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Joint Rotation Angle Sensing of Flexible Endoscopic Surgical Robots

Wenjie Lai¹, Lin Cao^{1*}, Phuoc Thien Phan², I-Wen Wu³, Swee Chuan Tjin⁴, Soo Jay Phee¹

Abstract—Accurate motion control of surgical robots is critical for the efficiency and safety of both state-of-the-art teleoperated robotic surgery and the ultimate autonomous robotic surgery. However, fine motion control for a flexible endoscopic surgical robot is highly challenging because of the shape-dependent and speed-dependent motion hysteresis of tendon-sheath mechanisms (TSMs) in the long, tortuous, and dynamically shape-changing robot body. Aiming to achieve precise closed-loop motion control, we propose a small and flexible sensor to directly sense the large and sharp rotations of the articulated joints of a flexible endoscopic surgical robot. The sensor—a Fiber Bragg Grating (FBG) eccentrically embedded in a thin and flexible epoxy substrate—can be significantly bent within a large bending angle range of $[-62.9^\circ, 75.5^\circ]$ and small bending radius of 6.9 mm. Mounted in-between the two pivot-connected links of a joint, the sensor will be bent once the joint is actuated, resulting in the wavelength shift of the FBG. In this study, the relationship between the wavelength shift and the rotation angle of the joint was theoretically modeled and then experimentally verified before and after installing the sensor into a robotic endoscopic grasper. The sensor, with the calibrated model, can track the bidirectional rotation of the robotic joint with an RMSE of 3.34°, good repeatability, high sensitivity (around 147.5 pm/degree), and low hysteresis (7.72%). It is suitable for surgical robots and manipulators whose articulated joints have a large rotation angle and small bending radius.

Index Terms— Flexible Robots, Surgical Robotics, Fiber Optics Sensor, Tendon Sheath Mechanism.

I. INTRODUCTION

Over the last two decades, robotic surgery has emerged as a new paradigm of minimally invasive surgery (MIS), for its unique features like motion scaling, tremor/fatigue reduction, clear vision, and enhanced dexterity. Flexible endoscopic surgical robots have been developed to further reduce the invasiveness of robotic surgery [1, 2]. Tendon-sheath mechanisms (TSMs) [3] are often employed as the transmission mechanism in flexible endoscopic surgical robots due to their flexibility, lightweight, small sizes, high force capacity, reliability, and safety. However, it is challenging to realize accurate motion control for TSM-driven surgical robots due to the inherent hysteresis and backlash of TSMs [4] from the friction and tendon elongation of TSMs. What is more, the hysteresis and backlash properties change with the shape of the robot body and the actuation speed [5]. These issues bring significant tool positioning errors during

surgery which could lead to potential injuries, poor hand-eye coordination, cognitive burdens to the operator, and defective surgical performance [6-8]. In addition, poor motion accuracy in surgical robots also hinders the development of autonomous surgery [9, 10] which is the ultimate goal of robotic surgery.

To improve motion accuracy, two categories of approaches have been developed: 1) elongation modelling of TSMs for motion compensation [5, 11, 12]; 2) motion sensing for closed-loop motion control with feedback, including vision-based pose tracking [13-15], photosensor [16], electro-conductive yarn [17], magnetic sensors [18], and fiber Bragg grating (FBG) sensors [19-24].

Agrawal *et al.* introduced a discretized transmission model of TSMs [11], which was only applicable for TSMs with a uniform shape. Do *et al.* overcame this limitation by proposing a nonlinear adaptive compensation model to enhance tracking performance for the TSMs, with a peak-to-peak error of 0.0836 rad (4.8°) [12]. The elongation of the TSMs was calculated based on a curve-fitted backlash profile of the TSMs. Sun *et al.* proposed a motion compensation scheme of TSMs, with an error of 3.3° [5], by calculating the tendon elongation based on the proximal pulling force and a pre-calibrated force-to-elongation ratio of the TSMs. The ratio may be updated with the shape change of the endoscope tip which is, however, unknown and has to be roughly estimated. All these model-based approaches do not require sensors in the small robotic instruments and thus are less costly and less limited by the space and size constraints. Nevertheless, these models are accurate mainly at certain operation conditions; once the shape of the robot body or the actuation speed is changed, the accuracy of these models drops dramatically since the pre-fitted or pre-calibrated parameters cannot reflect the current status of the running system.

