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MR Fluid Interface of Endovascular Catheterization Based on Haptic Sensation

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***Abstract*—During the teleoperated robot assisted catheter interventional neurosurgery, to improve the transparency between the patient side (slave side) and the remote control side (mast side) have been obtained plenty of attention. In this research we hypothesized that the slave catheter interventional robot fully complied with the dynamics command which comes from the master site. This paper is concerned with the design and implementation of a master side by Magnetorheological (MR) fluid, which can reflect the dynamic changes of the insertion resistance force, when the catheter goes through the blood vessel to the lesions. Experimental results showed that MR Fluid actuated haptic interface has a benefit to describe the blood viscous force during the endovascular catheterization.**

Keywords—MR fluid; Haptic Sensation; Minimally Invasive Surgery (MIS); Transparency

I. INTRODUCTION

Cardiovascular diseases and cerebrovascular diseases cause the greatest number of deaths worldwide each year according to the World Health Organization. Vascular catheterization is used to diagnose and treat cardiovascular diseases and cerebrovascular diseases. The catheter goes through the femoral artery to the lesions, and this is the most common procedures in vascular minimally interventional surgery. It had been shown some merits, such as more effective, less trauma, quickly rehabilitation and lower risks. However, compared with its advantages with traditional surgery, some limitations of minimally invasive surgery also introduced in. Such as lack of visual feedback, inevitable of hand tremor and longtime exposure of X-ray radiation[1].

In the last few decades, teleoperated robot-assisted catheter navigation systems had been adopted to address the some problems like, preventing the surgeon from being X-ray radiation and providing the comfortable operating environment to the surgeon. Moreover, it will be provided remote assistant to another medical center for telesurgery. As mentioned above, factors that prohibit the development are the lack of force sensing and haptic feedback to the operator. The terms of "haptics" refers to sensing and manipulation through the sense of touch [2]. It includes cutaneous touch and kinesthetic touch. Cutaneous touch sense refers to an awareness of the surfaces features and is usually sensed by the skin. Kinesthetic sensation refers to an awareness of limb positions and dynamics which arise within muscles and tendons. The main goal of a haptic teleoperated robot assisted catheter interventional surgery is transparency, which is defined as the ability to present undistorted dynamics of the remote environment to the human operator. Haptic technology is a promising research area in surgery. For example, some commercial robotic catheter systems, like Sigma 7, HD2and Premium 3.0 hand-controller had been added the haptic feedback [3].Tactile sensation of the remote organ has been provided to the surgeon for teleoperated medical application [4].Pacchierotti integrated haptic sensation (kinesthetic and vibratory information) in the master system for a teleoperated steering flexible needle surgery [5]. The target of these researches is to enhance the performance of teleoperated robot-assisted surgery system by introducing haptic technology.

Therefore, two challenges are existed to recreate the haptic sensation to the local human operator during teleoperated robot catheter surgery. The first one is lacking of effective haptic feedback which is most important. However, how to reflect the blood flow dynamics and delivery the viscous forces about different parts of vessel during catheterization to master side must be took into consider [6].

This study provides a novel catheter haptic master system for endovascular therapy. The MR fluid was exploited to provide haptic sensation (kinesthetic sensation) to the surgery during catheter interventional neurosurgery. It is assumed that the slave robotic catheter interventional system is fully complied with the catheter interventional dynamics commands which come from the master system. The characteristics of MR fluid can be changed reversibly in several milliseconds. The viscosity of MR fluid is proportional to an external electromagnetic power. The most important issue is that MR fluid actuated haptic interface produce a passive force. The surgeon operates the catheter through the MR fluid actuated catheter haptic interface and senses the resistance force (kinesthetic sensation). The surgeon can manipulate a real catheter rather than a manipulator or joystick to obtain a haptic sensation of blood vessel [7]. This dynamic resistance force had been changed with the blood viscous force. The sudden change is also needed when collision between catheter tip and blood vessel wall existed in the surgery practice.

The paper is organized as follows; MR fluid actuated haptic interface is presented in section II. The section III, experimental framework is presented. Section IV provides the experimental results which show the performance of the device. Finally, conclusions are given in Section V.

