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A Three-axial Force Sensor Based on Fiber Bragg Gratings for Surgical Robots

Wenjie Lai^{1*}, Lin Cao¹, Jiajun Liu¹, Swee Chuan Tjin², Soo Jay Phee¹

Abstract—Haptic feedback is absent in flexible endoscopic surgical robots due to the size constraint of installing sensors on the small robotic arms. Besides, inherent hysteresis caused by the nonlinear friction between tendons and sheaths makes it hard to estimate the distal force by modeling. In this work, we addressed this challenge by proposing a new three-axial force sensor. This standalone device can be seamlessly integrated into the endoscopic robotic arm. Three optical fibers with Fiber Bragg Gratings (FBGs) are embedded in the sensing structure, where one is located at the center hole of the structure ($\varnothing 1.4$ mm), and the other two are eccentrically placed around the structure at 90° apart from each other. This device can measure the pulling force and lateral forces of an articulated surgical instrument. Mechanics analysis has been studied to reveal the link between FBGs' wavelength shifts and forces caused by the elongation and the bending, with a temperature-compensation feature. The sensor has a lateral force sensitivity of 838.386 pm/N, with a measurement resolution of 1.19 mN. Performance comparison with a commercial force sensor Nano17 was made, with measurement errors from 4.50% to 6.18%. In the ex-vivo tests, we teleoperated the sensorized grasper to pull, steer and lift a piece of pig colon tissue. The tool-tissue interaction forces measured by the force sensor were displayed on the computer screen in real-time. In addition to the endoscopic robots, the force sensor can also be integrated with other surgical robots such as laparoscopic robots and catheters.

Index Terms— Force Sensor, Haptic Feedback, Fiber Bragg Gratings, Surgical Robots.

I. INTRODUCTION

ROBOT-ASSISTED minimally invasive surgery (RMIS) has been popular for the last two decades because of enhanced dexterity, less pain, and lower postoperative infection rates [1]. However, unlike traditional open surgery, the surgeon no longer directly touches the tissue in the surgical robots [2-4], and can only rely on the vision to estimate the tool-tissue interaction force. Force perception not only allows the surgeon to timely detect the proper implantation [5], tissue breakthrough [6], and suture breakage [7], but also provides the excellent potential to facilitate the learning of clinical skills and surgical performance

[8, 9]. Thus, haptic feedback is of paramount concern in the field of robotic surgery, including laparoscopic surgery [10-12], endoscopic surgery [13, 14], retinal/cataract surgery [5], and catheter manipulation [8]. Nevertheless, the haptic sensation is still absent in the commercial RMIS systems, which is the main obstacle in advancing teleoperated robot-assisted surgery [9] and autonomous surgery [15]. Weak force control in the flexible surgical robots would cause problems like cognitive burdens to the operator, potential injuries from excessive forces [16, 17], and poor surgical performance [18, 19]. Therefore, this work aims to develop a novel three-axial force sensor in a flexible endoscopic surgical robot and display the measured forces to the operator in real-time.

There are generally three ways of realizing distal force perception in the RMIS: 1) force estimation through modeling methods, 2) vision-based force prediction, and 3) direct force measurement by force sensors.

In flexible surgical robots [20, 21], tendon-sheath mechanisms (TSMs) are employed as the transmission system due to their flexibility, lightweight, and high force capacity. Many researchers have studied the nonlinear friction and the tendon elongation in TSMs to use the proximal force to estimate the distal force [22, 23]. Other than traditional modeling methods, convolutional neural networks and pre-trained long-short-term-memory models were used to predict the distal force in TSMs, without prior knowledge of sheath configurations [24, 25]. Among these studies, it cannot be neglected that dynamic changes in the robot's shape and velocity also change the parameters of the system. Consequently, it is challenging to make an accurate real-time estimation about the distal force purely based on the proximal data since the system states may vary during surgical operations.

The vision-based method can estimate the force from tissue deformation through a stereoscopic camera [26, 27] or multi-cameras [28] in laparoscopic robots, involving 3D map reconstruction. Computer vision is adaptive to the dynamic changes of the operation conditions of the system. However, uncertainties (such as cautery smoke, bloody liquid, and stains on the lens) may degrade the vision quality and affect results. Besides, instead of 3D cameras, a single 2D camera is generally used in the endoscopic robot, which lacks depth information. Therefore, the vision-based force prediction approach in [26-28]

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may not be practical in flexible endoscopic robots.

Various force sensors such as capacitive transducers [11], strain gauges [29], LVDT sensors [30], fiber optics sensors [31–33] have been evaluated to acquire distal force in the RMIS directly. Among these sensors, there is an increasing trend of integrating versatile Fiber Bragg Grating (FBG) sensors in surgical robots to detect the tool-tissue interaction force due to their weightless, thin, and flexible features. Wu *et al.* proposed an FBG-based triaxial force sensor for endoluminal optical biopsy with a force range of 0.5N [31]. The sensor, integrated with a continuum robot, can diagnose early-stage lung cancer, which requires a smaller force. However, the force range of the sensor is insufficient to cover the tool-tissue force in general surgical operations [11, 32, 34]. Lv *et al.* developed an FBG-based palpation force sensor to capture the tissue abnormality during the MIS [33], with a resolution of 2.55mN. However, the sensor can only provide force in one dimension. Most of the existing force sensors are intended for laparoscopic robot [29, 34, 35], needle [36], and catheter [37] that have different sizes and function requirements from the flexible endoscopic robot, and therefore they may not be suitable to apply in the endoscopic surgical robot.

