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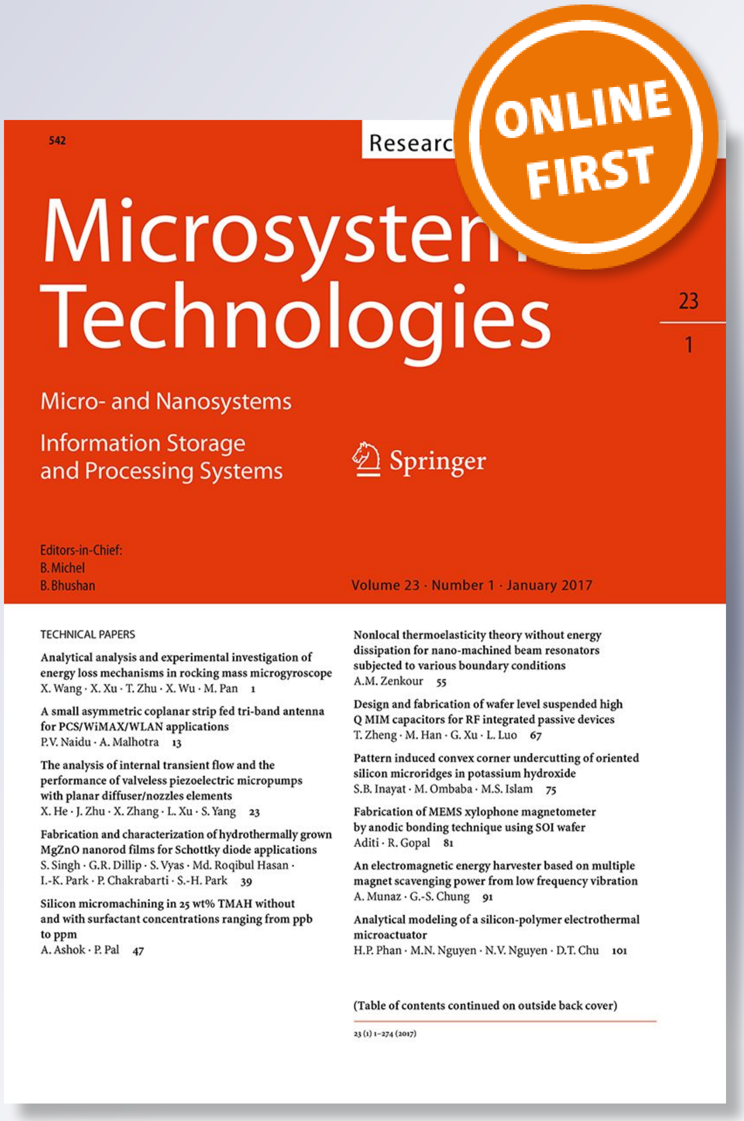
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TECHNICAL PAPER

Performance evaluation of a strain-gauge force sensor for a haptic robot-assisted catheter operating system

Linshuai Zhang[1,4] · Shuxiang Guo[2,3] · Huadong Yu[4] · Yu Song[1]

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Abstract A robotically assisted catheterization system can obviously reduce the radiation exposure to the surgeon and lesson the fatigue caused by standing for long time in protective clothing. However, effective detection and feedback of proximal force signals is essential to the success of a surgery. This paper presents a compact, cost-effective force-sensing device based on strain gauges, for our team developed slave side, to measure the proximal force signals of the input catheter. A significant advantage is that the proposed sensing device can detect the force signals directly without any mechanical transmission, and this can increase the measurement accuracy of force information. In addition, a haptic interface based on MR fluid is designed to provide a realistic sense of operation according to the force signals from the slave side, so as to improve the safety and success of a surgery. Comparing experimental results from an insertion test and a load cell, show that the average error for force measurement is less than 0.01 N. This research provides important insights into the design of compact, ergonomic, robotic catheter manipulators for intraoperative navigation using effective tactile sensing devices.

✉ Shuxiang Guo
guo@eng.kagawa-u.ac.jp

Linshuai Zhang
s16d502@stu.kagawa-u.ac.jp

Huadong Yu
yuhuadong@cust.edu.cn

Yu Song
Jasonsongrain@hotmail.com

1 Graduate School of Engineering, Kagawa University, Takamatsu, Kagawa, Japan

2 Key Laboratory of Convergence Medical Engineering System and Healthcare Technology, The Ministry of Industry and Information Technology, School of Life Science and Technology, Beijing Institute of Technology, Beijing, China

3 Department of Intelligent Mechanical Systems Engineering, Kagawa University, Takamatsu, Kagawa, Japan

4 School of Mechatronical Engineering, Changchun University of Science and Technology, Changchun, Jilin, China

1 Introduction

Cardiovascular and cerebrovascular diseases have become one of three major causes of human deaths, posing a serious threat to human health. Even in the developed countries, cardiovascular disease remains the major cause of mortality, accounting for 34% of deaths each year (Lloyd-Jones et al. 2010).

