Sensors and Actuators A 153 (2009) 230–236

Contents lists available at ScienceDirect

Sensors and Actuators A: Physical

journal homepage: www.elsevier.com/locate/sna

Fabrication and characterization of laser-micromachined polypyrrole-based artificial muscle actuated catheters

Kenneth K.C. Lee[a], Nigel R. Munce[b], Tina Shoa[c], Luc G. Charron[b], Graham A. Wright[b,e], John D. Madden[c], Victor X.D. Yang[d,e,f,*]

[a] Engineering Science Program, University of Toronto, Toronto, Canada
[b] Department of Medical Biophysics, University of Toronto, Toronto, Canada
[c] Department of Electrical and Computer Engineering, University of British Columbia, Vancouver, Canada
[d] Department of Electrical and Computer Engineering, Ryerson University, Toronto, Canada
[e] Imaging Research, Sunnybrook Health Sciences Center, Toronto, Canada
[f] Division of Neurosurgery, University of Toronto, Toronto, Canada

ARTICLE INFO

Article history:
Received 23 October 2008
Received in revised form 23 April 2009
Accepted 12 May 2009
Available online 22 May 2009

Keywords:
Polypyrrole
Active catheter
Laser micromachining

ABSTRACT

Minimally invasive surgical tools using catheter based technology plays an important role in biomedical diagnostics and treatments. Much research has focused on producing an actively controllable tip catheter to improve accuracy and efficiency over traditional passive catheters when navigating inside patients. In this work, we describe the design, fabrication and characterization of a novel laser-micromachined polypyrrole (PPy) based active catheter. Two-dimensional controlled bending motion of a four-electrode catheter is demonstrated. Combining such catheter with optical coherence tomography, which can provide subsurface visualization of biological tissue, imaging capability using the active catheter tip is also demonstrated.

© 2009 Elsevier B.V. All rights reserved.

1. Introduction

In recent years, there has been increasing interest and demand for minimally invasive surgical (MIS) tools to aid with medical diagnostics and treatments. Many of these MIS tools involve the use of catheters, which depend on translational force and torque applied at the proximal end and rely on the intrinsic mechanical property of long flexible catheters to transmit the motion to the distal tip [1]. Hysteresis and recoil may decrease the controllability of these catheters. Hence they lack the accuracy and efficiency in reaching their desired destinations, leading to long procedure times, peri-operative risk, and other procedural complications.

Many advanced catheter designs aiming to provide an actively controllable tip have been proposed and developed. Microelectromechanical systems (MEMS) [2–4], shape memory alloys (SMA) [5–8] and artificial muscle actuators (AMA) such as ionic polymer metal composites (IPMC) [9–11], and conducting polymers (CP) [12–16] are among the leading designs as they possess attractive characteristics including large strain, low operation voltage and potential for miniaturization [7,11,15,17]. However, the fabrication process of MEMS devices is often complex. And unlike shape memory alloys, conducting polymers do not require high operating current [8,15], making them more suitable for *in vivo* biomedical applications. Furthermore, conducting polymers offer higher stiffness than IPMCs, which is an attribute often important in catheter design [18].

Polypyrrole (PPy) is one such type of conducting polymer, which distinguishes itself by additional features such as ease of fabrication and biocompatibility [15]. The PPy actuator typically exhibits strains of between 2% and 8% with corresponding stress of 5 and 22 MPa respectively [15,17,19]. The strains are associated with the insertion and removal of ions to and from the polymer as it is charged and discharged. A PPy-based active catheter has been recently developed for controllable intravascular maneuvers [20] in which the actuators work well in an antagonistic configuration such that one polymer releases ions and contracts while the other accepts ions and expands.

Current active catheter designs lack effective two-dimensional control of motion. Such motion would be of tremendous value in procedures such as angiography, stent deployment, and coiling of cerebral aneurysms. Laser micromachining by ablation is a well-established technique used for the production of 3D features in a wide variety of materials. It has been used in the fabrication of competing SMA-based active catheter designs [7] and many other medical devices, such as commercially available intravascular

---

* Corresponding author at: Department of Electrical and Computer Engineering, Ryerson University, 350 Victoria Street, Toronto, Canada.
E-mail address: yangv@ee.ryerson.ca (V.X.D. Yang).

0924-4247/$ – see front matter © 2009 Elsevier B.V. All rights reserved.
doi:10.1016/j.sna.2009.05.005

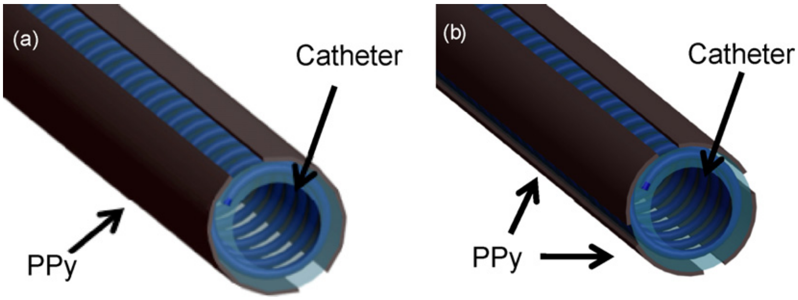

Fig. 1. Designs of PPy-based artificial muscle actuated catheters showing the PPy actuators grown onto a catheter with spiral skeletons and plastic substrates. (a) Two-electrode and (b) four-electrode versions.

stents. In most cases, these devices have much more complex geometries as compared to the linear micro-channel that is required for our proposed catheter design. We thus sought to investigate and characterize laser micromachining to enable precision manufacturing of active catheters capable of two-dimensional positional control.

