Towards a continuum variable-stiffness robotic arm for surgical applications

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TOWARDS A CONTINUUM VARIABLE-STIFFNESS ROBOTIC ARM FOR SURGICAL APPLICATIONS

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A thesis submitted to Nanyang Technological University in partial fulfilment of the
requirements for the degree of Doctor of Philosophy

2020

Statement of Originality

I hereby certify that the work embodied in this thesis is the result of original research, is free of plagiarised materials, and has not been submitted for a higher degree to any other University or Institution.

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~~*(A) This thesis **does not** contain any materials from papers published in peer-reviewed journals or from papers accepted at conferences in which I am listed as an author.~~

*(B) This thesis contains material from 3 papers published and 1 paper submitted in the following peer-reviewed journals/conference in which I am listed as an author.

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- I planned and prepared the manuscript.
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The contributions of the co-authors are as follows:

- I proposed the idea, formulated the problem, designed and conducted the experiments, and prepared the manuscript.
- Dr. L. Cao, Dr. T.N. Do, and Prof. S.J. Phee revised the manuscript.

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The contributions of the co-authors are as follows:

- I proposed the idea, initiated the design concept, designed and conducted the experiments, and prepared the manuscript.
- P.T. Phan helped to build the SolidWorks model and helped with the experiment.
- Dr. L. Cao, and Prof. S. J. Phee revised the manuscript.
- J. Lin helped with image processing.

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Table of Contents

Acknowledgements	i
Table of Contents.....	ii
Abstract.....	v
List of Figures.....	vii
List of Tables	xi
Chapter 1: Introduction.....	1
1.1: Background	1
1.2: Motivation.....	3
1.3: Objectives and Scope	4
1.4: Organization of Report	4
Chapter 2: Literature Review	6
2.1: Current surgical robotic systems.....	6
2.1.1: Laparoscopy	6
2.1.2: Flexible endoscopy.....	8
2.2: Actuators in surgical robots	10
2.2.1: Cable actuator.....	10
2.2.2: Fluidic actuator.....	14
2.2.3: Smart material actuator	17
2.2.4: Magnetic actuator	19
2.3: Overview of current approaches for variable stiffness	21
2.3.1: Structure-based methods	22
2.3.2: Material-based methods	31
2.3.3: Advantages and disadvantages of the current variable-stiffness approaches	37
Chapter 3: Conceptual Design, Preliminary Experiment, and Modelling of a Variable-Stiffness Manipulator	39
3.1: Conceptual design	39
3.1.1: Material selection	39
3.1.2: Design concept	41

3.2: Preliminary experiments and results	42
3.2.1: Variable stiffness tube (VST) bending tests	42
3.2.2: Commercialized endoscope bending tests	44
3.2.3: Bending results and discussion	45
3.3: Testing and modelling of stiffness	49
3.3.1: Dynamic Mechanical Analysis (DMA)	50
3.3.2: Three-point bending results.....	51
3.3.3: Heat transfer modeling and simulation results.....	53
3.3.4: Mathematical interpolation for flexural modulus	58
3.4: Discussion and conclusion.....	62
Chapter 4: Temperature-dependent, Variable-stiffness Endoscopic Manipulator with Active Heating and Cooling	63
4.1: Variable-stiffness robotic arm design	63
4.2: Variable stiffness characterization and validation	66
4.2.1: Variable stiffness characterization	66
4.2.2: Variable stiffness validation.....	69
4.3: Heating and cooling experiments.....	74
4.4: Heating and cooling models.....	78
4.4.1: Heating and ambient cooling models.....	79
4.4.2: Heating and active air cooling models	81
4.5: Discussion and conclusion.....	85
Chapter 5: Conclusions and Recommendations for Future Work	87
5.1: Summary of contributions.....	87
5.1.1: A novel variable stiffness approach using PET	87
5.1.2: A novel variable-stiffness manipulator with active heating and cooling.....	88
5.2: Advantages and disadvantages	89
5.2.1: Design.....	89
5.2.2: Modeling	90
5.2.3: Control.....	90
5.2.4: Experiment	90
5.3: Recommendations for Future Work.....	90

5.3.1: The design optimization	91
5.3.2: Implementation of a real-time stiffness control scheme	91
5.3.3: Evaluation of the design in a more realistic surgical scenario	91
List of Publications	93

Abstract

Natural Orifice Transluminal Endoscopic Surgery (NOTES) is a potential paradigm shift of Minimally Invasive Surgery that makes use of natural orifices to access the peritoneum for surgery without leaving visible scars. With NOTES, there are various benefits for patients such as fewer complications, improved precision, and faster recovery. In NOTES, the endoscope needs to be sufficiently flexible to go through the tortuous paths inside the human body to reach the target therapeutic site; once at the target site, the endoscope needs to be stiff enough to withstand external payloads as a stable platform during tissue manipulation without unwanted bending of the endoscope tip. Thus, an endoscope whose stiffness can be adjusted on command is desired.

This thesis presents a novel variable-stiffness manipulator/endoscope and its application in NOTES. Two main objectives are identified: (i) design, testing, and modeling of a novel approach for variable-stiffness manipulators, for proof of concepts, using a thermoplastic material whose stiffness is tunable through temperature. The temperature is adjusted through the joule heat generated by applying electric current into flexible metal coils inside the manipulator. A model that represents the relationship between the stiffness and the variables like dimensions of the design, heating time, and applied current will be studied and verified with experiments; and (ii) design, testing, and modeling of a variable-stiffness manipulator by considering critical requirements on sizes, and efficiency of real surgeries in flexible endoscopy and NOTES. Driven by 8 tendons, the robot has two bending sections and can be bent into “S” shapes. The active cooling mechanism is designed to shorten the activation time. The thermal insulation is also included to keep the temperature of the outer surface in the safe range.

The comparisons in terms of the stiffness changing ratio and stiffness itself between the proposed design and the commercial endoscopes are shown along with extensive experiments. In particular, the manipulator has a high stiffness-changing ratio (22) between rigid and flexible states while that of its commercial Olympus counterpart is only 1.59.

Heating and cooling mathematical models are also derived and evaluated using the experimental data. The cooling time with the active cooling mechanism is 11.9s while that of passive ambient cooling is 100.3s. The thermal insulation layer keeps the temperature of the outer surface within the safe range (below 41°C) during operation. The results confirm the feasibility of developing a compact variable-stiffness endoscopic manipulator with significantly high stiffness-changing ratio (low stiffness in the flexible mode and high stiffness in the rigid mode), shortened time for switching the modes, and safe thermal insulation. The proposed manipulator may be used as a stable endoscopic platform in general flexible endoscopy and key-hole surgery.

List of Figures

Figure 2. 1. Some major milestones in surgical robotic development with illustrated systems.....	7
Figure 2. 2. Da Vinci Surgical System (Intuitive Surgical, Inc., Sunnyvale, CA).....	8
Figure 2. 3. The MASTER system including actuator housing, microprocessor, master console, and slave manipulator.....	9
Figure 2. 4. Different configurations for actuator-driven surgical robots with (a) Actuators is directly installed at the end effector and (b) Actuator is put far away from the end effector	10
Figure 2. 5. Tendon-routed pulley (upper panel) and tendon-routed sheath (lower panel) [25].....	11
Figure 2. 6. (a) the cable paths and bending mechanism [18], (b) the view of EndoWrist with cables and pulleys [19].	12
Figure 2. 7. (a) The master and slave transluminal endoscopic robot (MASTER) system [24], (b) Pulley hinge design for gripper [25].	13
Figure 2. 8. (a) FMA's structure [46], (b) Bending and stretching based on the supplied pressure [46].	15
Figure 2. 9. Miniature McKibben actuators (a) Structure of McKibben actuators, (b) The working principle of actuators [36].	15
Figure 2. 10. Surgical instrument with miniature McKibben actuators [49]	16
Figure 2. 11. Minimally invasive neurosurgical intracranial robot; (a) The brass MINIR design, (b) The actuation mechanism.....	18
Figure 2. 12. The schematic representation of laparoscopic tissue retractor [74].....	21
Figure 2. 13. The locking mechanism design. (a) Compliant mode when the air tube is empty. (b) Stiff mode when air tube is pressurized. (c) The prototype. (d) Cross-section view of one segment [83].	22
Figure 2. 14. The design of FORGUIDE. (a) The cross-section overview. (b) Shaft in the compliant state. (c) Shaft in the rigid state [84].	23
Figure 2. 15. The design of a stiffness-controllable link. (a) Overview of the drawing. (b) In and out air flows. (c) Silicon/mesh wall view [85].	24
Figure 2. 16. Details of the NeoGuide locking structure. (a) Flexible mode when all the cables are free to move. (b) Rigid mode when all the cables are tensioned [3].	25