In our previous work, we developed a single-axis FBG-based force sensor [13, 14] for the TSM on endoscopic surgical robots. The small force sensor (length: 3mm) can be seamlessly integrated with the endoscopic robot without any design changes of the surgical instrument. A sensor-model integrated method was further proposed to capture distal tension forces on all TSMs using only one force sensor [38]. Both the sensor and the sensor-model approach can cover the tool-tissue force range in general surgical operations [11, 32, 34]. Apart from force sensing, we also implemented FBG sensors to sense bending in the flexible endoscopic robot. A rotation angle sensor was developed for an articulated surgical grasper with a measurable range of $[-62.9^\circ, 75.5^\circ]$ and a small bending radius of 6.9 mm [39]. This flexible and small sensor can be installed on the tool without affecting the tool's functions. However, all these sensors are for one-dimension measurement. To further save space and improve the measurement capability, we aim to realize multi-axis sensing in this work, i.e., directly measuring multiple forces using only one sensor.

In this paper, we proposed a new three-axial force sensor to measure the pulling force (0 to 6N), the steering force (-3.5N to 3N), and the lifting force ($\pm 4.5\text{N}$) in a flexible endoscopic surgical robot, whose measurement range is larger than/comparable with those in [11, 31, 32, 34]. The force sensor connects the flexible outer tube and the articulated robotic arm. It can sense any external forces on the robotic arm. Three FBGs were attached to the tubular sensor structure using a biocompatible epoxy, where one optical FBG was located at the center hole of the structure (along the x-axis) measuring the pulling force, and the other two were placed around the shaft eccentrically (along the y-axis and the z-axis) measuring the steering and lifting forces, respectively. The advantages of the proposed sensor are as follows: (1) external forces on the robotic arm from different degrees of freedom (DoF) can be directly measured using only one compact sensor; (2) the sensor

serves as the last link of the articulated robotic arm, without increasing the diameter of the robotic arm; in addition, separate from the robotic arm, the sensor can be generally applied to different instruments such as catheters or laparoscopic surgical robots (i.e., not limited one specific robotic arm); (3) the sensor is seamlessly integrated between the flexible outer and the robotic arm. The robotic arm can be easily replaced/repaired at the distal side when it is worn out, without re-installing the force sensor; (4) the three-axial sensor ($\varnothing 1.4$ mm) is smaller than those in [33, 37]; (5) the sensor has a lateral force sensitivity of 838.386 pm/N, with a resolution of 1.19 mN, which are better than/comparable with those in the literature [11, 33, 35]; (6) the sensor can offset the temperature effect during the real-time force measurements. Also, the sensor was applied to a flexible endoscopic surgical robot to demonstrate its effectiveness.

The paper is organized as follows: Section II introduces the development of the force sensor, including the sensor's design, fabrication, and integration with a grasper in the endoscopic surgical robot; Section III explains the working principles and modeling. Section IV provides verification tests and results of the force sensor, inclusive of force calibration, sensor hysteresis study, as well as temperature compensation test. Section V demonstrates the performance of the force sensor, consisting of a force comparison with a Nano17 sensor and **real-time force measurements during colon tissue manipulations in ex-vivo tests**. Section VI concludes this work.

II. DEVELOPMENT OF THE FORCE SENSOR

A. Design and Fabrication

Figure 1. Three-axial Force sensor: (a) Cross-section view of the structure; (b) Back view; (c) 3D model of the force sensor; (d) Design dimensions of the structure; (e) Prototype.

The force sensor has a structure, acting as a connection bridge between the articulated arm and the flexible outer tube, with a total length l of 13.1 mm (Fig. 1a and Fig. 1c). Three FBG optical fibers, labeled as fiber A (with single FBG_A), fiber B (with single FBG_B), and fiber C (with a dual-FBG array: FBG_{C1} and FBG_{C2}), are bonded with the shaft ($D = \varnothing 1.4$ mm) of the structure (Fig. 1a to Fig. 1d). The fiber A is located at the center hole ($d = \varnothing 0.3$ mm) of the shaft, and it is only sensitive to the elongation resulted from the pulling force. Fibers B and C (FBG_{C1}) are embedded in eccentric grooves ($t = 0.4$ mm) at 90° apart from each other (Fig. 1b and Fig. 1d). Typically, fiber

B is located at the left of the force sensor, and fiber C is located at the top. In addition to the elongation, they are also sensitive to the bending caused by the steering force and the lifting force. The FBG_{C2} in fiber C, located 10 mm away from the FBG_{C1}, is not attached to the structure, so that it is considered strain-free and designed as a pure temperature sensor. Working principles and modeling of the three-axial force sensor will be discussed in detail in Section III.

The force sensor was 3D printed using aluminum material *AlSi10Mg* through direct metal laser sintering by *Protolab Inc* (Fig. 1e). The FBG sensors were purchased from *Technica Inc*, with a central wavelength of 1540 ± 0.5 nm or 1550 ± 0.5 nm, an FWHM (full-width-half-maximum)/bandwidth of 0.6-0.7 nm, and a reflectivity larger than 30%. The fiber is recoated with heat-resistant polyimide material that allows the fiber to operate in environments up to 300°C. The effective grating length for each FBG is 1mm, and the gratings were inscribed along the single-mode fiber using a UV laser. The total cost of the force sensor is around USD540. During the manual attachment process, fibers were pasted to the structure in a sequence of fibers A, B, and C, using a biocompatible epoxy *EPO-TEK 353ND*. The epoxy was cured at 150 °C for more than 10 minutes for each FBG, with a dark amber color after curing (Fig. 1e).

B. Integration with a Grasper in the Endoscopic Robot

Figure 2. Force sensor integration with an articulated surgical grasper: (a) 3D model; (b) Cross-section view and back view; (c) Prototype.

After the fabrication, the proximal side of the force sensor is connected with a flexible outer tube of the endoscopic robot, whereas the distal side is linked to an articulated surgical grasper through a pin joint (Fig. 2a). The grasper (ø4.2 mm) has 5-DoF motions, including linear translation, rotation, pitch, yaw, and gripping [40]. The proximal end of the outer tube is connected to a linear slider and a rotation motor that control the translation movement and the rotation motion of the grasper,

respectively. A constraint tendon (Fig. 2b) with two crimping beads at both ends is used to restrain the undesired elongation of the flexible outer tube during the translation movement. One crimping bead is fixed with the linear slider at the proximal end, and the other one is fixed with the force sensor. Thus, when the grasper grips and pulls tissue at the distal end, the force sensor will experience elongation due to the pulling force. There are three pin joints in the grasper, for pitch (pin joint 1), yaw (pin joint 2), and gripping (pin joint 3) motions. Six TSMs (tendon's outer diameter: 0.4 mm, sheath's outer diameter: 0.9 mm) are installed to actuate these 3-DoF motions (Fig. 1d and Fig. 2b), where each DoF is driven by a pair of TSMs [38].