In this paper, we describe a method of micromachining the polypyrrole coatings using pulsed laser ablation. To investigate the feasibility of this catheter design in intravascular guidance and navigation, we also characterize the bending movement of a two-electrode PPy-based catheter (Fig. 1a). The characterization is performed by monitoring catheter's radius of curvature and curvature difference as a function of time. To the best of our knowledge, we demonstrate here the first working prototype of a four-electrode PPy-based catheter (Fig. 1b) that is capable of two-dimensional motion on a plane, which may be suitable for use in intravascular guidance and navigation. We also use a two-electrode catheter in conjunction with an optical coherence tomography (OCT) system as a preliminary demonstration and exploration of the use of PPy-based catheter in intravascular forward imaging.

2. Fabrication method

The basic catheter used in this study is the Ultraflow^{TM} HPC (Micro Therapeutics Inc., Irvine, CA) with a 0.5 mm OD and 0.28 mm ID, chosen for its flexibility. Polymerization of PPy onto the catheter was performed in two steps—electro-less deposition followed by electrochemical polymerization. In the first step, the catheter was uniformly coated with approximately 1 μm PPy layer, which acts as the seed layer in the electrochemical deposition step. The electrochemical deposition was accomplished by polymerizing the pyrrole monomer through electrochemical oxidation using the method of Yamaura et al. [12]. For a resulting PPy thickness of approximately 10 μm we employed a current density of 0.125 mA/cm^2 at −30 °C for 6 h of deposition.

The PPy-coated catheter was ablated by a laser in order to electrically isolate individual segments of the PPy, thus patterning 4 electrodes. The laser micromachining setup consists of an excimer Laser (Pulsemaster series, Lumonics, Novi, MI) with a center wavelength of 248 nm, beam-shaping optics (10× demagnification) and a XY motorized stage (ATS100 Aerotech, Pittsburg, PA) for sample translation. To create the individual electrodes, an aluminum mask of a 1 mm × 1 mm square was used to obtain a 100 μm channel cut on the samples. A static etch rate characterization for the polymer coating was performed on a glassy carbon slide coated with PPy as shown in Fig. 2 to determine the optimal laser parameters to be used. The ablation was performed by using 50, 100 and 200 pulses focused on a planar sheet of PPy at different laser powers. The laser power was measured after the aluminum mask with a power meter (COHERENT Sigma power meter, COHERENT J25LP-MB power head) and the target fluence was subsequently calculated. The depth of the 100 μm × 100 μm square holes were then measured by optical profilometery (Veeco Instrument Inc., Plainview, NY), which used white light to provide depth profilometry at 50 nm resolution.

A custom-built plastic V-groove was designed to be fixed onto the laser system's translation stage to etch the coated catheter. A guide wire was inserted in the catheter to keep it rigid during the micromachining process. The catheter was fixed on the plastic jig with adhesive tape. A simple G-CODE program was written to cut a 30–40 mm straight line at a speed of 10 mm/min by translating the motorized stage. The translation speed was calculated based on the etch rate characterization for completely removing the polymer coating while simultaneously minimizing the ablation of the catheter underneath. Once a single line was etched, the sample was rotated by 90° and the process repeated. The resultant catheter with four independent actuator strips (Fig. 3a and b) was then tested via resistance measurements to ensure no short circuits were present.

For easy electrical wiring to each individual PPy actuator at one end of the catheter, ablation is repeated several times over the same length to ablate through both the PPy and the catheter underneath. This creates strips of PPy actuators that can be bend away from the catheter as shown in Fig. 4.

3. Bending characterization

A two-electrode catheter (Fig. 1a) was electrically driven with a 0.005 Hz, ±0.5 V square waveform in an aqueous solution of 1 M $NaPF_6$ salt measured against an Ag/AgCl reference electrode (BASi, West Lafayette, IN) (see Fig. 5a). In the data analysis, the catheter was split into two sections, the half of the catheter closer to the electrical contacts called the "root" and the remaining half called the "tip" (see Fig. 5b). Pictures of the bending catheter in either direction are also shown in Fig. 5b and c. Here we have used low frequency actuation in an attempt to observe the maximum achievable bending curvature since large degree of bending is highly desirable

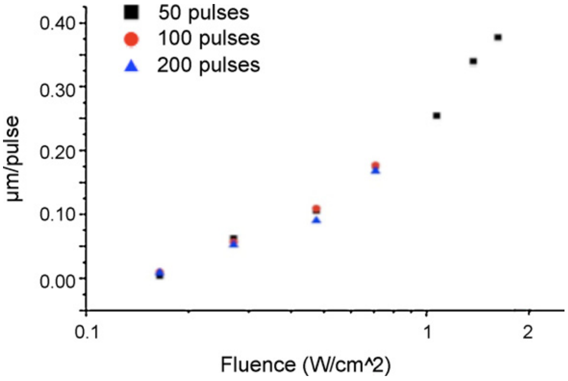

Fig. 2. Characterization of excimer laser ablation of PPy at different fluence rate and number of pulses used.