Figure 2. 17. Design of a Stiffness-Adjustable Hyper-redundant Manipulator. (a) The variable neutral-line mechanism. (b) The robot prototype [93].	26
Figure 2. 18. Design and working principle of variable stiffness mechanism using a set of nested segments and a centric wire. (a) Flexible state when the tension wire is relaxed. (b) Stiff state when tension wire is pulled. (c) Free body diagram of the forces in the system [3].	27
Figure 2. 19. The design of HARP. (a) The overview of the two concentric snakes. (b) The HARP prototype [89, 94].	28
Figure 2. 20. The design of tendon-based stiffening soft manipulator. (a) The cross-section view of a segment. (b) The antagonistic actuation control scheme [86].	28
Figure 2. 21. The working principle of variable stiffness using vacuum-packed particles. (a) Compliant mode without vacuum pressure. (b) Rigid mode with applied vacuum pressure [100].	29
Figure 2. 22. The design of the stiff-flop manipulator. (a) The overall architecture of the stiff-flop manipulator. (b) The details of the design with the cross-section view.	30
Figure 2. 23. Design of endoport. (a) Actuation unit design. (b) Transmission of the motor. (c) Airflow path during the experiment. (d) Cross-section showing the material filling area [101].	31
Figure 2. 24. The flexural modulus (stiffness) variation of the glass/phase change materials with different temperature.	32
Figure 2. 25. The cross-section view of a variable stiffness catheter. (1) The shaft. (2) The electrodes. (3) Electrorheological fluid channel. (4) Balloon channel. (5) Working channel [105].	33
Figure 2. 26. The stiffness switching mechanism of the insertion tube [107].	34
Figure 2. 27. Design of a variable-stiffness catheter. (a) The prototype. (b) Several configurations that catheter can adapt [108].	35
Figure 2. 28. The schematic view of the design. (a) One segment design. (b) Two segments design.	36
Figure 2. 29. The variable stiffness fibers. (a) The fiber composition. (b) PLA-based fiber. (c) ABS-based fiber [111].	37
Figure 3. 1. Structure and working principle of the variable stiffness tube	42
Figure 3. 2. VST bending tests. (a) Instron 5569 material testing machine. (b) Bending test with a single PET tube. (c) Bending test with VST in glass state. (d) Bending test with VST in rubbery state.	43

Figure 3. 3. Olympus endoscope bending tests. (a) The middle of the flexible section. (b) 20cm from the end of flexible tip position. (c) 40cm from the end of flexible tip position. (d) 60cm from the end of flexible tip position.	44
Figure 3. 4. Bending test results for VST.....	45
Figure 3. 5. Bending test results for (a) locked endoscope and (b) unlocked endoscope at different positions.	46
Figure 3. 6. Bending test results for the endoscope at different positions and modes. (a) For the flexible section. (b) For 20 cm position. (c) For 40 cm position. (d) For 60 cm position.	47
Figure 3. 7. Stiffness comparison between VST and the endoscope.	49
Figure 3. 8. PET flexural testing results from the DMA machine.	51
Figure 3. 9. The Instron bending result with 0.4 A current applied	52
Figure 3. 10. 3-point bending results when state changes from rubber to stiff one.	52
Figure 3. 11. The cross-section geometry of VST.....	53
Figure 3. 12. The geometry of the coil and the tube with the same length and thickness.	56
Figure 3. 13. The transient temperature at different times with 0.4 A direct current applied and the coil shape (a) and the tube shape (b).....	57
Figure 3. 14. Transient temperatures from COMSOL simulation and experiments	58
Figure 3. 15. The interpolation results and the error with two different proposed functions.	61
Figure 4. 1. Schematic view of the robot's design. (a) Overview of the robot. (b) Detailed inside components. (c) Cross-section of the disk. (d) Robot's prototype.	65
Figure 4. 2. The cooling system including inlet, connecting, and outlet channels.	66
Figure 4. 3. The 3-point bending experiment set up.	67
Figure 4. 4. Force and displacement data from the 3-point bending experiment of the robotic arm.....	67
Figure 4. 5. The validation experiment setup. (a) Robot in the original shape. (b) Robot in the changed shape after applying force F.	70
Figure 4. 6. The image processing scheme. (a) Color image. (b) Binary image. (c) Noise removed the image. (d) Marker points reconstruction.	71
Figure 4. 7. The data points in the event of 1 N applied force. (a) The robot without PET tubes. (b) The robot with PET tubes in the rubbery state.....	72

Figure 4. 8. The reconstructed dots along the robot body. (a) With 1 N force applied. (b) With 3.5 N force applied. (c) With 16 N force applied.	73
Figure 4. 9. The manipulator with temperature sensors (a) and the diagram of the heating and cooling experiment set up (b). Sensors 1, 2, 3 are to measure the temperatures of the cooling air at the inlet, PET tubes inside the manipulator, and the outer surface of the insulation layer, respectively.	75
Figure 4. 10. The temperature outside the PET tube in the heating and cooling phases. Different electric currents were tested.	76
Figure 4. 11. The temperature outside the PET tube when heating with 0.3A current, air, and ambient cooling.	77
Figure 4. 12. The temperature outside the Aerogel surface in 3 different cases.	78
Figure 4. 13. The energy flows during the heating and ambient cooling process.	79
Figure 4. 14. Experiment data and results from models in the case of ambient cooling ..	84
Figure 4. 15. Experiment data and results from models in the case of active cooling	85

List of Tables

Table 2. 1: The classification of variable stiffness approaches with their advantages and disadvantages.....	38
Table 3. 1: Property of common thermoplastics	40
Table 3. 2. The dimensions of the studied VST	43
Table 3. 3: Flexural modulus of the endoscope at different sections	48
Table 3. 4: Heat conductivity coefficient of SUS304 with different temperatures.....	55
Table 3. 5: Heat capacity coefficient of PET with different temperatures.....	55
Table 3. 6: The resistivity of SUS304 with different temperatures.....	57
Table 3. 7. The identified coefficients for the functions given by Eqs. (3. 13) and (3. 14).	60
Table 4. 1: The dimensions of major components	65
Table 4. 2: The performance comparisons of several variable-stiffness manipulators.	69
Table 4. 3: The values of coefficients of the above-derived models.....	83
Table 4. 4: Root mean square errors from the models.	85

Chapter 1: Introduction

This chapter presents the background of the thesis: *Towards a Continuum Variable-Stiffness Robotic Arm for Surgical Applications*. Section 1.1 describes current trends in surgical robots with the special attention given to the actuation in NOTES systems. Section 1.2 introduces the motivation of the PhD work, while section 1.3 outlines the objectives and scope of the research. Finally, the organization of the report is shown in section 1.4.

1.1: Background

Robots have been revolutionizing our lives in different ways. Apart from extensively being used in industrial areas, the advanced robots are now also creeping into other various sectors such as education, aerospace, and agriculture. In addition, over the course of the last few decades, robots have been integrated into clinical rooms to advance minimally invasive surgery (MIS) procedure with numerous benefits like shortened hospital time, minimized post-surgery effects, better cosmetics, and reduced patient's discomforts [1]. In general, the MIS procedure can be categorized into several types such as extraluminal, intraluminal, transluminal, and hybrid approaches.

With the support of modern technology and science advancements such as medical imaging, computer assistance, and design, MIS has gained significant success and popularity. However, there are many inherent obstacles that need to be dealt with. In the body's confined space, it is challenging to precisely accomplish the surgical tasks without using instruments that possess adequate maneuverability, triangulation, and flexibility. In addition, the lack of position and haptic feedback is much more challenging. Natural Orifice Transluminal Endoscopic Surgery (NOTES) is a new paradigm that makes use of natural orifices to access the peritoneum for surgery without leaving visible scars. Current technologies for NOTES system limit it to transoral, transvaginal, and transanal avenues of access, but with miniaturization, even more, maybe possible [2].

In order to provide motion and force for devices to manipulate the tissues or pass through the gastrointestinal (GI) tract to examine the organs, the conventional design concept usually embeds ferromagnetic actuator to actuate the end effector. However, this approach is inapplicable for surgical cases due to bulkiness, heaviness, and large size. For surgical robots, the power supplies and actuators are usually installed far away from the end effector and surgical sites, leading to simplicity, miniature, and lightweight for the instruments once they are inside the human body.

In NOTES, the surgical arms must be small and flexible in order to adapt to natural human orifices and complex GI tract. Compared to the embedded ferromagnetic approach where the actuators are directly installed at joints, the remote transmission system like cable actuator will reduce the induced force that plays an important role in surgery. So, the requirement of small, flexible, and high payload for actuators-driven surgical robots needs to be met. Up to date, several actuators have been used as the main mode of motion and force transmission in surgical robots such as cable system (tendon, wire), fluidic actuators, and magnetic actuators.

During NOTES procedures, instruments are used in combination with flexible endoscopes to access the abdominal cavity via natural orifices. A flexible endoscope (or manipulator) with a camera, a light source, and a channel for liquid or gas is used to transverse through the winding and narrow channels in human bodies. The endoscope also provides channels for the surgical robot end-effectors, e.g., graspers or electrocautery knives, enabling the doctors to perform treatments. The flexibility of the flexible endoscopes causes many challenges like undesired bending or buckling during insertion due to the extremely compliant nature of the lumens, unwanted movements of the endoscope tip during the tissue manipulations. During the procedures, instruments inserted through the flexible endoscope will grasp and pull the tissue toward the endoscope tip. Nevertheless, because of the flexibility of the endoscope, the exerted forces to pull the tissue could result in endoscope bending instead. To overcome these issues, researchers have come up with the idea of using variable stiffness.

1.2: Motivation

In NOTES, the combination of the instruments and flexible endoscopes is employed to access the abdominal cavity via natural orifices. To successfully do that, medical tools have to be soft in order to travel through sinuous paths inside the human body without damaging tissues or causing the patient's pain. However, the flexibility of the endoscopes also causes several concerns such as undesired bending or looping due to the nature of the colon, unwanted movements of the endoscope tip during applications making the procedures inaccurate. Tissue manipulations in which the instruments are required to grasp and pull the tissue toward the endoscope tip are commonly encountered during the surgery. But, if the endoscope is too compliant, the exerted force in transverse direction from the instruments to endoscope could cause it to bend alternatively. Therefore, flexible endoscopes should be designed in a way that the shaft would be kept stationary on a certain trajectory. This requirement is more critical in NOTES than other intraluminal endoscopic procedures because the endoscope may need to lift or manipulate large and heavy abdominal organs, as identified by the American Society for Gastrointestinal Endoscopy/Society of American Gastrointestinal and Endoscopic Surgeons [3]. The stiffness variation in these two cases is significant, and most of the currently available structures fail to meet these two conflicting requirements, while variable stiffness can address this issue. For that reason, there has been a growing interest in variable stiffness topics in recent years.

