An integrated sensor-model approach for haptic feedback of flexible endoscopic robots

Lai, Wenjie; Cao, Lin; Tan, Rex Xiao; Tan, Yung Chuen; Li, Xiaoguo; Phan, Phuoc Thien; Tiong, Anthony Meng Huat; Tjin, Swee Chuan; Phee, Soo Jay

2019

Lai, W., Cao, L., Tan, R. X., Tan, Y. C., Li, X., Phan, P. T., ... Phee, S. J. (2020). An integrated sensor-model approach for haptic feedback of flexible endoscopic robots. *Annals of Biomedical Engineering*, 48(1), 342-356. doi:10.1007/s10439-019-02352-8

<https://hdl.handle.net/10356/137930>

<https://doi.org/10.1007/s10439-019-02352-8>

© 2019 Biomedical Engineering Society. All rights reserved. This paper was published in *Annals of Biomedical Engineering* and is made available with permission of Biomedical Engineering Society.

Downloaded on 26 Apr 2023 18:03:51 SGT

An Integrated Sensor-Model Approach for Haptic Feedback of Flexible Endoscopic Robots

Wenjie Lai¹, Lin Cao¹, Rex Xiao Tan², Yung Chuen Tan³, Xiaoguo Li¹, Phuoc Thien Phan¹, Anthony Meng Huat Tiong¹, Swee Chuan Tjin², Soo Jay Phee¹

Affiliations:

¹ Robotics Research Center, School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore;

² The Photonics Institute, School of Electrical and Electronic Engineering, Nanyang Technological University, Singapore;

³ Temasek Laboratories @ Nanyang Technological University, Singapore.

Correspondence: Lin Cao, Robotics Research Center, School of Mechanical and Aerospace Engineering, Nanyang Technological University, Singapore. e-mail: lin.cao@ntu.edu.sg

Abstract

Haptic feedback for flexible endoscopic surgical robots is challenging due to space constraints for sensors and shape-dependent force hysteresis of tendon-sheath mechanisms (TSMs). This paper proposes (1) a single-axis fiber Bragg grating (FBG)-based force sensor for a TSM of a robotic arm and (2) an integrated sensor-model approach to estimate forces on other TSMs of that arm. With a robust and simple structure, a temperature-compensated sensor can be mounted on the distal sheath to measure forces applied by the TSM. This proposed sensor was integrated with a $\varnothing 4.2$ mm articulated robotic arm driven by six TSMs, with a measurement error of 0.37N in this work. The measurement from the single sensor was used to identify parameters in the force-transmission models of all other TSMs in the robot, realizing a one-sensor-for-all-distal-forces measurement method. The sensor-model approach could accurately estimate the distal force with an RMSE of 0.65N. An animal study was carried out to demonstrate the sensor's feasibility in real-life surgery. The sensor-model approach presented a robust, space-saving, and cost-effective solution for haptic feedback of endoscopic robots without any assumption on the shapes of the robot.

Key terms: Flexible Endoscopic Surgical Robots; Haptic Force Sensor; Fiber Bragg Grating; Optical sensors; Tendon-Sheath Mechanism Modelling.

1 Introduction

Robot-assisted Minimally Invasive Surgery (RMIS) has advanced substantially over the past decade. Following the maturity of laparoscopic robots with straight and rigid arms, flexible endoscopic surgical robots for natural orifice transluminal endoscopic surgery (NOTES) have also been developed^{1,2}. The lack of haptic feedback exists in both RMIS and NOTES due to the non-transparency brought by the robotic system between the surgeon and patient. Feedback during robotic surgery, especially force exerted by the system, is crucial and necessary for a safe operation³. This paper presents a novel and effective solution for haptic force sensing for flexible endoscopic robots driven by Tendon-Sheath Mechanisms (TSMs).

TSMs have been widely used in flexible endoscopic robots², steerable endovascular catheters⁴, and other continuum manipulators⁵. However, two general challenges exist when implementing haptic feedback solution with these robotic systems: 1) highly nonlinear shape-dependent friction between the tendon and sheath, which leads to motion backlash, pretension-dependent stability, and force hysteresis⁶; 2) critical space constraints due to small end-effectors, which makes sensor installation process arduous. Correspondingly, research studies have been conducted to realize haptic feedback in two directions: 1) tip force prediction through modeling; 2) tip force measurement using sensors directly on surgical instruments.

To obtain haptic feedback, researchers have proposed models to predict distal forces based on the measured proximal forces in a TSM. Kaneko *et al.* developed a numerical tension transmission model for a single TSM with uniform preloading and fixed curvature⁷. Although the assumption of constant pretension and curvature restrict its practicability, their finding in proximal-distal force relationship has been the foundation of subsequent TSM studies in flexible endoscopic robots^{4,8,9}. Li *et al.* also proposed a deep learning method to predict distal

force in TSMs based on measured proximal force¹⁰. There is a common limitation that the change in the route configuration of the robot varies the parameters of these models. Consequently, it is challenging to acquire accurate real-time distal forces purely based on proximal data due to the continuous shape evolution of the TSM during surgical operations.