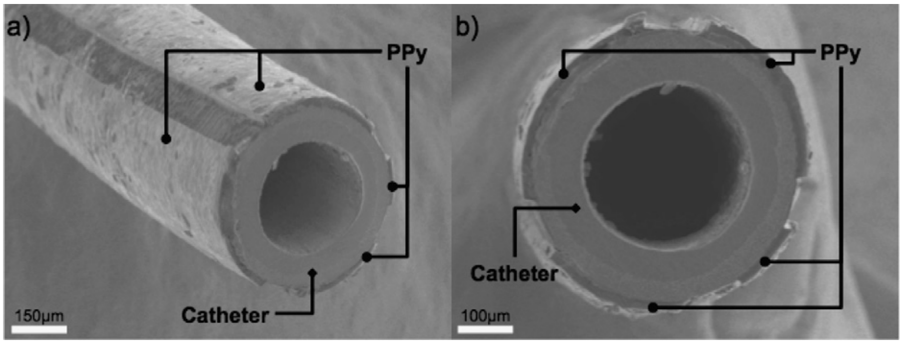

Fig. 3. SEM photographs of a four-electrode PPy-based active catheter, showing the catheter and the strips of polymer actuators after laser micromachining.

in intravascular navigation. The use of low voltage was to ensure the reusability of the same catheter for multiple trials as it has been previously reported that polymer degradation is likely to occur when voltage exceeds 1 V [16].

We captured the motion of the bending catheter with a digital video camera and then developed a LabVIEW^{TM} software program to analyze both the radius of curvature (R) of the tip and root segments as a function of time (see Fig. 5d). Immediately after the potential was applied, the root segment showed a decrease in R while the tip showed an increase in R in the negative direction, where the positive direction was taken to be to the right of the starting position. This was due to the initial catheter configuration in an S-shape as can be seen in Fig. 5a. The potential was reversed at $t = 100$ s, and both segments responded with immediate increase in R. In Fig. 5d, the gray areas are the transition bands. Each band indicates a time period in which the particular segment is switching between the negative and positive R, and vice versa. The minimum R of the root and the tip is 13 mm and 54 mm on the forward sweep, and 26 mm and 22 mm on the reverse sweep, respectively.

It has been shown that the curvature $((1/R)-(1/R_0))$ is linearly proportional to the induced polymer strain [21], and is an appropriate measure to analyze the electrochemical actuation response of the PPy-based catheter. Therefore, in Fig. 6a and b, we plotted both the experimental and the theoretical curvature as predicted by a simplified version of the variable resistance transmission line (VRTL) model. The model was developed employing different time constants to predict the time dependent response of PPy actuators [22]. It should be noted that the curvature is a relative measurement, which compares the curvature of the segment to its initial curvature. Therefore, a zero curvature does not necessary indicate that the segment is straight; instead it indicates that the segment has returned to the starting shape. In Fig. 6a and b, the discontinuities in the experimental curvatures correspond to the transition bands highlighted in Fig. 5d, during which the recorded radius crosses zero as the catheter changes its bending direction.

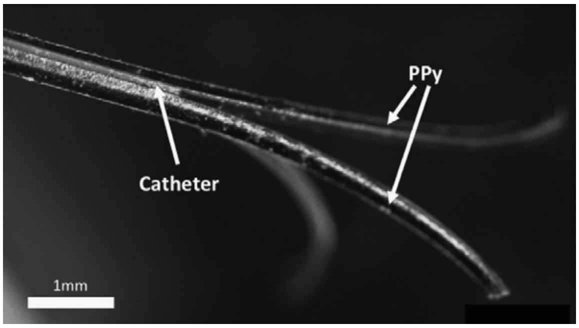

Fig. 4. One end of the PPy-based active catheter showing strips of PPy actuators ready for easy electrical wiring.

The bending of the root segment seems to be in close agreement with the model, except that the actual charging rate is slower than predicted (see Fig. 6a). For the tip, the amplitude of bending is less than expected, and the initial deflection of the tip starts earlier in time than expected in the model (see Fig. 6b). The overestimate of the tip bending appears to be due to a thinning of the active polypyrrole layer as the distance from the root is increased, which is likely the result of lower potential during electrodeposition. In the model it is assumed that the tip does not deflect until times exceeding a time constant proportional to the length squared and to the catheters electrochemical capacitance. The model assumes that the capacitance is the full capacitance of the catheter up to the start of the tip, which is an overestimate since in fact limited rates of ionic diffusion into the polypyrrole suggest that only a fraction of the total capacitance is accessible at short times and low frequencies. As a result, the time at which the tip bending begins is greatly overestimated by the model. The initially slower than expected deflection of the root is probably also due to the underestimate of the actively charging length of the catheter, with more charge going to the tip and less to the root than expected. Finally, there is a slow but steady change in deflection after the model predicts that steady state deflection is achieved. This is due to the use of a time constant to describe full charging, which in fact continues – albeit at a much reduced rate – at times after the charging time constant is reached. Furthermore, the model did not account for possible residual strain in the catheter at the start of the actuation. We hypothesize that the initial deflection of the tip was partially to overcome the initial residual strain and the bending was resulted from the bending of the root segment.

