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Force Sensing with 1mm Fiber Bragg Gratings for Flexible Endoscopic Surgical Robots

Wenjie Lai¹, Lin Cao^{1*}, Rex Xiao Tan², Phuoc Thien Phan³, Jianzhong Hao⁴, Swee Chuan Tjin², Soo Jay Phee¹

Abstract— This paper presents a novel force sensor to detect the distal force of tendon sheath mechanisms (TSMs) in flexible endoscopic surgical robots. We propose to measure the compression force on the sheath at the distal end so that the tension force on the tendon, which equals the compression force on the sheath, can be obtained. With this approach, a new force sensor made up of a 1mm Fiber Bragg Grating (FBG) attached to a 3mm long nitinol tube was developed to measure the compression force exerted on the sheath. Mechanics analysis and verification tests were conducted to characterize the relationship between tension and compression on a TSM. Force calibrations, hysteresis study, and temperature compensation verification tests on the sensor were carried out. The force sensor has a measurement error of 0.178 N and a sensitivity of 34.14 pm/N. Applications of the sensor in a TSM-driven robotic grasper and a tendon driven continuum robot were demonstrated. This force sensor has salient advantages: it is small, structurally simple, electrically passive, temperature-compensated, easy to assemble and disassemble, flexible, and biocompatible. This proposed approach with the new force sensor can also be applied to both TSM-driven systems and tendon-driven systems such as robotic fingers/hands, wearable devices, surgical catheters, and rehabilitation devices.

Index Terms— Haptic Force Sensing, Fiber Bragg Gratings, Tendon-Sheath Mechanisms, Endoscopic Surgical Robots.

I. INTRODUCTION

ROBOTIC surgical systems for Minimally Invasive Surgery (MIS) and Natural Orifice Transluminal Endoscopic Surgery (NOTES) are becoming increasingly popular due to promising advantages such as improved precision, better vision, enhanced dexterity, faster recovery, and better postoperative appearance [1-3]. However, due to space constraints for sensors, haptic feedback is often absent in the existing robotic surgical systems. Achieving accurate force feedback is a critical challenge for surgical robots [4-8], especially for flexible endoscopic surgical robots whose transmission systems are Tendon-Sheath Mechanisms (TSMs) [2, 9] with highly nonlinear and shape-dependent force hysteresis.

There are two ways to obtain force information at the distal ends of a surgical robot: 1) to estimate the distal force based on the force measured at the proximal end of the robot through mathematical models; 2) to directly mount force sensors at the

distal portions of the instruments.

For flexible endoscopic robots, modeling the force transmission of TSMs is challenging because the shape and/or configuration of the TSMs is unknown and varying during surgical procedures [10, 11]. Kaneko *et al.* analyzed the tension transmission in TSMs using the Coulomb friction model [12], with constant curvature and fixed pretension. This became the foundation of the subsequent studies on TSM [13, 14]. Wang *et al.* proposed a model to predict the tissue force reaction of a flexible surgical endoscopic robot, which outperformed the modeling methods, including the Voigt model and Kelvin model [15]. Do *et al.* developed a dynamic friction model to fit and predict nonlinear hysteresis behaviors of TSMs [10], with a mean square error of 0.027N and a peak-to-peak error of 1N. Li *et al.* used deep learning method to predict distal end force in TSMs, with a root mean square error of 0.0219N [16]. Apart from modelling on TSMs, researchers have also studied the force transmission of flexible endoscopes [6] and catheters [17] by considering the complex contacts and friction with the environment. These approaches are sensor-less and thus promising for surgical robots where small size and reliability are critical. However, these models assume fixed sheath configuration/shape and thus are not suitable for surgical procedures where the shape of the flexible instruments are unknown and change frequently.

Mounting force sensors directly to the distal portions of a surgical robot are straightforward solutions for obtaining haptic feedback. Various sensing principles have been explored for MIS. These principles are based on current [18], pressure [19], resistance [20, 21], capacitance [22], piezoelectricity [23], vibration [24] and optical properties [25-27]. Among existing force sensors, Microelectromechanical Systems (MEMS) have shown great potential because of their small sizes, good spatial resolution, and capabilities of multi-axis detections [28]. Commonly, when force sensors are directly embedded in a surgical instrument, a dedicated housing structure for that sensor is required. This additional structure occupies space in the instrument and also leads to tedious assembly/disassembly procedures and complex calibration processes [29]. Furthermore, maintaining small sizes, ease of sterilization, and

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¹W. Lai, L. Cao, and S. J. Phee are with Robotics Research Center, School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore. (e-mail: laiw0021@e.ntu.edu.sg, lin.cao@ntu.edu.sg, MSJPhee@ntu.edu.sg)

²R. X. Tan and S. C. Tjin are with Photonics Institute, School of Electrical and Electronic Engineering, Nanyang Technological University, Singapore. (e-mail: xtan024@e.ntu.edu.sg, ESCTJIN@ntu.edu.sg)

³P. T. Phan is with Graduate School of Biomedical Engineering, University of New South Wales (UNSW), Sydney, Australia. (e-mail: p.phan@student.unsw.edu.au)

⁴J. Hao is with Institute for Infocomm Research (PR), Agency for Science, Technology and Research (A*STAR), Singapore. (e-mail: haoemily@i2r.a-star.edu.sg)

robustness are still common problems faced by force sensors in surgical robots.