Penza *et al.* presented an enhanced vision system for robotic surgery with an algorithm for the 3D pose reconstruction of instruments [13]. However, heavy computation obstructs its practice [19]. Ma *et al.* realized automatic instruments tracking by using visual servoing to improve operation safety [14]. Du *et al.* performed image-based 2D pose estimation for multiple articulated instruments using a deep neural network [15]. Computer vision is relatively low cost because no additional sensor is needed, and it is adaptive to the dynamic changes of the operation conditions of the system. However, it may require 3D vision, high-resolution imaging, and is more prone to the uncertainties of the unstructured environment (e.g., the cautery smoke may degrade the vision quality and thus affect results).

Motion sensors designed particularly for the motion measurement in surgical robots are more desirable due to high accuracy, adaptiveness, and less computation consumption. Ko *et al.* used a photosensor detecting the tendon length

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variation to reflect the bending angle [16]. However, the linear relationship between the tendon elongation and the bending angle is only suitable for laparoscopic rigid instruments. For a flexible manipulator, Wurdemann *et al.* suggested to sense bending angle by using an electro-conductive yarn which is helically wrapped around an elastic strap [17]. This sensor can measure an angle up to 110° but has a maximum hysteresis of up to 45%. Magnetic tracking [18] is not practical because of the interferences caused by the existing magnetic field of the environment and conductive objects.

FBG sensor has recently become promising and popular for bending sensing in flexible surgical manipulators [19, 20], e.g., continuum neurosurgical robot [21, 22] and steerable needle [23, 24], due to its small size, flexibility, high sensitivity, and simplicity. Sheng *et al.* designed a bidirectional bend sensor for shape memory alloy (SMA) bending module by bonding two FBG fibers orthogonally with a superelastic nitinol strip [21]. However, this sensor has a maximum measurable angle up to 29° only and an apparent hysteresis problem. Rahman *et al.* developed a bending sensor module to capture the deflection of the MINIR-II robot by embedding an FBG within a Polydimethylsiloxane cylinder and attaching a superelastic nitinol spring [22]. The sensitivity varies from 3 pm/degree in the range of 0 to 20° to 5-7 pm/degree for 20° to 50° . Also, this sensor has a visible hysteresis effect, and the repeatability is varying. In addition, multiple FBG fibers located equally apart from each other are used to realize the shape sensing of flexible manipulator [23, 24]. Nevertheless, these sensors are not suitable for small flexible endoscopic robotic instruments [1] whose space is limited, bending radius is small, and the range of the rotation angle is large.

In this paper, we propose an FBG-based sensor to capture the rotation angle of an endoscopic surgical robot. The FBG is embedded in a flexible epoxy substrate, and there is an offset distance between the neutral axis of the substrate and that of the FBG. The bending angle of the substrate can be obtained from the wavelength shift of the FBG. This sensor can sense bidirectional angles ranging from -62.9° to 75.5° , which is larger than the measurable range in [21, 22]. After integration with a robotic surgical grasper, the sensor was bent with a minimum bending radius of 6.9 mm, which is much smaller than the one of 12.8 mm (converted from its maximum curvature of 77.87 m^{-1}) in [21]. A linear relationship between $\tan(\alpha/2)$ (α is the rotation angle) and Bragg wavelength shift is derived for this sensor, and the corresponding model is verified through experiment. It presents excellent repeatability during measurement, with 7.72% hysteresis, which is better than the one of 45% in [17]. After mounted in an endoscopic surgical grasper, the sensor, with the calibrated model, can reflect the grasper's bidirectional rotation angle in real time with an RMSE of 3.34° , which are comparable to the error from modeling method [5, 12].

