Intensity-modulated Bending Sensor Fabrication and Performance Test for Shape Sensing

Faisal ALJaber[1] and Kaspar Althoefer [2]

[1] Queen University of London, Mile End Rd E1 4NS, UK
[2] Springer Heidelberg, Tiergartenstr. 17, 69121 Heidelberg, Germany
lncs@springer.com

Abstract. Notable advancements in shape sensing for flexible continuum robot arms can be observed. With a keen interest to develop surgical and diagnostic tools that can advance further and further into inaccessible spaces along tortuous pathways, such as the human body, a need for the precise determination of the robot's pose arises. Whilst there have been techniques developed that use external sensors to observe the advancing robot from the outside to determine its location and orientation in space, there is an observable trend towards using integrated, internal sensors to measure these positional parameters. Especially in the medical world with its tough requirements on robot size, e.g., catheter-type robots, most pose-sensing approaches to date make use of a technique called Fiber Bragg Grating (FBG). FBG sensors make use of fibers that are grated, and the amount of bending can be determined with an appropriate optical interrogator. Although these fiber sensors have been successfully employed to measure the deformation and through advanced signal processing the pose of continuum catheters, they have a major drawback which is their exorbitant cost. To address this issue a different design and fabrication process is proposed to produce an affordable shape sensor that is highly flexible and can detect bending. The method of operation involves a segmented flexible robot arm with three waveguides in a 120-degrees configuration. The segments are made of silicon elastomer with channels that encapsulate light propagating internally, with a photodiode and light-emitting diode (LED) embedded in each individual channel. The prototype was developed and characterized for strain, and bending response detection.

Keywords: Shape sensing, Optical Fiber, Intensity modulated shape sensor, Soft robotics, Fiber Bragg Grating (FBG), PDMS.

1 Introduction

Minimally invasive surgery provides tremendous advantages compared to what conventional open surgery can offer in terms of blood loss, trauma, and recovery time. Therefore, laparoscopic techniques with multiple rigid surgical tools have found to be accepted and appreciated by the surgical community. Nevertheless, with the advancement of technology of laparoscopic techniques, nowadays it is more likely to design compact and highly flexible surgical instruments with high dexterity that can be deployed in Natural Orifice Transluminal Endoscopic Surgery (NOTES), additional reducing tissue damage and recovery time.

The main drawbacks associated with flexible robotic surgical instruments is the lack of two main features, (1) accurate position feedback and (2) information about the interaction force between the instrument and the tissue. Traditional optical and electro-magnetic tracking systems are of restricted applicability due to their larger size and line of sight limitations, and exposure to magnetic field and general susceptibility to noise, respectively. Hence the necessity for a precise and small sized sensor to obtain the reconstruction of the shape of a miniaturized and highly flexible robotic instrument arises. It is noted that position estimation is as important as to estimate the force applied to the tissue by the instrument, to reconstruct the true 3D trajectory and dynamic configurations of the robot arm or catheter.

In order to enable a flexible minimally invasive surgical continuum robot to possess an accurate steering, information about its shape in 3D environment is required. In the previous section, imaging techniques and shape reconstruction modalities were discussed. These modalities and techniques are divided to two main categories. The first category is concerning approaches allowing shape reconstruction using imaging-based modalities such as ultrasound, x-ray, computed tomography (CT) and magnetic resonance imaging (MRI). These imaging-based shape reconstruction techniques pose a lot of technical challenges. Ultrasound has low image resolution and tissue contrast, x-ray and CT will risk exposing the patients to high dosage of ionizing and radiation, and MRI has low refresh rate which makes it time consuming process. In addition, most of these modalities require a large workspace, their inherent characteristics limit their implementation in real-time applications, and they are not cost efficient. The second category includes optical fibers utilizing FBG shape-based sensors and EM-tracker for shape reconstruction and bending profile estimation. Unlike the pervious imagining-based modalities, FBG sensors and EM-trackers are miniaturized shape sensors, they have small footprint and they are integrated with the actual device, and, hence, they are suitable to be used in a confined workspace. Also, FBGs and other fiber-based approaches are well-known to be compatible with other magnetic or electrical shape reconstruction modalities.