For vascular tumors, thrombosis, vascular malformations, vascular contractions, vascular sclerosis, and other vascular diseases, vascular interventional surgery (VIS) is found to be the most effective treatments (Guo et al. 2015). Surgeons control a catheter (a rigid hose, with a guide wire) through the blood vessels to dissolve thrombi and expand narrow blood vessels under the guidance of a digital reduction shadow angiography (DSA) system. Compared with traditional surgery, VIS has many advantages: smaller incisions, quicker recovery and fewer complications. Thus, it has been widely adopted all over the world (Fu et al. 2011b; Guo et al. 2016a). Nonetheless, the operation has obvious disadvantages: Surgeries need to be carried out under the guidance of medical imaging equipment, causing damage to the surgeon's body (Kim et al. 2012; Mohapatra et al. 2013). There is a shortage of surgeons; Surgical training takes too long; It is very expensive (Zhang et al. 2011), and due to the high risks involved, the surgeon must be highly skilled and specialized. In other words, the success of the

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surgery will be reduced due to the surgeon's fatigue, physiological tremors and miss operation during fatigue.

In recent years, the use of medical and surgical robotic systems has become a hot study topic. Most of the systems currently used in medical surgery contain master–slave manipulators. However, to solve the VIS problems mentioned above, the combination of robot technology and vascular interventional technology is very important (Lu et al. 2013; Wang et al. 2015b; Xiao et al. 2012). Moreover, an efficiency tele-operated robotic catheter system should be adopted, which can assist the surgeon to operate the catheter interventional from a safe space (Guo et al. 2016b).The physiological tremors and miss operations of a surgeon can be filtered out through the system, increasing the success of surgery (Liu et al. 2011). Many research groups around the world are committed to the development of robotic catheter operating system. Yogesh et al. (2009) developed a novel remote catheter navigation system to reduce the physical stress and irradiation to the user during a fluoroscopic X-ray guided catheter intervention. The results of two experiments showed that the system had the ability to sense and replicate motion to within 1 mm and 1° in the axial and radial directions, respectively. A tele-operation master–slave minimally invasive vascular interventional surgical robot was designed by Ma et al. (2013). Sensors were used to detect the force signals in the axial direction and the torque signals in the radial direction, respectively. It has good maneuverability and can transmit the surgeon's operation to insert and rotate the catheter under the tele-operation. The dynamic and static performance of the system, and the synchronization between master and slave sides were evaluated (Ma et al. 2013). Fu et al. proposed a master–slave catheterization system, which included a steerable catheter with a positioning function and an insertion mechanism with force feedback (Fu et al. 2011a). The design concept of a human operator-centered haptic interface was firstly introduced by Yin (Yin et al. 2014, 2015). A new compact and sterilizable tele-robotic system with three degrees of freedom was proposed, which allowed the interventionalist to use conventional steerable catheters (Tavallaei et al. 2016). In order to simulate the vascular deformation more vividly, Wang et al. introduced a standard linear solid model to formulate its physical model and determine this model's parameters based on a vascular wall elasticity analysis (Wang et al. 2015a). Zhou et al. (2015) described the cardiovascular interventional surgery (CIS) virtual training platform, which was composed of a mechanical manipulation unit, a simulation platform and a user interface. The tests of translation and rotation showed that the accuracies improved by 50% and 32.5%. In addition, the measurement of force signals has been studied in the previous work on robotic catheter systems (Srimathveeravalli et al. 2010; Zhao and Duan 2011; Park et al. 2010; Wang et al. 2014). In these systems, the force sensors were mounted on the patient's side, allowing force signal detection. Compared to traditional catheter interventional methods, these systems can provide advantages such as improving stability and comfort, reducing radiation exposure to the surgeon and eliminating physiological tremors. On the downside, haptic sensations in the master site and force feedbacks in the slave site do not combine with each other. Therefore, it is extremely urgent for tele-operated robotic catheter systems to provide the surgeons with haptic cues during conventional catheterization procedures because excessive forces could rupture the blood vessel walls and result in bleeding.

In this paper, a novel master–slave robot-assisted catheter operating system with a haptic interface and a force-sensing device was developed. During a catheterization procedure, the surgeon operates a real catheter on the master side, and the information of operation will be transmitted to the slave manipulator, which will then control the catheter to do the exact forward, backward and rotational movements in the blood vessel on the slave side. Force signals will be transmitted to the master side from the slave side. A haptic interface, based on magnetorheological (MR) fluid, was proposed to provide a realistic sensation to the surgeon's hand. In addition, an inexpensive strain-gauge-based force sensor developed for the slave side is suitable to measure the proximal force signals of the catheter without any conventional force sensors. The proposed device can be customized to integrate with the catheter manipulator of the patient-side to measure the force signals for the tele-operated robotic catheterization system. The experiments are carried out to evaluate the measurement accuracy of the developed force sensor, and validate the strain-gauge-based method. What's more, the designed haptic interface will greatly increase the success rate and the safety of the surgery.