There is an increasing trend of applying Fiber Bragg Grating (FBG) sensors in surgical robots [29-33], because of their small sizes, high sensitivity, biocompatibility, immunity to electromagnetic noise, and electrical passivity. Compared to intensity-based and phase-modulated optical fiber sensors, FBG sensors are structurally simple and is not limited by phase discontinuity or light source intensity fluctuations [34]. Li *et al.* developed a tactile sensor based on FBG array for tissue palpation [30]. They also utilized FBGs to detect distal force for cardiac ablation [31]. Zarrin *et al.* embedded 3mm-grating and 8mm-grating FBG sensors in a robotic MIS gripper to measure grasping force and axial force [29]. Iordachita *et al.* employed FBG sensors for 2 Degrees-of-Freedom (DOF) force sensing in retinal surgery [32]. A 3-DOF force sensing pick instrument ($\phi 0.8$ mm) was further developed for vitreoretinal surgery, where four FBG sensors were used to decouple axial and transverse forces [33]. Apart from force sensing, FBG has been applied in shape sensing [35]. Park *et al.* introduced a biopsy needle integrated with FBGs to detect its bending deflection when inserted into the tissue [36]. Sefati *et al.* developed an FBG array for real-time large deflection shape sensing of a continuum dexterous manipulator [37].

For force sensing in a flexible surgical robot, space constraints and rigid length of the instrument become critical factors for the design of sensors. FBG sensors are promising solutions for endoscopic surgical robots due to their small size and inherent flexibility. However, FBG sensors do have some inherent limitations. FBG force sensors are subject to thermal cross-sensitivity, which, can be overcome with an additional strain-free FBG for temperature compensation. Optical fibers are fragile, but they can be protected by coating and by encasing the whole instrument with an outer protective tube. Although the signal interrogation equipment (interrogator or optical spectrum analyzer) is considerably bulky, it could be located far away from the tip of the robot without degradation of performance. The transmission loss of the optical fiber in such applications is negligible given the short fiber length (two to three meters). With these considerations, FBGs were determined to be a suitable sensing technique for the proposed force sensor in this work.

For distal end haptic sensing of TSMs, in this paper, for the first time, we propose to measure compression force on the sheath at the distal end so that the tension force on the tendon, which equals the compression force on the sheath (theoretically and experimentally confirmed in Section II.A), can be obtained. This new force sensor¹ is fabricated by bonding a 1mm FBG on to a 3mm-long nitinol tube using a biocompatible epoxy (EPO-TEK 353ND). The nitinol tube extends from the distal end of the sheath, and the tendon goes through the sheath and the nitinol tube. Pulling the tendon will compress the sheath as well as the nitinol tube whose axial deformation is captured by the FBG sensor. The wavelength shift of the FBG sensor reflects

the compression force on the nitinol tube and the sheath, which equals the tension force (magnitude) on the tendon. A secondary FBG is placed close to the force sensor for temperature compensation.

The proposed force sensing principle and force sensor presents five advantages: 1) total force applied to the end-effector by the transmission mechanism is measured, whereas other force sensors that are often directly mounted on the end-effector (such as grasper jaws) face challenges of force decoupling (to decouple axial forces and transverse forces) [29], design complexity, and higher risk of being damaged due to their exposure to tissue or the working environment; 2) as a standalone component independent of the instrument's structure, the proposed sensor can be calibrated before installation while other sensors [25, 29, 33] must be calibrated together with the instruments. This feature enables straightforward calibration and convenient replacement of sensors on the instrument; 3) the force on the sheath rather than the force on the tendon was measured, avoiding the need of decoupling the elongation and the bending/movement of the tendon; 4) the sensor can be seamlessly integrated with TSMs due to its similar diameter and hollow configuration with the sheath; 5) FBG optical fibers are flexible and commercially available.

This paper is an extension of a conference paper [38]. Additional contributions presented in this paper are as follows: 1) the length of the force sensor was reduced from 6mm to 3mm; 2) the length of the grating was reduced from 4mm to 1mm; 3) the sensitivity of the force sensor was increased from 24.28pm/N to 34.14pm/N; 4) force hysteresis and the temperature compensation studies were conducted; 5) the force sensor was demonstrated in a TSM-driven grasper and a tendon-driven continuum robot.

This paper is organized as follows. Section II introduces the methodology, including the mechanics of TSMs, the working principle of FBG, design and fabrication of the force sensor, and the integration of the sensor with a TSM-driven grasper and a tendon-driven continuum robot. Section III describes the experiments and results, including calibrations, hysteresis study, temperature compensation verification tests, and demonstrations on the TSM-driven surgical grasper and tendon-driven continuum robot. Section IV concludes the work.

II. METHODOLOGY

A. Mechanics of TSMs

1) Force Analysis of Tendon-sheath Transmission System

Fig. 1a is a schematic diagram of TSMs integrated with a surgical grasper driven by motors. Fig. 1b shows an infinitesimal segment of a tendon and sheath with constant curvature. C is the compression force on the sheath; T is the tension force on the tendon; f is the friction force applied by the sheath to the tendon; N is the normal force from the sheath to the tendon; f' is the friction force applied by the tendon to the sheath; N' is the normal force from the tendon to the sheath; μ

¹ A PCT patent application (PCT/SG2019/050271) has been filed for this force sensor in May 2019.

is the friction coefficient between the tendon and sheath; α is the curve angle of this configuration; r is the radius of the curve in tendon-sheath configuration.

Figure 1. (a) Schematic diagram of TSMs integrated with a surgical robot. (b) An infinitesimal segment of TSMs in section view (adapted from [39]).