II. MR FLUID ACTUATED HAPTIC INTERFACE

A. MR FLUID

MR fluid, controllable fluid, are capable of changing its rheological behavior when an external magnetic field is applied[8]. It has three components [9]:magnetizable micro-sized particles, additives, and carrier fluid (such as mineral oil, synthetic oil, water, ethylene glycol or vegetable oil).The magnetizable micro-sized particles can be polarized by the magnetic field, and then be the plastic viscosity fluid. The additives is made up of stabilizers and surfactants. The stabilizers was used to keep the particles suspended in the fluid, particle sedimentation would greatly decrease the behavior of MR fluid. The surfactants are adsorbed on the surface of the particles to enhance the polarization. The carrier fluid is a dispersed medium, the homogeneity particles dissolve in the carrier fluid.

In the absence of an external magnetic field, MR fluid displays Newtonian-like behavior. When the fluid was activated, the particles are held together to form chains parallel, aligning with the external magnetic field. In many cases, this effect is described as Bingham plastic model. An alternative to the Bingham model is Herschel-Bulkley plastic model which accounts for the post-yield shear thinning behavior of MR fluid. A variety of other models have also been used to describe the behavior of MR fluid, such as the biviscous model and Eyring plastic model. As well-known that MR fluid is operated in four different modes: squeeze mode, valve mode, direct shear mode and pinch mode.

Depending on those character, MR fluid actuated haptic interface have been extensively applied in engineering practice effectively. Haptic interface or rehabilitation device based on MR fluid was utilized in medical application and healthcare. The Haptic Black Box I and II (HBBI and HBB II) based on a freehand concept has been developed [8, 10], which the surgeons can immerse their bare hand in MR fluid, and obtain the mutual effect with virtual objects without mechanical constraints to acquire tactile sensation. An encountered-type of haptic interface through using MR fluid has been designed and evaluation [11] for surgical simulation. Ahmadkhanlou, F. et al. [12] developed two degree of freedom MR Fluid based haptic system for tele-robotic surgery. MR fluid actuated miniature tunable stiffness haptic interface has been designed and fabricated [13]. A haptic glove with MR brakes for virtual reality has been designed by [14].

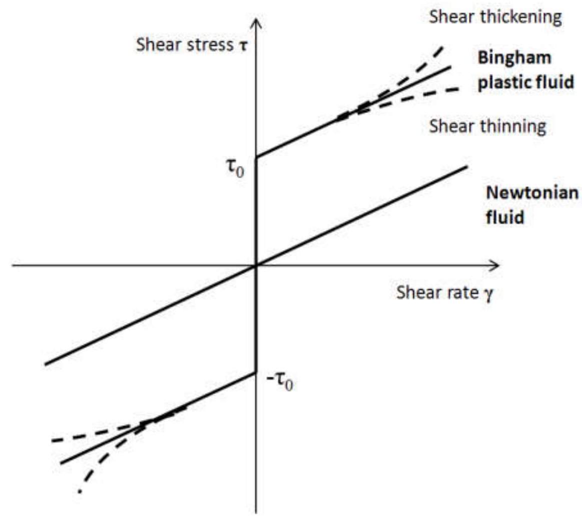


Fig. 1. Bingham plastic model

TABLE I
CHARACTERISTICS OF THE MRF-122EG

Main characteristics	Values
Viscosity	0.42±0.0020(Pa.s, 40^0C)
Density	2.28- 2.48(g/cm^3)
Solids fraction	72%
Flash point	>150^0C
Operating temperature	-40- +130^0C
Maximum magnetic permeability saturation	200-250(kA/m)

In our applications, we utilized Bingham plastic model to describe MR fluid field-dependent behavior[9], as Fig.1.This model assumes that the fluid presents shear stress proportional to shear rate and can be written as:

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

where τ is the shear stress in the fluid, τ_0 is the yield stress controlled by the applied magnetic field, η is the viscosity independent of the applied field, γ is the shear rate and sgn(.) is the signum function. When the shear stress is greater than the critical value τ_0, onset of flow will occur.

When operator inserts catheter through MR fluid, he/she will feel the viscous force (kinesthetic haptic sensation) by applied external magnetic field. Meanwhile, some particles will be moved along with the catheter, and their structure, arranged along the magnetic field lines, will be changed inevitably. Hence, the valve mode has been utilized as operation mode of robot-assisted catheter operation system. In our applications, we utilized a kind of commercial MR fluid named MRF-122EG produced by Lord crop, USA. The main magnetic and rheological characteristics shows in Table 1.

The MR fluid container is a cylinder which is in the center of two magnetic poles, its diameter is 10mm and length is 100mm. The catheter goes through the MR fluid, and it is coaxial with MR fluid container, as showed in Fig. 2.