After integration, a sensorized grasper was assembled (Fig. 2c). Then the grasper was installed to the flexible endoscopic surgical robot, where six tendons were mounted with motors at motor housing [38]. Pretension (~4N) was applied to each tendon at the proximal side to prevent its slacking along the flexible outer tube. The tension on the tendon cannot deform the sensing structure of the sensor because the tension force is balanced by the compression force on the sheath in each TSM [14, 38]. Near the motor housing, an interrogator (*Micron Optics SI-255*) was used to trace Bragg wavelengths of the three FBG fibers automatically at a scan rate of 1000 Hz. Note that the last two links of the assembled grasper (i.e the robotic arm) can be easily replaced/repared when it is worn out at the distal side, without any reinstallation of the force sensor.

III. WORKING PRINCIPLES AND MODELING

A. Mechanics Analysis

1) Bending due to Lifting Force and Steering Force

After installation, the flexible outer tube passes through and sits in a channel of the endoscope (Fig. 3a). When the grasper grips some tissue and intends to lift the tissue, it will experience a counterforce F_z (downwards), which makes the grasper to rotate at pin joint 1 that constraints the pitch movement only in the x-z plane. A free-body diagram of the grasper is drawn in the x-z plane for this scenario. The force sensor is labeled as ①, whereas the last two links of the grasper are treated as a single body and labeled as ②. The wall of the channel supports the proximal end of body ①.

Figure 3. Mechanics analysis about the lifting force: (a) Grasper model with a free body diagram in the x-z plane; (b) Simplified diagram of body ①: cantilever beam with a small deflection; (c) Sensors' locations.

Compared with the weight of the tissue, the weight of body ② is assumed negligible. Based on *Newton's Third Law*, (1) is obtained, where F_{12} is the force exerted from body ① to body ② whereas F_{21} is the force exerted from body ② to body ①.

$$F_z = F_{12} = F_{21} \quad (1)$$

For body ①, it is simplified as a cantilever beam with a single load at the end (Fig. 3b). Assuming body ① undergoes small deflection in the linearly elastic region and has a uniform cross-section. According to the deflection formula of the end-loaded cantilever beam, the total deflection W of body ① is $W = w(L) = \frac{F_z L^3}{3EI}$. Here, F_z is the vertical force, also called as the lifting force; L is the length of the body ①; E is the modulus of elasticity; I is the area moment of inertia of the shaft's cross-section, and $I = \frac{\pi D^4}{64}$; θ is the slope angle at $x=L$.

Assuming a constant curvature occurs during the bending of body ① owing to F_z . Based on the geometry, (2) is directly obtained. R is the bending radius; κ is the bending curvature and $\kappa = \frac{1}{R}$.

$$\tan \frac{\theta}{2} = \frac{W}{L} = \frac{F_z L^2}{3EI} \quad (2)$$

When body ① bends due to F_z , the axial elongation is very small (compared to the original length L) and can be neglected. Thus, the length of body ① maintains as L after the bending. Then we have $\theta = \kappa L$.

Using the small-angle approximation, we have $\tan \frac{\theta}{2} = \frac{\theta}{2}$ when θ is close to zero. Then we can express the bending curvature κ as follows.

$$\kappa = \frac{2F_z L}{3EI} \quad (3)$$

According to FBG's working principle, when there is no temperature change, a linear relationship exists between the Bragg wavelength shift $\Delta\lambda_B$ and strain ε of the fiber, as shown in (4) [13]. k_ε is the strain sensing coefficient of the fiber material; λ_B is the Bragg wavelength of the FBG.

$$\frac{\Delta\lambda_B}{\lambda_B} = k_\varepsilon \varepsilon \quad (4)$$

Figure 4. FBG sensor is eccentrically located on the shaft: (a) FBG is under tension; (b) FBG is under compression. The grey area stands for the shaft, while the yellow line with short black lines represents for FBG sensor.

The bending of body ① leads to the strain change of FBG_{C1} that is eccentrically located on the shaft (Fig. 3c). FBG_A and FBG_B are insensitive to the bending due to the lifting force F_z , because their offset distances to the neutral axis of the force sensor are zero in the x-z plane (verified in Section IV.A). The bending curvature κ is proportional to the Bragg wavelength shift $\Delta\lambda_B$, as shown in (5) [39], where t_{offset} indicates the

distance between the fiber C and the neutral axis of body ① (Fig. 4).

$$\kappa = \frac{1}{k_\varepsilon t_{offset}} \frac{\Delta\lambda_B}{\lambda_B} \quad (5)$$

Combining (3) and (5), a linear relationship between the Bragg wavelength shift $\Delta\lambda_{B,Fz}$ and the lifting force F_z is derived:

$$\Delta\lambda_{B,Fz} = \frac{2k_\varepsilon \lambda_{B,C1} t_{offset} L_{lifting}}{3EI} F_z, \quad (6)$$

where $\lambda_{B,C1}$ is the Bragg wavelength of FBG_{C1} in fiber C, and the length of body ① is $L = L_{lifting}$.