Studies have also been conducted to directly mount sensors on end-effectors using various sensing techniques, including sensing of displacement¹¹, pressure¹², resistance¹³, capacitance¹⁴, piezoelectricity¹⁵, vibration¹⁶, and optical properties^{17,18}. However, most of the force sensors are still too big for flexible endoscopic robots because they were intended for larger and rigid laparoscopic tools. Sterilization and robustness are also some common challenges for those electronic and structurally complicated sensors.

Integrating fiber optics sensors such as Fiber Bragg Gratings (FBGs) into medical instruments¹⁹⁻²² has been a recent trend because of FBGs' small sizes, flexibility, high sensitivity, biocompatibility, chemical inertness, electrical passivity, and sterilizability. To detect tri-axial tip contact forces in cardiac surgery, Gao *et al.* embedded an FBG-based force sensor inside a catheter²⁰. Zarrin *et al.* used FBGs in a grasper to measure grasping force and pulling force²¹. However, the grasper needs a dedicated design of the sensing structure, which significantly increases the complexity of sensor installations. Although adding FBG sensors at the tooltip is preferred, the end-effector structure, fiber alignments, sensor calibrations, and the fiber attachment procedure become complicated when multiple fibers are involved. Hence, for flexible endoscopic surgical robots, it is desired to use a minimum number of sensors to obtain maximum information at the distal side.

In this study, we designed a novel FBG-based force sensor (a PCT patent application PCT/SG2019/050271 has been filed) mounted at the distal end of a flexible endoscopic surgical robot. The proposed force sensor consists of a 1mm-grating FBG bonded to a 3mm-long nitinol tube that is welded between two sheaths. Pulling the tendon will result in compression from the sheath to the nitinol tube and thus induce a strain on the FBG. The compression force on the sheath is reflected through the corresponding Bragg wavelength shift. As a result, the associated tension force on the tendon, which is equal to the compression force on the sheath²³, can be observed.

Furthermore, we propose a sensor-model integrated approach to realize force sensing of all TSMs by using only one force sensor. The requirement of only one force sensor saves space and improves the robustness of the system. Inside an endoscopic surgical robot, all the TSMs are in the same route configuration and hence share the same parameters in force-transmission models^{6,7,9}. The measured force from this sensor is employed to identify the parameters in force-transmission models of other TSMs in the robot to realize the one-sensor-for-all-distal-forces design.

2 Materials and Methods

This section presents the development of the force sensor and the integrated sensor-model approach.

Subsection 2.1 introduces the fundamentals of FBG, the sensor's working principle, and

approaches for sensor calibration and temperature compensation. Subsection 2.2 proposes the integrated sensor-model approach. A proposition and associated verification tests are detailed after the tension transmission model of TSMs. Subsection 2.3 presents an endoscopic robotic grasper with the developed force sensor. The first test confirmed the sensor's functionality and demonstrated how the measured force reflects the interaction force between the grasper and the biological tissue it interfaced with. The second test was to verify the sensor-model approach in this robotic grasper. Lastly, an in-vivo animal trial demonstrates the sensor's feasibility in real-life surgery.

2.1 FBG-based Force Sensor

2.1.1 Fundamentals of FBG

FBG can reflect a specific wavelength of light, called Bragg wavelength λ_B . The Bragg wavelength shifts with respect to changes in temperature and strain due to physical deformation of the optical fiber hosting the FBG's periodic structures. Bragg wavelength is defined by the following equation²⁴:

$$\lambda_B = 2n_{eff} \Lambda \quad (1)$$

where n_{eff} and Λ are the modal effective refractive index of the optical fiber and the pitch of the periodic refractive index modulation induced in the core of the optical fiber, respectively. Λ of an FBG can vary with temperature and strain changes experienced by the host optical fiber.

Equation (2) expresses how the Bragg wavelength shifts in response to strain and temperature variations²⁵:

$$\frac{\Delta\lambda_B}{\lambda_B} = K_T \Delta T + K_S \Delta \epsilon \quad (2)$$

where ΔT , $\Delta \epsilon$, K_T , and K_S are the changes in temperature, axial strain along the fiber, thermal sensing coefficient, and strain sensing coefficient of the fiber material, respectively.

A dual FBG array is commonly adopted²⁶ to offset the cross-sensitivity of FBGs to temperature while the primary physical measurement of interest in strain, where FBG1 serves as a pure temperature sensor while the FBG2 attached to a testing object captures the combined strain due to external force F and fluctuations in temperature. The proportional relationship, as shown in expression (3), acts as the fundamental of temperature compensation study in this work:

$$\Delta F \propto (\Delta\lambda_{B2} - r\Delta\lambda_{B1}) \quad (3)$$

where r is the temperature sensitivity ratio of FBG2 to FBG1. Once the temperature sensitivity and the wavelength shift of FBG1 and FBG2 are known, force information could be easily computed. In this paper, Bragg wavelength shifts $\Delta\lambda_{B1}$ and $\Delta\lambda_{B2}$ were measured using a commercial optical spectrum analyzer (Micron Optics Interrogator si255). This interrogator

was selected for its real-time interrogation capability.