2 The master–slave robotic catheter operating system

2.1 Overview of the robotic catheter system

The conceptual diagram of the robotic catheter system, shown in Fig. 1, describes the flow chart of the operational process (Guo et al. 2016c). The robotic catheter system consists of five subsystems: master subsystem (master manipulator), slave subsystem (slave manipulator), local control subsystem on the master side and the slave side, and communication subsystem via the internet. When the surgeon operates the master manipulator to insert the catheter, the information of the movement will be obtained by the DSP controller of master side and will be transmitted to

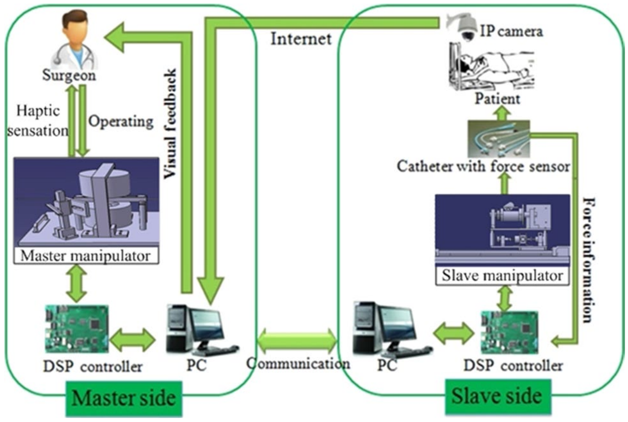

Fig. 1 The conceptual diagram of the robotic catheter system

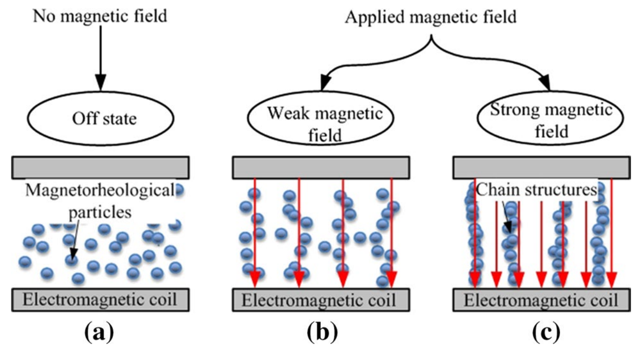

Fig. 2 The magnetic field characteristic of magnetorheological particles: **a** magnetorheological fluid no magnetic field, **b** magnetorheological fluid applied weak magnetic field, and **c** magnetorheological fluid applied strong magnetic field (Tsujita et al. 2013; Yin et al. 2014)

the DSP controller of slave side through the TCP/TP communication protocol (Guo et al. 2016c). The slave manipulator will then insert the catheter into the blood vessel of the patient according to the control information from the master side. At the same time, an IP camera is utilized to monitor the information of the operation process and transmitted to the computer monitor on the master side. The force sensor in the slave manipulator will detect the contact force between the catheter and the blood vessel and the force feedback from the slave side will be reflected to the surgeon's hand through the haptic interface on the master side, improving the safety of the operation.

2.2 The design of master haptic device

Haptic devices have been proposed based on the development of intelligent materials. The most representative application is MR fluid. It is a kind of suspension mixture composed of tiny soft magnetic particles with high permeability and low hysteresis in a non-magnetic liquid. This suspension shows Newton fluid characteristics of low viscosity at zero field conditions, but in a strong magnetic field, the magnetorheological particles will generate chain structures showing the Binghan characteristics of high viscosity and low fluidity. Figure 2 describes the magnetic field characteristics of MR fluid (Yin et al. 2014). Rizzo et al. proposed the Haptic Black Box I and II (HBB I and HBB II) with the concept of freehand to acquire tactile sensation (Sgambelluri et al. 2006; Rizzo et al. 2007). Tsujita et al. (2013) designed a novel encountered-type haptic interface with MR fluids to increase a sense of reality in surgical simulators. Blake and Gurocak (2009) developed a haptic glove with MR brakes for virtual reality.

Taking into account the magnetic field characteristics of the magnetorheological fluid and the requirement of a catheter robotic system, the haptic feedback device based on MR fluid is designed to achieve a realistic sensation,

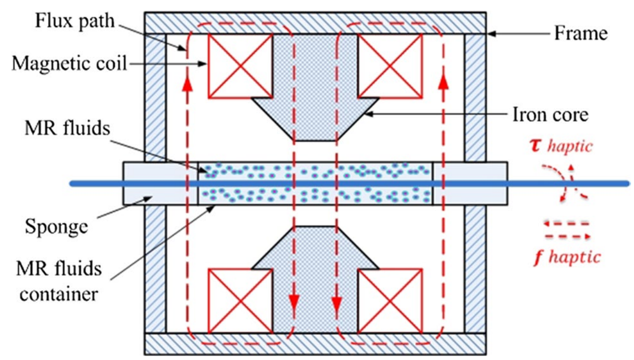

Fig. 3 Schematic of the haptic device

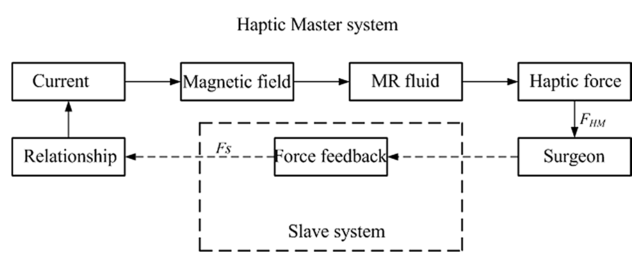

Fig. 4 Control block diagram of the master haptic interface

shown in Fig. 3. According to Fig. 2, the magnetorheological particles will be the chain structures when the magnetic field is applied. When the surgeon operates a catheter in the MR fluid (applied magnetic field), he/she will feel subtle resistance forces caused by the viscosity of the MR fluid, similar to operating a catheter inside the blood vessel of a patient in vascular surgery (Yin et al. 2014). The control block diagram of the master haptic interface is shown in Fig. 4. The force feedback from the slave side will be displayed in the form of current, which can control the magnetic field intensity. And then the