Therefore, the lifting force sensitivity K_{fz} of FBG_{C1} , with a unit of pm/N , can be expressed as in (7):

$$K_{fz} = \frac{2k_\varepsilon \lambda_{B,C1} t_{offset} L_{lifting}}{3EI} \quad (7)$$

Similarly, when the grasper grips some tissue and intends to steer it in the x-y plane, it will experience a horizontal counterforce F_y (named as the steering force), which makes the grasper rotate at pin joint 2 that constraints the yaw movement only in the x-y plane (Fig. 5). A free body diagram of the grasper is drawn in the x-y plane for this scenario, where the force sensor and the second last link of the grasper are treated as a single body and labeled as ③, and the last link of the grasper is labeled as ④. In this case, only FBG_B is sensitive to the bending caused by the steering force. FBG_A and FBG_{C1} are insensitive to the bending due to the steering force F_y , because they do not have offset distance to the neutral axis of the force sensor in the x-y plane (verified in Section IV.A).

Figure 5. Mechanics analysis about the steering force: Grasper model with a free body diagram in the x-y plane.

For body ③, except that its length L becomes longer, the rest of the mechanics analysis is like the case about the lifting force. Thus, there is a linear relationship between the Bragg wavelength shift $\Delta\lambda_{B,Fy}$ and the steering force F_y , where $\lambda_{B,B}$ is the Bragg wavelength of FBG_B , and $L_{steering}$ is the length of body ③.

$$\Delta\lambda_{B,Fy} = \frac{2k_\varepsilon \lambda_{B,B} t_{offset} L_{steering}}{3EI} F_y \quad (8)$$

The steering force sensitivity K_{fy} can be expressed as:

$$K_{fy} = \frac{2k_\varepsilon \lambda_{B,B} t_{offset} L_{steering}}{3EI} \quad (9)$$

Since all parameters in (7) and (9) are constant, both the lifting force sensitivity and the steering force sensitivity are constant as well, which will be obtained during force calibration tests in Section IV.A. When designing such a force sensor, we can increase the sensitivity in the y-axis and z-axis by increasing the value of t_{offset} or L , depending on the application.

Since $L_{lifting} < L_{steering}$ and the rest parameters have almost identical values in each case, we derive the following relationship straightforwardly.

$$K_{fz} < K_{fy} \quad (10)$$

2) Elongation due to Pulling Force

When the grasper grips some tissue and intends to pull back along the x -direction, it will experience a counter forward force F_x (Fig. 6). A free-body diagram is illustrated for the shaft of the force sensor. ε indicates the strain of the shaft due to elongation δL resulted from the pulling force. A is the area of the shaft, and $A = \frac{\pi D^2}{4}$.

Figure 6. Mechanics analysis on the shaft: elongation caused by the pulling force.

From Hooke's law and definition of stress and strain, we have $\varepsilon = \frac{F_x}{AE}$ in the elastic region. Substituting it into (4), we obtained (11) after rearrangement:

$$\Delta\lambda_B = \frac{k_\varepsilon \lambda_B}{AE} F_x \quad (11)$$

A linear relationship between the Bragg wavelength shift $\Delta\lambda_B$ and the pulling force F_x is derived. The pulling force sensitivity K_{fx} can be expressed as in (12).

$$K_{fx} = \frac{k_\varepsilon \lambda_B}{AE} \quad (12)$$

Since $\frac{K_{fx}}{K_{fy}} = \frac{3D^2}{32t_{offset}L_{steering}} \ll 1$, $\frac{K_{fx}}{K_{fz}} = \frac{3D^2}{32t_{offset}L_{lifting}} \ll 1$, the following relationships can be implied.

$$K_{fx} \ll K_{fy}, \quad K_{fx} \ll K_{fz} \quad (13-14)$$

Note that FBG_A, FBG_B, and FBG_{C1} can all detect the elongation due to the pulling force. Values of associated pulling force sensitivities $K_{fx,A}$, $K_{fx,B}$, and $K_{fx,C1}$ are obtained in Section IV.A. Relationships (10), (13), and (14) are also verified by the experiment result in Section IV.A. A model is built to decouple the elongation and the bending effect in FBG_B and FBG_{C1} in the next subsection.

B. Temperature Compensation and Decoupling between Bending and Elongation

In the mechanics analysis, the temperature change is assumed to be zero. However, in real surgery, the ambient temperature around the surgical tool may change during the surgical procedure. Both strain change and temperature variation will affect the FBG's wavelength shift [13]. In this subsection, a temperature compensation study is conducted to cancel the undesired temperature effect. A modeling study to decouple the effects of bending and elongation is also presented.

We define the temperature-induced wavelength shifts of FBG_A, FBG_B, FBG_{C1}, and FBG_{C2} as $\Delta\lambda_{B,AT}$, $\Delta\lambda_{B,BT}$, $\Delta\lambda_{B,C1T}$, and $\Delta\lambda_{B,C2T}$, respectively (Fig. 1a). The strain-free FBG_{C2} is treated as a pure temperature sensor.

According to FBG's working principle, when there is only temperature change, a linear relationship exists between the

Bragg wavelength $\Delta\lambda_B$ and temperature variation ΔT [13]. Thus, (15) to (18) are obtained. We define $K_{T,A}$, $K_{T,B}$, $K_{T,C1}$ and $K_{T,C2}$ as the temperature sensitivity of FBG_A, FBG_B, FBG_{C1}, and FBG_{C2}. $r_{T,A}$, $r_{T,B}$, and $r_{T,C1}$ are temperature sensitivity ratios from FBG_A, FBG_B, FBG_{C1} to FBG_{C2}. Their values can be obtained from the temperature compensation test in Section IV.C.

$$\Delta\lambda_{B,C2T} = K_{T,C2}\Delta T \quad (15)$$

$$\Delta\lambda_{B,AT} = K_{T,A}\Delta T = r_{T,A}\Delta\lambda_{B,C2T} \quad (16)$$

$$\Delta\lambda_{B,BT} = K_{T,B}\Delta T = r_{T,B}\Delta\lambda_{B,C2T} \quad (17)$$

$$\Delta\lambda_{B,C1T} = K_{T,C1}\Delta T = r_{T,C1}\Delta\lambda_{B,C2T} \quad (18)$$