2.1.2 Working Principle and Design of Force Sensor

The force sensor is designed to measure the compression force C on the sheath, which equals the tension force T on the tendon²³ (Fig. 1a). A tube is welded between a long sheath (length: more than 2 meters) and a short sheath (length: 3cm). When the tendon at the proximal end is pulled, the compression force will be transmitted from the long sheath to the tube, then to the short sheath at the distal end. These transmitted compression forces have equal magnitudes at the welding area between the sheath and the tube, such as cross-section 2 and 3 (Fig. 1a). The expression $T = -C$ is always valid at any cross-section in the structure, like cross-section 1 to 4 (Fig. 1a).

FBG2 (Fig. 1a) of a dual FBG array (Technica, USA) is bonded with a super-elastic nitinol tube (length: 3mm, OD: 1.27 mm) using a biocompatible epoxy (EPO-TEK 353ND). FBG1 (Fig. 1a) is strain-free and located 10mm away from the FBG2 for temperature compensation. Both fiber and epoxy can perform continuously at temperature 200°C so that the force sensor can bear the high temperature (134°C) during a steam sterilization process.

2.1.3 Force Calibration and Hysteresis

During force calibration and hysteresis study, a tendon is driven by a motor at the proximal end, passing through the tortuous welded structure, and connected with a spring at the distal side (Fig. 1b). The tendon, welded structure, and fiber are inside a transparent outer tube, which route configuration is maintained by sticky tapes. The proximal end and distal end of the welded structure are stopped by a sheath stopper and a load cell, respectively. The load cell detects the compression force on the sheath, providing the ground truth of the actual force. In this test, both the distal force and the associated wavelength shift were recorded, when the motor was driven to apply a distal force from 0N to 20 N, then unloaded from 20 N to 0 N. To minimize the temperature cross-sensitivity, the force calibration and hysteresis study were carried out at constant laboratory temperature.

2.1.4 Temperature Compensation

Electrocautery (cutting tissue using electrically generated heat) may be involved during a surgery, which may increase the local temperature around the cutting incision inside the body. A dual FBG array was used to offset the temperature effect during force measurement.

During temperature calibration (Fig. 1c), water in the glass was heated from 25 °C to 85 °C and then left to cool from 85 °C to 25 °C, both the temperature from the thermometer and the wavelength shifts of FBG1 and FBG2 were recorded. The temperature sensitivity ratio of FBG1 and FBG2 was then calculated from the collected data.

After obtaining the temperature sensitivity ratio of FBG1 to FBG2, a temperature compensation test was conducted. The platform is similar to force calibration (Fig. 1b), with the spring being replaced by a 548-grams weight (equivalent to 5.38 N). Before any temperature change, the distal compression force due to the weight was measured by the force sensor. Then, a heat gun was used to increase ambient air temperature surrounding

the sensor from room temperature (25 °C). After heating for a few minutes, the heat gun was switched off, and the sensor was left to cool down. All the associated wavelengths of FBG1 and FBG2 were recorded during the heating and self-cooling process. By applying the compensation technique in section 2.1.1, the temperature effect was compensated during the force measurement.

2.2 Sensor-Model Approach

2.2.1 Basics for Tension Transmission Modeling

A tension transmission study on a single TSM was conducted (Fig. 2a). When a tendon is pulled and released by a motor at the proximal end, both proximal tension force F_p and distal tension force F_d are captured by load cells. In the proximal-distal force profile, F_p and F_d always have a linear relationship at the pulling phase (slope: K_p) and the releasing phase (slope: K_r), while F_d is constant at the two transition phases⁷ (Fig.2b).

Tension distribution along the entire length of a tendon⁷ is expressed in equation (4).

$$F(x) = F_p e^{\frac{-\langle \mu \rangle \cdot x \cdot \text{sgn} \xi}{R}} \quad (4)$$

L is the total length of a tendon and x could be any value from 0 to L , indicating the expected location along the tendon from the proximal end. $F(x)$ is the tension force at a particular location x . R and ξ are tendon radius of curvature and tendon displacement, respectively. $\text{sgn} \xi$ is given by,

$$\text{sgn} \xi = \begin{cases} 1 & (\xi \geq 0) \\ -1 & (\xi < 0) \end{cases} \quad (5)$$

where $\xi > 0$ and $\xi < 0$ indicate the pulling phase and releasing phase.