For the tendon, by applying the force equilibrium equations for a small portion dx , with a corresponding angle $d\alpha$, we obtain the following set of four equations [12].

$$T d\alpha = -N, d\alpha = dx/r, f = \mu N \text{ and } dT = f, \quad (1-4)$$

where the loss of tension force is due to the friction [12, 40] between the tendon and the sheath.

Similar relationships also exist for the sheath:

$$C d\alpha = -N', d\alpha = dx/r, f' = \mu N' \text{ and } dC = f'. \quad (5-8)$$

Based on Newton's Third Law,

$$N = -N', f = -f'. \quad (9-10)$$

Thus,

$$T d\alpha = -C d\alpha. \quad (11)$$

The crucial fundamental equations for this work are then obtained as:

$$T = -C; dT = -dC. \quad (12-13)$$

This means that the magnitude of the compression on the sheath equals that of the tension on the tendon at the same cross-section. This relationship was further experimentally verified in the following subsection.

2) $T = -C$ Verification Experiment

Figure 2. (a) Schematic diagram of the experimental setup for $T = -C$ test. (b) Simplified diagram of the experimental setup for $T = -C$ test

A test was conducted to verify Eq. (12). Fig. 2a illustrates the platform for the test. Fig. 2b shows a schematic diagram of the setup. Four hollow compression load cells (denoted as ' L ') were utilized to measure the compression force on the sheath and the tension force on the tendon at both the proximal end and the distal end. Given that the load cells are only capable of

measuring compression, the tension on the tendon was converted to compression using a linear slider. A spring was affixed to the tendon at the distal end and a fixture, and a motor was employed to pull the tendon at the proximal end. The flexible configuration of the sheath and tendon was arbitrarily set and fixed by pins onto the foam.

A sinewave signal, with a frequency of 0.3 Hz and a converted motion amplitude of $\pm 90^\circ$, was sent to the motor (with a $\varnothing 25\text{mm}$ pulley) to drive the tendon. The load cells were used to measure the associated forces. Root Mean Square Error (RMSE) between the tension force and the compression force was 0.618N for the proximal end and 0.332N for the distal end. As shown in Fig. 2b, the tendon passes through the four load cells. Due to the limitation of the experimental setup, the tendon could not be precisely aligned in parallel to the measuring axes of the load cells. The small angle of misalignment may result in a decomposition of the transmitted force, leading to the above errors. In addition, the load cells used in the experiment have a resolution of 0.1N, which may also contribute to the errors.

Figure 3. $T = -C$ test result at the proximal end and the distal end

As shown in Fig. 3, the experimental results verified that the compression force (C) of the sheath has almost the same magnitude as the tension force (T) of the tendon at both the distal end and the proximal end, respectively. Thus, we propose to measure the compression on the sheath at the distal end using an FBG-based force sensor so that the tension on the tendon at the distal end can be obtained.

B. Working Principle of FBG

Figure 4. FBG working principle

FBGs are distributed Bragg reflectors constructed in a short

segment of an optical fiber through periodic modulation of fiber core refractive index. The grating period of an FBG can be customized to reflect a specific spectral band of light. As shown in Fig. 4, this reflected spectrum shifts in response to variations of temperature and/or strain due to the physical deformation of the host fiber. The central wavelength of the FBG reflection band is called the Bragg wavelength λ_B , given by the Bragg equation [41]:

$$\lambda_B = 2n_{eff}\Lambda, \quad (14)$$

where n_{eff} is the effective refractive index of the fundamental guided mode for the optical fiber and Λ is the pitch of refractive index modulation. Consequently, Λ of an FBG is dependent on temperature and strain that the host fiber experiences. Changes of n_{eff} due to strain and temperature are negligible; thus, n_{eff} can be usually considered as a constant. The wavelength shift with respect to strain and temperature variations is as follows [42]:

$$\frac{\Delta\lambda_B}{\lambda_B} = K_T\Delta T + K_S\Delta\varepsilon, \quad (15)$$

where ΔT , $\Delta\varepsilon$, K_T , and K_S are the temperature change, axial strain change along the fiber, thermal coefficient, and strain coefficient of the fiber material, respectively.

Figure 5. Diagram for dual FBG temperature compensation technique

In Fig. 5, a force F is applied to a testing object bonded with FBG2 while a strain-free FBG1 is added to offset FBG2's cross-sensitivity to temperature. Equation (16) is obtained by using the dual-FBG temperature compensation technique [43].

$$\Delta\varepsilon_2 = c_2\Delta\lambda_{B2} - c_1\Delta\lambda_{B1}, \quad (16)$$

where $\Delta\varepsilon_2$ is the strain change in FBG2; λ_{B1} and λ_{B2} are the central wavelengths of FBG1 and FBG2, respectively; Here, $c_1 = (K_{T2}\lambda_{B2}) / [(K_{T1}\lambda_{B1})(K_{S2}\lambda_{B2})]$ and $c_2 = 1 / (K_{S2}\lambda_{B2})$. $K_{T1}' = K_{T1}\lambda_{B1}$ and $K_{T2}' = K_{T2}\lambda_{B2}$ are the temperature sensitivities (unit: pm/°C) of FBG1 and FBG2, which can be obtained through calibrations by subjecting the FBGs to heating, such as immersing both FBGs in a water bath heated by a hotplate. The temperature sensitivity ratio of FBG2 to FBG1 is a constant: $r = K_{T2}' / K_{T1}'$.