4. Initial demonstration of 2-directional actuation

For intravascular navigation in a 3D environment, torsional force would be required to provide additional degrees of freedom when using a passive catheter. A paradigm change will occur if one considers the use of active catheters. Each opposing pair of PPy actuators allows for one directional movement perpendicular to the length of the catheter. In order to minimize the need for torsional force, we proposed a four-electrode catheter, which would move in two independent directions under control. The two opposing actuator pairs of a four-electrode catheter (Fig. 3a) were electrically driven independently by 0.5 Hz, ±0.5 V square waveform and 0.02 Hz, ±0.5 V sine waveform in the x and y direction respectively. The salt solution and the reference electrode used were identical as above.

Four cycles of the actuation were recorded using a digital video camera, and analyzed using a modified version of the software program mentioned above, which tracked and recorded the center of the catheter tip. Fig. 7a shows the reconstructed trajectory of the catheter with distance measurements. The time interval between each point is 166 ms. A plot of x and y displacement as a function

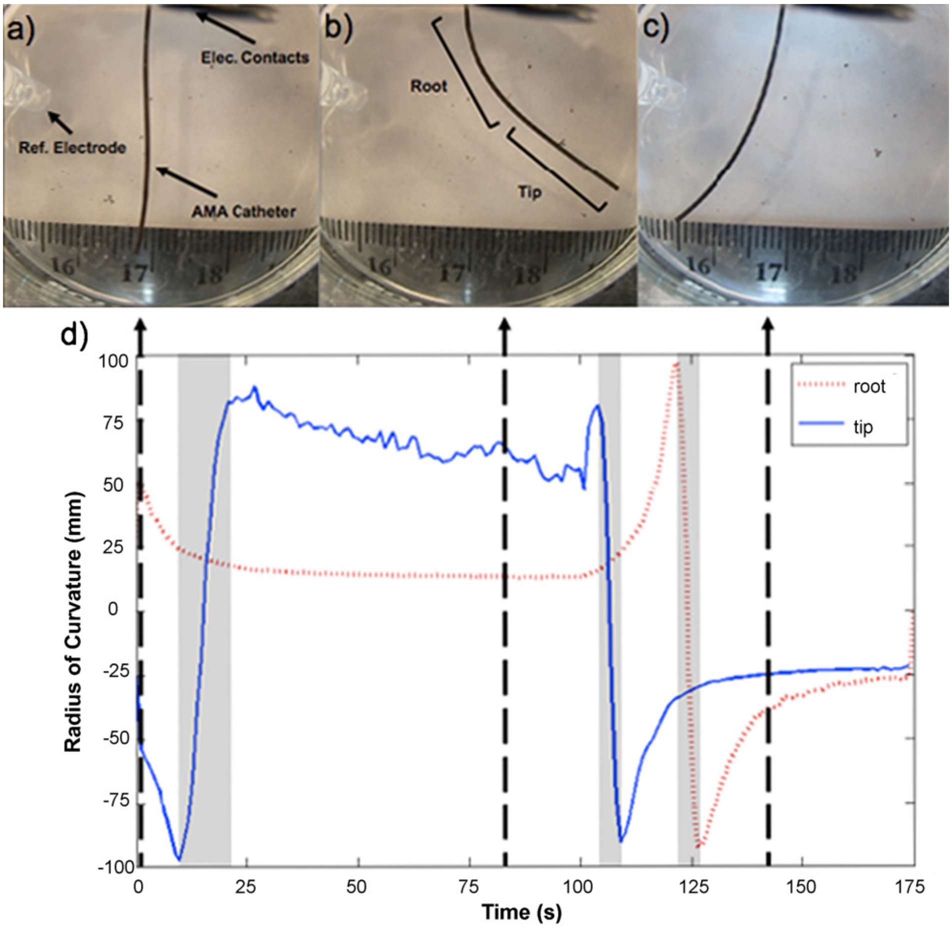

Fig. 5. Individual video frames of a two-electrode catheter at different times during actuation. (a) Starting shape, (b, c) shapes during the forward and reverse sweep, respectively. (d) Measured radius of curvature of the tip and the root of the catheter. The radius of curvature and the corresponding shape shown in (a–c) are indicated by the dashed arrows.

of time (Fig. 7b) displays the same frequencies as the driving waveforms.