$\langle \mu \rangle$ is the frictional coefficient depending on the status of tendon movement, and defined as

$$\langle \mu \rangle = \begin{cases} \mu_s & (\xi = 0) \\ \mu_d & (\xi \neq 0) \end{cases} \quad (6)$$

Consequently, setting $a = \frac{\langle \mu \rangle \cdot L}{R}$, equation (7) could be derived.

$$\frac{\Delta F_d}{\Delta F_p} = \begin{cases} e^{-a} \\ e^a \end{cases} = \begin{cases} K_p, \text{ pulling phase} \\ K_r, \text{ releasing phase} \end{cases} \quad (7)$$

which infers the critical point of tension transmission on a TSM, i.e., the proportional relationship between the proximal and distal tension forces exists for both the pulling phase and releasing phase for a fixed route configuration.

The tension transmission is independent of the route configuration as long as the cumulated curve angles are the same⁹. In an endoscopic robot, all TSMs of an endoscopic

instrument will have the identical route configuration or cumulated curve angles, since they pass through the same channel, which leads to our proposition.

Proposition: All tendons in an endoscopic surgical robot have identical slope parameters at the pulling phase and releasing phase, respectively, in their proximal-distal force profiles.

For any two different TSMs of an endoscopic instrument (Fig. 2c), due to the same route configuration or cumulated angles, they present the same force profiles and share identical parameters, i.e., $K_{p1} = K_{p2}$ and $K_{r1} = K_{r2}$ (Fig. 2b).

Generally, in an endoscopic robot, load cells are located near the motors to trace F_p of all TSMs. By adding only one force sensor to capture F_d at the tip of any TSM in the surgical tool, slope values of the force profile can be obtained. Since the parameters are identical on all TSMs, the distal tension force on other TSMs can be then estimated by using the force profile. This is the proposed sensor-model approach.

2.2.2 Verification and Comparison

A verification test was conducted to verify the proposition, i.e., to prove $K_{p1} = K_{p2}$ and $K_{r1} = K_{r2}$. In this test, each TSM is driven by one motor at the proximal side and loaded with a spring at the distal side (Fig. 2d). To keep the cumulated angle identical to each other, TSMs in Lane1 and Lane2 are symmetrical at two ends. In Lane1, a load cell at the proximal end is utilized to detect F_p while the FBG-based force sensor at the distal end is used to measure F_d . In Lane 2, two load cells are implemented to sense the proximal and distal tension force, serving as a reference.

Pretension around 5N was applied for both TSMs to prevent tendon slacking. When the tendons were pulled and released by the motors, force measurements from the load cells and the force sensor were recorded accordingly. Force profiles were plotted, and the parameters for these two TSMs were compared at both pulling and releasing phases. By using parameters from Lane1, the sensor-model approach was applied to estimate the distal tension in Lane2. A comparison study between the actual force and estimated force in Lane2 was then carried out. The logic flow is presented to illustrate this study procedure (Fig. 2e).

2.3 Tests in an Endoscopic Robot

2.3.1 Integration with a TSM-driven Grasper

The force sensor is integrated with a TSM-driven grasper of a flexible endoscopic robot (Fig. 3a). The grasper (Fig. 3b) has three Degree of Freedoms (DOFs), where each DOF is driven by two TSMs, and the position of each TSM is allocated in a sequence (Fig. 3c). Two force sensors were integrated with this grasper, with one on the TSM to move up the grasper and the other one on the TSM to close the jaw (Fig. 3d).

2.3.2 Lifting a Colon Tissue

In this ex-vivo test, to confirm the functionality of the force sensor and to demonstrate how

the measured force reflects the interaction force between the grasper and the biological tissue, the grasper was used to lift a piece of pig colon tissue.

Only one force sensor was placed on the TSM to move up the grasper. A flexible protection tube (OD: 1.5mm, length: 3cm) was used to cover this sensor to prevent potential abrasion from the other sheaths. The tissue was placed on a plate connected with a load cell. The lifting force on the tissue can be reflected by the weight decrement from the load cell. When the colon was lifted, the tension force F_t from the force sensor and the lifting force F_l were recorded in pairs. The average force ratio between F_t and F_l was obtained for four different runs.

2.3.3 Verification for the Sensor-Model Approach

A test was conducted to prove the effectiveness of the sensor-model approach. Two force sensors (Fig. 3b and 3d) were assembled with the grasper in the robot to measure F_d , with one on the TSM to close the jaw (TSM1) and the other on the TSM to move up the grasper (TSM2). The nitinol tubes were welded at a different position on each TSM so that any abrasion between these sensors was avoided. Load cells at the motor housing side were used to collect F_p for these TSMs (Fig.3a). The robotic grasper was teleoperated by a haptic device (Omega 7) which controls the pitch and yaw motions of the grasper. When the grasper was pitching and yawing, F_p and F_d on TSM1 and TSM2 were recorded in pairs. Force profiles of TSM1 and TSM2 were drawn out, and errors in their slope values were analyzed. By using parameters from TSM1, the sensor-model approach was applied to estimate the distal tension for TSM2. A comparison study between the actual force and estimated force for TSM2 was then carried out.