It is clear that during the NOTES interventions challenges still exist since the endoscopes with constant stiffness do not have the right stiffness at all instances. In an attempt to address this paradox, variable stiffness endoscopes have been developed. It is reported that the main mechanism to adjust the stiffness is "through the channel" stiffening wires. By tensioning or loosening the central cable which is surrounded by a helical coil, the helical coil will be compressed or decompressed respectively, making the stiffness modifiable. By applying this technique, the variable stiffness colonoscope from Olympus (model XCF-SH230L) can achieve the stiffness ratio of 5 between two states, while the model CF-240AL of the same brand yields the ratio of 6 [4]. Although

variable stiffness endoscopes could improve the situation, the detailed designs and analysis are still lacking [4-6].

1.3: Objectives and Scope

The stiffness of flexible endoscope or surgical instruments is designed as a compromise between being compliant enough to adapt with complex, nonlinear colon curves and being stiff enough to enable pushing forward, bearing loads, and providing force for surgical tasks. Previous studies reported in the literature only consider the design concepts of variable stiffness. In addition, modelling, control, and application design problems of variable stiffness which have not considered yet play a more important role in surgical robotic applications. Therefore, the objectives of the PhD work are two folds:

1. To develop a new variable stiffness approach using thermoplastic materials which have different stiffness under changing temperature. In this research, PET (Polyethylene Terephthalate) is selected as the variable stiffness material and stainless steel spring is considered as a heating agent. Then, a variable stiffness model that defines the relationship between the stiffness (flexural modulus) and the input variables such as geometry dimensions, time, and applied current is investigated through the combination of experimental data and heat transfer analysis. Extensive experiments will be carried out to prove the feasibility of the proposed design concept.
2. Based on the design concept from the first objective, the second aim is to develop a continuum variable-stiffness tendon-driven manipulator for surgical applications. A series of experiments will be conducted to examine the effectiveness of the design. Moreover, the heating and cooling models are established based on heat transfer analysis and data collected from measurements. Experiments will be carried out to illustrate the performance of the design with the interaction force within the surgical range.

1.4: Organization of Report

The dissertation includes five chapters in total. Their organization is as follow:

Chapter 1 introduces the background of the surgical robot in recent years and the motivation of the research. The objectives and scope are also presented with the organization of the report.

Chapter 2 discusses the literature review of the study. It contains three major sections: current development in the surgical robotic systems, actuators in surgical robots, and overview of present approaches for variable stiffness.

Chapter 3 outlines the conceptual design that comprises the material section and the design concept. Several preliminary experiment setups are described along with the results and discussion. Stiffness testing and modeling are also reported with the support of Dynamic Mechanical Analysis (DMA), 3-point bending experiments, and heat transfer modeling.

Chapter 4 is devoted to showing the design of a variable stiffness manipulator with cooling and heating insulation. The details of the design are highlighted along with the characterization and the validation of the robotic arm. The analysis of heating and cooling will be also discussed with a series of experiments. After that, the heating and cooling models are derived and verified based on the heat transfer analysis and data collected from measurements.

Chapter 5 summarizes the conclusions with the contributions, and the recommendations for future work of the PhD research.

Chapter 2: Literature Review

This chapter discusses the literature review of the PhD work. It is divided into four main sections. In section 2.1, an overview of the current surgical robotic systems which can be categorized into two main groups, laparoscopy and flexible endoscopy, will be written. The review of actuators in the surgical robotic sector with their advantages and disadvantages along with special attention to cable actuation will be presented in section 2.2. Lastly, the review of current approaches in variable stiffness is introduced in section 2.3.

2.1: Current surgical robotic systems

Robot-assisted surgeries have been integrated and leading to a paradigm shift in surgical fields. With the emergence of Minimally Invasive Surgery (MIS), especially Natural Orifice Transluminal Endoscopic Surgery (NOTES), there are various benefits such as a minimization of side effects, enhancement of precise surgical procedures, and faster recovery after the surgery that patients can take from. In general, the MIS procedure can be categorized into several types such as extraluminal, intraluminal, transluminal, and hybrid approaches. An overview of the development of MIS is summarized in Figure 2. 1.

2.1.1: *Laparoscopy*

Based on the applied tools, surgical robotic systems can be divided into two main groups: laparoscopy and flexible endoscopy [7]. Take laparoscopic surgery into consideration, the instruments are usually rigid and long shaft with cameras integrated for visual display, which all enter the human body via incisions in the abdominal walls. In contrast, flexible endoscopy utilizes flexible endoscope with integrated camera and tool channels to allow flexible arms to access the human body via natural orifices or small incision. In the most common embodiment, instruments emerge from the tip of the endoscope into the camera's field of vision. These instruments are similar to those used in laparoscopic surgery, such as tissue graspers, electrocautery devices, and wire loops. Some typical laparoscopic systems have been commercialized such as Da Vinci surgical system [8], RAVEN and MiroSurge robot [1, 9], FreeHand and Telelap ALF-

X teleoperated surgical system [1], NeroArm and MrBot robot [1]. For illustration, Da Vinci surgical system which is the most famous platform will be introduced.

Figure 2. 1. Some major milestones in surgical robotic development with illustrated systems

Da Vinci surgical system which is shown in Figure 2. 2 from Intuitive Surgical Inc. is the only surgical robot with a thousand systems installed all over the world. It consists of surgeon console where the surgeon can sit and remotely control the surgical tools, patient-side cart which contains the robotic arms to hold and exchange the tools easily, vision system for observation enhancement during operation period, and EndoWrist instruments which are driven by cable pulley system and can provide up to 7 degrees of freedom (DOFs) with 90^0 of articulation. Initially, the Da Vinci Surgical System was cleared for laparoscopy and is now cleared by FDA for other different procedures [10, 11]. Although Da Vinci surgical system has been gaining huge success since initialized, it still presents some limitations like high-associated cost, the heaviness and bulkiness, lack of position and force feedback for the surgeon and sometimes inefficiency in flexibility because of rigid articulated tools, which is addressed by flexible endoscopy.

Figure 2. 2. Da Vinci Surgical System (Intuitive Surgical, Inc., Sunnyvale, CA)

(a) Surgical Console; (b) Patient Side-Cart; (c) 3-D vision; (d) EndoWrist instruments

To limit the port-site complications, reduce the discomforts, and eliminate the scars from laparoscopy, Natural Orifice Transluminal Endoscopic Surgery (NOTES) has been proposed. With NOTES, robotic arms are implemented in flexible endoscope to increase their effectiveness and safety, as well as to augment their therapeutic capabilities. The NOTES approach allows flexible tools to access the human body via natural orifices such as mouth, anus, vagina, and then follow the bending, tiny tract to do certain surgical tasks. The devices intended for NOTES are often flexible tubes integrated with instruments and visual systems inside. There is a diversity of surgical systems for NOTES. For example, Master and Slave Transluminal Endoscopic Robot MASTER system [12], Medrobotics Flex System [13], Viacath [14], USGI Cobra and R-Scope [15], Scorpion shaped robot [16], IREP [17], Hominis robot from MEMIC [18]. In this section, the MASTER system will be presented as a typical example of NOTES.

2.1.2: Flexible endoscopy

The MASTER system has been developed at Nanyang Technological University and commercialized under the company EndoMaster Pte Ltd for NOTES and Endoscopic Submucosal Dissection (EDS) applications (See Figure 2. 3). It utilizes a tendon-sheath mechanisms-driven flexible robotic manipulator to actuate multi DOFs of the robotic

arms integrated at the tip of a flexible endoscope. To operate the robot during operation, an endoscopist and a robotic operator are required.

Figure 2. 3. The MASTER system including actuator housing, microprocessor, master console, and slave manipulator.

The MASTER robot consists of a master console, a microprocessor and actuator housing, and a slave manipulator. The slave manipulator contains a camera to provide visual feedback to the surgeon and two flexible robotic arms including a grasper with the size of about 3.7mm and a cauterizing hook with the size around 2.8mm to perform surgical procedures. The master console maps the natural DOFs for rotation and translation motion of the human's arms to the slave manipulator via a motor housing. By attaching suitable sensors, this console can sense the movement from the surgeon and produce rough haptic feedback to the surgeon. The microprocessor, which is connected to the motion controller, can interpret the signals from encoders and loadcells located at the proximal end. The actuator housing, which contains tendon-sheath mechanisms, rotating drums, loadcells, and motors, is an intermediate interface between the actuator and the slave manipulator. The actuator housing receives the signals from the master console and directly drives the slave manipulator to carry out the necessary

treatments for patients such as cutting and suturing. Clinical demonstrations have been performed, such as NOTES liver resection and endoscopic submucosal dissection (ESD).

2.2: Actuators in surgical robots

In order to provide motion and force for devices to manipulate the tissues or pass through the gastrointestinal (GI) tract to examine the organs, the conventional design concept usually embeds ferromagnetic actuator to actuate the end effector. However, this approach is inapplicable for surgical cases due to bulkiness, heaviness, and large size. For surgical robots, the power supplies and actuators are usually installed far away from the end effector and surgical sites, leading to simplicity, miniature, and lightweight for the instruments once they are inside the human body (see Figure 2. 4).