As aforementioned, when the grasper grips some tissue and lifts it, a lifting force F_z is exerted from the tissue to the tooltip, making the shaft bend bidirectionally about the pin joint 1. The deformation of the shaft due to the bending will be detected by FBG_{C1}, whereas FBG_A and FBG_B are insensitive to F_z because they do not have the offset distance. We define the F_z -induced wavelength shifts of FBG_{C1} as $\Delta\lambda_{B,C1Fz}$. Similarly, when steering force F_y is exerted on the tooltip, the shaft bends bidirectionally about the pin joint 2, and the deformation will be detected by FBG_B, whereas FBG_A and FBG_{C1} are insensitive to F_y . We define the F_y -induced wavelength shifts of FBG_B as $\Delta\lambda_{B,BFy}$. When pulling force F_x is exerted on the end of the grasper, the shaft will experience strain change due to the tension, which will be captured by FBG_A, FBG_B, and FBG_{C1}. We defined the relevant F_x -induced wavelength shifts as $\Delta\lambda_{B,AFx}$, $\Delta\lambda_{B,BFx}$, and $\Delta\lambda_{B,C1Fx}$.

Equations (19) to (22) summarize all contribution in FBGs' wavelength shifts $\Delta\lambda_{B,A}$, $\Delta\lambda_{B,B}$, $\Delta\lambda_{B,C1}$ and $\Delta\lambda_{B,C2}$ of FBG_A, FBG_B, FBG_{C1} and FBG_{C2}.

$$\Delta\lambda_{B,A} = \Delta\lambda_{B,AT} + \Delta\lambda_{B,AFx} \quad (19)$$

$$\Delta\lambda_{B,B} = \Delta\lambda_{B,BT} + \Delta\lambda_{B,BFy} + \Delta\lambda_{B,BFx} \quad (20)$$

$$\Delta\lambda_{B,C1} = \Delta\lambda_{B,C1T} + \Delta\lambda_{B,C1Fx} + \Delta\lambda_{B,C1Fz} \quad (21)$$

$$\Delta\lambda_{B,C2} = \Delta\lambda_{B,C2T} \quad (22)$$

Substituting (15)-(18) into (19)-(22), after rearrangement, expressions of F_x , F_y , and F_z are gained as follows.

$$F_x = \frac{1}{K_{fx,A}} (\Delta\lambda_{B,A} - r_{T,A}\Delta\lambda_{B,C2}) \quad (23)$$

$$F_y = \frac{1}{K_{fy}} \left(\Delta\lambda_{B,B} - r_{T,B}\Delta\lambda_{B,C2} - \frac{K_{fx,B}}{K_{fx,A}} (\Delta\lambda_{B,A} - r_{T,A}\Delta\lambda_{B,C2}) \right) \quad (24)$$

$$F_z = \frac{1}{K_{fz}} \left(\Delta\lambda_{B,C1} - r_{T,C1}\Delta\lambda_{B,C2} - \frac{K_{fx,C1}}{K_{fx,A}} (\Delta\lambda_{B,A} - r_{T,A}\Delta\lambda_{B,C2}) \right) \quad (25)$$

Wavelength shifts $\Delta\lambda_{B,A}$, $\Delta\lambda_{B,B}$, $\Delta\lambda_{B,C1}$ and $\Delta\lambda_{B,C2}$ can be all captured by the interrogator in real-time. Using (23) to (25), we can figure out forces F_x , F_y , and F_z , with temperature compensation feature and decoupling effect from elongation and bending for FBG_B and FBG_{C1}. Note that when there is only F_x on the force sensor, (23) is also adaptive to FBG_B and FBG_{C1}, i.e., $F_x = \frac{1}{K_{fx,B}} (\Delta\lambda_{B,B} - r_{T,B}\Delta\lambda_{B,C2})$, $F_x = \frac{1}{K_{fx,C1}} (\Delta\lambda_{B,C1} - r_{T,C1}\Delta\lambda_{B,C2})$. Relevant verification test results are shown in Section IV.C.

IV. VERIFICATION EXPERIMENTS

A. Force Calibration Test

1) Experimental Setup

A test was conducted to obtain the pulling force sensitivities $K_{f_{x,A}}$, $K_{f_{x,B}}$, and $K_{f_{x,C1}}$ of the assembled grasper, where a load cell *Futek FSH03963/LTH300* was used to collect the pulling force. Wavelength shifts of FBG_A, FBG_B, and FBG_{C1} were collected through the interrogator *Micron Optics SI-255* (accuracy: 1 pm).

Figure 7. The platform for the pulling force calibration test.

In the platform for the pulling force calibration test (Fig. 7), the proximal end of the force sensor is fixed using a clamp holder, and the rest of the grasper lies along with a slot of a constraint part. The constraint part is designed to restrain the movement of the grasper only in the x-direction. The grasper bites a tendon that extends to a load cell in a linear slider. The distal side of the linear slider connects to tendons, a spring, and a pulley. As the orange arrows depict, when the pulley rotates, the linear slider will move forwards and pull the tendon that links to the grasper. Inside the load cell, the tendon is restrained using a crimping bead, so that the tension force on the tendon is converted to the compression force on the load cell. Thus, the load cell can reflect the pulling force.

Another two tests were carried out to obtain lifting force sensitivity K_{f_z} and the steering force sensitivity K_{f_y} of the assembled grasper, where the lifting/steering force was collected by a 6-DOF Force/Torque sensor *ATI Nano17*. Wavelength shifts of FBG_A, FBG_B, and FBG_{C1} were collected through the interrogator *Micron Optics SI-255*.

Figure 8. The platform for the lifting/steering force calibration tests: (a) Closer view at the distal side during the lifting force calibration; (b) Top view at the distal side.