2.3.4 In-vivo Animal Study

One force sensor was used during an in-vivo animal study, which was mainly to test a novel flexible endoscopic suturing device²⁷, with approval from the Institutional Animal Care and Use Committee (NO.: INH2018/017) in Singapore. The sensor was on the TSM to move up the surgical grasper. When the surgical operation was conducted inside the rectum of a live pig under general anesthesia, distal force data from the force sensor and the associated proximal force data from the load cells at the motor housing side were collected in pairs, in return, provided a relevant force profile.

3 Results

3.1 FBG-based Force sensor

3.1.1 Force Calibration and Hysteresis Result

The force sensor presents good linearity between force and wavelength shift, with a 4.798% hysteresis error (Fig.4a). The ratio of the absolute value of the wavelength shift to the compression force change is 34.12 pm/N with an R-square of 0.99700. The root mean square error (RMSE) of the wavelength shift was 12.59 pm, indicating a measurement error of $12.59/34.12 = 0.37\%$.

3.1.2 Temperature Compensation Result

Without any force loading, the force sensor presents good linearity between temperature change and wavelength shift, with no apparent hysteresis (Fig.4b). From the temperature calibration results, the temperature sensitivity of FBG1 is 11.23 pm/°C, with an R-square of 0.99703 while the temperature sensitivity of FBG2 is 25.62 pm/°C, with an R-square of 0.99431. Therefore, the temperature sensitivity ratio r is 2.28. The difference in temperature sensitivity of both FBGs is likely due to the different amounts of epoxy used for the adhesion process, which would affect the strain transfer (caused by thermal expansion) from nitinol tube to FBG.

For the temperature compensation result (Fig.4c), the compression force was measured as a negative value in the figure. The measured weight was 5.44N from the force sensor, which has a difference of 0.06N from the actual weight of 5.38N, well within the measurement error of 0.37N. The corresponding temperature change is plotted at the right y-axis, with a range of 24.21°C. Without temperature compensation, the reflected force was largely off from the measured weight, with a maximum error of 18.44N. With temperature compensation, the force fluctuated around the measured weight, with a maximum error of 0.94N. This noise can be further reduced to 0.68N by using a second-order low pass filter. According to the test results, this force sensor can offset the temperature effect and reflect accurate force information.

3.2 Verification: Sensor-Model Approach on Two TSMs

In the force profile for one pulling-releasing cycle in Lane1 (Fig.5a), the pulling slope K_{p1} is 0.60909 (R-square of 0.99432), and the releasing slope K_{r1} is 1.55751 (R-square of 0.99358). In the force profile for Lane2 (Fig.5b), K_{p2} is 0.58750 (R-square of 0.99610), and K_{r2} is 1.62565 (R-square of 0.99334). Taking the slope values in Lane2 as a reference, the relative errors of the slope values of the two lanes are 3.67% at pulling phase and 4.19% at releasing phase, respectively. Due to the limitation of the experimental setup, the route configurations at two end portions were not perfectly symmetrical. Also, the marginal curvature difference in these two TSMs along the outer tube may lead to a slight difference in slope values. Hence the accumulated angles in Lane1 and Lane2 were not completely identical to each other, which may result in the small differences in slopes at both the pulling and releasing phases.

A comparison study between real distal force and estimated distal force was conducted to verify the effectiveness of the sensor-model approach. Force data from two continuous pull-release cycles in Lane2 were used for this analysis. The actual distal tension force for Lane2 was collected through the load cell at the distal side. The estimated distal tension force for Lane2 was calculated by using parameters at the pulling and releasing phases from Lane1. The estimated force follows closely with the actual force at both pulling and releasing phases, with an RMSE of 0.36N (Fig.5c). There was a peak force gap at the transition phase, with a maximum error of 0.61N. Generally, in an endoscopic surgical robot, the distal tension force can reach as high as 40 N. Therefore, the percentages are 0.9% for RMSE and 1.53% for maximum peak force error, which are acceptable errors in this application.

3.3 Tests in an Endoscopic Robot

3.3.1 Lifting a Colon Tissue

In this test (Fig.6a to 6c), after four continuous runs to lift the colon tissue (weight: 945 grams), the force ratio of the distal tension force F_t to the lifting force F_l was obtained as 21.21, with an R-square of 0.98339 (Fig. 6d). The lifting force on the tissue has a range of 0 to 0.5N. At the initial region from 0 to 0.1N, tissue was lifted from a stationary status, where the associated acceleration causes the non-linear force pulse. However, data becomes stable and shows good linearity within the range of 0.1 to 0.5N lifting force.