Periodic position drifts can be observed in both scanning directions. More apparently, the frequency of the drift in the *x*-direction matches closely to the actuation frequency in the *y*-direction. This is mainly due to the imperfections during fabrication leading to the coupling of the motions in the *x* and *y* direction. As can been seen in Fig. 3, the four strips of PPy actuators are not equal in size and/or separated by exactly 90°. In addition, drift and hysteresis of PPy-based actuators have been observed in both fixed positioning [23,24] and oscillation experiments [24]. It has been suggested that if the drift is reproducible, then the level of DC bias required to counteract the drift can be experimentally determined for future uses of the same catheter. A PID controller based on a position feedback using a laser sensor has also been developed to minimize the drift [25].

5. Preliminary imaging and scan conversion

Previously, electroactive ionic polymer–metal composite cantilever has been reported for optical coherence tomography (OCT) [11]. Here we demonstrate laser micromachined PPy-based active

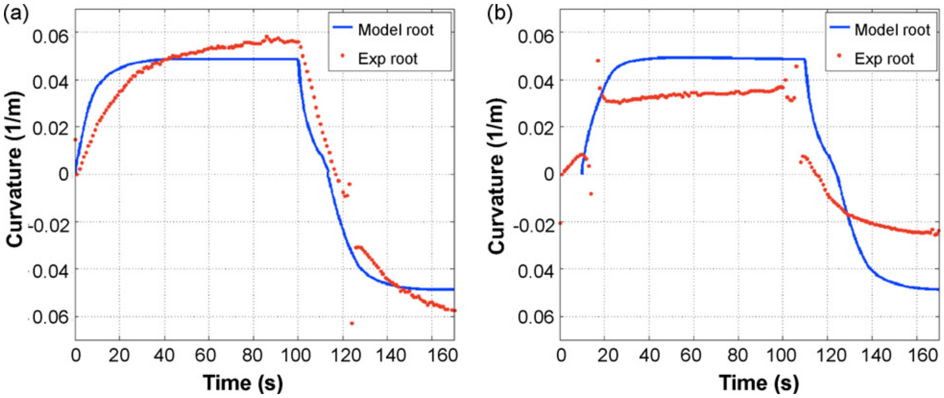

Fig. 6. Theoretical and experimental curvature difference of the root (a) and the tip (b) of the active catheter.

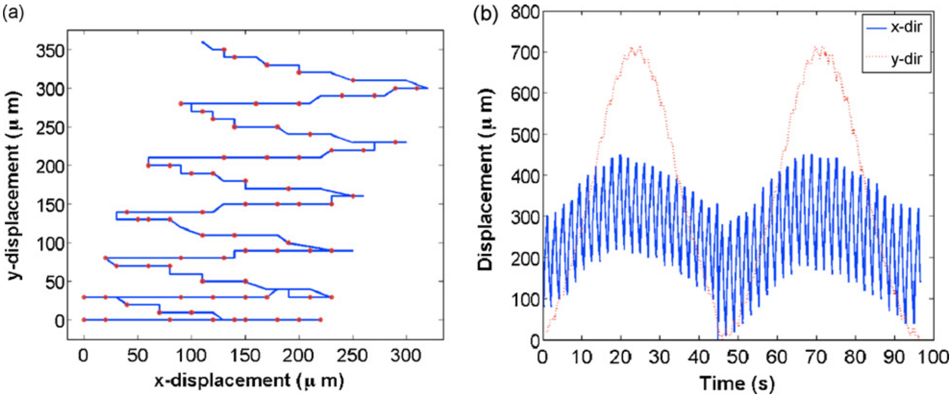

Fig. 7. (a) Reconstructed trajectory of the tip of a four-electrode PPy-based active catheter moving in two (*x* and *y*) directions to form a raster scan pattern. (b) Tip displacement in both the *x*- and *y*-direction as a function of time.

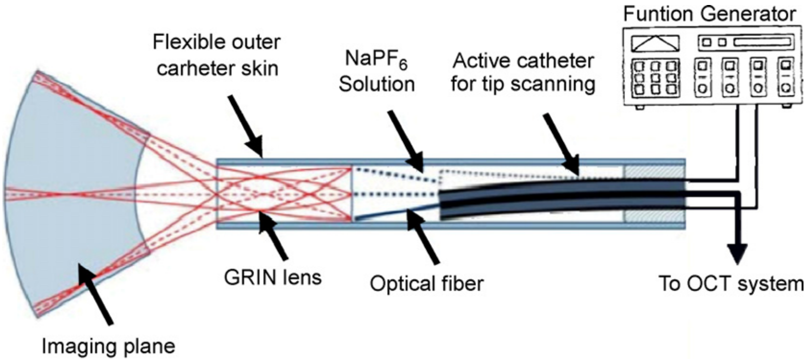

Fig. 8. Scheme of forward OCT imaging using PPy-based active catheter.

microcatheter can also be used for forward viewing optical coherence tomography (OCT) imaging. A single mode fiber (125 μm in diameter) was fitted through and fixed to the inside of the PPy-based catheter (280 μm inner diameter). The catheter and the fiber are then placed inside another catheter that has a GRIN lens mounted at the end to provide focusing of the light (see Fig. 8). A 1325 nm swept source OCT system (Thorlabs, Newton, NJ) was used to acquire, process and display data.