Figure 2. 4. Different configurations for actuator-driven surgical robots with (a) Actuators is directly installed at the end effector and (b) Actuator is put far away from the end effector

Up to date, several actuators have been used as the main mode of motion and force transmission in surgical robots such as cable system (tendon, wire), fluidic actuators, magnetic actuators, and smart materials (shape memory alloy (SMA) or piezoelectricity).

2.2.1: Cable actuator

As a result of the development of flexible transmission systems, it is able to place the joints and application points far away from the actuator sites. Cable-driven actuation is capable of conveying position/force from external actuators to remote sites via flexible tubes or pulleys. This type of actuation can across confined space and manipulate

objects in complex environments. Due to their lightweight, simplicity, biocompatibility, safety, and flexibility cable systems have been used for various surgical robotic systems such as Da Vinci surgical system [8, 19, 20], endoscopes [20, 21], Medrobotics Flex System [13, 22, 23], MASTER system [24, 25], and Viacath system [14].

Conceptual design

In general, there are two main types of cable (tendon, wire) actuator configurations namely tendon-routed sheath and cable-routed pulley as shown in Figure 2. 5. Normally, one side of the cable is attached to the actuator and the other side is connected to the actuated joint. Once the cable is pulled at one side, it will slide inside the sheath or over the pulleys and transmit the position/force to the other side.

Figure 2. 5. Tendon-routed pulley (upper panel) and tendon-routed sheath (lower panel) [26]

Tendon-sheath mechanism (TSM) basically includes a hollow coil wire as a sheath and an inside cable as a tendon. The TSM not only operates in small working areas but also passes through long narrow and tortuous paths, and allows for simpler mechanical design as well as less spatial constraints on the operating environment. In addition, it does not require high electrical power or actuator to operate the joints. The need for flexible actuation with low bulkiness, simple design, high maneuverability, small size, and lightweight has made the TSM a very potential candidate for the power transmission in surgical systems [26]. As a result of many advantages, it has been widely used in many robotic applications and flexible surgical devices [14, 24, 27-29]. However, high

friction force and backlash hysteresis are still the underlying problems in the use of TSM.

On the other hand, tendon-routed pulley uses pulleys to support the tendon and transmit the force/motion to the remote joint. The main advantages of this design compared to the TSM are that it can provide higher force due to minimized friction effects and more predictable control. Although tendon-routed pulley offers advantages over the TSM [30-33], its rigidity and bulkiness result in the heavy systems and complexity. Hence, the use of tendon-routed pulley is limited in some surgical applications, especially in NOTES systems.

Cable-driven surgical devices

The Da Vinci slave manipulator has up to 7 DOFs (3 for orientations, 3 for translations, and one for grip) [8]. In its design, a majority of disks or vertebrae are stacked together to build the wrist mechanism.

Figure 2. 6. (a) the cable paths and bending mechanism [19], (b) the view of EndoWrist with cables and pulleys [20].

The multiple cables that can extend from the proximal vertebra to the intermediate or to distal vertebra are used for actuation. The holes or grooves with small or large radius can be seen as the actuator mechanism (see Figure 2. 6 (a)). In basic designs, the instrument's tip is controlled by looped cables and pulley systems, providing 2 rotational DOFs in two perpendicular planes as shown in Figure 2. 6 (b). There will be no backlash seen when the cables are fully tensioned and the EndoWrist can be considered as a stiff system since the steel cables and rigid links are highly stiff.

The MASTER robot which is given in the previous section utilizes TSM as the main mode of transmission. A recent prototype that is commercialized at EndoMaster is shown in Figure 2. 7. The slave manipulator includes two separate arms; one is an electrocautery hook for cutting and the other (the right arm in Figure 2. 7) is a grasper for manipulating the flesh of the patient. The two robotic arms with nine DOFs in total are fixed to the distal end of the endoscope (GIF-2T160, Olympus Medical Systems Corporation, Japan) and driven by tendon-sheath mechanism. At the distal end, the rotational joint is created by attaching the tendons to a pulley (Figure 2. 7 (b)).

Figure 2. 7. (a) The master and slave transluminal endoscopic robot (MASTER) system [25], (b) Pulley hinge design for gripper [26].

Advantages and disadvantages of cable actuator

According to a recent literature review, compared to other kinds of actuators, cable is most prominent candidate in surgical robots. In addition to lightweight, miniature, and flexibility, cable is very safe when being applied to the human body. On the one hand, it can be made from bio-material like stainless steel. On the other hand, in the worst case of failure as broken cable, the safety still can be guaranteed. In contrast, fluidic flexible actuators which will be given in the next section are operated with high pressure or smart material actuators that need high temperature or voltage. These types of actuators make them still challenges in surgical applications. Apart from many undeniable benefits of tendon-sheath abovementioned, there are still some underlying drawbacks as its nonlinear friction, backlash hysteresis, and poor force delivery resulting in the difficulties in modeling as well as control problems [34]. In addition, stiffness enhancements for surgical tools are also the main issues.

2.2.2: Fluidic actuator

Hold the same properties as cable actuators, flexible fluidic actuators also can provide high flexibility with the power supply being installed away from the joints. Otherwise, owing to high compliance and bio-compatibility with non-energized parts, they are excellent candidates for various medical applications such as miniaturized surgical manipulators [35-39], catheters [40, 41], graspers [42], positioning system [43], tissue removal tool [44], fluidic actuator for neurosurgical instrument [45], and hydraulic hood for ESD [46].

Conceptual design

Flexible fluidic actuators usually consist of flexible structures that can be deformed under the applied pressure from fluidic (i.e. pneumatic or hydraulic) actuation. Based on studies in [36], flexible fluidic actuators are categorized into two major groups that are bending due to internal chambers differently pressurized and anisotropic rigidity. Most of devices in the former group contain an elongated shape closed at one end and hindered radial deformation. When a chamber is pressured, its length increase while that of others is unchanged, making the whole device bending. In contrast, the tools in the latter category provide an elongated shape closed at one end and a rigid area that is stiffer than that of the rest. After being pressured, the length of stiffer area grows less than the rest, leading the desired bending motion.

Fluid-driven surgical devices

An electro-pneumatically or electro-hydraulically driven flexible micro-actuator (FMA) is introduced in [47, 48]. This actuator has total three DOFs (pitch, yaw, and stretch) with the diameter ranging from 1mm to 20mm. It is made of fiber-reinforced rubber and shaped in a cylinder with three internal chambers (see Figure 2. 8 (a)). The pressures inside internal chambers are independently controlled by external pressurized sources via flexible tubes and valves. The fibers are embedded inside the rubber in the circular direction with the purpose of preventing the actuator from deformation in the radial direction. The actuators will be bent in desired directions due to different or

unequal pressures in the three chambers. For example, if the chamber 1 is pressurized, the actuator will bend in the y-direction (see Figure 2. 8 (b)).

Figure 2. 8. (a) FMA's structure [47], (b) Bending and stretching based on the supplied pressure [47].

In different approaches, McKibben actuators were invented in the 1950s by J.L. McKibben and Gaylord [49]. As a result of the presence of large stroke, high force, and compliance, it has been used throughout in humanoid robotics, human-machine interfaces, prostheses, and medical instruments [37]. The basic structure of McKibben actuators includes a woven reinforced elastic tube as shown in Figure 2. 9(a). The working principle is briefly described in Figure 2. 9(b). Due to the constraint of woven, the length of actuator is shortened after being pressured.

Figure 2. 9. Miniature McKibben actuators (a) Structure of McKibben actuators, (b) The working principle of actuators [37].

These well-known actuators are scaled down to a diameter of 1.5mm and a length of 22mm and they still provide the high force densities [37]. To incorporate these

dedicated fluidic actuators, Moers and his colleges [50] presented a surgical instrument with three fluidic actuators, spacer disks, and an elastic tube (see Figure 2. 10).

Figure 2. 10. Surgical instrument with miniature McKibben actuators [50]

Advantages and disadvantages of the fluidic actuator

The advantages and disadvantages of flexible fluidic actuators were partially discussed in the literature [36]. With respect to human safety, fluidic actuators made of biocompatible materials have no energized parts that opposite to electricity-related or Shape Memory Alloys and thermal actuators. They can handle the tissue without any impairment due to their compliant property. Compared to the traditional mechanisms, flexible fluidic actuators are easy to change their shapes since they are conducive to enter and retract from the human body as well as being able to operate in confined spaces of the human GI tract. In addition, flexible fluidic actuators possess abilities to provide high force at small sizes, especially at microscale. In a recent study, fluidic actuators deliver among the highest force and power densities compared to other types of actuators [35]. With these promising properties, ever the last few decades, numerous micro-hydraulic and micro-pneumatic actuators have been applied in some surgical applications. However, their main limitations are bulky, complex, and high power supply requirements. Safety is also the main challenge. From the surgical points, these heavy systems can be installed outside the human body, which does not have negative effect on the surgical quality. The key challenge can be identified as the difficulties in model and control problems due to nonlinear friction and backlash hysteresis. Since the properties of flexible materials are considerably different from the conventional and rigid ones, most of current approaches are not able to provide accurate models and

efficient controllers. This significantly degrades the system's performance. Another drawback of fluidic actuators is the need for high pressure in the operation process. If this high pressure is not efficiently controlled, human safety is put in danger.

2.2.3: Smart material actuator

Smart materials such as shape memory alloys (SMAs) and piezoelectric actuators will be investigated in this review in terms of property, design, advantages, and disadvantages.