Furthermore, both techniques are easy to be integrated into intraoperative procedures for their accessibility and free from line-of-sight restrictions. However, both techniques have their own limitations and disadvantages. EM-tracker technique is sensitive to distortions imposed by metallic objects and electrical/magnetic noise close to their field of generation, the workspace is constrained by the field of generation generated by EM tracker generator, and they tested to have better accuracy when used in combination with continuum robots with simplified design and computational load [1-5]. FBGs are well-known for being sensitive to both strain and temperature, and they measure mainly the axial strain. Moreover, FBG accuracy depends on the number of sensors, the placement position of the sensors along the instrument they are attached to, the manufacturing precision, the surrounding environment, and the stiffness of the instrument they are attached to. Finally, FBGs-based shape sensors tend to be very expensive which limits their use in general.

2 Background

The recent development and advent of soft robotics came as result of the necessity of sophisticated robotics systems that are developed to have close interaction with human beings within friendly and safe environment. The material selection and the components used in soft robots are different from those in conventional robots that are typically made of rigid components [6], [7]. The performance of sensors is considered generally to be one of the most essential elements in robotics and soft robotics. The fabrication of soft robots and associated sensors is largely based on a range of polymer materials, such as Polydimethylsiloxane (PDMS) and other types of commercialized silicone rubbers [25].

Nevertheless, one of the disadvantages of polymers is that they are not conductive; hence, integrated conductive media are required to communicate electrical signals along a soft material robot without compromising the flexibility and elasticity of the used rubber material. A most preferred conductive medium is based on liquid conductors due to their electrically-conductive property and high elasticity [8-10]. The liquid conductors can be easily sealed and encapsulated by closed microchannels inside a highly deformable soft sensor [25]. The encapsulated liquid inside the microchannels behaves like a sensing element by changing the resistance with respect to deformation of the sensor shape [11], [12]. For strain, pressure [9], [13], curvature and multi-axis shear forces detection many soft sensors developed utilize liquid metals [14], [15]. Nonmetallic liquid conductors, such as ionic liquids were proposed as a biocompatible option; however, their limited conductivity has constrained their use in applications [16-18].

Despite the many advantages of liquid conductors, there are several drawbacks that they are suffer from when they are coupled with soft materials. Liquid conductors require high pressure injection when introduced to a microchannel with a very thin needle; this is due to the hydrophobic nature of polymers that is injected inside a narrow space. The air captured during fabrication inside the microchannels makes it very challenging to control the injection process. Therefore, the manufacturing of such sensors consisting of microchannels is complicated by the fact that appropriate sealing of the injection ports needs to be ensured. Also, there is the risk of a mechanical failure that might result in leakage, and in case of using liquid metal this could be harmful and toxic if getting in direct contact with a living organism [25].

As an alternative to the microfluidic-based sensing approach, this paper proposes a soft sensor with channels that can encapsulate optical fibers. These fibers are used to transfer light from an LED at one end to a photoreceiver on the other end; fiber bending leads to a change in light intensity which is being detected by the photoreceiver. The fabrication for the proposed sensor aims to ease the manufacturing complexity and operational risks without compromising the mechanical properties desired. The idea behind this sensor is to exploit the optical fiber light conducting principles to estimate bending. The contribution of the paper is to propose an approach that novelty is the easy fabrication process of the soft sensor, setup of the segmented sensor and the range of the miniaturized sizes that is provided by the fabrication process. Optical fiber-based sensors are regarded to be a method suitable for shape deformation estimation and preferred to other approaches due to its electromagnetic interference immunity, as mentioned in the literature [19-21].

Besides their bio-compatibility, small size and, thus, suitability for various MIS applications, and compatibly with other magnetic or electrical systems, FBG shape-based sensors pose a lot of challenges. These challenges include loss of strain inaccuracies due to the surrounding environments that can directly interfere with wavelength detection, calibration issues, reduction in reflected wavelength transfer, geometrical inaccuracies, and high cost of interrogator equipment limiting the technology's accessibility and real-life implementations. These limitations are the main motive to develop a new sensor that is affordable, easy to manufacture, and capable of measuring the deformation of the flexible soft manipulator that the fibers are embedded with.