Thus, the strain change in FBG1 has a proportional relationship with $(\Delta\lambda_{B2} - r\Delta\lambda_{B1})$, as shown below:

$$\Delta\varepsilon_2 \propto (\Delta\lambda_{B2} - r\Delta\lambda_{B1}) \quad (17)$$

When the strain is in the elastic range of the testing material, the change of applied force ΔF is proportional to the strain change in the testing object $\Delta\varepsilon_{ob}$. The strain transfer efficiency between the testing object and FBG2 is $\eta = \Delta\varepsilon_2 / \Delta\varepsilon_{ob}$. Therefore, there is a linear relationship between the change of applied force and $(\Delta\lambda_{B2} - r\Delta\lambda_{B1})$, as shown in (18).

$$\Delta F \propto (\Delta\lambda_{B2} - r\Delta\lambda_{B1}) \quad (18)$$

As a result, temperature influence can be effectively offset from the measurement of the loading force by tracking the variance of Bragg wavelength shifts.

In this work, Bragg wavelength shifts $\Delta\lambda_{B1}$ and $\Delta\lambda_{B2}$ were interrogated through a Micron Optics Interrogator SM130, and the ratio r was obtained through temperature calibration in Section III. B. The interrogator has the wavelength repeatability of 1 pm, and wavelength stability of 2 pm typically and 5 pm as a maximum. Information of the components used in this work is provided in the Appendix.

C. Design and Fabrication of the Force Sensor

Fig. 6a illustrates the design of the force sensor. This sensor consists of a 3mm-long nitinol tube and a 1mm-grating FBG fiber, bonded by epoxy.

Figure 6. (a) A simplified design of the force sensor. (b) The flat surface on the nitinol tube. (c) The fabricated force sensor. (d) Microscope photograph of the force sensor.

Two force sensors were fabricated in this work. The fabrication includes three stages: (1) pre-processing: to provide more contact area between the fiber and the tube, the outside surface of the nitinol tube was ground flat, as shown in Fig. 6b. The fiber was cleaned using a lint-free wipe and precision cleaning fluid (from MicroCare Sticklers). Similarly, the tube was cleaned using the wipe; (2) adhesion: the tube was fixed by tweezers. One syringe with a Ø0.3mm needle was employed to apply a small amount of epoxy on the surface of the tube. Translation stages were adjusted to align the fiber to the tube (with epoxy) and straighten the fiber with a pretension. More epoxy was applied, affixing the fiber to the tube when the tube and fiber were well aligned. The epoxy was cured using a heat gun; (3) post-processing: after the epoxy was cured, the extra end of the fiber was cut at a slightly inclined angle (within 10°) to reduce back coupling of light at the fibre end. A fabricated force sensor is presented in Fig. 6c. A microscope photograph of the force sensor is shown in Fig. 6d.

To ensure strong bonding, it is crucial to choose suitable epoxy for attaching the FBG fiber to the nitinol tube. Based on trials, common epoxy designed for metal like Selleys' Super Strong / Super Steel Epoxy, Araldite's 90min Standard Epoxy, and Elmer's Super-Fast Epoxy did not grip well for the nitinol tube. Epoxy with weak bonding can be easily peeled off and may introduce considerable noise and creeping effect (wavelength shift drops even when the object is under a constant load) to the sensor, as illustrated in Fig. 7a and Fig. 7b. After several trials, a stable and biocompatible epoxy, EPO-

TEK 353ND, was chosen. The epoxy EPO-TEK 353ND, complied with USP (U.S. Pharmacopeial Convention) Class VI biocompatibility standards, is found to be able to strongly bond the optical fiber to nitinol. A color change from amber to dark red (Fig. 6d) was observed on the epoxy after curing. No creeping occurred when force calibration was conducted with the EPO-TEK 353ND, as shown in Fig. 7c. The FBGs are commercial products from Samyon, Beijing, China. They were fabricated using a phase mask based FBG fabrication setup, followed by polyimide recoating.

Figure 7. (a) Force sensor with improper epoxy would easily peel off. (b) Creeping occurs when unsuitable epoxy is applied. (c) No creeping occurs when suitable epoxy is used.

D. Force Sensor Integration in Flexible Robots

This proposed force sensor is designed to be utilized in flexible surgical robots. This section describes its integration with a TSM-driven grasper and a tendon-driven continuum robot.

1) Integration with a Tendon-Sheath Driven Grasper

Fig. 8a shows a prototype of this 3mm-long force sensor with a TSM. A nitinol tube was placed at the distal end of the coiled sheath, and the FBG sensor adhered to the outer surface of the nitinol tube by epoxy. The tendon goes through both the sheath and the nitinol tube. Fig. 8b depicts the force analysis of the force sensor and a TSM in a sectional view. When integrated with the end-effector, the “m” end of the nitinol tube is fixed to the end-effector. Pulling the tendon at the proximal end causes a compression force from the sheath to the nitinol tube. This compression causes a strain on the FBG strain sensor, resulting in a corresponding Bragg wavelength shift. Based on the Bragg wavelength shift, the compression force on the sheath at the distal end can be calculated. Thus, the tension force on the tendon at the distal extremity can be obtained.