The rest of this paper is organized as follows. Section II introduces the design, fabrication, working principle, and modeling of the sensor, as well as its integration in a flexible endoscopic robot. Section III presents the results of modeling validation, pre-calibration and repeatability test, and sensor's performance in a surgical instrument. At last, Section IV concludes the work.

II. SENSOR DEVELOPMENT AND MODELING

In this section, the design and fabrication of the rotation angle sensor are introduced, followed by the sensing principle of the FBG with an epoxy substrate. A general model is proposed to describe the relationship between the wavelength shift and the rotation angle for the sensor before its installation in a surgical instrument. Then, the integration of the sensor and a TSM-drive robotic grasper ($\varnothing 4.2 \text{ mm}$) is presented, and the model is further developed for the sensorized robotic grasper.

A. Sensor Design and Fabrication

The rotation angle sensor consists of a single 5mm-grating FBG fiber with polyimide coating (Samyon Inc., Beijing) and a thin and flexible epoxy substrate (Araldite 90min standard high-performance epoxy), and the fiber was located off the neutral axis of the substrate (Fig. 1a). Note that it is crucial to locate the FBG fiber off the neutral axis of the epoxy substrate; otherwise, the FBG will be insensitive to the bending of the substrate (refer to section II.B for the working principle of this sensor).

In the sensor fabrication process, plasticine was selected as the molding material because it is detachable from epoxy [25]. A stainless steel rectangular rod was used to create a slot in the mold. An FBG fiber was placed at the bottom of the slot, which was then filled with epoxy (Fig. 1b). When the epoxy was cured, the sensor (FBG with the epoxy substrate) was removed from the plasticine mold, with a cross-section size of $0.9 \text{ mm} \times 0.5 \text{ mm}$ (Fig. 1c).

Figure 1. A rotation angle sensor: (a) schematic design, (b) fabrication process, (c) prototype.

Other materials such as half-circle stainless steel tube as the mold and a silicone substrate (Ecoflex 00-20) have also been explored. However, it was hard to peel off the cured epoxy from the tube, and the sensor was easily broken during the peeling process. For the silicone substrate, the sensor appeared to be insensitive to bending since silicone is too soft to provide strong enough bonding between itself and the fiber.

B. Bending Theory

The specific wavelength of light reflected by the FBG is called Bragg wavelength λ_B [3], which shifts with respect to strain and temperature changes, as in:

$$\frac{\Delta\lambda_B}{\lambda_B} = k_T \Delta T + k_\epsilon \epsilon \quad (1)$$

where k_T , ΔT , k_ϵ and ϵ are the thermal sensing coefficient, temperature change, strain sensing coefficient of the fiber material, and axial strain along the fiber, respectively.

In this study, tests were conducted at constant laboratory room temperature ($24.3^\circ\text{C} \pm 0.1^\circ\text{C}$) to minimize the temperature cross-sensitivity. Thus, the wavelength shift becomes proportional to strain change, as in:

$$\varepsilon = \frac{\Delta\lambda_B}{k_e \lambda_B} \quad (2)$$

When the proposed sensor bends, the embedded FBG is either under compression or in tension, depending on the bending direction (Fig. 2), where t is the thickness of the sensor; t_{offset} is the distance between the fiber and the neutral axis of the substrate; L is the length of the neutral axis of the epoxy substrate; R is the bending radius; θ is the bending angle of the sensor, defined as $\theta = L/R$.

Figure 2. Bending theory on the proposed sensor: (a) under compression, (b) in tension. The blue area stands for an epoxy substrate, while the yellow line with short black lines represents for FBG sensor.

