and magnetic field generator, consist of the haptic interface. The structure of this haptic interface is as follows. The MR fluids container is located in the center space of two magnetic poles, which is made of low permeability materials and can never be magnetized during the external magnetic field, shows in Fig.6. The input catheter plays like a piston rode, goes through the MR fluids container, which is overwhelmed by the MR fluids. In order to prevent MR fluids leakage, special seal method is utilized. Two permanent magnets (their magnetic poles attract each other) are fixed inside one container seal, which placed both above and below the input catheter. The sponge makes up the gap between the two permanent magnets. Two couples of core and coil are equipped separately on the left and right side of the MR fluids container. The detailed design of magnetic field generator was analyzed in our previous works [6], [15].

Two hall-effect sensors (TLE 4990, Infineon Technologies, Germany) were placed in MR fluids container to measure the magnetic flux density in real time. The placement of hall sensors strongly influences the accuracy of its readings. Hence, the sensor should be perpendicular to the magnetic field lines, and little variation of magnetic flux in both sides of the sensor. The Gauss meter (TM701; KANETEC, Japan) was utilized to measure the distribution of magnetic field flux density along the direction of two cores in different applied currents, and the result is shown in Fig. 7. For the above reasons, two hall sensors are embedded at the center positions of container's inner walls, which the magnetic lines are evenly distributed and across two sensors vertically. Meanwhile, it has the great effect on the shear stress of MR fluids and directly affects the resistance force of catheter insertion. The commercial MR fluids (MRF-122EG, Lord Crop, USA) was used in this design.

The MR fluids exhibits a linear magnetic behavior and its hysteresis can be neglected that due to the soft irons used in the fluids suspension. The relationship between the input current and generated the magnetic field in a certain range is shown in Fig. 8. The applied current is increased from 0 A to 5 A in 0.1 A steps. Enough time is provided to establish the steady-state magnetic field in the certain constant current. As

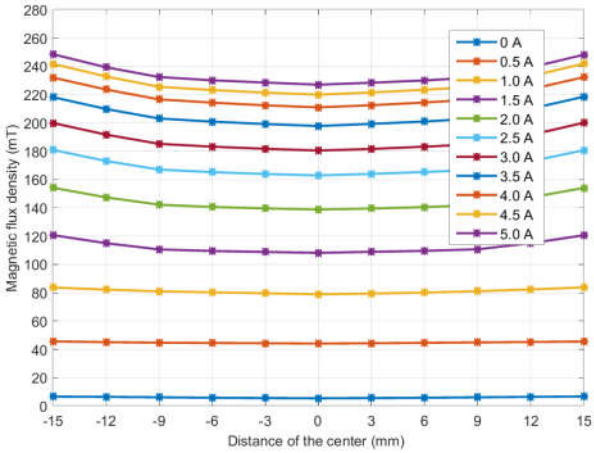

Fig. 7 Magnetic field distribution along the direction of two cores

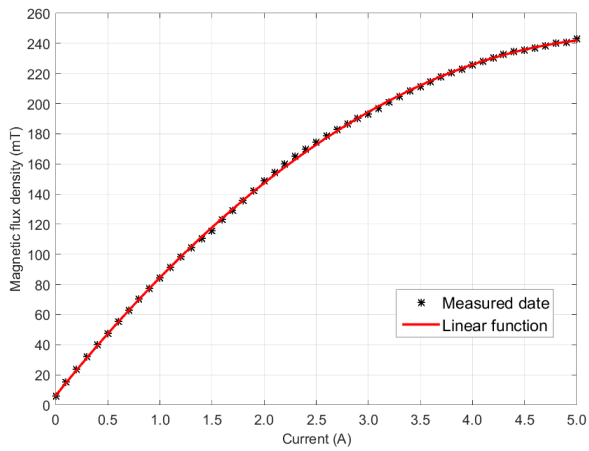

Fig. 8 Relationship between the input current and generated magnetic field

obvious, the one-to-one relationship between the applied current and generated magnetic flux density in steady state can be expressed as follows:

$$B=(-7.80)*I^2+86.11*I+6.26 \quad 0A\leq I\leq 5A \qquad (2)$$

III. Experimental Setup

A. Evaluation of Motion Sensor

The axial accuracy of the motion sensor in measuring axial motion was evaluated by advancing a 7 F input catheter. The catheter was advanced, and then retracted, in the different constant speeds, 2 mm/s, 10 mm/s, 20 mm/s, 100 mm/s and 200 mm/s, which the range of catheter insertion speeds meets the clinic commands, and the motion profile is 100 mm. Each insertion speed was repeated 10 times and the actual length of catheter moved was measured. The input catheter is directly connected to the hollow encoder. Hence, the accuracy of radial position measurements is related to the measurement precision of encoder.