A B-mode OCT image of an infrared card was recorded at 0.1 fps, containing a collection of 20,000 axial lines shown in Fig. 9a and b in rectangular and sector scan displays, respectively. Sector conversion is necessary as the catheter scans in an arc as opposed to a straight line. In our sector conversion algorithm, we assumed the catheter is a stiff rod with a pivot point located 10 mm away from the sample, and the scanning, or actuation rate, was constant. We also approximated that the radial distance between the probe when perpendicular to the sample surface as 1 mm, and the effective sector scan angle as 25°. These assumptions and approximations may have been an oversimplification of the scan conversion process resulting in distorted sector display, as the curvature of the top surface layer of the infrared card was not fully corrected. Nonetheless, the structural information is retrained, and features of the infrared card are easily distinguishable.

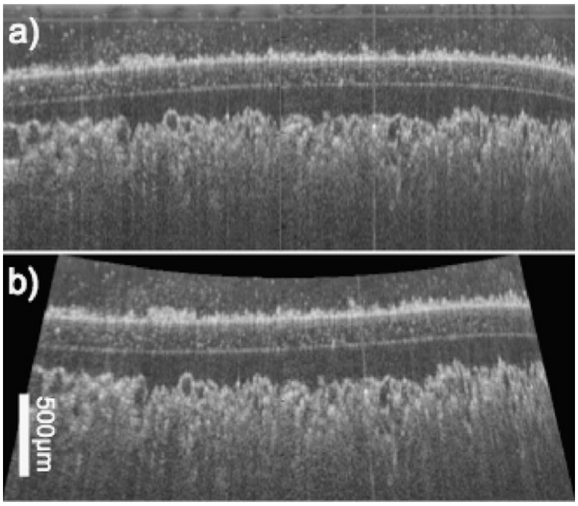

Fig. 9. (a) Rectangular and (b) converted sector optical coherence tomography (OCT) images of an infrared card.

An improved algorithm, which takes into account the non-uniformity of the actuation, can be developed with the help of a precise mathematical model and/or a position feedback control mechanism. In addition, position tracking methods either using digital video camera mentioned above or laser sensor [25] can be used for *ex vivo* applications, but will not be suitable for *in vivo* applications. Therefore, we propose implementing fiber Bragg gratings (FBG), which react to strain through optical spectral shift in reflectivity, onto the four-electrode catheter to provide real-time *in vivo* capable strain measurements for opto-mechanical feedback control of the bending actuation.

6. Discussion

Currently there exist limited intravascular forward OCT imaging techniques, some of which have been proposed previously by our group [26] and others [27,28] including the popular uses of MEMS mirrors in OCT endoscopes [29–31]. We presented our PPy-based active catheter as a novel technique but recognized that the slow actuation speed might be unsuitable for this particular application.

Therefore we intend to explore the possible use of resonant frequency to drastically increasing the scanning speed up to previously reported speed of 35–40 Hz [32,33].

PPy actuator's ability to operate in an aqueous environment is one of its advantages over competing materials. Together with its proven biocompatibility [15], it was thought to be able to actuate in blood. Though actuation in blood has not been demonstrated, it has been shown that the performance of PPy actuator is generally conserved in various complex buffer media [34]. In our video-based characterization, we have chosen to use saline because blood obscures the catheter thus making the analysis difficult. Furthermore, in case of poor performance in blood and/or for better biocompatibility, our group has considered encapsulation measures [20].

To address some of the major drawbacks of PPy actuator, namely speed and controllability, future work will include more elaborated modeling and geometry optimization. In this work, we have attempted to apply the 1D VRTL model to a cylindrical geometry and the results had shown areas that require improvement as we continue to modify the model to better fit a more complex and 3D geometry. At the same time, we will explore another existing phenomenological model developed using finite element method (FEM) for a very similar tri-layer PPy geometry [35]. Since the actuation speed is highly associated with the rate of ion flux in and out of the polypyrrole, we will explore ways to increase the porosity of the actuator to improve its response time. We will also attempt to develop a position feedback control mechanism based on fiber Bragg gratings, which can be used for *in vivo* applications to increase the controllability of the catheter by reducing or eliminating drift/creep and hysteresis.

7. Conclusion

We have demonstrated a method for producing PPy-based catheters, which are capable of two-dimensional active steering, using laser micromachining. The segmented bending characteristics of a two-electrode catheter show non-uniform activation of the PPy. The results were compared with the transmission line model developed for the PPy-based catheter. The curvature of the root segment was found to be in close agreement of the model, whereas, the curvature of the tip segment suggested that the residual strain and the bending coupling with the root segment should be included in the model to better predict the motion of the tip segment. Nonetheless, a similar two-electrode catheter was used with OCT to demonstrate the imaging capability of this design. Furthermore, the motion analysis of a four-electrode catheter reveals coupling between individual directional motions. By implementing a position feedback mechanism, independent uni-directional motions may be achieved.

With rapid improvement in both fabrication process and control mechanism, PPy-based catheter may have an important role in steering imaging probes or guide wires to enhance intravascular navigation, reduce operative time, and minimize vascular damage, during interventional angiographic procedures for cerebral, cardiac, and peripheral vascular diseases.