The following expression shows the deformed length L' due to bending (plus sign for tension, minus sign for compression):

$$L' = L \pm \varepsilon L = L \pm t_{offset} \theta \quad (3)$$

Since the bending curvature is defined as $\kappa = 1/R$, the bending angle θ can also be expressed as $\theta = \kappa L$. A linear relationship between the bending curvature κ and Bragg wavelength shift $\Delta\lambda_B$ can be revealed, as in:

$$\kappa = \frac{1}{k_e \cdot t_{offset} \cdot \lambda_B} \Delta\lambda_B \quad (4)$$

C. Relationship between Bidirectional Rotation Angle and Bragg Wavelength Shift

Figure 3. Schematic diagram to illustrate: modeling between the rotation angle and the Bragg wavelength shift (upwards rotation: under compression).

The sensor is used to measure the bidirectional rotation angle between two articulated links (link 1 and link 2) with a pivot motion center C (Fig. 3). One end of the sensor is fixed to link 1; the other end is fixed in a way that it can rotate with link 2 but free to move along the longitudinal axis of link 2. This free movement prevents the sensor from being broken due to the rotation of link 2. Point A refers to the exit of the hole 1, and point B (D) is the entry of the hole 2. As link 2 rotates upwards, from position B to position D, the sensor is under compression, and the rotation angle is set as negative; as link 2 rotates downwards, the sensor is in tension, and the rotation angle is set as positive. To link the rotation angle α with the Bragg wavelength shift $\Delta\lambda_B$, a unique geometry feature, equal distance from the rotating joint towards holes in link 1 and link 2 ($AC = CB = CD = m$, m is a constant), is required. The curvature along the fiber section AD is assumed uniform. Note that this curvature changes as link 2 rotates.

Based on this geometry, angle relationships below can be obtained:

$$\angle BCD = \angle AED = \alpha \quad (5)$$

$$\angle AEC = \angle DEC = 1/2 \alpha \quad (6)$$

In the schematic diagram, the bending radius is $R = AE = DE$. By definition, $\tan(\alpha/2)$ can be then expressed as follows:

$$\tan \frac{\alpha}{2} = \frac{AC}{AE} = \frac{CD}{DE} = \frac{m}{R} \quad (7)$$

The bending curvature κ is equivalent to $1/R$. A linear relationship between Bragg wavelength shift $\Delta\lambda_B$ and $\tan(\alpha/2)$ is derived, after combining Eq. (4) and Eq. (7).

$$\Delta\lambda_B = \frac{k_e \cdot t_{offset} \cdot \lambda_B}{m} \tan \frac{\alpha}{2} = c \tan \frac{\alpha}{2}, \quad (8)$$

where c is the sensitivity, which is a constant because parameters m , k_e , t_{offset} , and λ_B all have fixed value. Equation (8) indicates that a higher sensitivity can be achieved by reducing the joint-hole distance m or increasing the offset distance t_{offset} . Also, the sensitivity is not affected by the grating length of the FBG.

Equation (9) is obtained by switching the position of $\Delta\lambda_B$ and $\tan(\alpha/2)$ in Eq. (8). Equation (10) can be directly calculated from Eq. (9). Equation (10) indicates that, as long as the wavelength shift is captured, the corresponding rotation angle of the sensor can be calculated.

$$\tan \frac{\alpha}{2} = \frac{m}{k_e t_{offset} \lambda_B} \Delta\lambda_B = \frac{1}{c} \Delta\lambda_B \quad (9)$$

$$\alpha = 2 \tan^{-1} \left(\frac{m}{k_e t_{offset} \lambda_B} \Delta\lambda_B \right) \quad (10)$$

Equations (5) to (10) are always valid no matter the sensor is under compression or in tension, which entails that the rotation direction will not affect coefficients in this modeling.

A testbed with the aforementioned geometry feature was used to verify the modeling, and the corresponding result is presented in Section III.A. The calibration and repeatability test for the sensor before its installation in a surgical instrument were conducted, with the result shown in Section III.B.