Conceptual design

SMAs belong to the group of shape memory materials (SMMs) which can memorize and recover to their previous forms under the change of temperature or magnetic stimulus. This phenomenon is known as shape memory effect (SME). Because of high work density, large force and displacement, bio and magnetic compatibility, numerous surgical robotic systems have exploited SMAs as the main mode of actuation such as robotic catheter [51, 52], endoscopes [53, 54], surgical graspers [55, 56], capsule endoscopes [57], and Neurosurgical Robot and Steerable Cannula [58-60]. By comparison, piezoelectric materials that are used to build ultrasonic actuators can induce the movement under certain voltage. As opposed to electronic actuators, ultrasonic actuators from smart materials offer higher power density, higher torque for lower speed, and precise motions. As a result, they are selected as the actuation means for surgical applications like surgical tools [61] and endoscopes [62, 63].

Basically, SMAs exist in two phases namely martensite at low temperature and austenite at high temperature. By selecting suitable training, heating, and cooling methods, the transformation between the two phases can be achieved. For more details, further reading in some studies [64-66] is needed. In some applications, the most method of actuation is to use external force to deform the robotic structures and then force these structures to recover or contract to their original shapes using internal or external heating sources or magnetic fields. Because of instability, impracticability, and poor thermal-mechanic performance of Fe-Mn-Si, Cu-Zn-Al, and Cu-Al-Ni materials, Ni-Ti based SMAs is much popular for practical applications [64]. SMAs have been widely

applied for numerous fields such as industrial applications, automotive, aerospace, MEMs devices, robotics, and medical devices [64, 67-70].

Smart material-driven surgical devices

Various surgical devices have been used smart materials as the main mode of transmission. For example, robotic catheters and endoscopes are actuated by the SMA cables that are fixed around the main structure and can pull in order to bend the instruments in different directions [51, 53, 54, 71, 72].

The researchers also designed a minimally invasive neurosurgical intracranial robot (MINIR) using SMA wires as actuators [58]. The prototype of the robot has nine revolute joints actuated by 0.5mm diameter SMA wires (see Figure 2. 11). The two adjacent joints are shown in Figure 2. 11 (a). Larger holes are milled for the passage of the SMA wires while smaller holes are designed for the columns. The SMA actuators push against these columns and transmit the force to actuate the revolute joints. Figure 2. 11 (b) shows the working principle of their proposed SMA actuators. Two antagonistic SMA wires are used for each joint so that each joint is able to move backward and forward directions. In stationary state, these two SMA wires play a role of keeping the links to be straight. To rotate the joint in clockwise direction, the left SMA wire will be actuated to reform to its original shape.

Figure 2. 11. Minimally invasive neurosurgical intracranial robot; (a) The brass MINIR design, (b) The actuation mechanism

Advantages and disadvantages of the smart material actuator

As a result of many excellent technical properties like high corrosion resistance, bio-compatible, and non-magnetic [64], SMA actuators have been selected for numerous biomedical applications. Apart from this, research work has shown that SMA offers very high work density, high strength, significant displacement, and large force. For example, for SASI system [73], it can provide up to 30N with small size and is able to act in three-dimensional manners (extension, bending, and twisting). In addition, as given in the aforementioned review, the diversity of actuator shapes (cable, spring, plate, tube, and ring) for SMA is another benefit. However, some limitations including relative small strain, low actuation frequency, and difficulty in accurate control problems are much more challenging in the use of SMA. For example, the applied current to create high-temperature transformation is about 600°C. For safety, temperature and current isolation are required for human use. The heating and cooling methods inducing the phase transformation also have a considerable effect on the system response. According to [74], heating process can be done in several ways like conductive heat transfer, radiative heat transfer with microwaves or infrared light, inductive heating, and direct Ohmic heating. It also reported in few studies that a very high-speed response SMA actuator can be achieved by using large electrical currents. As a result, the main obstacle of the narrow bandwidth of SMA actuators is lying in the long cooling time that depends upon the size, shape, and design mechanisms of actuators. With a number of available techniques such as water immersion, heat sinking, and fluidic medium, the cooling time can be cut down. However, the heating time may grow and leakage problems must be considered. In addition, the durability and reliability of SMA actuators after experiencing multiple transformations are other challenges. Researchers have shown that the thermal effect is linked to the fatigue-life of SMA actuators which is vital for surgical applications.

2.2.4: Magnetic actuator

Most of the transmission systems for surgical applications such as cables, fluidic actuators, and shape memory materials require “link” from power supply to drive surgical tools.

Conceptual design

Magnetic actuation uses the magnetic field without requiring “link” to actuate surgical tools/joints. This type of actuator completely removes the transmission means. In surgical application using magnetic actuator as the main mode of transmission, the use of moving external permanent magnet/electromagnet to drive the internal magnet [75, 76] or MRI gradients to induce motion to ferromagnetic components [77-79] is the basic working principles. In these systems, the power and torque are magnetically transmitted from the external units through the abdominal wall into the internal units, leading the fact that the onboard electromagnetic motors, connection cables, and links are completely eliminated.

Magnetic driven surgical devices

Various surgical applications using magnetic actuator such as laparoscopic tissue retractor [75], surgical manipulator [80], MRI-powered actuator [77], capsules [81, 82], and magnetic air capsule for painless colonoscopy [83], will be introduced and discussed in this section.

Base on the local magnetic actuation (LMA), one single-DOF laparoscopic tissue retractor was introduced in [75]. As depicted in Figure 2. 12, the device consists of an anchoring unit and an actuation unit. As a supporter, the anchoring unit includes a pair of external and internal square-section permanent anchoring magnets (EAM and IAM) that stay in the abdominal wall to reinforce the stability of the system. By contrast, the actuation unit has an external driving and internal driven magnet (EDM and IDM). When a motor rotates the EDM, the IDM then also rotates, followed by the actuation of inside mechanism that is attached to IDM. The last component of this system is the offset crank mechanism that is used to produce the translational movement from the rotation of the IDM.

Figure 2. 12. The schematic representation of laparoscopic tissue retractor [75]

Advantages and disadvantages of the magnetic actuator

Like other kinds of actuators, magnetic actuators also hold both advantages and disadvantages in the surgical robotic scenario. Taking advantages into account, with the magnetic power in terms of forces and torques being wirelessly transmitted via abdominal wall, the need for connecting cables or linked mechanisms is completely eliminated, followed by simpler and more lightweight systems. Otherwise, the surgical tools with high speed, capacity, and dexterity can be achieved by large-power external driving magnets or MRI machines. However, some limitations still need to be solved. For instance, the magnetic field exposes the high hysteresis, nonlinearity, and uncertainty in the models. Furthermore, since a number of coupled magnets are required for multiple-DOF systems, as a consequence, there is a presence of their interactions force, which raises the challenge in optimizing the magnet's position. Last but not least, the high-associated cost and high risk of tissue damages also need to be addressed.

2.3: Overview of current approaches for variable stiffness

In the literature, the principles of variable stiffness are twofold, namely structures and materials. In the former category, stiffness is adjustable by reorganizing and/or reconnecting parts which must be linked in an attachable/detachable way inside the

structure. By doing that, compliance occurs once parts are detached and high stiffness is achieved when parts are reattached [3].

2.3.1: Structure-based methods

Traditionally, various methods have been used to increase the structural stiffness in the engineering fields, e.g., welding a diagonal bar in a frame. However, most of them are for permanent change of stiffness. When it comes to tunable stiffness, parts inside the structure must be linked in a nonpermanent way. By doing that, the design can be switched between flexible and stiff modes by disconnecting and reconnecting these parts [3]. For instance, the slider linkage lock mechanism using air pressure is developed for outer sheath for endoscopic surgery [84]. In this design, multiple cylindrical pieces that consist of the sheath, link, sliders, and channels are connected serially (see Figure 2. 13). When the fluid channel is empty, all the parts move in compliant mode. By contrast, once we apply air into fluid channel, the friction force will be generated between the piece and the slider resulting in rigid state.

Figure 2. 13. The locking mechanism design. (a) Compliant mode when the air tube is empty. (b) Stiff mode when air tube is pressurized. (c) The prototype. (d) Cross-section view of one segment [84].

In another study, An inflated tube is also deployed to rigidify a FORGUIDE shaft in [85]. In the compliant state, the expandable tube is deflated so that cables are able to slide between spring and tube with small friction. But when pressure is introduced into the tube, high stiffness is achieved due to the friction between a ring of cables which are located between the tube and the spring.

Figure 2. 14. The design of FORGUIDE. (a) The cross-section overview. (b) Shaft in the compliant state. (c) Shaft in the rigid state [85].

Agostino Stilli et al. presented a novel concept for safe, variable-stiffness robot links in the paper [86]. The link includes two main elements: a plastic mesh that is embedded inside a layer of silicon in the cylindrical shape (Figure 2. 15). When the pressure inside the chamber is in low state, the robot's link stays flexible. With the support of the mesh, the cylindrical link can maintain the shape without collapsing. Once being inflated with air to high pressure, the silicone chamber will perform like a stiff structure. The stiffness performance tests show that robot can bear up to 40N force along the axial direction.

Figure 2. 15. The design of a stiffness-controllable link. (a) Overview of the drawing. (b) In and out air flows. (c) Silicon/mesh wall view [86].

Fluid solutions are also applied to control stiffness in several other research [87, 88]. The main disadvantages of these approaches are the complexity in design with numerous small parts involved and the limited stiffness since it mainly relies on the air pressure.

As discussed previously, cable (tendon) has been widely used as actuator in surgical robots. Apart from that, it is also exploited as a medium to control the stiffness [3, 89-94]. One of the first examples is the curve length locking system made by NeoGuide [3, 92]. In this design, fixed at the disk of each segment at the one end, multiple wires are connected with the clamping actuators at the other end (Figure 2. 16).