A sectional view of the force sensor integration in a TSM-driven grasper is presented in Fig. 8c. The grasper consists of two pivotally connected jaws, and two TSMs are involved in controlling this DOF. The lower jaw cannot be rotated and thus serves as the body structure of the grasper, while the upper jaw can be bi-directionally rotated by the two tendons which are

fixed to the jaw via a crimping bead. Each tendon extends from the upper jaw to its associated force sensor and sheath. The force sensor measures the compression force on its associated sheath, and thus the tension force on the associated tendon can be obtained. Fig. 8d shows the integrated grasper model with both fibers and TSMs, which was covered and protected by a flexible outer tube. The grasper was 3D printed using stainless steel 316L through Direct Metal Laser Sintering. Fig. 8f depicts a 3D model of the FBG-integrated grasper inside an endoscope.

Figure 8. (a) A prototype of the force sensor with a TSM. (b) Force analysis between the force sensor and TSM. (c) Sectional view of the integrated grasper. (d) Transparent 3D Model of the integrated grasper. (e) Sizes of the grasper. (f) The force-sensor-integrated grasper sits inside an endoscope.

The force sensor, made of a 1mm-grating FBG fiber ($\varnothing 0.245$ mm) and a 3mm-long nitinol tube (outer diameter or OD: 1.27 mm; inner diameter or ID: 0.97 mm), was then inserted into a $\varnothing 2$ mm step hole on the body structure ($\varnothing 4.4$ mm) of the grasper (Fig. 8e). The sheath (OD: 1.189 mm, ID: 0.72 mm) directly connects with the nitinol tube, and a 1mm-long segment of the sheath stays inside the step hole to ensure good contact with the tube. A tendon ($\varnothing 0.5$ mm) passes through both the tube and the sheath. A dual FBG array fiber is adopted in this work to offset FBG cross-sensitivity to the temperature variance. The first 1mm grating is near the fiber tip, bonded to the nitinol tube for strain sensing; the second 1mm-grating is 10mm away from the first FBG and is for temperature sensing (not bonded to anything).

Constraints on the diameters and rigid lengths of components, including sensors, are critical in a flexible surgical robot. This force sensor was designed to sit in a small rigid portion of the instrument (Fig. 8c and Fig. 8d). Increasing the length of the grating would increase the rigid lengths of the nitinol tube and the instrument, which would reduce the

flexibility and maneuverability of the instrument in the tortuous tool channel of the endoscope. Thus, a short, 1mm grating was chosen in this work.

In this work, the nitinol tube was selected because of its superelasticity and biocompatibility. Compared to other biocompatible metal materials such as stainless steel 304/316, with the same dimensions and loads, the strain of the nitinol tube is much larger, and consequently, increases the sensitivity of the force sensor. The superelasticity of nitinol also provides the sensor with a larger strain limit and longer fatigue life. In addition, both the nitinol tube and the sheath are hollow and have similar dimensions. Hence, the integration of the sensor with the grasper and the TSM can be compact. The wall thickness of the nitinol tube was chosen to be 0.15mm to provide a sufficient contact area between the nitinol tube and sheath for distributed force transmission.

2) Integration with a Tendon-Driven Continuum Robot

The force sensor can also be used for tendon-driven robots [44, 45]. Fig. 9a depicts a tendon-driven 2-DOF continuum robot, with two channels (OD: 4 mm) capable of inserting surgical tools. This continuum robot is made of 26 disks (thickness: 2mm, OD: 16 mm), a 20cm-long super-elastic spine (a nitinol tube with an OD of 1.59 mm), four tendons (OD: 0.5 mm), and a stand for the fixture. The four tendons drive the continuum robot's up/down and left/right motions. All the tendons are terminated by crimping beads at the distal end. One force sensor is integrated with the robot for distal force sensing on a tendon. The force sensor is compressed by the last disk and stopper. Slots at the edge of disks allow the FBG fiber (highlighted in blue) to pass through the robot smoothly.

Figure 9. (a) Force sensor integrated with a tendon-driven continuum robot. (b) Force analysis of force sensor, last disk, and stopper of continuum robot.

Fig. 9b depicts relevant force analysis between the force sensor, the last disk, and the stopper of this continuum robot. When the tendon is pulled, the nitinol tube between the last disk and the stopper is compressed. The strain change due to the compression force on the nitinol tube will be measured by the FBG, and hence the tension force on the tendon at the distal extremity can be obtained.

III. EXPERIMENTS AND RESULTS

A. Force Sensor Calibrations and Hysteresis Study

Fig. 10a illustrates the experimental setup to study the sensitivities and hysteresis of the force sensors. In each lane, a motor was used to pull the tendon which passed through a

sheath stopper, the TSM and terminates at a load cell holder at the distal end. The distal side of the tendon was connected to a spring with constant stiffness of 0.52N/mm. The force sensor, along with the sheath, was inserted into the step hole of a customized stud inside the load cell holder. A detailed depiction of the load cell holder is shown in Fig. 10b, and relevant force analysis in Fig. 10c. When the tendon was pulled in tension by the motor, both the sheath and the attached force sensor were compressed; the compression force transferred to the stud was captured by the hollow load cell at the same time. The magnitude of compression force measured by the load cell is equal to the compression force detected by the force sensor, which also reflects the tension force of the tendon as given in Eq. (12). Compression forces from 0 N to 25N were applied in each lane, and the corresponding Bragg wavelength shifts of the sensors were recorded. Considering the arm ratio or mechanical advantage from the distal tension force to the reaction force on tissue, the calibration force range, 0-25N, sufficiently covers the normal distal force range of surgical robots, which is generally within 3N [25, 46]. With the recorded compression forces and Bragg wavelength shifts, the FBG force sensitivity (unit of pm/N) was obtained. The calibration tests were run four times for the force sensors to ensure repeatability. To minimize temperature cross-sensitivity, the calibration experiments were carried out at constant laboratory temperature.