D. Grasper Design and Integration with the Sensor

This sensor is integrated with a TSM-driven endoscopic surgical grasper (Ø4.2 mm) (Fig. 4a) [1]. The grasper has three degree-of-freedom (DOF), including pitch, yaw, and gripping. Each DOF is actuated by a pair of TSMs (tendon OD: 0.4mm, sheath OD: 0.96mm) and thus in total, six TSMs are used in this grasper. The grasper was manufactured in direct metal laser sintering process using stainless steel 316L by *Protolab Inc, US*. After the installation of six TSMs, available space for the sensor becomes very limited, and only two locations (right-up and right-down corner) are left spare (Fig. 4c). Considering the minimum wall thickness requirement (0.2mm) for 3D printing, the maximum allowable cross-section size for the sensor is 1.0 mm * 0.615 mm. In this study, the pitch rotation angle is measured using this proposed sensor for demonstration. The sensor was placed in the hole at the right-up corner of the grasper (Fig. 4c). A simplified diagram (Fig. 4b) illustrates the inner design of the grasper in a vertical projection plane, where point O is the center of the joint; the distance between hole and the interested joint are equal, i.e., FO = OG = 3.5mm; HF and IG refer to the distance

from the bottom of the hole to the centerline of the link 1 and link 2, respectively, and $HF = IG = 0.325\text{mm}$.

The sensor bends and is under compression when the grasper moves up, with a bending radius $R = AE$ (Fig. 5a), while the sensor bends and is in tension when the grasper moves down, with a bending radius $R = AE'$ (Fig. 5b). Points A, C, D , and E (E') match those points in Fig. 3 ($AC = CD = m$, $\angle AEC = \angle DEC = 1/2\alpha$); points F, O and G match those points in Fig. 4b ($AF = DG$, and $FO = OG$); l_1 and l_3 represent the centreline of the sensor in the link 1 and link 2; l_2 and l_4 are the centreline of the link 1 and link 2, respectively.

Figure 4. A robotic endoscopic grasper with the rotation angle sensor: (a) integration with the sensor, (b) a simplified diagram to illustrate the design details, (c) location and maximum allowable size for the sensor.

Figure 5. Correlation between rotation angles α and β when the grasper (a) rotates up, (b) rotates down (α : rotation angle of the sensor, β : rotation angle of the grasper).

Based on parallel relationships, $l_1 \parallel l_2$ and $l_3 \parallel l_4$, the rotation angle of the sensor α is equivalent to the rotation angle of the grasper joint β , which is defined as a positive value when the sensor is in tension while a negative value under compression.

$$\alpha = \beta \quad (11)$$

Based on the geometry, the following equation can be directly obtained (plus sign for upward movement; minus sign for downward movement; AE (AE') = R):

$$\tan \frac{\alpha}{2} = \frac{FO}{EF} = \frac{FO}{E'F} = \frac{FO}{R \pm AF} = \frac{FO}{\frac{1}{\kappa} \pm AF} \quad (12)$$

Combining equations (11), (12), and (4), the bidirectional rotation angle of the grasper joint can be calculated as long as the wavelength shift is known, as in:

$$\beta = 2 \tan^{-1} \left(\frac{FO}{\frac{k_e t_{offset} \lambda_B}{\Delta \lambda_B} \pm AF} \right) = 2 \tan^{-1} \left(\frac{c_1}{\frac{c_2}{\Delta \lambda_B} \pm c_3} \right) \quad (13)$$

where FO, AF, k_e, t_{offset} , and λ_B all have a fixed value, and they are replaced by constant c_1, c_2 , and c_3 , whose values can be obtained through the design and experiment.

According to Eq. (13), with the same rotation angle magnitude, the upward motion would result into more significant wavelength shift than the downward motion, i.e., this sensor has a higher sensitivity when the grasper is moving up. This different behavior is caused by the off-center mounting position of the sensor in the grasper. A demonstration was conducted on a sensor-integrated surgical grasper to verify Eq. (13) and test its

functionality/performance, and the result is presented in Section III.C.