Figure 2. 16. Details of the NeoGuide locking structure. (a) Flexible mode when all the cables are free to move. (b) Rigid mode when all the cables are tensioned [3].

When flexibility is required, all the wires are free to move. By contrast, when all the wires are tensioned by the clamping actuators at the base, the distance between two successive segments is constrained, making the design stiff. However, the design is still highly complicated with many mechanical parts and wires, which causes difficulty in miniaturization. In addition, a complex controller is needed for the system, making it even more expensive to build. The durability of the cables and links is also needed to put into consideration.

To improve the dexterity and stiffness of the robot, a variable neutral-line mechanism is proposed [94]. Unlike the traditional snake-like manipulators, the neutral line (backbone) of the robot is not fixed at the center in this design (Figure 2. 17). By doing that, the authors proved that stiffness can be trolled by cable tensions. The relationship between the stiffness and cable tensions is also derived and proximately linear.

Figure 2. 17. Design of a Stiffness-Adjustable Hyper-redundant Manipulator. (a) The variable neutral-line mechanism. (b) The robot prototype [94].

Using a quite similar concept, but instead of using multiple wires, a number of patents which only employ a tension cable running through all the segments except the last one of the design (see Figure 2. 18). In the compliant mode, the cable at the center is relaxed, the segments can slide over each other with little friction forces. Once the center cable is stretched, segments are pressed together generating high friction forces between them. These forces prevent them from sliding over each other resulting the rigidified structure.

Figure 2. 18. Design and working principle of variable stiffness mechanism using a set of nested segments and a centric wire. (a) Flexible state when the tension wire is relaxed. (b) Stiff state when tension wire is pulled. (c) Free body diagram of the forces in the system [3].

Applying both multiple and center wires concept, a highly articulated robotic probe (HARP) intended for MIS was developed [90, 95]. The details of the design were described in section 2.2. In this example, totally 4 cables (3 cables for the outer tube and 1 cable for the inner tube) made of polyethylene are exploited for both shape and stiffness control (Figure 2. 19). When the stiffness is required, cables are pulled towards the back of the mechanism increasing the friction forces between the links which are connected by the spherical joints. By contrast, cables' relaxation will decrease the mentioned friction forces making the outer or inner snakes limp.

Figure 2. 19. The design of HARP. (a) The overview of the two concentric snakes. (b) The HARP prototype [90, 95].

Inspired by muscular structures of octopus, a variable stiffness MIS silicon-based manipulator was developed using tendon and antagonistic actuation scheme [87].

Figure 2. 20. The design of tendon-based stiffening soft manipulator. (a) The cross-section view of a segment. (b) The antagonistic actuation control scheme [87].

In this design, a silicon body is embedded with three pairs of pneumatically actuated chambers and three tendons (Figure 2. 20 (a)). The stiffness of the system is varied by an appropriate control scheme of the two opposing actuation means (Figure 2. 20 (b)).

Their experiments have shown the capability of increasing the load-bearing of the antagonistic mechanism.

In recent years, particle jamming technology based on granular materials, such as dry sand or coffee beans which has phase-transition from a liquid-like state to solid state with applied fluids, is applied to tune the stiffness in the continuum robots [38, 96-102]. The working principle is illustrated in Figure 2. 21. In the compliant mode, there is a balance between the pressures inside and outside the foil tube. As a result, the particles can freely move inside the tube. Conversely, the pressure outside will be higher than that of the inside when the vacuum pump is switched on, creating a compression force to restrain particles from moving. The benefits of using jamming particles are short activation time and dramatic change in stiffness between two states. Yet, high stiffness requires substantial volume of granular materials and bulky structure designs.

Figure 2. 21. The working principle of variable stiffness using vacuum-packed particles. (a) Compliant mode without vacuum pressure. (b) Rigid mode with applied vacuum pressure [101].

Using the abovementioned granular jamming solution, a stiff-flop surgical manipulator was designed by researchers in [38]. The combination of flexible fluidic actuators and the jamming phenomenon is to enable the omnidirectional bending and stiffness changing. The coffee powder was used as the granular material and stacked into a 8mm stiffness channel located at the center of the robot (Figure 2. 22). The stiffness is controlled by the level of applied vacuum.

Figure 2. 22. The design of the stiff-flop manipulator. (a) The overall architecture of the stiff-flop manipulator. (b) The details of the design with the cross-section view.

E. Amanov et al. [102] introduced a flexible variable stiffness endoport single site partial nephrectomy using granular and layer jamming. With two tendon-driven segments and several working channels, the robot has 120mm in length and 26mm in outer diameter. To maximize the jamming effects, the whole space that is not utilized for working channels is filled with granule (see Figure 2. 23). The authors also did experiments with multiple materials such as sucrose, lactose, collagen, and coffee to choose the best candidate.

Figure 2. 23. Design of endoport. (a) Actuation unit design. (b) Transmission of the motor. (c) Airflow path during the experiment. (d) Cross-section showing the material filling area [102].

2.3.2: Material-based methods

Unlike structure-based methods, material-based ones make use of phase-change materials, such as electrorheological and magnetorheological fluids [103-106], phase changing alloy (PCA) [107-110], and thermoplastic polymer [111-113], which can exist in various states under certain stimuli to regulate the stiffness.

Figure 2. 24. The flexural modulus (stiffness) variation of the glass/phase change materials with different temperature.

Figure 2. 24 describes the flexural modulus (stiffness) variation of these glass/phase change materials. When the temperature stays below the transitional level (T_g), materials are in the glass/stiff state. In contrast, they will transform into the rubber/flexible state once the temperature reaches above the T_g .

Electrorheological and magnetorheological fluids, which change their states between liquid and quasi-solid states in electric and magnetic fields, respectively, have been used for catheters and prosthesis penile [103-106].

Figure 2. 25. The cross-section view of a variable stiffness catheter. (1) The shaft. (2) The electrodes. (3) Electrorheological fluid channel. (4) Balloon channel. (5) Working channel [106].

R. Brenneman et al. [106] developed a variable stiffness catheter using Electrorheological fluid whose strength can go from 0 to 5 kPa by exposing to the electric field (Figure 2. 25). Although the rigidity is massively changed between liquid and quasi-solid states, it is still too weak for stiffening mechanism since the stiffness of the colonoscope is reported to be 11.4 MPa at least [114]. Furthermore, high voltage and current would be dangerous to human body.

Figure 2. 26. The stiffness switching mechanism of the insertion tube [108].

A group of researchers from China recently developed a variable stiffness manually operating platform for laparoendoscopic single-site surgery (LESS) using liquid metal (a mixture of indium, gallium, and stannum) which has melting point around 40°C for stiffness adjustment [108]. The details of the mechanism are exhibited in Figure 2. 26. There are still several limitations in this design; the stiffness difference between compliant and rigid mode is only four times; both the time of the phase change from rigid to flexible mode and the time of the reverse phase change are considerable, 22s with 89°C hot water and 15s with 18°C cold water respectively.

Recently, a low melting point alloy (Cerrolow 117), which can transform from solid to liquid phase upon joule heating, was used for a variable-stiffness catheter [109]. The authors have selected copper wire which was designed in coil shape as the heater (Figure 2. 27). With 0.8-1A current applied to the copper wire, heating time takes 2 to 10 s. Addition, cooling relies on convective heat transfer without any integrated cooling system and cooling time being reported.

Figure 2. 27. Design of a variable-stiffness catheter. (a) The prototype. (b) Several configurations that catheter can adapt [109].

A continuum manipulator was designed and prototyped using Field's metal an alloy of bismuth (32.5 wt%), indium (51 wt%), and tin (16.5 wt%), with a relatively low melting temperature (62°C), low viscosity in the liquid state, and high stiffness in the solid-state. In this design, Joule heating using a resistive stainless steel spring and water cooling are selected for transitioning the phase of the alloys (see Figure 2. 28). A high applied current is the main drawback for this design (4A) [107].

Figure 2. 28. The schematic view of the design. (a) One segment design. (b) Two segments design

In another study, Huan et al. [115] developed a stiffness varying mechanism using a low melting point polymer, Polycaprolactone (PCL). This material was melted at about 60°C with the heat transmitted via copper wire and braided stainless steel tube. The paper reported that the achieved flexural modulus of the manipulator in rigid state was 225 MPa that is relatively low.

Using a similar method but different materials, researchers in [112, 113] have employed thermoplastic polymers PLA (polylactic acid) and ABS (acrylonitrile butadiene styrene) as variable stiffness solutions and shape memory alloy as heating method for changing stiffness of fabrics (see Figure 2. 29). Long heating and cooling durations are

the main limitations of the design. It was reported that heating takes up to 60s and complete cooling costs up to 5 minutes. Besides, the shortcoming of PLA is brittleness [116], and that of ABS is high glass transition temperature (105°C).

Figure 2. 29. The variable stiffness fibers. (a) The fiber composition. (b) PLA-based fiber. (c) ABS-based fiber [112].

2.3.3: Advantages and disadvantages of the current variable-stiffness approaches

Both structure-based and material-based approaches have been widely studied to vary the stiffness in the last decades. While having their own benefits, there still exist several drawbacks.

The advantages of the former techniques are the ability of fast and significant stiffness changes between two different states. It is because the applied pressure or the cable tension can be changed instantly. However, several limitations are still realized. For instance, slider linkage or cable-based designs are complex with multiple small custom-made parts and cables required to embed the stiffness controlling mechanism. Therefore, these mechanisms have to be simplified to be easy to scale and produce. In addition, cable-rigidified systems (e.g., NeoGuide) require very large and complicated controllers, resulting in great production costs. Jamming technology has the potential of controlling stiffness. But the substantial volume and high pressure are needed, making it challenging to be miniaturized and safe.