Figure 10. (a) Platform setup for sensor calibrations. (b) Detail inside the load cell holder. (c) Force analysis inside the load cell holder

The calibration results for the two force sensors are shown in Fig. 11a and Fig. 11b, where the force sensor 1 is with a single FBG fiber while force sensor 2 is with a dual FBG array. For force sensor 2, FBG1 is strain-free for temperature measurement, and FBG2 is attached to the nitinol tube.

The average sensitivity of force sensor 1 is 34.14pm/N with an R^2 (or R-squared) error of 0.99264, and the average sensitivity of force sensor 2 is 25.17pm/N with an R^2 of 0.96235. When the force sensors were under a constant force or freeloading, the FBG peak wavelength readings from the interrogator fluctuated within a certain range, which resulted in reading errors of the force sensor. According to the experimental results, the wavelength fluctuation range is 6.06pm for force sensor 1 whose sensitivity is 34.14 pm/N, and thus the reading error of force sensor 1 is $6.06 / 34.14 = 0.178\text{N}$. Similarly, the wavelength fluctuation range is 13.32pm for force sensor 2 whose sensitivity is 25.17 pm/N, and then the reading error of force sensor 2 is $13.32 / 25.17 = 0.529\text{N}$. The

variance of sensitivities and errors between two force sensors might be caused by different epoxy distribution/amount, epoxy lifespan, pre-treatment of the tube and fiber, and curing conditions. The applied amount of epoxy affects the strain transfer efficiency from the tube to the FBG strain sensor. Fig. 12a reflects the associated spectrum behavior of the force sensor during calibration.

Figure 11. (a) Calibration result for the force sensor 1. (b) Calibration result for the force sensor 2 (R1 is short for Run1, and so on)

Figure 12. (a) Spectrum behavior of force sensor 1 during calibration. (b) Hysteresis study on force sensor 1 ('+' for loading, '-' for unloading)

Hysteresis study was carried out on the force sensors. Bragg wavelength shifts were recorded under loading (0N to 25N) and unloading (25N to 0N) conditions for both force sensors. The result for force sensor 1 given in Fig. 12b. It presents excellent linearity between the wavelength shift and the applied force, without apparent hysteresis. Further improvement can still be made if uncoated FBGs were used for the sensors, due to the fact that silicon optical fiber glass structure is very resistant to permanent deformation. Hence, any hysteresis if observed for a polymer coated fiber is likely to be due to inelasticity of the coating polymer.

B. Verification Test for Temperature Compensation

Diathermy cutting (cutting using electrically generated heat) may be used during surgery to cut tissue and may increase the

local temperature around the instruments. A dual FBG array was used for the force sensor so that the Bragg wavelength shift due to temperature variations can be compensated. This subsection presents the results of temperature calibration and verification tests of the compensation.

Temperature calibration for the force sensor with a dual FBG array fiber was executed. The force sensor was immersed in a water glass heated by an electrical hotplate shown in Fig. 13. The Bragg wavelengths of the two FBGs were recorded when the water temperature increased from 25°C to 85°C and decreased from 85°C to 25°C. According to the Bragg wavelength shifts in relation to temperature changes, the temperature sensitivities for both FBG1 and FBG2 were obtained, with a unit of pm/°C.

Figure 13. Schematic diagram of temperature calibration test.

The result is shown in Fig. 14, where the temperature sensitivity of FBG1 is 10.61 pm/°C with an R^2 error of 0.9895, and the temperature sensitivity of FBG2 is 21.37 pm/°C with an R^2 of 0.9994. Both FBG1 and FBG2 present good linearity without obvious hysteresis. The temperature sensitivity ratio of FBG2 to FBG1 is $r = 21.37/10.61 = 2.014$. The difference in temperature sensitivity of both FBGs is likely to be due to the volume and distribution of epoxy applied in the manual adhesion process. The volume and distribution of epoxy affect the resulted strain and the strain-transfer (strain is caused by thermal expansion) from the nitinol tube and epoxy to the FBG.

Figure 14. Temperature sensitivity results for force sensor 2

As gas solubility in water decreases with the increase of temperature, formation of air bubbles was observed around 80°C and the bubbles started to attach onto the nitinol tube. The small misalignment of FBG2 linearity at 85°C in Fig. 14 might have been caused by these bubbles.

Upon obtaining the temperature sensitivity ratio r of FBG2 to FBG1, a temperature compensation verification test was

carried out to demonstrate the force sensor's capability of offsetting the temperature effect. The test platform is shown in Fig. 15. In this test, a constant force was applied to the force sensor by hanging a weight to the tendon at the distal end. A heat gun was then used to heat up ambient air around the force sensor.

Figure 15. Schematic diagram of temperature compensation verification test.

The result of the verification test for temperature compensation is presented in Fig 16. The raw data was collected from the interrogator at a frequency of 500 Hz and was post-processed by a second order low pass Butterworth filter. After filtering, the maximum force measured with temperature compensation was 1.83N, and the minimum was 1.58N, which indicates a reading noise of $1.83 - 1.58 = 0.25\text{N}$. The average value of the force with temperature compensation was 1.71N. The difference between the average force and the applied constant force was $1.71 - 1.55 = 0.16\text{N}$, which was within the error (0.529N) of force sensor 2. The noise and error may be caused by both the vibration induced by the rapid flow of air from the heat gun and the reading noise of the interrogator. Without temperature compensation, the force measurement will be significantly different from the real value when the temperature changes. Therefore, the sensor's capability to compensate the influence of temperature changes was demonstrated.