III. EXPERIMENTS AND RESULTS

A. Model Verification

This test was to verify the model described in Section II.C. The setup consists of an encoder, a link fixed to the ground, and a link that can be rotated around pivot motion center C (Fig. 6). In the test, link 2 rotated three continuous cycles, from the neutral position to one side (where the sensor was in tension) and then returned to the other side (where the sensor was under compression). Each cycle can be divided into four stages: 1) in tension – loading, 2) in tension – unloading, 3) under compression – loading, and 4) under compression – unloading. To show the consistency between the testbed and the geometry feature mentioned in Section II.C, points A, C, D , and holes in link 1 and link 2 (Fig. 6c) are mapped with those in Fig. 3, and $AC = CD = 3.5\text{mm}$.

Figure 6. Test process for model verification: (a) neutral position, (b) one side where the sensor is in tension, (c) the other side where the sensor is under compression. The sequence for one cycle is labelled as (a)-(b)-(a)-(c)-(a).

During rotation, the rotation angle of the joint was recorded by an encoder (*Scancon 2RM3600D*), and the wavelength data was collected by an interrogator (*Micron Optics SI-255*) in real-time. The wavelength shift $\Delta \lambda_B$ and the rotation angle α for these three cycles are plotted in Fig. 7. As can be seen, the wavelength shift can closely track the rotation angle.

Figure 7. Sensor's response to the rotation angle

Data for the first cycle was then used for model verification. Linear fitting, $\tan(\alpha/2) = k\Delta \lambda_B + b$, was applied for this cycle, where k was obtained as 0.1052 and b was -0.0114 (Fig. 8). This is consistent with Eq. (9), i.e., $1/c$ keeps as a constant 0.1058 for this fabricated sensor no matter which direction it rotates to. In this cycle, the RMSE was 0.0273, and the maximum hysteresis was 4.82% ($\sim 5.56^\circ$).

A dead zone $[-3.50^\circ, 3.52^\circ]$ was observed when $\tan(\alpha/2)$ is near to zero, where the wavelength shifts little while the rotation angle changes. This is caused by the clearance between the sensor (width: 0.9mm, height: 0.5mm) and the hole (inner diameter: 1 mm) in link 2: the sensor will only start to move when it touches the inner wall of holes. This is also the reason for RMSE, hysteresis, and the unexpected intercept

value b . Meanwhile, the bonding condition between the fiber and epoxy may also cause the hysteresis problem.

After converting $\tan(\alpha/2)$ into α using Eq. (10), a comparison between the calculated angle through modeling and the actual angle was carried out for the first cycle (Fig. 9), with an RMSE of 2.99° , which is less than the error in [5, 12].

Figure 8. Verification test result: $\tan(\alpha/2)$ with respect to $\Delta\lambda_B$.

Figure 9. Comparison result for 1st cycle: calculated angle v.s actual angle

B. Pre-calibration and Repeatability Test

Figure 10. Sensor pre-calibration and repeatability test result

Before its installation in a surgical instrument, a calibration and repeatability test were run for the rotation angle sensor using the same testbed. Another three cycles were conducted with a rotation range of $[-62.9^\circ, 75.5^\circ]$, which is larger than the measurable range in [17, 21, 22]. The result (Fig. 10) shows excellent repeatability of this sensor, with a hysteresis value of 6.81%. The calibrated sensitivity c in Eq. (8) was 9.1024 nm (note: $\tan(\alpha/2)$ is unitless), with an RMSE of 0.2184 nm indicating that the measurement angle error was $2\arctan(0.2184/9.1024) = 2.75^\circ$.