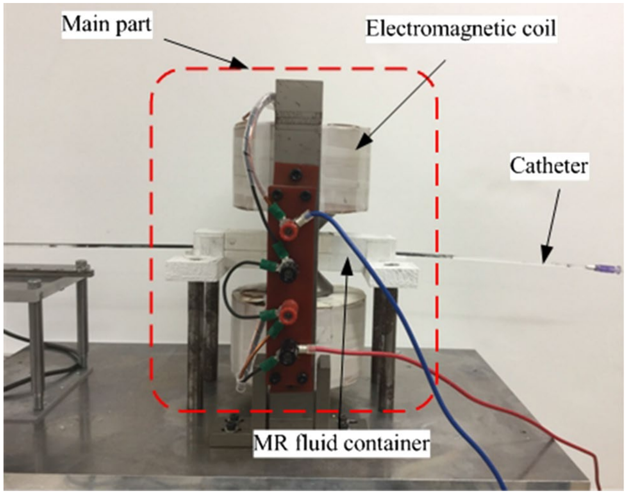

Fig. 5 The fabricated haptic master manipulator

viscosity of MR fluid will be affected so that the shearing force caused by the insertion of a catheter into the MR fluid will be transmitted to the operator in the form of a haptic force.

The prototype of the haptic master manipulator is shown in Fig. 5. In this haptic interface, the haptic force transmitted to the surgeon's hand consists of two parts: the resistance force caused by the viscosity of MR fluid and the friction force between the catheter and the container. Since the total force is very small in a catheter interventional surgery, this friction force has a great influence on the total force. In order to minimize the friction force, sponges with MR fluid are utilized to brace the catheter to avoid contact with the edge of the container. When the surgeon operates inserts or rotates a real catheter in the MR fluid, he/she can feel a realistic sensation similar to that felt in a conventional catheter interventional surgery.

2.3 The design of slave force device

In order for the tele-operated robotic catheter to provide a surgeon with the haptic cues necessary during conventional catheterization procedures, the cooperation between the master haptic interface and the application of force sensing on the slave side is extremely critical. When we use additional sensors, the corresponding structures of mechanical transmission need to be designed to connect the catheter and the sensor. The gap and vibration between the transmission mechanisms will cause a measurement error. However, the proposed sensor on the slave side can directly contact with the catheter to measure the contact force avoiding measurement errors caused by mechanical transmission. In this section, an inexpensive strain-gauge-based force sensor is developed, which is suitable to measure the proximal force signals of the catheter without any conventional force sensors. The proposed device can be customized to integrate with the catheter manipulator of the patient-side to measure the force signals for the tele-operated robotic catheterization.

The schematic diagram of the proposed force device, shown in Fig. 6, is designed to consist of four strain gauges, which are respectively pasted on both sides of the elastomer capable of measuring the deformation of the elastomer. The elastomer is fabricated from A2024 aluminum alloy and has good linearity of load-deformation characteristics and repeatability, small elastic hysteresis and creep, high specific strength and specific stiffness. Compared to the other force sensing methods for remote robotic catheter systems, the proposed design allows the size of the elastomer to be adjusted to the measurement range and resolution according to specific applications. An elastomer with a lower rigidity can provide a higher force measurement resolution but a decreased sensing range.

The insertion force diagram of the proposed force device is shown in Fig. 6. Considering mechanics, the strain ε can be computed according to (1):

$$\varepsilon = |\varepsilon_1| = |\varepsilon_2| = \frac{\sigma}{E} \tag{1}$$

where σ is the elongation rate of the strain gauge, E is the elastic modulus, ε_1 and ε_2 represent the strain gauge 1 and strain gauge 2 in Fig. 6, respectively. Also,

$$\sigma = \frac{M}{W} \tag{2}$$

where M is the torque of the strain gauge in the paste part and W is the bending modulus. From Fig. 6 we get:

$$M = \frac{F \times L}{4} \tag{3}$$

$$W = \frac{b \times t^2}{6} \tag{4}$$

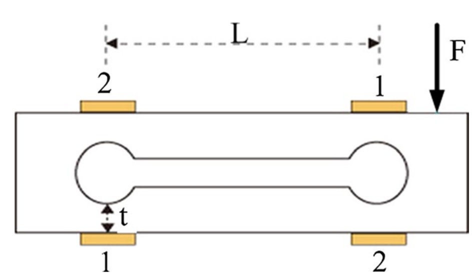

Fig.6 The force diagram of proposed force device

where F is the insertion force which feedbacks from the catheter, L is the centre distance between strain gauge 1 and strain gauge 2 on one side, and b is the thickness of the elastomer, t is the distance from the edge of the small hole to the strain gauge as shown in Fig. 6. From the above equations the strain of strain gauge can be obtained:

$$\varepsilon = \frac{3 \times F \times L}{2 \times b \times t^2 \times E} \tag{5}$$

The strain gauge, made of Cu-Ni alloy with good stability under repeated loading, is used to generate the strain in the axial direction by extension and compression. The strain gauge we choose can be used at least 10 million times and has the operating temperature range from 10 to 100 °C, which is suitable for catheterization procedures. The connection mode of the four strain gauges is shown in Fig. 7. The four strain gauges are named R_1, R_2, R_3, R_4, respectively and their position pasted on the elastomer is shown in Fig. 6. When the insertion force of the catheter is not exerted on the elastomer, according to Ohm's law and Kirchhoff's law of current, we get:

$$e_{out} = \frac{R_1 R_3 - R_2 R_4}{(R_1 + R_2)(R_3 + R_4)} e_{in} \tag{6}$$

where e_{in} is the input voltage of the bridge, and e_{out} is the output voltage. When the insertion force of the catheter transfers to the elastomer, and the resistance value of the strain gauges will be changed slightly.