3 Design and fabrication of shape sensor

3.1 The Fabrication of Soft Silicone Segments

PDMS is a commonly used polymer especially because of its transparency and optical characteristics compatibility [22-24]. Nevertheless, PDMS is also known for its relatively low elongation at breaking point which is considered to be one of its main limitations. Even though, the availability of highly stretchable commercialized elastomers is relatively high, it is difficult to find one with high optical transparency. Therefore, the challenge here was to create a design which addresses the balance between optical transparency and stretchability. The approach employed by Celeste et al. is based on creating a hybrid material by mixing PDMS and commercially available silicone gel (EcoFlex Gel, Smooth-On) [25]. The mixing ratio of the material created was 9 parts of optical transparent PDMS and 1 part of soft stretchable silicone gel [25]. As Celeste et al. report, their five-hour fabrication approach consists of a three-step process for creating the sensor. Firstly, two separate moulds are used to fabricate a semi-circular waveguide and the bottom layer of the sensor. Secondly, an LED and a photodiode are embedded at opposite ends before curing the waveguide. Then, the waveguide gets coated by a thin gold leaf layer after curing and the complete waveguide is then placed on top of the bottom layer. Lastly, more substrate of the 9:1 mixture of EcoFlex Gel and PDMS is poured to cover the waveguide and complete the sensor housing [25].

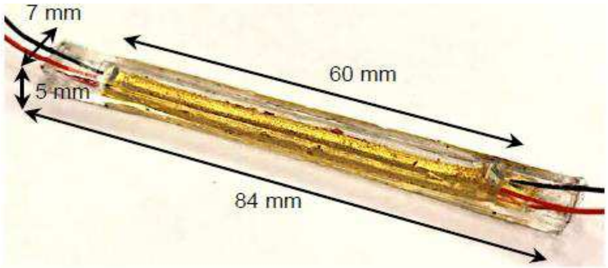

Fig. 1. Photo showing the soft pressure, curvature, and strain sensor by Celeste et al [25].

Although, Celeste et al attempted to design and fabricate a highly flexible optical sensor based on waveguide technique for strain, pressure, and curvature measurements. This sensor potentially was developed as an alternative for other stretchable sensors for shape reconstruction which will impose any harm if it came in contact with any human being. And the sensor design suggested was an attempt to replace the highly stretchable conductive liquid sensors, too. It is noted that, the liquid-based deflection sensors proposed by Celeste et al are toxic in nature unless nonmetallic liquid is used which reduces conductivity significantly. In addition, there is always a risk of leakage which might harm the patient if used in MIS procedures. Furthermore, the complicated liquid sensor fabrication process is considered to be time consuming, particularly when it comes to injecting the liquid to the microchannels. However, the waveguide-based highly stretchable sensor suffers from several constraints, such as the sensor rectangular design shape and size makes it considerably large for MIS applications. Also, the thin gold layer which plays the role of the reflective interface is susceptible to microcracks allowing partial escape of the light resulting in light intensity reduction [25]. These microcracks created by stretching the sensor will be permanent making the gold coating unsustainable for long term usage of the sensor.

The mixture ratio of the materials used to create our proposed flexible soft arm is 1A:1B of two silicone rubber (Dragon Skin 10 Fast, Smooth-On) a soft silicone rubber (A) and fast cure silicone (B), respectively. The mixture ratio was achieved by using a volumetric plastic measuring cup and a digital scale to ensure equal added volume of both silicone rubber A, and B. 30ml of each induvial material is added, then a silicone thinner was added and the mixture stirred thoroughly before dispensing. The amount of silicone thinner added is 10% of the total amount of A+B added together (which is 6ml). It is advisable for the silicone thinner not to exceed the recommended 10%. Table 1 below shows the properties of the mixture before and after adding the silicone thinner. Using the silicone thinner has some advantages, such as lower mixed viscosity of the created material which means fast de-air of the silicone while vacuuming. The silicone thinner helps the mixture to flow better over the details of the mold. Also, the life expectancy and working time is proportional to the silicone

6

thinner applied. However, the tensile strength and the ultimate tear will be reduced proportionally to the amount of silicone thinner added.

Table 1. The silicone material mixture used properties before and after adding the silicone thinner.