Figure 16. The result of the temperature compensation verification test

C. Demonstrations

1) Grasper Driven by TSMs

To demonstrate the feasibility of the sensor's applications in flexible surgical robots, the force sensor was installed in a TSM-driven grasper. Fig. 17a depicts the platform and Fig. 17b shows a prototype of the grasper. Fig. 17c shows the top view of the platform at the distal end. The relationship between the gripping force of the grasper and the corresponding tension force on the tendon was observed in this test. The force sensors, inserted into the step holes on the body structure of the grasper,

were used to measure the tension force on the tendon. The load cell, connected with the upper jaw of the grasper using a short string, measured the gripping force.

Figure 17. (a) Demonstration platform of a grasper integrated with force sensors, (b) A prototype of the grasper, (c) Top view at the distal end.

Pretension was applied on the tendon to close the jaw, ensuring that both the tendon and string were in tension. The tendon for the opening movement was left loose as it does not affect this investigation. During the demonstration, when the grasper was manipulated to close, the force readings from the load cell and the force sensor 1 (connected with the tendon to close the grasper) were collected in pairs. Two runs of tests were performed, with the gripping force from 0N to 5N.

The result in Fig. 18 revealed a linear force relationship, with a slope of 2.788. With this relationship, once the distal end tension force was measured from the force sensor, the related grasping force can be obtained.

In other works [47, 48], this force sensor was used on a five-DOF robotic grasper for the data collection of the grasping force and lifting force.

Figure 18. Force relationship between distal tension force and grasping force

2) Continuum Robot Driven by Tendons

Another application demonstration of the force sensor on a 2-DOF tendon-driven continuum robot was carried out, as shown in Fig. 19a. In this test, we focused on one DOF (up/down movements) of the continuum robot. One force sensor, located at the end of the top tendon, was utilized to measure the tension force. The top/bottom tendons were kept in tension so that the force sensor was always compressed. When the continuum robot was lifted from the bottom to the top position (Fig. 19b to Fig. 19e), the top tendon would experience

an increasing tension force. At the same time, the compression force was increased on the nitinol tube. The associated Bragg wavelength shifts of the force sensor were recorded as the continuum robot was moving up.

Figure 19. (a) Demonstration schematic diagram of force sensor integrated with a continuum robot, (b) Bottom position; (c) Neutral position; (d) Top position 1; (e) Top position 2.

Fig. 20 shows the result of this test. The measured tension forces on the top tendon at the bottom, neutral, and two top positions were 3.31N, 4.13N, 5.62N, and 6.68N, respectively. Based on this demonstration, the proposed force sensor is confirmed to be suitable for applications in tendon-driven robots to provide the distal force information.

Figure 20. Spectrum Behavior of Continuum Robot Demonstration

IV. CONCLUSION

In this paper, we theoretically derived and experimentally confirmed that the compression force on the sheath and the tension force on the tendon at the same cross-section of a TSM are equal in magnitudes. As such, we propose to measure the distal compression force on the sheath using a new FBG-based force sensor for haptic force sensing in TSM-driven surgical robots. The force sensor consists of a 3mm long nitinol tube and a 1mm-grating FBG fiber. One calibrated force sensor with a single-FBG fiber has a sensitivity of 34.14 pm/N and the other force sensor with a dual-FBG fiber (with temperature compensation feature) has a sensitivity of 25.17pm/N. The force sensors have good linearity without apparent hysteresis. For the temperature-compensated sensor, the effect of temperature variations is compensated with an error of 0.16 N. A suitable epoxy appeared to be critical to such designs as an improper selection of epoxy may lead to a creeping effect and considerable noise to the sensor. After several trials, the biocompatible epoxy EPO-TEK 353ND was chosen as the adhesive. We also demonstrated that the sensor could be used for both TSM-driven robots and tendon-driven robots.

The proposed approach and the new FBG force sensor have

several salient features such as small size, simple structure, high sensitivity and resolution, large force range, temperature compensation, compact integration with TSMs, flexibility, biocompatibility, and ease of fabrication and assembly.

For this study, we examined the distal end force sensing capability of the force sensor in flexible surgical robots. However, the performance of the fiber and FBGs may degrade or be altered during the steam sterilization or autoclaving process. The sterilizability of the proposed sensor needs to be evaluated. In the case of poor sterilizability, the sensor can be made disposable because of its ease in replacement on the instrument. Currently, the fiber is recoated with polyimide to reduce the chance of fiber breakage. With improved precision tools for sensor fabrication and assembly, uncoated FBGs may be applied to increase sensitivity. With the current design, epoxy on the sensor may be slowly scratched off due to the very small movements (i.e., the “disturbance” to the sensor) of nearby sheaths. Hence, the sensor may not sustain its normal sensitivity performance. This robustness and reliability [49] issue may be overcome by using stronger epoxy and by covering the sensor with a thin polymer tube which prevents the direct contact between the sensor and other nearby sheaths.

For future work, the nitinol tube and sheath will be welded into a whole part for secure contact. Interrogators with real-time data acquisition, e.g., Micron Optics Si255, will be used. The sensed forces will be displayed, in real time, to the operator of the surgical robot so that the benefits of haptic feedback in the flexible endoscopic surgery may be studied.