The change values are $(+\Delta R_1)$, $(-\Delta R_2)$, $(+\Delta R_3)$ and $(-\Delta R_4)$. Then we can obtain the output voltage (Δe):

$$\Delta e = \left\{ \frac{R_1 R_2}{(R_1 + R_2)^2} \left(\frac{\Delta R_1}{R_1} + \frac{\Delta R_2}{R_2} \right) \right\} e_{in} + \left\{ \frac{R_3 R_4}{(R_3 + R_4)^2} \left(\frac{\Delta R_3}{R_3} + \frac{\Delta R_4}{R_4} \right) \right\} e_{in} \tag{7}$$

Moreover, we take $R_1 = R_2 = R_3 = R_4 = R$ with $|\Delta R_1| = |\Delta R_2| = |\Delta R_3| = |\Delta R_4| = \Delta R$ from (7) to get:

$$\Delta e = \frac{\Delta R}{R} e_{in} \tag{8}$$

where R is the original resistance value of the strain gauge, ΔR is the resistance change caused by stretching or compression. Also, there is

$$\frac{\Delta R}{R} = K \times \varepsilon \tag{9}$$

where, K is the sensitivity coefficient of the strain gauge. Then from Eq. (8) and (9), we get:

$$\varepsilon = \frac{\Delta e}{K \times e_{in}} \tag{10}$$

Theoretically, from Eq. (5) and (10) we can get the insertion force of the catheter F:

$$F = \frac{2 \times b \times t^2 \times E \times \Delta e}{3 \times L \times K \times e_{in}} \tag{11}$$

The slave manipulator with the proposed sensor, shown in Fig. 8, is designed to consist of a clamping device, a measuring device and a driving device. One end of the proposed sensor is fixed on the bottom plate in front of the

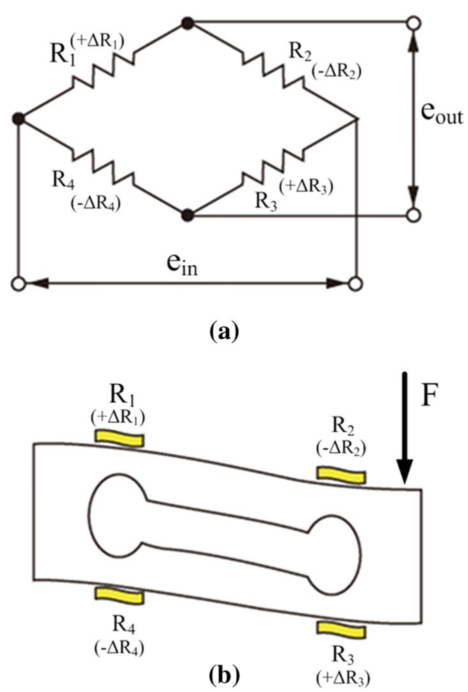

Fig. 7 Schematic diagram of **a** the connection mode and **b** force deformation of the strain gauges

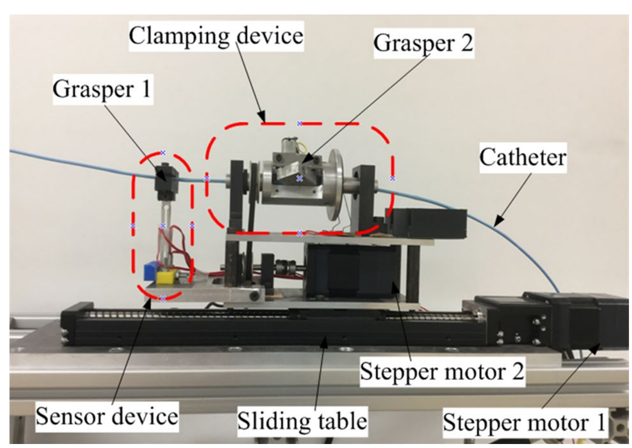

Fig. 8 The assembly of the proposed force sensor in slave manipulator

slave manipulator, and the other end is connected with the clamping device to the catheter, which is fixed with the proposed sensor. The force acting on the catheter will be transmitted directly to the sensor. The whole structure is fixed on a supported frame, which can change the interventional angle from 0° to 45°. Two stepper motors are adopted to realize the insertion and rotation of the catheter, respectively. The clamping action of a surgeon is mimicked by the two designed graspers.

When grasper 1 is closed, grasper 2 is opened, the stepper motor 1 will drive the catheter into the blood vessel, and then the proposed sensor device can measure the force signals transmitted to the master side. When grasper 1 is opened, and grasper 2 is closed, the catheter can be rotated to find the correct direction of insertion by the clamping device and stepper motor 2.

3 Experimental results

In order to validate the sensing device described in the previous sections, three kinds of experiments were performed. The force signals in the axial direction can be calculated according to the measured voltage signals from the proposed sensor based on Eq. (11).