In the lifting/steering force calibration platform (Fig. 8), the proximal end of the force sensor is fixed with a clamp holder that is mounted on a plate of a manual 3-axis XYZ translation stage *Photonik PPT3B* (resolution: 1 μ m; range: 25 mm). The last link of the grasper (opening/closing jaw) is inserted into a hole of a supporting part that is mounted with the *Nano17* sensor. There is some clearance between the hole and the jaw, and the contact surface is smooth so that it can slide in the x-axis and gets free from the force F_x . Both the support part and

the *Photonik PPT3B* are fixed on the ground. When the z-axis knob of the *Photonik PPT3B* is rotated, the assembled grasper mounted on the plate will move up/down and cause the bending of the force sensor in the x-z plane. The grasper will experience the vertical force F_z . Similarly, when the y-axis knob of the *Photonik PPT3B* is rotated, the assembled grasper mounted on the plate will steer to left/right and cause the bending of the force sensor in the x-y plane. The grasper will experience the horizontal force F_y .

2) Experimental Results

Figure 9. Pulling force calibration results: F_x v.s Wavelength shifts (LF is short for linear fitting).

Using the platform in Fig. 7, a pulling force calibration test was conducted, where the pulley was rotated to increase the value of the pulling force from 0N to 6 N. All data from the load cell *Futek FSH03963/LTH300* and the interrogator *Micron Optics SI-255* were collected at a frequency of 1000 Hz.

The test result (Fig. 9) shows a linear relationship between the Bragg wavelength shift and the pulling force for FBG_A, FBG_B, and FBG_{C1}, which conforms to (11) and (12). The pulling force sensitivity $K_{f_{x,A}}$ for FBG_A is 11.441 pm/N, with a root mean square error (RMSE) of 0.193 N and R^2 (R-square) of 0.992. The pulling force sensitivity $K_{f_{x,B}}$ for FBG_B is 13.561 pm/N, with an RMSE of 0.468 N and R^2 of 0.959. The pulling force sensitivity $K_{f_{x,C}}$ for FBG_{C1} is 26.690 pm/N, with an RMSE of 0.313 N and R^2 of 0.975. The difference in pulling force sensitivity of FBG_A, FBG_B, and FBG_{C1} may be due to the volume and distribution of epoxy applied in the manual adhesion process. Small clearance exists between the hole in the constraint part and the grasper (Fig.7). When the tendon pulls the grasper, the grasper's tip will undesirably move in the y-axis and the z-axis because of the small misalignments between the linear slider and the constraint part (due to manufacturing errors). These unwanted movements cause the bending of the force sensor, resulting in ill-favored measurement noises on FBG_B and FBG_{C1}. When the pulling force increases, the noises/errors will become larger, explaining why larger errors occur after 3.5 N for FBG_B and 4.5 N for FBG_{C1}.

Figure 10. Steering force calibration result: F_y v.s Wavelength shifts.

Figure 11. Steering force calibration result.

Using the platform in Fig. 8, a steering force calibration test was carried out, where the y-axis knob of the translation stage was tuned to steer the assembled grasper from the neutral axis to the left and the right, which leads to the shaft bending of the force sensor. The deformation along the shaft due to the bending caused by the steering force was reflected through the wavelength shifts of the FBG_B. FBG_A and FBG_{C1} were relatively insensitive to the bending due to the steering force F_y , as can be seen in the test results (Fig. 10). The good linear relationship between the Bragg wavelength shift of FBG_B and the steering force conforms to (8) and (9) in Section III.A. After the calibration with a steering force range from -3.5 N to 3 N, the RMSE was calculated (Fig. 11). The force sensor has a high steering force sensitivity K_{fy} of 1757.438 pm/N, R^2 of 0.998, and an RMSE of 0.088 N. The interrogator system has a resolution of 1 pm; thus, the force sensor has a steering force resolution of $\left\{\frac{1}{K_{fy}}\right\} = 0.57$ mN, which is better than/comparable with the resolution in the existing literature [11, 33, 35]. The errors of FBG_B mainly resulted from the undesired small vibrations during our manual tuning of the y-axis knob of the XYZ translation stage.

Figure 12. Lifting force calibration result: F_z v.s Wavelength shifts.

Figure 13. Lifting force calibration result.

Similarly, a lifting force calibration test was performed (Fig. 8a), where the z-axis knob of the translation stage was tuned to move the assembled grasper up and down, which leads to the shaft bending of the force sensor. The deformation along the shaft due to the bending caused by the lifting force was reflected through the wavelength shifts of FBG_{C1}. FBG_A and FBG_B are insensitive to bending due to the lifting force F_z , as can be seen in the test results (Fig. 12). From the result, the good linear relationship conforms to (6) and (7) in Section III.A. After the calibration with a lifting force range from -4.5 N to 4.5 N, the RMSE was calculated (Fig. 13). The force sensor has a high

lifting force sensitivity K_{fz} of 838.386 pm/N, R^2 of 0.994, and an RMSE of 0.156 N. The lifting force resolution of the force sensor is $\left\{\frac{1}{K_{fz}}\right\} = 1.19$ mN, which is better than/comparable with the existing literature [11, 33, 35].

Comparing force sensitivities values obtained from the calibration tests, we have the relationship of $K_{fx} \ll K_{fz} < K_{fy}$, which agrees with (10), (13), and (14) in Section III.A. Besides, various amount of epoxy applied to fibers B and C during the manual fabrication process leads to the various actual value of t_{offset} , which may contribute to the difference of lateral force sensitivities.

Additionally, the calibration force ranges (F_x : 0 N to 6 N, F_y : -3.5 N to 3 N, and F_z : ± 4.5 N) for this force sensor sufficiently covers the measured pulling force (0.18 N-1.09 N) and lifting force (0.36 N-1.43 N) during in-vivo tests in [34], the maximum pulling force of 1.45 N in [11], and the axial force range is 2.5 N in [32] whereas 5 to 6 N in [41, 42] for MIS tools. The polyimide-recoated optical fiber and the epoxy can work continuously up to 200 °C, and the AlSi10Mg sensing structure maintains performance within 200 °C [43], so that the force sensor may bear the high temperature (132 °C) during the steam sterilization. In the case of poor sterilizability, the sensor can be disposable.