Acknowledgements

This research was supported by grants from the Canadian Institutes of Health Research (CIHR) and Natural Science and Engineering Research Council (NSERC). The authors would also like to thank Dr. Beau Standish for technical discussion and comments on the manuscript.

References

[1] Y. Haga, M. Esashi, Biomedical microsystems for minimally invasive diagnosis and treatment, Proceedings of the IEEE 92 (1) (2004) 98–114.

[2] K.-T. Park, M. Esashi, Active catheter with integrated circuit for communication and control, in: Proceedings of the IEEE Micro Electro Mechanical Systems (MEMS), IEEE, Orlando, FL, USA, 1999.

[3] K.J. Rebello, Applications of MEMS in surgery, Proceedings of the IEEE 92 (1) (2004) 43–55.

[4] H. Takizawa, et al., Development of a microfine active bending catheter equipped with MIF tactile sensors, in: Proceedings of the IEEE Micro Electro Mechanical Systems (MEMS), IEEE, Orlando, FL, USA, 1999.

[5] Y. Haga, et al., Minimally invasive diagnostics and treatment using micro/nano machining, Minimally Invasive Therapy and Allied Technologies 15 (4) (2006) 218–225.

[6] S.K. Lee, et al., Biomedical applications of electroactive polymers and shape memory alloys, in: Proceedings of SPIE—The International Society for Optical Engineering, San Diego, CA, 2002.

[7] A.T. Tung, et al., Laser-machined shape memory alloy sensors for position feedback in active catheters, Sensors and Actuators A: Physical 147 (1) (2008) 83–92.

[8] Y. Haga, Y. Tanahashi, M. Esashi, Small diameter active catheter using shape memory alloy, in: Proceedings of the IEEE Micro Electro Mechanical Systems (MEMS), IEEE, Heidelberg, Germany, 1998.

[9] B.C. Lavu, M.P. Schoen, A. Mahajan, Adaptive intelligent control of ionic polymer–metal composites, Smart Materials and Structures 14 (4) (2005) 466–474.

[10] K. Onishi, et al., Morphology of electrodes and bending response of the polymer electrolyte actuator, Electrochimica Acta 46 (5) (2000) 737–743.

[11] Y. Wang, et al., Low-voltage polymer-based scanning cantilever for in vivo optical coherence tomography, Optics Letters 30 (1) (2005) 53–55.

[12] M. Yamaura, T. Hagiwara, K. Iwata, Enhancement of electrical conductivity of polypyrrole film by stretching: counter ion effect, Synthetic Metals 26 (3) (1988) 209–224.

[13] A. Della Santa, A. Mazzoldi, D. De Rossi, Steerable microcatheters actuated by embedded conducting polymer structures, Journal of Intelligent Material Systems and Structures 7 (3) (1996) 292–300.

[14] A. Mazzoldi, D. De Rossi, Conductive polymer based structures for a steerable catheter, in: Proceedings of SPIE—The International Society for Optical Engineering, vol. 3987, 2000, pp. 273–280.

[15] J.D.W. Madden, et al., Artificial muscle technology: physical principles and naval prospects, IEEE Journal of Oceanic Engineering 29 (3) (2004) 706–728.

[16] T. Shoa, et al., Polypyrrole operating voltage limits in aqueous sodium hexafluorophosphate, in: Proceedings of SPIE—The International Society for Optical Engineering, San Diego, CA, 2007.

[17] J.D.W. Madden, P.G.A. Madden, I.W. Hunter, Conducting polymer actuators as engineering materials, in: Proceedings of SPIE—The International Society for Optical Engineering, San Diego, CA, 2002.

[18] J. Carey, A. Fahim, M. Munro, Design of braided composite cardiovascular catheters based on required axial, flexural, and torsional rigidities, Journal of Biomedical Materials Research—Part B Applied Biomaterials 70 (1) (2004) 73–81.

[19] M. Cole, J.D. Madden, The effect of temperature exposure on polypyrrole actuation, in: Materials Research Society Symposium Proceedings, Boston, MA, 2006.

[20] T. Shoa, et al., in: Engineering in Medicine and Biology Society, 2008. EMBS 2008, 30th Annual International Conference of the IEEE, Conducting polymer based active catheter for minimally invasive interventions inside arteries (2008).

[21] M. Christophersen, B. Shapiro, E. Smela, Characterization and modeling of PPy bilayer microactuators. Part 1. Curvature, Sensors and Actuators, B: Chemical 115 (2) (2006) 596–609.

[22] T. Shoa, et al., Rate limits in conducting polymers, Journal of Advances in Science and Technology 61 (2008) 26–33.

[23] G. Alici, N.N. Huynh, Performance quantification of conducting polymer actuators for real applications: a microgripping system, IEEE/ASME Transactions on Mechatronics 12 (1) (2007) 73–84.

[24] J.D. Madden, et al., Creep and cycle life in polypyrrole actuators., Sensors and Actuators A: Physical 133 (1) (2007) 210–217.