B. Evaluation of Passive Resistance Force Generation

In the real application, the passive resistance force is generated when the master side received the force signal which measured by the force sensor of the catheter manipulator or catheter tip force sensor. In order to find the relationship

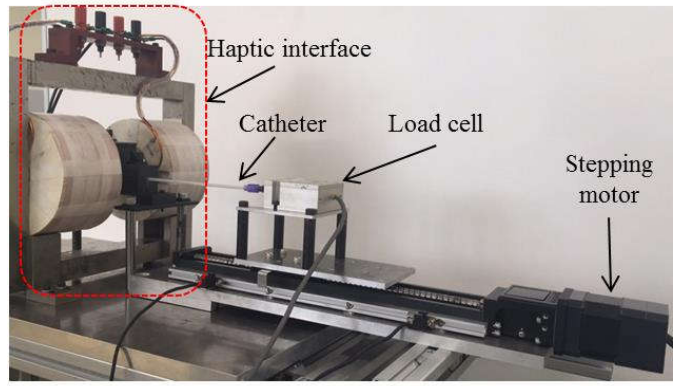

Fig. 9 Experimental setup

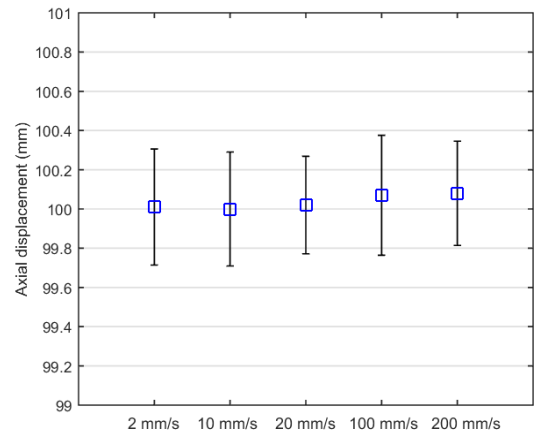

Fig. 10 Evaluation of axial motion for positioning accuracy

between passive resistance force of the input catheter manipulation and the applied magnetic field. We utilized the load cell (TU-UJ, TEAC, Japan) to measure the resistance force of input catheter manipulation in the axial direction. one end of input catheter is connected to the load cell, which navigated by the stepping motor (ASM46AA, Oriental Motor, Japan) in the horizontal direction, experimental setup shows in Fig. 9. The generated magnetic flux density is increased from 0 mT to 240 mT in 20 mT steps. In the procedure of endovascular surgery, the insertion speeds of the patient catheter are based on the complexity and vessel sizes, perceived tactile sensation, and the distances between the catheter tip and lesion location. In order to imitate those different situations in the master haptic interface, two different modes of insertion speeds are separated as low speed and high speed. The low-speed range is from 0 mm/s to 20 mm/s, and the high-speed range is from 20 mm/s to 200 mm/s. The velocity of catheter insertion meets the clinic commands, and the motion profile is 30 mm. During the different speed situations, the increased intervals are 2 mm/s step and 20 mm/s step, respectively.

IV. Experimental Results

A. Evaluation of Motion Sensor

It was found that for cumulative trials the mean slip or error in insertion was 0.04 mm (precision, ±0.05 mm) for 100 mm length of advancing and retracting, the results can be seen in Fig. 10.

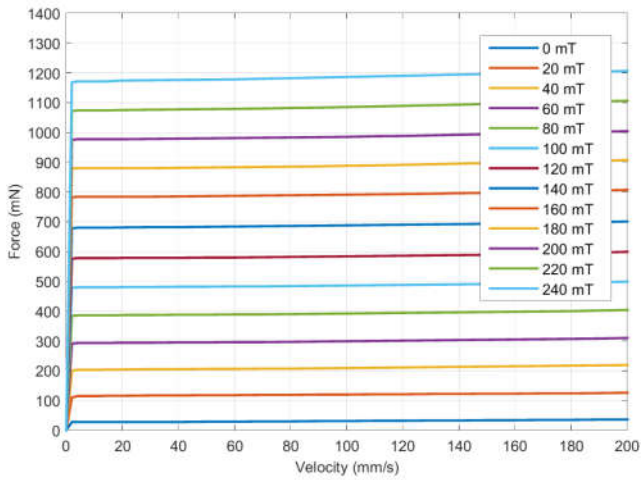

Fig. 11 Measured force-velocity behaviors.

B. Evaluation of Passive Resistance Force Generation

Fig. 11 presents the measurement of output resistance force for applied step increased magnetic field. The considerable range of generated passive resistance force is obtained from 28 mN to 1206 mN. The difference of resistance force between minimum velocity (2 mm/s) and maximum velocity (200 mm/s) in the same magnetic field intensity is from 5 mN to 41 mN. One can note that the measured forces under certain magnetic field are different by different insertion speeds, and such differences are associated with the strength of the applied magnetic field. On the other hand, it should be noted that the measured force values in this test are for a certain step magnetic field generated using a certain value of current. Hence, the results here not include the effects of the hysteresis phenomenon within the haptic interface. And it is mainly associated with ferromagnetic components including iron cores and coils. The little or no hysteresis is observed in the B-H curve of MR fluids. Based on these limitations, it can hardly realize the one-to-one relationship between the perceived haptic forces and input currents of the magnetic field generator.

V. CONCLUSION

In this paper, an MR fluids-based master haptic interface was designed for robot-assisted endovascular catheterization procedure. The interface can provide physicians with the ability to use their conventional bedside catheterization skills during practice. Experimental results showed that the mean accuracy of axial motion is 0.04 mm (precision, ±0.05 mm) for 100 mm length of catheter insertion. And the considerable range of generated passive resistance force is from 28 mN to 1206 mN. However, the MR fluids based haptic interface can only generate the resistance force in longitudinal direction along the input catheter. In the next study, the method to produce the torque information of radial motions in the master haptic interface should be taken into account.

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