Dragon Skin 10 Fast (A+B)	The hybrid mixture with 0% Silicone Thinner	The hybrid mixture with 10 % Silicone Thinner
Shore A Hardness *(ASTM D-2240)*	10A	>decrease
Tensile Strength *(ASTM D-412)*	475 psi	>decrease
Elongation at Break % *(ASTM D-412)*	1000%	<increase
Tear Strength (Die B) *(ASTM D-624)*	102 pli	>decrease
Mixed Viscosity (A+B) *(ASTM D-2393)*	23000 cps	>decrease

3.2 Mold Design

For us to be able to create the soft segments of the soft sensor molds were designed using computer-aided design software (SolidWorks). Five different molds were created with different diameters ranged between 7 mm to 3.2 cm, however only the segment with the largest diameter was used for this performance test. The mold has the following features as Fig.2 shows: there are channels created to accommodate optical fibers, magnet attachments, and the center supporting tensile spring which have the following diameters 3.6 mm, 5 mm, and 4 mm, respectively. The molds are designed in a way to make it easy to extract them after the curing time has passed and easy to be reused. The silicone segments are disposable and their diameter differ individually as Fig.3(a) shows. A single LED was fixed on a movable holder pointing towards a photodiode. At the initial position when there is zero distance between both the LED and the photodiode, the photodiode will read approximately 5 V. This experiment of selecting the suitable length of the segment for bending involves two parameters. Those parameters are the voltage is incremented by 0.5 V received by the photodiode and the corresponding distance between the LED and the photodiode. The distance is reverse proportional to the voltage signal received by the photodiode so the voltage target was set to be between 3 V and 1.5 V and it can be less than 1.5 V for better bending. The experiment was repeated four times for consistency and the results suggested that the best length is between 5 cm to 7 cm maximum because any length below 5 cm will make it hard to bend a single segment to maximum angle of 90-degrees.

7

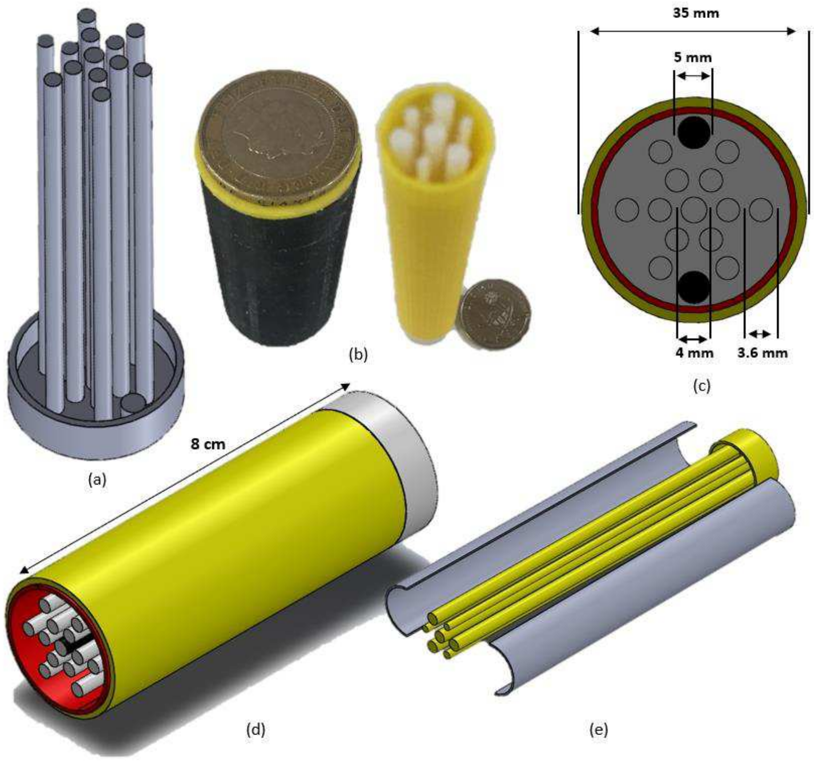

Fig. 2. (a) mold CAD design showing the columns that will create the channels. (b) different dimensions of molds created. (c) complete design of the mold with its parts diameters. (d) length of the mold. (e) inner part of the mold.