[25] Q. Yao, G. Alici, G.M. Spinks, Feedback control of tri-layer polymer actuators to improve their positioning ability and speed of response, Sensors and Actuators A: Physical 144 (1) (2008) 176–184.

[26] N.R. Munce, et al., Electrostatic forward-viewing scanning probe for Doppler optical coherence tomography using a dissipative polymer catheter, Optics Letters 33 (7) (2008) 657–659.

[27] S. Han, et al., Handheld forward-imaging needle endoscope for ophthalmic optical coherence tomography inspection, Journal of biomedical optics 13 (2) (2008) 020505.

[28] Y. Pan, T. Xie, Z. Wang, Detection of bladder tumors using optical coherence tomography, in: Proceedings of SPIE—The International Society for Optical Engineering, San Jose, CA, 2004.

[29] J.M. Zara, et al., Scanning mirror for optical coherence tomography using an electrostatic MEMS actuator, in: Proceedings of SPIE—The International Society for Optical Engineering, San Jose, CA, 2003.

[30] K. Kumar, T.E. Milner, J.J. Zhang, MEMS scanners enable in vivo 3-D OCT, Laser Focus World 44 (8) (2008) 87–90.

[31] L. Wu, H. Xie, A dual reflective electrothermal MEMS micromirror for full circumferential scanning endoscopic imaging, in: Proceedings of SPIE—The International Society for Optical Engineering, San Jose, CA, 2008.

[32] S.W. John, G. Alici, C.D. Cook, Validation of resonant frequency model for polypyrrole trilayer actuators, IEEE/ASME Transactions on Mechatronics 13 (4) (2008) 401–409.

[33] Y. Wu, et al., Fast trilayer polypyrrole bending actuators for high speed applications, Synthetic Metals 156 (16–17) (2006) 1017–1022.

[34] S.S. Pandey, W. Takashima, K. Kaneto, Conserved electrochemomechanical activities of polypyrrole film in complex buffer media, Sensors and Actuators B: Chemical 102 (1) (2004) 142–147.

[35] P. Metz, G. Alici, G.M. Spinks, A finite element model for bending behaviour of conducting polymer electromechanical actuators, Sensors and Actuators A: Physical 130–131 (special issue) (2006) 1–11.

Biographies

Kenneth Lee is an undergraduate engineering science student at the University of Toronto (UofT). He has been working as a research assistant on optical coherence tomography (OCT) related projects in the Department of Medical Biophysics at the UofT since the summer of 2007. He learned about electroactive polymer as artificial muscle actuator during his visit to the University of British Columbia during the summer of 2008. His interests include opto-electronics and surgical robotics. He is currently involved in the development of multichannel OCT system and polypyrrole-based active catheter.

Nigel Munce is a PhD candidate at the University of Toronto whose interests include vascular imaging and biology as well a medical device development. He obtained his Master of Science from the Department of Medical Biophysics at University of Toronto in 2004 on the development of a “lab-on-a-chip” device for parallel single cell capillary electrophoresis. He is currently finishing writing his PhD thesis.

Tina Shoa was born in Tehran, Iran in 1977. She received a BSc degree in electrical engineering from Iran University of Science and Technology (IUST) in 2000 and an MSc degree in electrical engineering from University of Manitoba (U of M) in 2004. She is currently a PhD student at the University of British Columbia (UBC) where she is exploring conducting polymer actuators.

Luc Charron is a PhD candidate at the University of Toronto in the Department of medical biophysics. He obtained his MSc and BSc in physics at Ottawa University where he studied optical properties of semiconductors. His current research involves developing microfluidic/optofluidic technologies for single cell analysis. He specializes in laser microfabrication techniques such as excimer laser ablation and ultrafast photoresist-exposures.

Graham A. Wright received his PhD in electrical engineering from Stanford University in 1991 and has been on the faculty of the Department of Medical Biophysics at the University of Toronto since 1993, currently at the rank of professor. He is also the research director of the Schulich Heart Program at Sunnybrook Health Sciences Centre and the Chair of the Ontario Consortium in Imaging for Cardiovascular Therapeutics. He has made substantial contributions to the characterization of vascular and myocardial pathophysiology with MRI. His current research focus is imaging for cardiovascular interventions.

John D. Madden was born in Ottawa, Canada in 1968. He received a BSc in physics from the University of British Columbia (UBC) in 1991, an MEng in biomedical engineering from McGill University, Montreal, in 1995 and a PhD from the BioInstrumentation Lab at MIT in 2000. Dr. Madden left MIT for UBC in 2002 and is currently an assistant professor of electrical and computer engineering. His research involves fundamental studies and applications of polymer and nanotube actuators and electronic devices.

Victor X.D. Yang received his BASc in biomedical engineering, MASc in electrical and computer engineering, PhD in medical biophysics, and MD in medicine, all from University of Toronto in 1997, 1998, 2004, and 2006, respectively. He is currently an assistant professor in the Department of Electrical and Computer Engineering at Ryerson University, Toronto, Canada. Since 2008, he holds the Tier II Canada Research Chair in Biophotonics and Bioengineering. He is a member of the Canadian Medical Association and IEEE, with keen interest in interventional neuroradiology.