3.1 Experiment evaluation of step response

To validate the step response of the proposed force sensor, a load cell is adopted to do the contrast test. Figure 9 shows the experimental setup. During the step response experiments, a rod was fixed on the developed force-sensing device and connected with the load cell. The proposed sensor and the load cell collect the force signals at the same time when step loads are exerted on the front of the rod. At the beginning of the experiment, we adjusted the input voltage, which made the initial value of the proposed sensor the same as that of the load cell. The whole structure was driven to do forward and backward movements by the stepper motor. Meanwhile, the proposed force sensor and the load cell were given a load and the load cell was interfaced with the data acquisition software to record the benchmark force signals. When the load cell reaches the preset value, the load applied to the proposed sensor will be recorded in the form of output voltage (Δe). The experiments for each load were repeated ten times and the average value was calculated. The average output voltage was then introduced into Eq. (11), and the average value of the force signal was obtained. The calculated values were compared with the values measured by the load cell.

The steady state of step response is shown in Fig. 10. The force magnitudes from 0.1 to 1.0 N with step 0.1 N

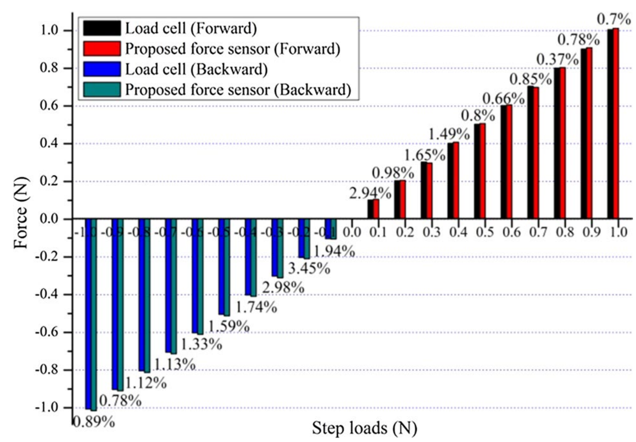

Fig.10 Theexperimental results for force step loads

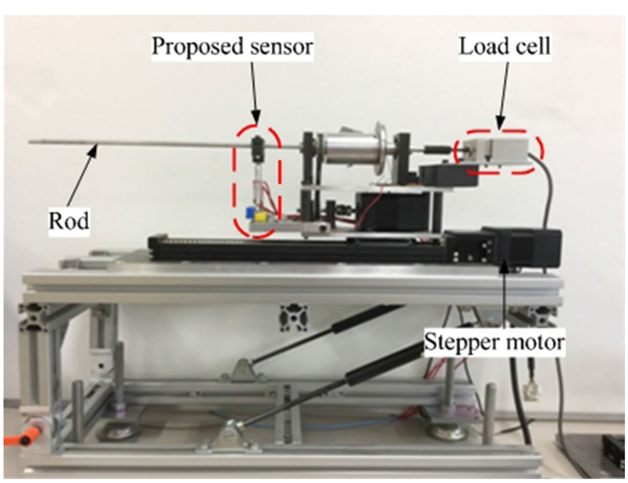

Fig.9 The experimental setup for calibration of measurement accuracy

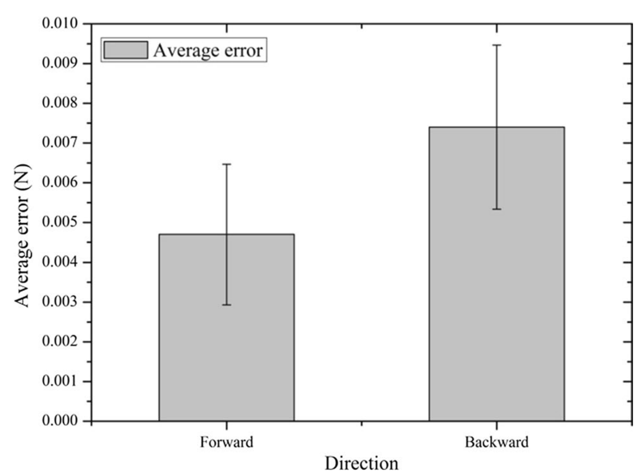

Fig.11 The average error of force signals between proposed sensor and load cell

increases were exerted on the proposed force sensor in the axial direction. The relative errors of force signals are summarized in Fig. 10. The average error of force signals between the proposed sensing device and the load cell is shown in Fig. 11. The experimental results show that the measurement accuracy of the proposed sensor is smaller than 5%, and the average error of the measured value between the proposed force sensor and the load cell is less than 0.01 N, whether it is forward or backward. In actual operation, the accuracy of this measurement is acceptable (Polygerinos et al. 2013).

4 Experiment evaluation of measurement stability

For a new developed force-sensing device, it is essential to test the stability of the measurement. Therefore, we conducted stability experiments for the force device from 0.2 to 1.0 N with step 0.2 N increases. The test for each step load was run 10 times, whether it was forward or backward. The test results, shown in Fig. 12, indicate that whether it is a positive pressure test or a negative pressure test, the measurement of the proposed force sensor has good stability when the applied force is less than 1.0 N. Furthermore, the average error of each step load, shown in Fig. 13, was calculated and the data is expressed as means ± variance. From (a) and (b) we can see that the average error of forward and backward movements were less than 0.01 N, when the exerted force was less than 1.0 N. Such a small error is acceptable in an actual operation. In addition, the contact force between the tip of a catheter and the blood vessel of more than 1 N has been a danger value in actual vascular interventional surgery, therefore, the tests with more than 1 N force exerted is not discussed here.