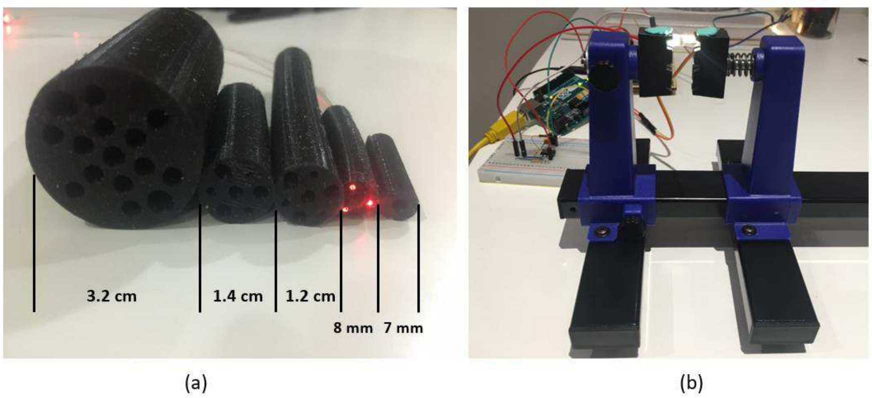

Fig. 3. (a) soft silicone segments with different diameters. (b) holder used to hole both the LED and to photodiode to decide the length of the silicone segment.

4 Concept

The principle of operation of our sensor depends on optical fibers since the sensor utilizes optical fibers to sense the change of shape of any flexible soft instrument that has the ability to bend. A single optical fiber consists of three main components: the core, the cladding, and the coating or buffer. Fig. 4 (a) shows the basic structure of an optical fiber. The core of an optical fiber is a cylinder-shaped rod made of dielectric material and is typically made of glass. Light beam propagates primarily along the core of the fiber [26].

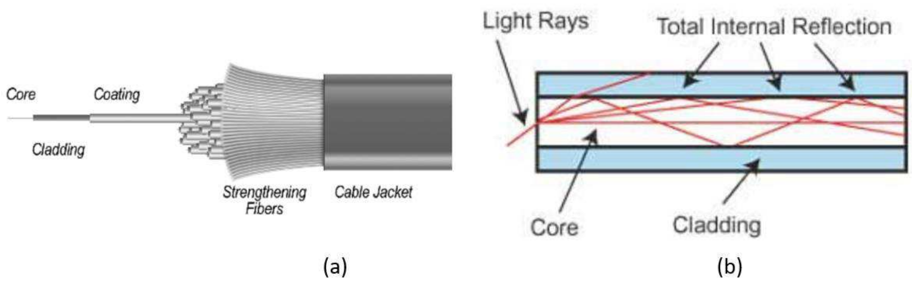

Fig. 4. (a) Basic structure of an optical fiber. (b) Total internal reflection in an optical fiber [26].

The cladding layer consists of a dielectric material. The cladding is generally made of glass or plastic and its main functionality is to execute such tasks by limiting the loss of light from the core into the air, decreasing the scattering loss at the core surface, and preserving the fibers from absorbing the surface impurities [26], [27].

The optical fiber is coated by a layer of material applied to protect it from any external impact or any physical damage. Usually the material used as a coating is a type of plastic and the coating is elastic in nature to avoid abrasions {Abrasions of what?} [26], [28-30]. Optical fibers achieve total internal reflection by using two different refractive indices between the waveguide and the outer cladding. At any angle of incidence that is greater than the critical angle, the light is totally reflected back into the glass medium (see Fig. 13). The total reflection of the light back into the glass medium would happen if the angle of incidence is greater than the critical angle (see Fig.4(b)) [26], [28-30].

4.1 Light Intensity Modulation

The sensor we are developing mainly based on the principle of light intensity modulation [31-33]. The intensity of the light emitted can possibly be modulated when an external force is exerted on the sensor designed. The measurements of the deformation were defined as the result of bending along each channel that occupies a fiber optic. For light detection and emission, a silicon PIN photodiode (OSRAM Opto Semiconductors, SFH 229) with a maximum wavelength sensitivity of 860 nm and maximum sensitivity of 1100 nm and a warm white color light-emitting diode (LED) (Hebei) were used, respectively. The surface mount of both LED and the photodiode is 3 mm facilitating a miniaturized sensor design. All photodiodes utilized in designing the sensor were operating in forward bias in which both the voltage signals output and the light intensity has a logarithmic relationship shown in the following equation [34]:

$$V = \frac{kT}{q}\left(\ln\frac{i}{i_s} + 1\right) \tag{1}$$

where V, k, T, q, i and i_s are the voltage output, Boltzman's constant, temperature, current, and saturation current, respectively. It is known to us that the current is linearly proportional to the light intensity. For our design of the soft silicon that goes under bending, the assumption can be made that any increase of the soft silicone structure deformation is inversely proportional to the light intensity, and that the ratio of the current can be replaced by the original and deformed surface areas ratio of the sensor. The assumption can be made that the surface area change is subjected to the change in length, the ratio is reduced further to that of lengths. The original length and length after deformation can be denoted by l_o and l, respectively. In addition, the expression $\frac{kT}{q}$ can be replaced by a single notation which is the initial voltage V_o for simplification. Finally, the theoretical model equation can be written as [34]:

$$V = V_o\left(\ln\frac{i}{i_s} + 1\right). \tag{2}$$

5 Results

The soft segment is tested by labeling each channel that contains an optical fiber and set between a photodiode from the bottom and an LED from the top end.

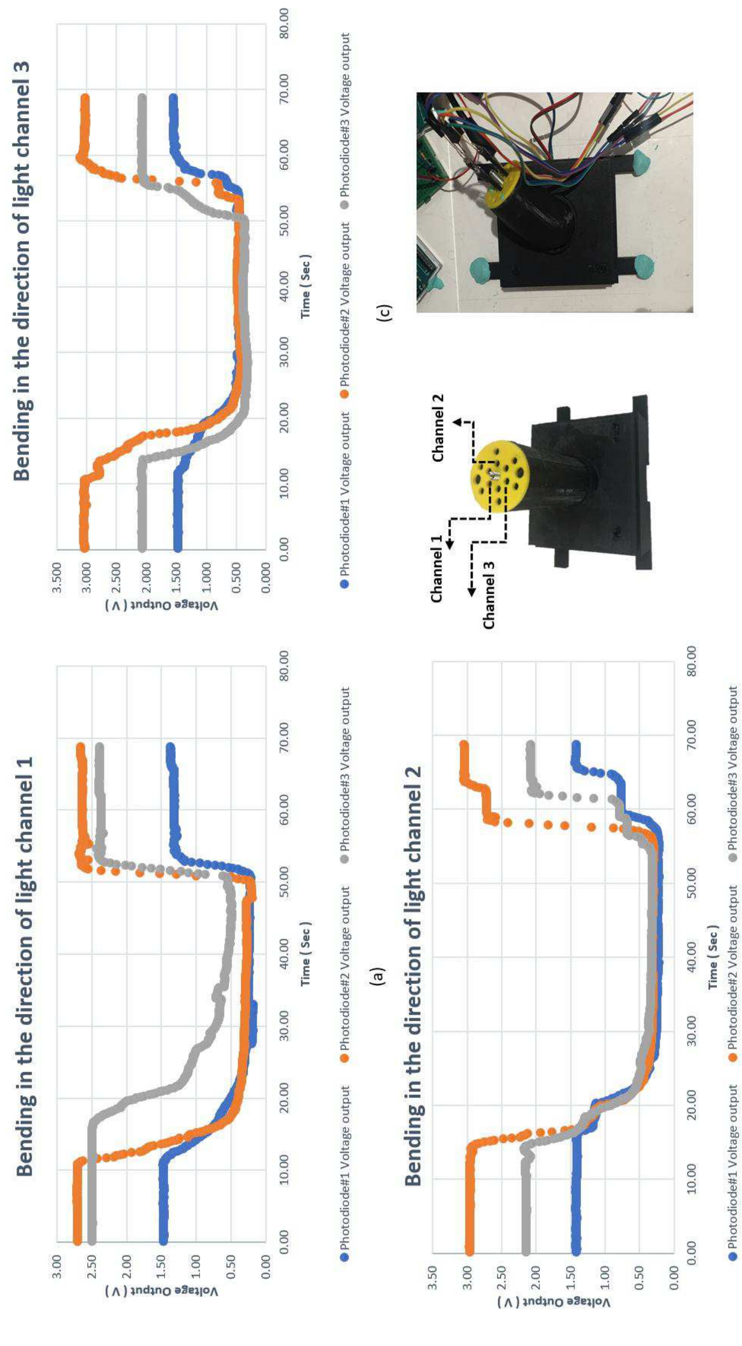

Fig. 5. (a), (b), and (c) is the voltage outputs of channels one, two, and three when the segment is bent. (d) channels locations. (e) the soft segment is bent 90-degrees.