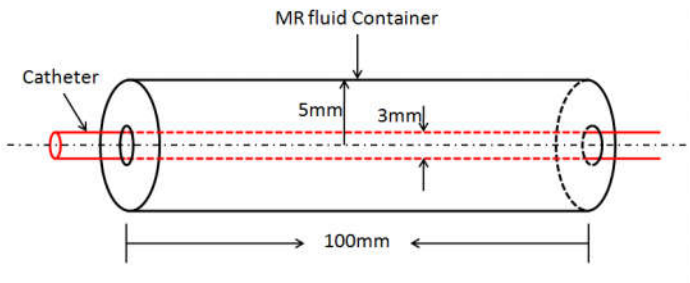


Fig. 2. MR fluid container

B. Design of Electromagnetic Field

MR fluid actuated system is a coupled analysis problem: electromagnetic analysis and fluid system analysis. The purpose of the modeling of fluid actuated catheter haptic interface is to find the relation between the applied electric power (by actuator), usually the current applied to the coils, and output resistance force of the catheter. In general, the magnetic circuit can be analyzed using the magnetic Kirchoff's law as follows[15]:

$$\sum H_i l_i = NI \tag{2}$$

where H_i is the magnetic field intensity in i link of the circuit. l_i is the effective length of magnetic path, N is the number of turns of coil, and I is the applied current. The magnetic flux is determined:

$$\Phi = B_i A_i \tag{3}$$

where the magnetic flux of the two coils is Φ, magnetic flux density of the magnetic path is B_i , and A_i is the cross sectional area. But at low magnetic field, the magnetic flux density is as follows:

$$B_i = \mu H_i = \mu_0 \mu_i \mu_f H_i \tag{4}$$

where μ_0 is the magnetic permeability of vacuum ($\mu_0 = 4\pi 10^{-7}$Tm/A) and μ_i is the relative permeability of the MR fluid container (assuming the relative permeability of the structural materials of container is similar). For non-ferromagnetic materials, $\mu \approx 1$. For MR fluid μ_f typical value is about 5. It is well known that the magnetic field becomes large, the MR fluid will be polarized, and then it will be almost magnetically saturated. However, for modeling purposes, it is hypothesis to note that the relationship between B and H is linear behavior, before the saturation. The magnetic field intensity of the MR fluid can be approximated calculated by

$$H_f = NI / l_f \tag{5}$$

The design idea is showed in Fig. 3, iron cores inside two coils separately are assembled together to generate the electromagnetic field, so the magnetic field intensity changed quickly by the actuator. As shown in Fig.4, an external magnetic field is applied perpendicular to the catheterization direction. The resistance of the catheter insertion can be adjusted by varying the intensity of the applied magnetic field.

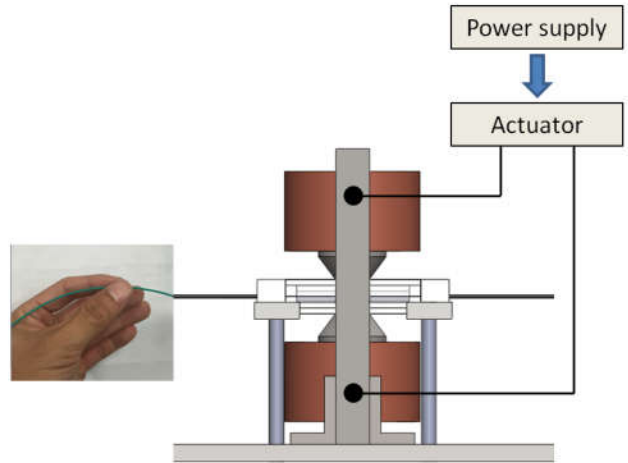


Fig. 3. MR fluid actuated catheter insertion system

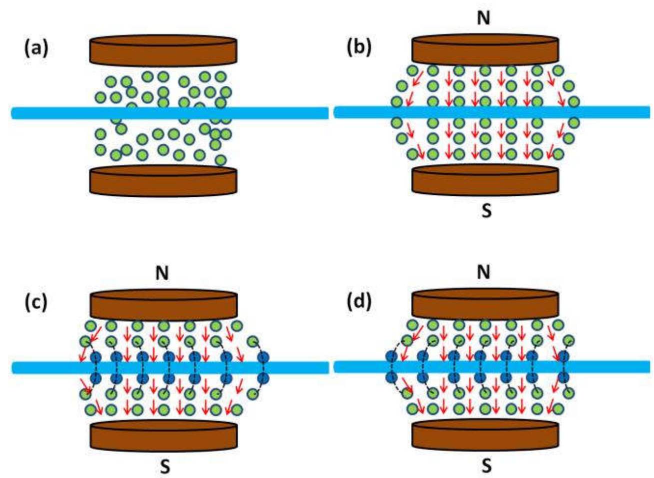


Fig.4. The arrangement state of particles when the magnetic field was provided (b, c, d) or not(a), (b) the catheter was not moved, (c) the catheter was moved forward, (d) the catheter was moved backward.