3.3.2 Verification: Sensor-Model Approach on the Robot

Two force sensors were integrated into the endoscopic grasper in this verification test (Fig.7a). In this work, we mainly focused on pulling and releasing phases (Fig.7b and 7c). At the pulling phase, slope values for TSM1 (0.45888) and TSM2 (0.45686) are significantly close to each other. At the releasing phase, slope values for TSM1 (2.29464) and TSM2 (2.48407), are also sufficiently close to each other. Taking the slope values for TSM2 as a reference, relative errors are 0.44% for the pulling phase and 7.63% for the releasing phase. Due to the limitation of fabrication platforms, there will be a slightly different hysteresis error among the force sensors. The force sensor on TSM1 presented a relatively large force hysteresis of 5.80% in its calibration, which may make a contribution to the relative error (Fig.7b). Also, the marginal curvature difference in these two TSMs along the endoscope may lead to a slight difference in slope values. This estimated force was compared to the actual force for the TSM2 in one pull-release cycle, with an RMSE of 0.65N (Fig.7d), which are mainly induced by the relative errors and hysteresis problem. Thus, the effectiveness of the approach to be applied in an endoscopic robot was proven.

3.3.3 In-vivo Animal Study

The animal study was conducted inside the rectum of a live pig under general anesthesia, where a force sensor was installed on the grasper (Fig.8a). In the corresponding force profile for one pulling-releasing cycle during this in-vivo test, the slope at the pulling phase K_p is 0.34583 with an R-square of 0.97986, and the slope at releasing phase K_r is 1.74501 with an R-square of 0.99386 (Fig.8b). The proximal force was limited to 40N, and motor movement was restricted when it reaches the force limit, which causes the proximal force fluctuating around 40N in the results. In the transition phases, the distal force was unchanged, which agrees with the mechanics of TSMs. The result implies that the force sensor can reliably measure real-time distal tension force on the TSM in real surgical scenarios.

4 Discussion

For the sensor, it can detect the distal tension force on the TSM with a sensitivity of 34.12 pm/N and an error of 0.37N. It was able to compensate temperature effect during force measurement, with a maximum filtered error of 0.68N. The force sensor can also accurately reflect the lifting force at the tip of the end-effector exerted on the tissue. The feasibility and functionality of the force sensor were also proven during the animal trial.

For the proposed integrated sensor-model approach, the experimental tests on two TSMs have verified the proposition, with a relative error of 4.19% in slope values and an RMSE of 0.36N between the real force and the estimated force. With this sensor-model approach, haptic feedback on all TSMs can be achieved through one force sensor. This was further verified on the robot, where TSMs for gripping and pitch motions were chosen for demonstration. The approach can be extended to TSMs for other DOFs as long as they stay in the same endoscope.

Sheaths were selected as the mounting location of the force sensor due to the following reason: Firstly, if the sensor is attached to the tendon, both the elongation and bending of the tendon are coupled, which would result in significant measurement errors of the sensor. Moreover, the frequent movement of the tendon may also lead to wearing of the fiber. Secondly, if the sensor is on the end-effector, it may be damaged, or its measurement may be altered by the interaction between the moving end-effector and the external unstructured surgical environment. In contrast, mounting the force sensor on the sheath can avoid the above issues because (1) the sensor on the welded tube-sheath structure only detects the single-axis strain change on the nitinol tube, avoiding the coupling issue; (2) the sheath barely move during operations, and thus the sensor is unlikely to be damaged from the movements of the tendon or end-effector; (3) the sensor and sheaths are all inside the tubular body of the instrument, which shields the sensor from direct contact with the unstructured surgical environment; (4) there is more space inside the tubular body of the instrument, where sheaths are located, than the end-effector.

Though reliable real-time distal force sensing was validated in this study, the force sensor poses some measurement limitations. The force sensor has a hysteresis of 4.798%, due to recoating as well as the bonding condition between the epoxy and optical fiber. Although FBG without recoating would reduce hysteresis and improve sensitivity, the uncoated fiber is fragile and hence, easy to break in this application. With improved precision and reliable tools for fiber fabrication and assembly, uncoated FBGs may be considered. Also, since the force sensor is located 3 cm away from the end of the sheath, its measurement cannot reflect the friction forces between (1) the tendon and the pulley at the joint, and (2) the tendon and the sheath in the 3 cm region. Further study needs to be carried out to overcome this inherent limitation.

Even though the estimated force traced the actual force closely using the sensor-model approach at the pulling phase and releasing phase, there was a peak force gap of 0.61N at the transition phase. Small force growth exists at the transition phase, because of the complex nature of TSMs when changing movement or friction directions, as also observed in Refs.^{8,10}. This error might be due to the simplicity of the current model. An improved model¹⁰ which can precisely capture the special feature in the transition phase might be able to resolve this issue.

In this study, the obtained force is not displayed to the master console of the robot. To further study the benefits of haptic feedback to flexible endoscopic surgery, subject studies on the robot with and without the haptic feedback will be conducted in the future.