The soft segment went under 90-degrees bending three times towards the direction of each channel. The performance test was done as following: (1) the recording of the voltage output starts from the initial position at 0-degree (2) then it is bent gradually until it reaches 90-degrees (3) finally, the soft segment is released to it is initial position. Each experiment is repeated five times for 75 seconds which explains the behavior of the plots shown in Fig 5 (a), (b), and (c). All three plots show similar behavior to bending when an external force is applied with minor delays of the third photodiode of plot (a) in Fig.5 and the first photodiode on plot (b) when it is rested after bending.

6 Conclusion

This design and fabrication of the soft silicone with channels that can contain optical fiber has the potential to test more than bending. We currently working on a simulation that can mimic the movement of the segment while goes under different type of deformations.

12

References

1. C. Shi, X. Luo and T. Fukuda, "Shape Sensing Techniques for Continuum Robots in Minimally Invasive Surgery: A Survey," IEEE Transactions on Biomedical Engineering, vol. PP, no. 99, 27 October 2016.
2. R. J. Roesthuis, M. Kemp and J. J. van den Dobbelsteen, "Three-Dimensional Needle Shape Reconstruction Using an Array of Fiber Bragg Grating Sensors," IEEE/ASME Transactions on Mechatronics, vol. 19, no. 4, pp. 1115 - 1126, 12 July 2013.
3. R. J. Roesthuis, S. Janssen and S. Misra, "On using an Array of Fiber Bragg Grating Sensors for Closed-Loop control of flexible minimally invasive surgical instruments," in IEEE/RSJ International Conference on Intelligent Robots and Systems, Tokyo, 2013.
4. S. Chang Ryu and P. E. Dupont, "FBG-based Shape Sensing Tubes for Continuum Robots," in IEEE International Conference on Robotics and Automation (ICRA), Hong Kong, 2014.
5. M. Abayazid, M. Kemp and S. Misra, "3D Flexible Needle Steering in Soft-Tissue Phantoms using Fiber Bragg Grating Sensors," in 2013 IEEE International Conference on Robotics and Automation, Karlsruhe, 2013.
6. S. Kim, C. Laschi, and B. Trimmer, "Soft robotics: a bioinspired evolution in robotics," Trends Biotechnol., vol. 31, no. 5, pp. 287–294, 2013.
7. C. Majidi, "Soft robotics: a perspective–current trends and prospects for the future," Soft Rob, vol. 1, no. 1, pp. 5–11, 2014.
8. Y.-L. Park, C. Majidi, R. Kramer, P. Berard, and R. J. Wood, "Hyperelastic pressure sensing with a liquid-embedded elastomer," J.Micromech. Microeng., vol. 20, no. 12, p. 125029, 2010.
9. Y.-L. Park, B. Chen, and R. J. Wood, "Design and fabrication of soft artificial skin using embedded microchannels and liquid conductors," IEEE Sens. J., vol. 12, no. 8, pp. 2711–2718, 2012.
10. C.-Y. Wu, W. H. Liao, and Y.-C. Tung, "Integrated ionic liquid-based electrofluidic circuits for pressure sensing within polydimethylsiloxane microfluidic systems," Lab Chip, vol. 11, no. 207890, pp. 1740–1746, 2011.
11. M. D. Dickey, R. C. Chiechi, R. J. Larsen, E. a. Weiss, D. a. Weitz, and G. M. Whitesides, "Eutectic gallium-indium (EGaIn): A liquid metal alloy for the formation of stable structures in microchannels at room temperature," Advanced Functional Materials, vol. 18, pp. 1097–1104, 2008.
12. T. Liu, P. Sen, and C. Kim, "Characterization of nontoxic liquid-metal alloy galinstan for applications in microdevices," J. Microelectromech. Syst., vol. 21, pp. 443–450, 2012.
13. R. P. Wong, J. Posner, and V. Santos, "Flexible microfluidic normal force sensor skin for tactile feedback," Sens. Actuators, A, vol. 179, pp. 62–69, 2012.
14. D. M. Vogt, Y.-L. Park, and R. J. Wood, "Design and Characterization of a Soft Multi-Axis Force Sensor Using Embedded Microfluidic Channels," IEEE Sensors Journal, vol. 13, no. 10, pp. 4056–4064, Oct. 2013.
15. C. Majidi, R. Kramer, and R. J. Wood, "A non-differential elastomer curvature sensor for softer-than-skin electronics," Smart Mater. Struct., vol. 20, p. 105017, 2011.
16. A. Visser, N. Bridges, and R. Rogers, Ionic Liquids: Science and Applications. American Chemical Society, 2012, vol. 1117.
17. J.-B. Chossat, Y.-L. Park, R. J. Wood, and V. Duchaine, "A Soft Strain Sensor Based on Ionic and Metal Liquids," IEEE Sens. J., vol. 13, no. 9, pp. 3405–3414, Sept. 2013.