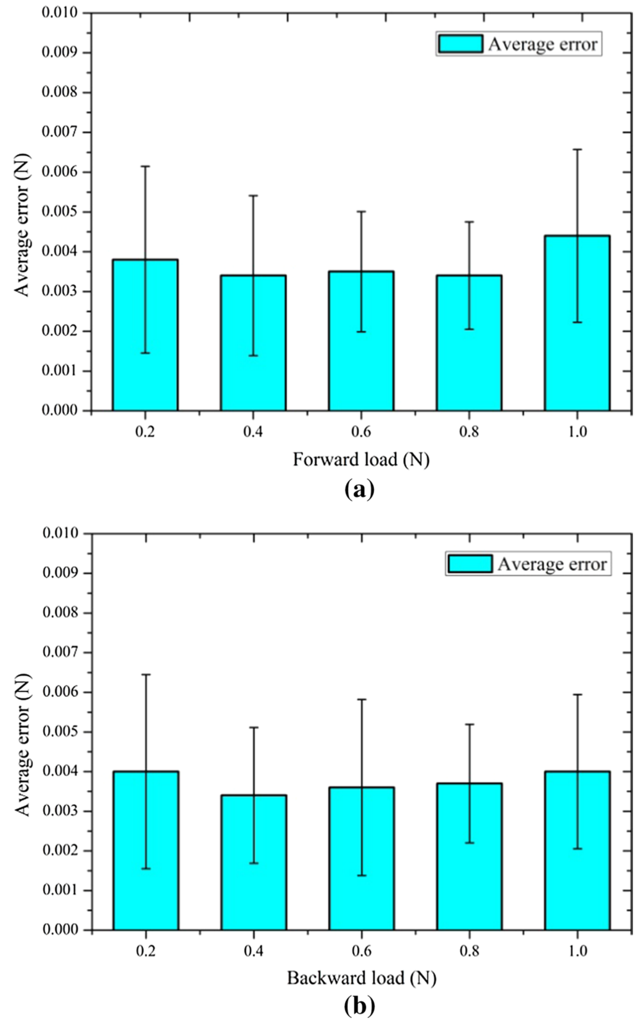

Fig.13 The average error of measuring force in each step load: **a** the average error of forward, **b** the average error of backward