In the absence of electromagnetic field, the particles are statistically distributed, which can be viewed as a Newtonian fluid, it is shown in Fig. 4(a). The particles were magnetized to the formation of chain-like structures that roughly aligned parallel along the direction of the external electromagnetic field as Fig. 4(b), when exposed to a magnetic field. Fig. 4(c) and Fig. 4(d) showed that the particles, near the catheter, moved along the catheter motion direction, instead of aligned with the external field.

III. Experimental Setup

According to design concept of catheter haptic interface, the catheter was inserted through MR fluid to obtain resistant force sensation just like the catheter inserted into the vascular system. Not only the system should reflect the dynamic changes of the insertion resistance forces, when the catheter goes through the blood to the lesions, but also can detect the collision between the catheter tip and the vessel.

The blood flow dynamics involved geometric and mechanical characteristics, it is very complex[16]. Thus, it was necessary to simplify the dynamic model. Assuming that the flow is a Newtonian incompressible fluid, and the vessel can be divided into many short rigid cylindrical tubes, the length and diameter is constant. The resistance force of hydrodynamic can be simplified as follows[17]:

$$F_f = -\frac{1}{2}\rho v^2 ACl \tag{6}$$

where v is the relative velocity of the catheter tip with respect to the fluid, l is the length of the inserted catheter, ρ denotes the density of the blood, A is the frontal area of the catheter tip and C is the drag coefficient. In our study, we assume the flow to be laminar, the drag coefficient is given by[15]

$$C = \frac{24}{R_e} + \frac{6}{1+\sqrt{R_e}} + 0.4 \tag{7}$$

where R_e is Reynolds number and a dimensionless positive number. In a low Reynolds number, the drag coefficient is approximated by $C=24/R_e$, the Reynolds number can be expressed as follows:

$$R_e = \frac{\rho v d}{\upsilon} \tag{8}$$

where d is the diameter of the catheter tip, υ is the blood's viscosity, as Newtonian model υ is 0.0345P.

The stepping motor and load cell formed the dynamometric system. Constant speed was provided by the stepping motor, and the load cell was used to detect the insertion force. The manipulative catheter is rigid cylindrical tube; its diameter is 2mm, length is 400mm. Tne end of the catheter was connected with the load cell, fixed above the stepping motor, as Fig. 5. The axial movement (forward and backward) of the catheter is controlled by the stepping motor motion controller.

During catheter intervention, the human operator will feel the resistance force F, which is called haptic sensation. It contains of two parts: one is the dynamic friction F_{MR}, caused by the relative motion between the catheter and the MR fluid. Another part of dynamic friction F_{seal} is caused by the seal, when the container is equipped with nothing, it's a constant value. And the haptic sensation equation is:

$$F = F_{MR} + F_{seal} \tag{9}$$

Constant insertion speed is given by the stepping motor motion controller. Five times haptic sensation was measured in forward and backward separately at different insertion velocity, when MR fluid is in 'off-state' situation. Dynamic friction force was measured, the operating speeds were 5mm/s and 10mm/s, separately. During each operating speed, the catheter was manipulated forward and backward 5 times. Results are shown in Table 2, The average of dynamic friction force is about 0.012N.

TABLE II

DYNAMIC FRICTION FORCE OF CATHETER INSERTION IN OFF-STATE MR FLUID

N / V (mm/s)	Number of trials									
	Forward(10^{-3}N)					Backward(10^{-3}N)				
	1	2	3	4	5	1	2	3	4	5
5	11	13	11	12	13	11	10	12	13	11
10	12	13	10	9	12	12	13	13	11	11