B. Sensor Hysteresis Study

Figure 14. Force hysteresis results: (a) FBG_A; (b) FBG_B; (c) FBG_{C1}.

Using the same platform, as shown in Fig. 8, we did a hysteresis study of the force sensor. Three separate tests were conducted for FBG_A, FBG_B, and FBG_{C1}, and data were collected at a frequency of 1000 Hz.

During the hysteresis test for FBG_A, only the pulling force was changed by turning the x-axis knob of the XYZ linear stage. The force was loaded from 0 N to 4 N and then unloaded from 4 N to 0 N. The result shows good linearity and repeatability of FBG_A, with a maximum hysteresis value of 9.65% (Fig. 14a).

During the hysteresis test for FBG_B, only the steering force was changed by turning the y-axis knob. The force was loaded from -3.5 N to 3 N and then unloaded from 3 N to -3.5 N. The

result shows good linearity and repeatability of FBG_B , with a relatively larger maximum hysteresis value of 14.12% (Fig. 14b). The hysteresis problem of FBG_B may be caused by the bonding condition between epoxy and fiber B.

During the hysteresis test for FBG_{C1} , only the lifting force was changed by turning the z-axis knob. The force was loaded from -1.5 N to 3 N and then unloaded from 3 N to -1.5 N. The result shows good linearity and repeatability of FBG_{C1} , with a maximum hysteresis value of 5.99% (Fig. 14c).

Note that all force calibrations and force hysteresis studies were executed at a constant laboratory temperature ($\sim 25^\circ\text{C}$). The difference in the applied force between F_y and F_z is caused by the movement limit of knobs in the XYZ linear stage. The variance of hysteresis and errors between two force sensors can be caused by different epoxy distribution/amount, epoxy lifespan, pre-treatment of the tube and fiber, fiber recoating condition, and curing conditions. The epoxy amount affects both the strain transfer efficiency from tube to FBG strain sensor and the reflectivity of the light.

C. Temperature Compensation Tests

1) Experimental Setup

To take temperature effect into account, we design a temperature calibration test to obtain the temperature sensitivities $K_{T,A}$, $K_{T,B}$, $K_{T,C1}$, and $K_{T,C2}$ of the force sensor. In the temperature calibration setup (Fig. 15), the force sensor is immersed in the water in a glass that sits on a heater plate. The heater can heat the water from room temperature 25°C to 50°C . After heating, the water is left to cool down by itself from 50°C to room temperature 25°C . The temperature is measured through a digital thermometer, and the interrogator captures the corresponding wavelength shifts of FBG_A , FBG_B , FBG_{C1} , and FBG_{C2} .

Figure 15. Temperature Calibration Setup.

Figure 16. Temperature Compensation Setup.

A temperature compensation test is designed to verify the temperature compensation modeling in Section III.V2. In the platform (Fig. 16), the proximal end of the assembled grasper is fixed by a fixture, and the rest of the assembled grasper lies

along with a slot of a constraint part. The grasper tip is wound with a tendon that connects to a constant weight (1N) at the distal side, which implies that the force sensor only experiences a constant pulling force in this case. A heat gun is used to increase the ambient temperature around the force sensor (that sits inside the constraint part) from room temperature. During the heating process, all the wavelength shifts for FBG_A , FBG_B , FBG_{C1} , and FBG_{C2} are captured by the interrogator.

2) Experimental Results

The obtained temperature sensitivities $K_{T,A}$, $K_{T,B}$, $K_{T,C1}$, and $K_{T,C2}$ are $31\text{pm}/^\circ\text{C}$, $27\text{pm}/^\circ\text{C}$, $30\text{pm}/^\circ\text{C}$, $11\text{pm}/^\circ\text{C}$, with a R^2 of 0.996, 0.985, 0.997, 0.998, respectively. The force sensor has good linearity between temperature changes and wavelength shifts without any obvious temperature hysteresis (Fig. 17), which agrees with (15) to (18). The difference in temperature sensitivity of four FBGs may be owing to the different volume/distribution of epoxy applied in the manual adhesion process.

Figure 17. Temperature calibration results: (a) FBG_A ; (b) FBG_B ; (c) FBG_{C1} and FBG_{C2} . (+: temperature increasing; -: temperature decreasing; LF is short for linear fitting).

Figure 18. Temperature compensation result (TC is short for temperature compensation).

After obtaining temperature sensitivity ratios $r_{T,A}$, $r_{T,B}$ and $r_{T,C1}$ from the experiment, we used (23) for FBG_A , FBG_B , and FBG_{C1} in the temperature compensation test. In the result (right y-axis in Fig. 18), the force sensor experienced a

temperature change of 3.7 °C, measured by the FBG_{C2}. Without the temperature compensation feature, the measured force changed from 1N to 6.93 N for FBG_A, from 1N to 7.27 N for FBG_B, and from 1N to 4.52 N for FBG_{C1} (left y-axis in Fig. 18). With the temperature compensation, all measured forces from the force sensor fluctuated around the weight value of 1N, with an RMSE of 0.204 N for FBG_A, 0.292 N for FBG_B, and 0.197 N for FBG_{C1}. After the temperature compensation, the errors are mainly from the pulling force measurement errors for FBG_A, FBG_B, and FBG_{C1} (0.313N, 0.468N, and 0.193N from Section IV.A2).

V. PERFORMANCE DEMONSTRATIONS

A. Force Comparison Study

After force calibrations, another three runs of tests were conducted on the force sensor using the same platform shown in Fig. 8 to compare the measured force from the force sensor and the reference force from the *Nano17* sensor. For the pulling case, screws were tightened to eliminate the clearance between the grasper jaw and the hole in the supporting part. The measured force was very close to the reference force, especially for the pulling force and lifting force, reflecting that the force sensor has high repeatability and can maintain its accuracy after multiple usages.

Figure 19. Force measurement comparison results.