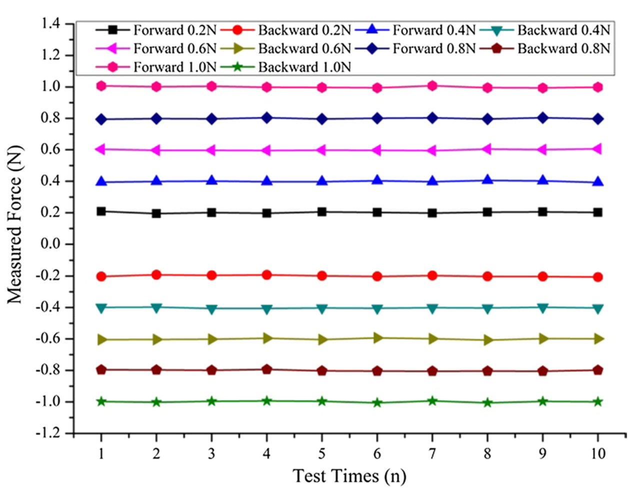

Fig.12 The measurement results of proposed sensing device with step loads

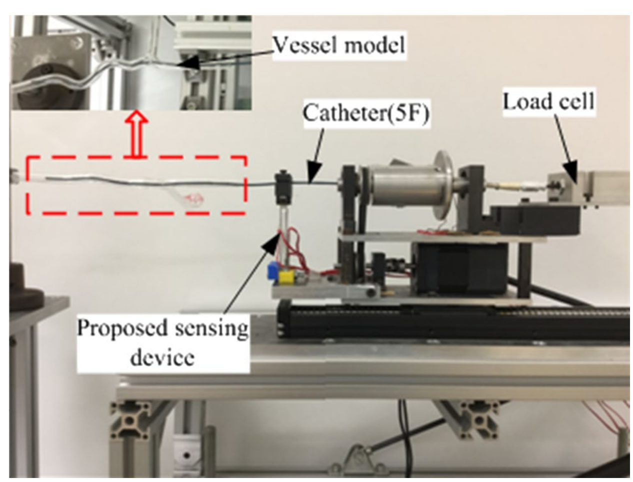

Fig.14 The experimental setup of insertion force in a vessel model

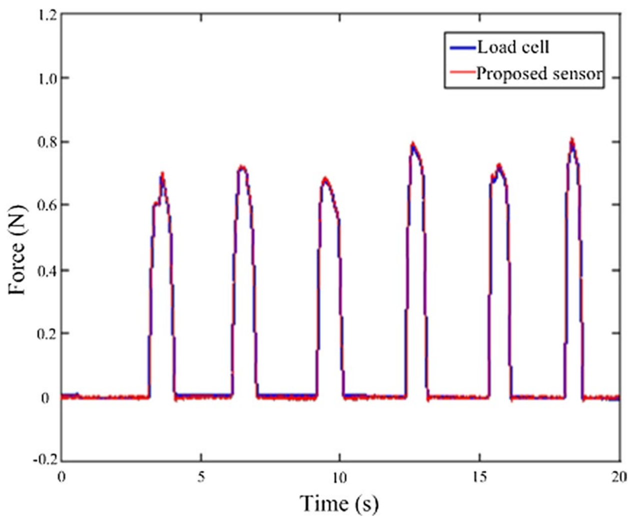

Fig.15 The measurement of insertion force in a vessel model

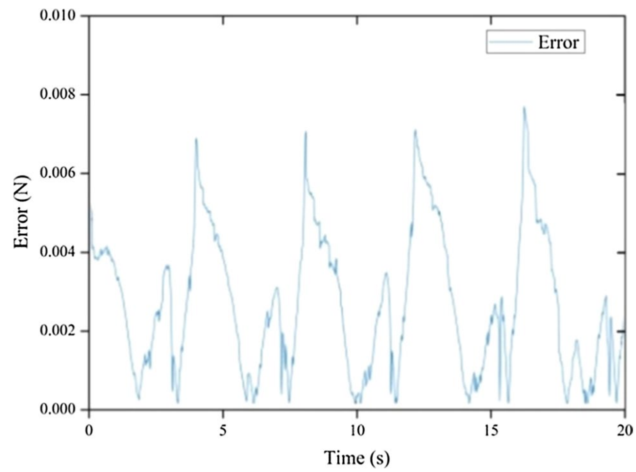

Fig.16 The error of insertion force in a vessel model

4.1 Experiment evaluation of insertion force in vitro

In order to test the ability of the designed sensing device to detect the actual insertion force, a catheter insertion test was carried out. The experimental setup is shown in Fig. 14. In the experiment, a model of an artificial vessel was used to simulate the blood vessel of a human. A 5F catheter, clamped by the proposed sensing device and connected with a load cell, was used for insertion. While the catheter was inserted into the vessel model, the proposed force sensor and the load cell collected the force signals when it went through the bifurcation and bending region of the vessel model. Figure 15 describes the experiment results. From 0 to 4 s, the manipulator moved the catheter forward to the first bending point; from 4 to 7 s, the catheter continued moving forward to the second bending point; from 7 to 10 s, it reached the third bending point; from 10 to 13 s, it reached the forth bending point; from 13 to 16 s, it moved to the fifth bending point; from 16 to 18 s, the catheter reached the last bending point; from 18 to 20 s, the catheter passed through all bending points. At the bending point, the force increased to a peak value because the tip of the catheter is subject to the maximum resistance force.

From the results we can see that the force signals measured by the designed force are consistent with the measured value of the load cell. However, it is obvious that errors remain. For vascular interventional surgery, even a small error may lead to great risks. Therefore, the error of the insertion experiment is calculated. Figure 16 shows the calculation results of the error. From this figure we can see that the maximum error is less than 0.01 N, which is acceptable in the experiment. The experimental results show that the proposed sensing device can be used to record the force signals instead of the load cell so as to eliminate errors caused by mechanical transmission.

5 Discussion

Endovascular robotic technology is an effective method to reduce a surgeon's radiation exposure and fatigue during vascular interventional surgery. However, a realistic sense of operation and an effective force feedback are necessary to prevent excessive forces which could rupture blood vessel walls and result in bleeding. With this feedback, the RMS force could be reduced by 30–60% and the peak value could be dropped by 20–60%. In addition, the operation time could be reduced by 30% and the error rate reduced by 60% (Van der Meijden and Schijven 2009). Therefore, the application of force sensing is extremely critical for a robotic catheter system.

Previous research on a cardiac ablation catheter-tip force sensor has shown that the measurement accuracy needs to be smaller than 5% (Polygeribos et al. 2013). In our proposed device, the maximum value of the measurement accuracy is about 2.98%, which is accurate enough to measure the subtle change of the force signals during the surgery.

In the present work, a compact and cost-effective force-sensing device based on strain gauges was proposed to obtain the force feedback. It can detect the force signals directly without any mechanical transmission, thereby avoiding transmission errors. Furthermore, an added haptic interface based on MR fluid is able to achieve a real sense of operation.

There are however, limitations concerning the proposed sensing system. The current design of the clamp structure is not flexible to the diameters of catheters. In addition, the clamping device is not efficient, as the user needs to tighten and loosen the catheter with screws. In the future, we plan to improve the clamping efficiency of the device using an automatic and flexible clamping structure, which will be suitable for catheters with different diameters.

6 Conclusions

In this paper, a new force sensor for detecting the proximal force signals of the input catheter during a tele-operation catheterization procedure has been proposed based on elastomer and strain gauge. The significant advantage of the proposed device is the resolution and range can be customized according to the specific circumstances by using deferent elastomers and strain gauges. The sensing range of the proposed force sensor can also be adjusted by changing the size and material. A higher resolution can be obtained by using a more sensitive strain gauge. Moreover, the proposed force sensor can detect the insertion force of a catheter without any mechanical transmission reducing measurement errors.

Additionally, the proposed sensing device combines with the haptic interface on the master side to make the surgeon feel a realistic force feedback. The measured force signals from the slave side will be converted to voltage applied to the haptic interface on the master side. The viscosity of the MR fluid can be changed according to the applied voltage, so that surgeons can feel a realistic operating environment.

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