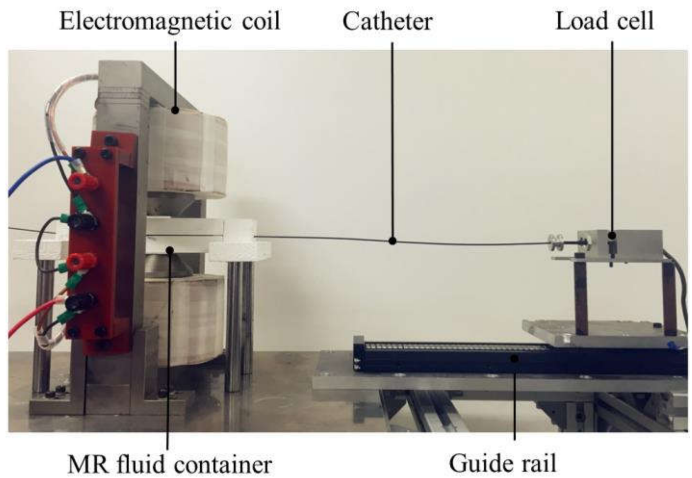


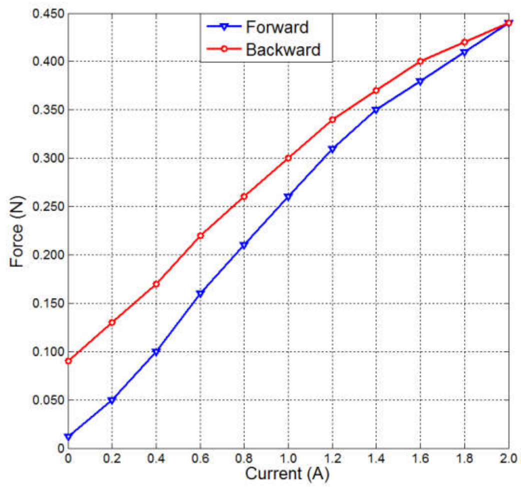
Fig. 5. Experiment setup for obtaining haptic sensation

IV. EXPERIMENTAL RESULTS

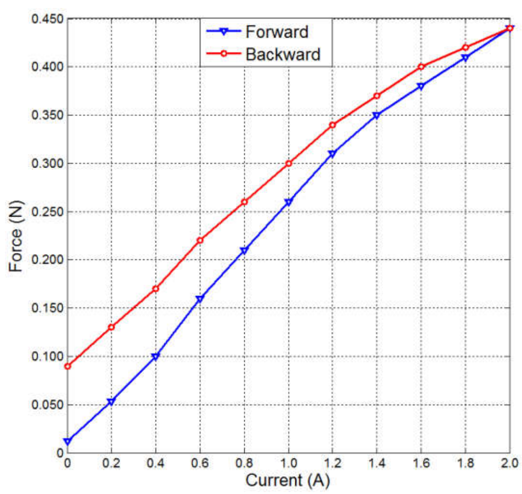
In order to reflect the dynamic changes of the blood viscous force to the MR fluid interface, different currents in the range of 0-2(A) were applied to the coils. The current was changed with constant time interval by the certain step (0.2A, changed 20 times). The real endovascular catheterization movement contains two parts, axial and radial motion. However, here we just analyzed the forward and backward manipulation.

The catheter was navigated by the stepping motor, and the guide rail is 200mm. That means when the operating speed is 5mm/s, the performance time of the catheter is 40s. The current was not altered in 2s. In the same way, when the operating speed is 10 mm/s, the performance time is 20s, and time interval is 1s. During each operating speed, the catheter was manipulated forward and backward 5 times. The Fig. 6 shows the average of forward and backward performance of the catheter movement through the MR fluid device at different speeds (5mm/s and 10mm/s, separately).

At low electric current, the resistance force which MR fluid created was increased linearly, when the catheter was moved forward. After that, decreasing the current and extracting the catheter simultaneously, the resistance force is in a linear decrease. Nonetheless, two curves were not overlap, and the difference was much more significant, with the increase of time. At high speed (10mm/s), the difference is more evident.



(a) Operating speed is 5mm/s



(b) Operating speed is 10mm/s

Fig. 6. The resistance force of on-state MR fluid, during forward and backward manipulation.

V. CONCLUSION

In this paper, a MR fluid actuated haptic interface is designed and experimented with step current. By using the step current, the liner change of insertion and extraction resistance force can be provided. The surgeon operates a real catheter rather than a manipulator or rod to obtain a real haptic sensation of blood vessel. The experimental results indicated that MR Fluid actuated haptic interface has a benefit to describe the blood viscous force during the endovascular catheterization. But at same speed of insertion and extraction, the resistance force curves are different, especially at the high speed (10mm/s). The MR fluid and iron cores show hysteresis phenomenon, the state change of MR fluid is behind with the change of the applied current. Moreover, the inductive time constant of coils should also be taken into consider. The average just noticeable difference (JND) for viscous force is more than 10%, which meet the condition of Tan and Durlach (1991) report that JND lies between 5% and 10% for pinching motions between finger and thumb[18].

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