5 Conclusion

In this paper, we proposed (1) a novel force sensor for TSM-driven surgical robots for haptic force sensing, and (2) an integrated sensor-model approach to obtain distal force on all TSMs in an endoscopic surgical robot by using only one sensor. The sensor can offset the temperature effects and reflect distal tension force on the tendon in real-time, with a measurement error of 0.37N. The estimated force by using the sensor-model approach follows the actual force accurately in the endoscopic robot, with an RMSE of 0.65N.

This small force sensor has salient advantages: it is robust with a simple structure, small, electrically passive, temperature-compensated, easy to assemble and disassemble, flexible, biocompatible, and sterilizable. Since only one sensor is needed for the robotic arm, the proposed integrated sensor-model approach also represents a robust, cost-effective, and space-saving solution for the real-time distal force feedback of flexible endoscopic robot without assumptions on the shapes of TSMs.

In the future, the sensor may be improved by using uncoated FBGs and being closer to the end-effector. The integrated sensor-model approach may also be improved by incorporating other more sophisticated models^{8,10} which can more precisely capture the transition phase of TSMs. To study the benefits of haptic feedback for robotic endoscopic surgery, subject studies will be conducted. Applications of this force sensor to other TSM-driven systems will also be explored, e.g., robotic fingers/hands, wearable devices, surgical catheters, and rehabilitation devices.

Acknowledgements

This work was supported by the National Research Foundation (NRF) Singapore (NRFI2016-07).

References

1. Gu X, Li C, Xiao X, Lim CM, Ren H. A compliant transoral surgical robotic system based on a parallel flexible mechanism. *Annals of Biomedical Engineering*, 2019.
2. Cao L, Li X, Phan PT, Tiong AMH, Kaan HL, Ho KY, Chiu PWY, Phee SJ. A flexible endoscopic robotic suturing system for gastrointestinal perforations: animal study. *Hamlyn Symposium on Medical Robotics*, 2019.
3. Wagner CR, Stylopoulos N, Jackson PG, Howe RD. The benefit of force Feedback in surgery: examination of blunt dissection. *16(3):252-262*, 2007.
4. Khoshnam M, Patel RV. Robotics-assisted control of steerable ablation catheters based on the analysis of tendon-sheath transmission mechanisms. *IEEE/ASME Transactions on Mechatronics*. *22(3):1473-1484*, 2017.
5. Le HM, Do TN, Cao L, Phee SJ. Towards active variable stiffness manipulators for surgical robots. Paper presented at: 2017 IEEE International Conference on Robotics and Automation (ICRA); 2017.
6. Agrawal V, Peine WJ, Yao B. Modeling of Transmission Characteristics across a cable-conduit system. *IEEE Transactions on Robotics*. *26(5):914-924*, 2010.

7. Kaneko M, Yamashita T, Tanie K. Basic considerations on transmission characteristics for tendon drive robots. Paper presented at: Advanced Robotics, 1991. 'Robots in Unstructured Environments', 91 ICAR., Fifth International Conference, 1991.
8. Do TN, Tjahjowidodo T, Lau MWS, Yamamoto T, Phee SJ. Hysteresis modeling and position control of tendon-sheath mechanism in flexible endoscopic systems. *Mechatronics*. 24(1):12-22, 2014.
9. Sun Z, Wang Z, Phee SJ. Elongation modeling and compensation for the flexible tendon-sheath system. *IEEE/ASME Transactions on Mechatronics*. 19(4):1243-1250, 2014.
10. Li X, Cao L, Tiong AMH, Phan PT, Phee SJ. Distal-end force prediction of tendon-sheath mechanisms for flexible endoscopic surgical robots using deep learning. *Mechanism and Machine Theory*. 134:323-337, 2019.
11. McKinley S, Garg A, Sen S, et al. A single-use haptic palpation probe for locating subcutaneous blood vessels in robot-assisted minimally invasive surgery. Paper presented at: 2015 IEEE International Conference on Automation Science and Engineering (CASE), 2015.
12. Arabagi V, Felfoul O, Gosline AH, Wood RJ, Dupont PE. Biocompatible pressure sensing skins for minimally invasive surgical instruments. *IEEE Sensors Journal*. 16(5):1294-1303, 2016.
13. Srivastava A, Xu R, Escoto A, Ward C, Patel RV. Design of an ultra thin strain sensor using superelastic nitinol for applications in minimally invasive surgery. Paper presented at: 2016 IEEE International Conference on Advanced Intelligent Mechatronics (AIM); 2016.
14. Kim U, Kim YB, Seok DY, So J, Choi HR. A surgical palpation probe with 6-Axis force/torque sensing capability for minimally invasive surgery. *IEEE Transactions on Industrial Electronics*. 65(3):2755-2765, 2018.
15. Feng J, Yahui Y, Zhao Z, et al. A variable-impedance piezoelectric tactile sensor with tunable sensing performance for tissue hardness sensing in robotic tumor palpation. *Smart Materials and Structures*. 27(11):115039, 2018.
16. Pacchierotti C, Prattichizzo D, Kuchenbecker KJ. Cutaneous feedback of fingertip deformation and vibration for palpation in robotic surgery. *IEEE Transactions on Biomedical Engineering*. 63(2):278-287, 2016.
17. Li T, Shi C, Ren H. A high-sensitivity tactile sensor array based on fiber bragg grating sensing for tissue palpation in minimally invasive surgery. *IEEE/ASME Transactions on Mechatronics*. 2018:1-1.
18. Su H, Shang W, Li G, Patel N, Fischer GS. An MRI-Guided telesurgery system using a Fabry-Perot interferometry force sensor and a pneumatic haptic device. *Annals of Biomedical Engineering*. 45(8):1917-1928. 2017.
19. He X, Handa J, Gehlbach P, Taylor R, Iordachita I. A submillimetric 3-DOF force sensing instrument with integrated fiber bragg grating for retinal microsurgery. *IEEE Transactions on Biomedical Engineering*. 61(2):522-534, 2014.
20. Gao A, Zhou Y, Cao L, Wang Z, Liu H. Fiber bragg grating-based triaxial force sensor With parallel flexure hinges. *IEEE Transactions on Industrial Electronics*. 65(10):8215-8223, 2018.