When the pulling force (0 N to 6 N) was acted on the grasper, the RMSE between the measured force and the reference force was 0.270 N, where the percentage error was 4.50% (Fig. 19a). When the steering force (-4 N to 2 N) was exerted on the grasper (i.e., the grasper was steered to the left and then the right), the RMSE was 0.371 N, where the percentage error was 6.18% (Fig. 19b). When the lifting force (-1.3 N to 3.9 N) was acted on the grasper, the RMSE was 0.256 N, where the percentage error was 4.93% (Fig. 19c). In the force comparison study, the percentage error of the steering force was slightly larger than

the other 2-DoF forces, which was caused by the more significant hysteresis of FBG_B (see Section IV.B).

B. Ex-vivo Tests to Manipulate Colon Tissue

In the ex-vivo test, a piece of pig colon tissue was located below the endoscope. Two screws crimped one end of tissue on a mounting structure, and the other end was gripped by the sensorized grasper, which was inserted into a channel ($\varnothing$ 5mm) attached to an endoscope *Olympus Evis Exera* (Fig.20a). A haptic device *Omega7* was used to teleoperate the grasper. We teleoperated the grasper to pull, steer and lift the hanging tissue (Fig.20b). At the meantime, real-time force readings of F_x , F_y , F_z were displayed to the operator on the computer screen and saved after the test. The direction definition of three-axial forces acting on the grasper is shown in Fig. 20c.

Figure 20. Ex-vivo test about colon tissue manipulation: (a) Gripping tissue; (b) Manipulating tissue; (c) Three-axial forces' directions.

Figure 21. Ex-vivo test results: (a) pulling the pig colon tissue; (b) steering the pig colon tissue; (c) lifting the pig colon tissue.

Since the tissue was below the grasper, when the grasper gripped and pulled it along the x-axis, the grasper experienced a forward force $+F_x$ and a downward force $-F_z$. In the result (Fig.21a), the pulling force F_x was up to 0.859 N, while lifting force $|F_z|$ was up to 1.397 N. Small horizontal force to the right $-F_y$ up to -0.113 N was observed in this scenario. It is because the rest of the tissue was placed a little bit to the right of the

grasper, when it was pulled in tension, it exerted a right force to the grasper.

When the grasper gripped the tissue in tension and then steered it in the x-y plane, the grasper experienced a forward force $+F_x$ and a horizontal force $\pm F_y$. In the result (Fig.21b), the pulling force F_x was up to 0.547 N, while steering force F_y was varying from -0.477 N to 0.401 N. Small variance in the vertical force F_z from -0.268 N to 0.103 N was observed in this scenario because of the small undesired movement of the operator's hand in the vertical plane.

Similarly, when the grasper gripped the tissue in tension and then lifted it in the x-z plane, the grasper experienced a forward force $+F_x$ and a downward force $-F_z$. In the result (Fig.21c), the pulling force F_x was up to 0.442 N, while lifting force F_z was varying up to -1.015 N. Small variance in the horizontal force F_y from -0.021 N to 0.142 N was observed in this scenario because of the small undesired movement of the operator's hand in the horizontal plane.

Note that all the measured forces during the ex-vivo tests are within the range shown in [11, 32, 34]. The force behavior patterns are maintained during three pulling actions, two cycles of steering motions, and four lifting actions, which indicates the repeatability and stability of the force sensor.

In total, more than 40 runs of tests (total time > 380 minutes) were conducted on the force sensor, including calibrations, hysteresis study, the force comparison, and the haptic display study. The lifetime of the force sensor is considered sufficient to cover general surgical procedures.

VI. CONCLUSION

We proposed a new three-axial force sensor in a flexible endoscopic surgical robot. Three FBG optical fibers are attached to the sensing structure using a biocompatible epoxy. The calibrated force sensor can measure a pulling force F_x from 0 N to 6 N, a steering force F_y from -3.5 N to 3 N, and a lifting force F_z from -4.5 N to 4.5 N, which sufficiently covers the tool-tissue force range in general surgical operations [11, 32, 34]. The pulling, steering and lifting force sensitivities are 11.441 pm/N, 1757.438 pm/N, and 838.386 pm/N. The steering force resolution is 0.57 mN, and the lifting force resolution is 1.19 mN, which are better than/comparable with the resolutions in the existing literature [11, 33, 35]. The force sensor has force hysteresis of 9.65% on the x-axis, 14.12% on the y-axis, and 5.99% on the z-axis. Also, the force sensor can compensate for the temperature effect, with a maximum root-mean-square-error of 0.292 N. In the performance comparison study, its measurement errors are from 4.50% to 6.18%. Also, this sensor was applied in a flexible endoscopic surgical robot to manipulate a piece of pig colon tissue, and its effectiveness has been demonstrated in ex-vivo tests.

The small and biocompatible force sensor is seamlessly integrated with the flexible endoscopic robot. The articulated arm can be easily replaced/repared at the distal side when worn out, without any reinstallation of the force sensor. Besides, this high-sensitive device can compensate for the temperature effect during the real-time force measurement. Apart from endoscopic robots, the standalone force sensor can also be integrated with

other surgical robots such as laparoscopic robots and catheters.

Although the force sensor presents high force resolutions and force sensitivities in the y-axis and the z-axis, it has a low force sensitivity on the x-axis. It is advised to improve the force sensitivity in the x-axis, either by modifying the design of the shaft in the force sensor or changing the material of the force sensor with a lower young's modulus.

In the future, we will display measured three-axial forces directly to the haptic devices so that the surgeon can feel the real-time tool-tissue interaction forces during suturing [40]. Our previous single-axis force sensor [38] will be implemented together to measure the gripping force, achieving 4-DoF force measurements. An outer tube or corrosion-resistant coating may be applied outside the sensor to improve its robustness and maintain its performance in a harsh environment (such as blood and liquid during a real surgery). Also, the manual fabrication process can be improved to an automated one to secure the optical fibers' precise locations on the sensor structure, and achieve uniform distribution of the epoxy.

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