In addition to the flexible endoscopic robot and the continuum robot, this force sensor can also be applied in a variety of either TSMs-driven or tendon driven systems [45], such as robotic fingers/hands [50], wearable devices, surgical catheters, and rehabilitation devices.

APPENDIX

TABLE 1. COMPONENTS LIST

Component	Model	Company	Country
Motor controller	QPIDe	Quanser	Canada
Motor	2642W024CR 354	Faulhaber Schonaich	Germany
Motor driver	30/1S 134:1	Faulhaber Minimotor SA	Swiss
Motor encoder	HEDS-5540 A14		
Motor amplifier	Brush PWM Servo	Advanced Motion Controls	USA
Load cell controller	Uno	Arduino	Italy
Load cell	LW1025-25	Interface	USA
Load cell amplifier	DCA Strain Gage		
Load cell	LTH300	Futek	USA
Load cell amplifier	IAA100		
Spring	6843	Misumi	Japan
Nitinol tube (force sensor)	NI207130	Goodfellow	England
Nitinol tube (continuum robot)	Niti tubing	Memry	German
Epoxy	EPO-TEK 353ND	Epoxy Technology Inc.	USA

Sheath	X0620116	Asahi Intecc	Japan
Tendon	0.5mm, Stainless steel	Asahi Intecc	Japan

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Wenjie Lai graduated with B.Eng (Hons) from Nanyang Technological University (NTU), Singapore in 2015 on an SM3 scholarship, specialized in mechatronics. She is now pursuing her Ph.D. at NTU. She is concurrently a research assistant in Robotics Research Center, MAE, NTU. Ms. Lai's research interests include optical fiber Bragg grating sensors, haptic feedback and position tracking in robotic surgical systems.

Lin Cao obtained his Ph.D. degree in Mechanical Engineering from the University of Saskatchewan, Canada, in 2015. He is currently a Research Fellow in the Robotics Research Center at Nanyang Technological University, Singapore. He was also a visiting scholar at the Department of Precision and Microsystems Engineering at the Delft University of Technology, the Netherlands, in 2013.

Dr. Cao's research interests include surgical robots, soft robots, and compliant systems.

Rex Xiao Tan graduated with B.Eng. (Hons) from Nanyang Technological University (NTU), Singapore in 2014, specialized in intelligent systems and control engineering and recently completed his Joint Ph.D. between the University of Southampton's Optoelectronics Research Centre (UoS-ORC) and Nanyang Technological University's Centre for Optical Fiber Technology (NTU-COFT).

He currently serves as Chief Technology Officer in Aevice Health Pte Ltd, a medical device company span off from NTU, focusing on medical devices and bio-signal processing research and development. Rex Tan's research interests include optical fiber Bragg grating sensors, wearable technologies and medical devices.

Phuoc Thien Phan received the B.Eng. degree in Mechanical Engineering from Vietnam National University, Ho Chi Minh City University of Technology (HCMUT) in 2015.

He is now a Ph.D. student at Graduate School of Biomedical Engineering, University of New South Wales (UNSW) Sydney, Australia.

Mr. Phan's research interests include the development of magnetic soft endoscopic capsule for obesity treatment, and mechanical design and development of medical devices, flexible surgical robotic systems.

Jianzhong Hao received her Ph.D in 2001 from Nanyang Technological University, Singapore, majored in fibre optic sensors.

She is currently a Senior Scientist III at Institute for Infocomm Research (I2R), Agency for Science, Technology, and Research (A*STAR).

Dr. Hao's research interests include fibre optic sensors and applications, sensor packaging, bio-photonics, and system integration. She has published 2 book chapters and 90 over papers in prestigious journals and conferences. She holds 14 patents, with 6 granted and 6 licensed. She is the key inventor of award winning and patent granted technology - FBG-based Perimeter Intrusion Detection System (PIDS).

Swee Chuan Tjin graduated from University of New England with the BSc (Hons) in 1987. He obtained his Ph.D. from the University of Tasmania, Australia in 1992. He is currently the Associate Provost (Graduate Education and Lifelong Learning) at NTU. He is concurrently a Professor of Optics and Photonics in the School of Electrical and Electronic Engineering, the Co-Director of The Photonics Institute (TPI). He was appointed the Founding Chair of the National Research Foundation (NRF) funded, LUX Photonics Consortium.

Professor Tjin's research interests are in fibre optic sensors, biomedical engineering and biophotonics. Over the years, he has published more than 300 refereed journal papers, and has filed 40 patents in fibre optic sensors, biomedical engineering and biophotonics.

Soo Jay Phee graduated from Nanyang Technological University, Singapore (NTU) with the B.Eng (Hons) and M.Eng degrees in 1996 and 1999 respectively. He obtained his Ph.D. from Scuola Superiore Sant'Anna, Pisa, Italy in 2002 on a European Union scholarship.

He is currently the Dean of the College of Engineering at NTU. Prior to his Deanship, he has served 3 years as Chair of the School of Mechanical & Aerospace Engineering. He was the founding CEO of EndoMaster Pte Ltd, a company he co-founded to commercialize a surgical robotic system he developed.

Professor Phee's research interests include Medical Robotics and Mechatronics in Medicine. He is currently a recipient of the prestigious National Research Foundation (NRF) Investigator Award. He was awarded the Young Scientist Award (2006), the Outstanding Young Persons of Singapore Award (2007), the Nanyang Outstanding Young Alumni Award (2011), the President's Technology Award (2012), the Nanyang Innovation and Entrepreneurship Award (2013) and the Nanyang Alumni Achievement Award (2017).