21. Zarrin PS, Escoto A, Xu R, Patel RV, Naish MD, Trejos AL. Development of a 2-DOF sensorized surgical grasper for grasping and axial force measurements. *IEEE Sensors Journal*. 18(7):2816-2826, 2018.
22. Li T, Shi C, Ren H. Three-dimensional catheter distal force sensing for cardiac ablation based on fiber Bragg grating. *IEEE/ASME Transactions on Mechatronics*. 23(5):2316-2327, 2018.
23. Lai W, Cao L, Xu Z, Phan PT, Shum P, Phee SJ. Distal end force sensing with optical fiber Bragg gratings for tendon-sheath mechanisms in flexible endoscopic robots. Paper presented at: 2018 IEEE International Conference on Robotics and Automation (ICRA), 2018.
24. Hill KO, Meltz G. Fiber Bragg grating technology fundamentals and overview. *Journal of Lightwave Technology*. 15(8):1263-1276, 1997.
25. Ren F, Zhang W, Li Y, Lan Y, Xie Y, Dai W. The temperature compensation of FBG sensor for monitoring the stress on hole-edge. *IEEE Photonics Journal*. 10(4):1-9, 2018.
26. Zhou Z, Ou J. Techniques of temperature compensation for FBG strain sensors used in long-term structural monitoring. Paper presented at: Fundamental Problems of Optoelectronics and Microelectronics II, 2005.
27. Cao L, Li X, Phan P, Tiong A, Liu J, Phee SJ. A novel robotic suturing system for flexible endoscopic surgery. *International Conference on Robotics and Automation*. Montreal, Canada, 2019.

FIGURE 1. FBG-based force sensor: (a) Design and working principle, where C represents the compression force on the sheath, T the tension force on the tendon, and EE end-effector. (b) Platform for force calibration and hysteresis tests. A commercial load cell was used nearby the force sensor for ground truth measurement. (c) Temperature calibration setup: the force sensor was immersed into water in a glass container with a thermometer.

FIGURE 2. Tension transmission model on TSMs: (a) Study on a single TSM. (b) Force profile: proximal force F_p v.s distal force F_d . (c) Two TSMs of an endoscopic instrument stay in the same channel. (d) Verification test platform: two TSMs were constrained by an outer tube with a fixed route configuration. (e) Logic flow of verification test and force comparison for the sensor-model approach.

FIGURE 3. Force sensors' integration with a TSMs-driven grasper in an endoscopic robot: (a) Schematic diagram of the robot. (b) 3D model of the grasper with three DOFs: gripping, yaw, and pitch. (c) Position arrangement of each TSM. (d) Cross-section view of the grasper with two force sensors.

FIGURE 4. Results in (a) Force calibration and hysteresis. (b) Temperature calibration. (c) Temperature compensation verification.

FIGURE 5. Verification results of the sensor-model approach: (a) Force profile in Lane1 with a FBG-based force sensor. (b) Force profile in Lane2 with two load cells. (c) Force comparison.

FIGURE 6. Test results of lifting up a colon tissue: (a) the grasper with a sensor. (b) Setup: a robotic endoscopic grasper with one force sensor to lift a colon tissue. (c) The grasper lifting a piece of tissue. (d) Force ratios of four different runs.

FIGURE 7. Verification results of the sensor-model approach in a grasper of an endoscopic robot. (a) A prototype of the integrated grasper. (b) Obtained force profile on TSM1. (c) Obtained force profile on TSM2. (d) Force comparison.

FIGURE 8. (a) An in-vivo animal study inside the rectum of a live pig. (b) Force profile of one pulling-releasing cycle.