Compared to the structure-based techniques, material-based ones tend to be simpler in design by using a relatively uncomplicated state switching mechanism: heating and cooling, which makes the designs easy to fabricate, miniaturize, and scale. Besides, this category could obtain a significantly high stiffness ratio since the flexural modulus of the material is vastly different between the two states. However, there are still various downsides such as long activation time, too high transition temperature or even too low, and low stiffness in rigid state (refer to Table 2. 1 for more details).

Table 2. 1: The classification of variable stiffness approaches with their advantages and disadvantages.

Classification	Principle	Advantages	Disadvantages	References
Structure-based methods	Mechanical, Fluids	<ul style="list-style-type: none"> The ability to switch between flexible and stiff mode fast High stiffness-changing ratio 	<ul style="list-style-type: none"> Complexity in design 	[84-88]
	Cables		<ul style="list-style-type: none"> The need for durable cables and links Complex control system The coupling effect of the cable tensioning and position of the robot 	[3, 87, 89-95]
	Jamming technology		<ul style="list-style-type: none"> Requirement for substantial volume and high pressure 	[38, 96-102]
Material-based methods	Electrorheological and magnetorheological fluids	<ul style="list-style-type: none"> Simple in design with less small parts involved Higher stiffness-changing ratio 	<ul style="list-style-type: none"> Long activation time High applied voltages and currents High transition temperature Low stiffness in rigid states 	[103-106]
	Phase changing alloys			[107-110]
	Thermoplastic polymers			[111-113]

Chapter 3: Conceptual Design, Preliminary Experiment, and Modelling of a Variable-Stiffness Manipulator

In this chapter, the conceptual design, some preliminary experiment results, and the variable stiffness modelling will be given. Section 3.1 outlines the material sections and the details of the mechanism. The comparison between different materials and the working principle of the design will be analyzed. Section 3.2 describes the bending experiments set up with the results and discussion. Section 3.3 presents the testing and modeling of stiffness, which includes the Dynamic Mechanical Analysis (DMA), the 3-point bending experiments, and the heat transfer model with simulation results.

3.1: Conceptual design

Although available variable-stiffness approaches have both advantages and disadvantages, the material-based method is selected in this research due to its simplicity in design and large stiffness ratio between flexible and stiff states. Furthermore, the shortcomings such as long activation time, improper transition temperature, and undesirable stiffness can be addressed by suitable material selection and appropriate active heating and cooling design. This section presents the design concept with the material selection based on mechanical and thermal properties.

3.1.1: Material selection

Thermoplastic materials become flexible upon heating and hard upon cooling, which is potential for variable stiffness structures. In this work, we selected the material for surgical variable stiffness manipulators based on the following criteria: (1) stiffness variation ratio; (2) glass transition temperature, i.e., the temperature at which the stiffness of the material changes dramatically; (3) biocompatibility; (4) strength. Table 3. 1 shows these properties of common thermoplastic materials [117-119]. Although the flexural modulus data in rubbery state is not available, with these common materials we can consider them soft and flexible in rubbery state [120]. Among the reviewed

thermoplastic materials, Nylon and PET have lower temperature and higher stiffness ratio in comparison with others. However, Nylon is unstable with humidity-dependent properties. Because of that, PET is the final selection.

Table 3. 1: Property of common thermoplastics

Acronym	Polymer	Glass transition temperature T_g ($^{\circ}$ C)	Flexural modulus (GPa) in the glassy state	Flexural strength (MPa)	Biocompatible
ABS	Acrylonitrile butadiene styrene	110-125	2.07-4.14	50-80	Yes
PMMA	Poly(methyl methacrylate)	85-110	2.24-3.17	70-127	Yes
PLA	<u>Poly</u> lactic acid	60-65	2.39-4.93	48-110	Yes
Nylon 6	Nylon 6	47-57 (humidity dependent)	0.7-2.83 (humidity dependent)	35-108 (humidity dependent)	Yes
Nylon 6,6	Nylon 6,6	-15-77 (humidity dependent)	1.21-2.96 (humidity dependent)	42-123 (humidity dependent)	Yes
PET	Polyethylene terephthalate	68-80	2.41-3.1	82-124	Yes
PVC	Polyvinyl Chloride	75-105	2.07-3.45	65-94	Yes
PC	Polycarbonate	150	2.34	93.1	Yes

Developed in the 1940's, PET (Polyethylene terephthalate) is a thermoplastic (or thermosoftening) material that turns to the rubbery state from the glass state once its temperature goes beyond the glass transition temperature (67 $^{\circ}$ C – as data from Vention Medical Inc.). PET is considerably stiff in the glass state but highly flexible in the rubbery state and relative strong compared to other materials in the group. For example, in glass state, the flexural modulus of PET is around 2.41-3.1 GPa, while that of Nylon 6 and PC are 0.7-2.83 GPa and 2.34 respectively; PET has the flexural strength of 82-124 MPa, while that of ABS is only 50-80 MPa. In this study, we took advantage of this thermoplastic feature to design a new type of variable stiffness tube. Furthermore, PET poses some other properties that make it a great candidate for medical applications such

as biocompatibility, clearness, lightweight, high strength, stiffness, favorable creep characteristics, low flavor absorption, high chemical resistance, barrier properties, low cost, and relatively low transition temperature [121]. The PET tube used in this study is commercialized and supplied by Vention Medical Inc.

Regarding heat transmission, Thomas et al. [113] employed a shape memory alloy (SMA) cable with the current being supplied. The disadvantage of this idea is that the design will have undesirable shape changes due to SMA's shape memory effect. In our proposed design, we used a flexible stainless steel coiled sheath from Asahi Intecc Japan, Inc. Compared to SMA used in [113], stainless steel does not have shape memory effect and thus does not result in undesired shape changes. Stainless steel is also biocompatible and has high electrical resistance which results in low applied current. Moreover, the chosen stainless steel coiled sheath is highly flexible and thus has little effect on the stiffness of the manipulator in the rubbery state.

3.1.2: Design concept

The design and working principle are depicted in Figure 3. 1. The variable stiffness tube (VST) is made of two different tubes including an outer PET tube and an inner flexible stainless steel sheath (coiled tube). At room temperature, the PET tube is in its glass state and thus is stiff. If the current is applied to the stainless steel sheath, due to its electrical resistance, the sheath and PET tube will be heated. Once the temperature reaches the glass transition threshold, PET tube will change from the glassy state to the rubbery state, making the VST flexible.

Figure 3. 1. Structure and working principle of the variable stiffness tube

3.2: Preliminary experiments and results

This section describes some preliminary bending experiments set up with the results and discussions.

3.2.1: *Variable stiffness tube (VST) bending tests*

Experiments were conducted to compare the flexural modulus of the proposed VST and a commercial endoscope in compliant and rigid states.

Three-point bending tests were performed using the Instron 5569 material testing machine with 500N and 50N loadcells from Instron Singapore ITW Pte Ltd.

The experimental setup is shown in Figure 3. 2. Three kinds of tubes were tested: a single PET tube, a VST in the glass state, and a VST in the rubbery state. All the components are off-the-shelf items, and their sizes are as follows: the lengths of the PET tube and sheath were 115 mm and 120 mm, respectively, and the OD and thickness of the sheath are 1.45 mm and 0.3 mm, respectively. The OD and ID of the proposed VST equal to the PET tube's OD and the flexible sheath's ID, respectively (Table 3. 2). Note that the flexural modulus is independent of the sizes of components.

Table 3. 2. The dimensions of the studied VST

	PET tube (mm)	Stainless steel sheath (mm)
Length	115	120
OD	2.22	1.45
ID	1.45	0.85

The support span was set at 50 mm for all the tests. The samples were loaded at 2 mm/min until the deflection reached 7 mm for the single PET tube and VST in glass state and 15 mm for the VST in the rubbery state. The bending test for glassy VST was conducted at room temperature, while that of rubbery VST started after supplying 0.3A into the stainless steel sheath for 30 seconds. This was to ensure enough time for the VST change from the glassy state to the rubbery state.

Figure 3. 2. VST bending tests. (a) Instron 5569 material testing machine. (b) Bending test with a single PET tube. (c) Bending test with VST in glass state. (d) Bending test with VST in rubbery state.

3.2.2: Commercialized endoscope bending tests

A commonly used endoscope, GIF-2T240 from Olympus Co., was also tested to compare with the proposed VST in terms of stiffness and flexibility. Bending test was carried out at four different positions in the endoscope: the middle of the flexible tip, and the positions which are 20cm, 40cm, and 60cm away from the end of the flexible tip. For each position, the endoscope was tested in two different modes: locked mode and unlocked mode. The endoscope is stiff in locked mode and compliant in unlocked mode. The experiment setup is shown in Figure 3. 3. The endoscope was loaded with a speed of 2mm/min, and the maximum deflection was set as 4 mm.

Figure 3. 3. Olympus endoscope bending tests. (a) The middle of the flexible section. (b) 20cm from the end of flexible tip position. (c) 40cm from the end of flexible tip position. (d) 60cm from the end of flexible tip position.

3.2.3: Bending results and discussion

Test results for VST, locked endoscope, and unlocked endoscope are plotted in Figure 3. 4, Figure 3. 5, and Figure 3. 6, respectively. In each case, the measurement data is presented in terms of force versus deflection.

Figure 3. 4. Bending test results for VST.