18. J.-B. Chossat, H.-S. Shin, Y.-L. Park, and V. Duchaine, "Design and Manufacturing of Soft Tactile Skin using an Embedded Ionic Liquid and Tomographic Imaging," J Mech Robot, no. c, 2014.
19. K. Li, I.-M. Chen, S. H. Yeo, and C. K. Lim, "Development of fingermotion capturing device based on optical linear encoder," J. Rehabil. Res. Dev., vol. 48, no. 1, p. 69, 2011.
20. P. Kampmann and F. Kirchner, "Integration of fiber-optic sensor arrays into a multi-modal tactile sensor processing system for robotic endeffectors." Sensors, vol. 14, no. 4, pp. 6854–76, Jan. 2014.
21. A. Grillet, D. Kinet, J. Witt, M. Schukar, K. Krebber, F. Pirotte, and A. Depr´e, "Optical fiber sensors embedded into medical textiles for healthcare monitoring," IEEE Sens. J., vol. 8, no. 7, pp. 1215–1222, 2008.
22. S. Kopetz, D. Cai, E. Rabe, and A. Neyer, "PDMS-based optical waveguide layer for integration in electrical–optical circuit boards," AEU Int. J. Electron. Commun, vol. 61, no. 3, pp. 163–167, Mar. 2007.
23. D. Chang-Yen, R. Eich, and B. Gale, "A monolithic PDMS waveguide system fabricated using soft-lithography techniques," Journal of Lightwave Technology, vol. 23, no. 6, pp. 2088–2093, June 2005.
24. J. S. Kee, D. P. Poenar, P. Neuzil, and L. Yobas, "Monolithic integration of poly(dimethylsiloxane) waveguides and microfluidics for on-chip absorbance measurements," Sens. Actuators, B, vol. 134, no. 2, pp. 532–538, Sept. 2008.
25. C. To, T. L. Hellebrekers and Y.-L. Park, "Highly stretchable optical sensors for pressure, strain, and curvature measurement," in IEEE/RSJ International Conference on Intelligent Robots and Systems (IROS), Hamburg, 2015.
26. Fidanboylu, K. FIBER OPTIC SENSORS AND THEIR APPLICATIONS. 2009.
27. Méndez, A., Overview of fiber optic sensors for NDT applications, IV NDT Panamerican Conference, 1-11, 2007.
28. Jones, D., Introduction to Fiber Optics, Naval Education and Training Professional Development and Technology Center, 1998.
29. Jenny, R., Fundemantals of Fiber Optics: An Introduction for Beginners, Volpi Manufacturing USA Co., New York, 2000.
30. Tracey, P. M., Intrinsic Fiber-Optic Sensors, IEEE Transactions on Industry Applications, 27, 1, 1991.
31. P. Polygerinos, A. Ataollahi, T. Schaeffter, R. Razavi, L. D. Seneviratne, and K. Althoefer, "Mri-compatible intensity-modulated force sensor for cardiac catheterization procedures," IEEE Trans. Biomed. Eng., vol. 58, no. 3, pp. 721–6, Mar. 2011.
32. P. Polygerinos, L. D. Seneviratne, and K. Althoefer, "Modeling of light intensity-modulated fiber-optic displacement sensors," IEEE Trans.Instrum. Meas., vol. 60, no. 4, pp. 1408–1415, Apr. 2011.
33. P. Puangmali, K. Althoefer, and L. D. Seneviratne, "Mathematical modeling of intensity-modulated bent-tip optical fiber displacement sensors," IEEE Trans. Intrum. Meas., vol. 59, no. 2, pp. 283–291, 2010.
34. J. Fraden, Handbook of Modern Sensors, 3rd ed. New York: Springer- Verlag, 2004.