C. Integration with a Grasper in a Surgical Robot

The sensor was integrated with a robotic flexible endoscopic grasper (Fig. 11a). The rotation angle of the grasper was recorded by an encoder (Fig. 11b) and the corresponding wavelength information was collected by the aforementioned interrogator. The grasper made the pitch motion for five continuous times. This integrated sensor shows good repeatability, with a hysteresis value of 7.72% (Fig. 12). The measurable range of the rotation angle was from -47.8° to 39.1° after integration to the grasper, which is found to be smaller than that in the pre-calibration result. After installation, the sensor's movement is affected by the

surrounding sheaths, which may make it less flexible. In this test, the minimum bending radius was 6.9 mm, which is much smaller than the one of 12.8 mm in [21].

When the grasper rotated from the neutral position to -47.8° upwards, wavelength shift changed from zero to -7052 pm while when the grasper rotated from the neutral position to 39.1° downwards, wavelength shift changed from zero to 4914 pm. This indicates a rough sensitivity of 147.5 pm/degree for the upward motion and 125.7 pm/degree for the downward motion of the grasper, which agrees with the statement in Section II.D that this sensor has a higher sensitivity when the grasper is moving up. This sensitivity is higher than the one of 7 pm/degree in [22].

Figure 11. (a) Sensor's integration with an endoscopic surgical grasper, (b) a testbed for this integration.

Figure 12. Modeling on five continuous runs on the grasper

Model fitting (Eq. (13)) was applied to this integrated sensor (Fig. 12). After obtaining the values of c_1 , c_2 , and c_3 , a comparison study was conducted for two continuous cycles of pitch motion of the grasper, between the computed angle from the modeling and the actual rotation angle β . The RMSE was 3.34° (Fig. 13), which may come from the assumption of the modeling, i.e., in practice, the curvature radius is not constant because the sensor's movement will be affected by the surrounding sheaths. Nevertheless, this sensor still shows its capability to reflect the distal rotation angle in real-time.

Figure 13. Comparison result on the sensor-integrated grasper

IV. CONCLUSION

In this work, a bidirectional rotation angle sensor for the joints of a TSM-driven flexible endoscopic surgical robot was proposed. The sensor is a single FBG fiber embedded in a thin and flexible epoxy substrate. Modeling was conducted on this sensor: a linear relationship between $\tan(\alpha/2)$ and wavelength

shift was derived. This model was further verified using a testbed, with an RMSE of 2.99° in the comparison test. A calibration and repeatability test were carried out for the sensor before its integration in a robotic surgical grasper, with a high sensitivity of 9.1024 nm (note: $\tan(\alpha/2)$ is unitless), broad measurable range of $[-62.9^\circ, 75.5^\circ]$, and an RMSE of 2.75° . After installation in the grasper, it can reliably measure the bidirectional rotation angle of a joint, with a hysteresis of 7.72% , a minimal bending radius of 6.9 mm , and an RMSE of 3.34° . This proposed small sensor has a simple structure, flexibility, high repeatability, relatively low hysteresis, high sensitivity, and large measurable range. It is comparable to those sensors in the recent literature [21, 22], in terms of sensitivity, minimal bending radius and measurable range.

The measurable range of the sensor drops to $[-47.8^\circ, 39.1^\circ]$ after being integrated with the grasper. In the future, a more flexible epoxy material will be explored to increase the measurable angle range of the sensor. Sensors with shorter gratings (e.g., 1 mm) will be used to shorten the link length of the grasper further. A dual-FBG array will be applied to compensate for the temperature effect. Another sensor will be integrated into the grasper to sense the yaw motion so that the spatial motion of the grasper can be sensed. Meanwhile, in the models, constant curvature along the body of the sensor is assumed, which lead to angle errors. These errors can be reduced by considering non-constant curvature bend using large deformation beam theory. In addition, closed-loop motion control of the TSM-driven grasper will be carried out. This small and flexible sensor is suitable for surgical robots and manipulators whose articulated joints have a large rotation angle and small bending radius. It can also be used in soft robots with large bending deflection.

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