From Figure 3. 4, it is noted that the stiffness (slopes of the lines in the figure) of the VST in glass state and rubbery state are significantly different. During the experiment, we observed that the stiffness of the PET tube and the glassy VST were close, meaning that the flexible stainless steel sheath was very flexible and did not contribute much to the stiffness of the VST in glass state for the single PET tube and glassy VST. The flexural modulus was calculated using the Eq. (3. 1) of the middle point deflection of an elastic beam of length L loaded by a central force F in three-point bending test [122]:

$$E = \frac{FL^3}{48\delta I} \quad (3.1)$$

Where δ is the mid-point deflection, F is the central load, L is the flexural modulus, and I is the second moment of area.

Based on Eq. (3. 1), the flexural modulus of glassy and rubbery VST are 2.141 GPa

$$\left(E_{glass} = \frac{1.918 \times 50^3}{48 \times 2 \times \pi \left(\frac{2.22^4 - 0.85^4}{64} \right)} \right) \quad \text{and} \quad 38.88 \quad \text{MPa} \quad \left(E_{rubber} = \frac{0.0 \times 50^3}{48 \times 2 \times \pi \left(\frac{2.22^4 - 0.85^4}{64} \right)} \right)$$

respectively. It can be seen that the stiffness of VST drops 55 times when shifting from the glassy state to the rubbery state. Compared to the liquid metal used in [108] with four times difference in stiffness between rigid and flexible states, the stiffness of the proposed PET-based VST can change more significantly.

Figure 3. 5. Bending test results for (a) locked endoscope and (b) unlocked endoscope at different positions.

Figure 3. 5 shows the bending test results of the unlocked and locked endoscope at different positions. In addition, the comparison at each position in two different modes

is shown in Figure 3. 6. It can be seen that the endoscope is slightly stiffer in locked mode than in unlocked mode due to the cable tension. This indicates that cable tension in the endoscope cannot significantly increase the stiffness of the endoscope. It is also noted that the further the section is from the distal end, the stiffer it will be, for instance, among the 4 testing points, the 60 cm point in the locked endoscope has the largest stiffness and the mid-point in the unlocked flexible tip has the smallest stiffness. The flexural modulus values at these two critical positions are used to compare with the proposed VST. The second moment of area for the endoscope is calculated based on the endoscope specifications from Olympus [123]. The principal moments of inertia of area for the endoscope are 144.5 mm^4 and 155.68 mm^4 . Finally, with the bending formula and testing results, the flexural modulus of different points in the endoscope are given in Table 3. 3.

Figure 3. 6. Bending test results for the endoscope at different positions and modes. (a) For the flexible section. (b) For 20 cm position. (c) For 40 cm position. (d) For 60 cm position.

Table 3. 3: Flexural modulus of the endoscope at different sections

	Locked endoscope	Unlocked endoscope
E_{Tip}	45–49 MPa	37–40 MPa
E_{20cm}	103–111 MPa	83–90 MPa
E_{40cm}	117–126 MPa	93–100 MPa
E_{60cm}	170–183 MPa	172–186 MPa

Compared to the tip of the endoscope, the rubbery VST is as flexible as the unlocked endoscope's tip, while the glassy VST is about 40 times stiffer than the locked endoscope. Compared to the body of the endoscope, the rubbery VST is approximately 2.5 times (38 MPa versus 83-90 MPa) more flexible than the unlocked endoscope (20 cm position). Meanwhile, the glassy VST is 9 times stiffer than the locked endoscope (at 60 cm position). Thus, the following conclusions can be made: when flexibility is desired, the proposed VST is at least as flexible as the most flexible part of the currently commercialized endoscope; when stiffness is desired, the proposed VST is 9 times stiffer than the stiffest part of the endoscope (Figure 3. 7). Therefore, if a VST with similar size or geometry of the endoscope is employed to construct an endoscope or used as an over-tube of the endoscope, theoretically, this VST-based endoscope will be about nine times stiffer than the existing one. For example, suppose that the VST's OD is also 11.8 mm (the same with the endoscope's OD) and the ID is 11.28 mm. So the area moment of inertia will be $I_t = \frac{\pi}{64} (D^4 - d^4) = \frac{\pi}{64} (11.8^4 - 11.28^4) = 157 \text{ mm}^4$, which is the same as that of the endoscope. Thus, the VST, in this case, is 9 times stiffer than the endoscope ($I_t E_t = 9 I_e E_e$) since they have the same area moment of inertia but the flexural modulus values are different.

Figure 3. 7. Stiffness comparison between VST and the endoscope.

3.3: Testing and modelling of stiffness

This section investigates the relationship between the stiffness (flexural modulus, E) and other parameters of the system such as geometric dimension (d), time (t), and the applied current (I) through modeling and experiments. Eq. (3. 2) depicts this relationship:

$$\begin{cases} E = f_1(T) \\ T = f_2(d, t, I) \end{cases} \quad \text{or} \quad E = f_1(f_2(d, t, I)) \quad (3. 2)$$

Where T is the temperature of the PET tube. The main problem is divided into two sub-problems. The first one is to figure out the flexural modulus function in terms of temperature T (function f_1), and the second one is to calculate the temperature distribution in terms of geometry or design dimensions (d), time (t), and current (I) (function f_2).

Dynamic Mechanical Analysis (DMA) tests, the three-point bending tests, and heat transfer modeling were used to formulate two above sub-problems. The reason we use

both DMA and three-point bending tests is to verify the final results. The details are presented in the following sub-sections.

3.3.1: Dynamic Mechanical Analysis (DMA)

Dynamic Mechanical Analysis (DMA) is a widely employed approach to characterizing the mechanical properties of polymers or composites upon temperature effect [124-126]. In a DMA test, an oscillating force that plays a similar role as the bending force is applied to a sample, and the material's responses to that force were recorded and analyzed. Based on that, the viscosity and stiffness (modulus) over a temperature range are calculated from phase lag and sample recovery, respectively. In DMA, the flexural modulus was calculated using Eq. (3. 3) which includes the real part (E') as the storage modulus and the imaginary part (E'') as the loss modulus with its magnitude [127].

$$\begin{cases} E = E' + iE'' \\ |E| = \sqrt{(E')^2 + (E'')^2} \end{cases} \quad (3. 3)$$

In this study, the tests were conducted in DMA Q800 from TA Instrument, the USA with three-point bending mode, multi-frequency-strain module, temperature ramp method. The temperature was increased from 30 °C to 100 °C at a rate of 5 °C/min. The outer diameter and inner diameter of the PET tube are 1.95 mm and 1.47 mm, respectively. The support span is 20 mm long. The testing results (Figure 3. 8) from DMA machines with three different runs illustrate that the flexural modulus of the PET tube at the two states (glass state and rubbery state) is immensely different. It starts decreasing at about 65 °C–70 °C and going to the plateau rubbery region at about 80 °C–85 °C.

Figure 3. 8. PET flexural testing results from the DMA machine.

3.3.2: *Three-point bending results*

The dynamic flexural modulus varied with time was also measured and computed from the three-point bending test. In this measurement, bending and heating are conducted simultaneously. In addition, the time-dependent flexural modulus is also calculated from Eq. (3. 1) and depicted in Figure 3. 9 with three different trials and 0.4 A current applied. Furthermore, the reverse direction from rubbery to glassy state is also examined by 3-point bending and plotted in Figure 3. 10. The tests show that with only ambient cooling, it takes up to 80 s–100 s to change the state from flexible to stiff.

Figure 3. 9. The Instron bending result with 0.4 A current applied

Figure 3. 10. 3-point bending results when state changes from rubber to stiff one.

3.3.3: Heat transfer modeling and simulation results

In the proposed design, resistive heating from the stainless-steel sheath has been used to control the PET tube's stiffness. Due to non-linear and time-dependent heat transfer coefficients from both stainless steel and PET materials, the heat transfer model is highly non-linear so that it needs to be solved numerically with simulation software. The mathematical heat transfer model is based on the approach for electrical cables [128, 129] with different boundary conditions. Figure 3. 11 shows the cross-section geometry of the VST consisting of stainless steel and PET layers. We denote hollow radius by r_0 , outer sheath radius by r_1 , and outer PET tube radius by r_2 . The VST's heat transfer model includes the heat transfer equation for the sheath (Eq. (3. 4)) and the homogeneous non-linear heat conduction equation for the PET layer (Eq. (3. 5)), which relate to each other by the conjugation conditions at $r = r_1$ (Eq. (3. 6)), the boundary conditions at $r = r_0$ and $r = r_2$ (Eq. (3. 7)), and the initial conditions (Eq. (3. 8)).

Figure 3. 11. The cross-section geometry of VST

$$c_0 \frac{\partial T_0}{\partial t} = \frac{1}{r} \frac{\partial}{\partial r} \left(r k_0 \frac{\partial T_0}{\partial r} \right) + f_0(r, t) \quad r \in (r_0, r_1) \quad (3. 4)$$

$$c \frac{\partial T}{\partial t} = \frac{1}{r} \frac{\partial}{\partial r} \left(rk \frac{\partial T}{\partial r} \right) \quad r \in (r_1, r_2) \quad (3.5)$$

$$\begin{cases} T_0|_{r=r_1} = T|_{r=r_1} \\ k_0 \frac{\partial T_0}{\partial r} \Big|_{r=r_1-0} = k_0 \frac{\partial T_0}{\partial r} \Big|_{r=r_1+0} \end{cases} \quad (3.6)$$

$$\begin{cases} r = r_0: k_0 \frac{\partial T_0}{\partial r} \Big|_{r=r_0} + \varepsilon_0 \sigma (T_{r=r_0}^4 - T_{\text{env}}^4) = 0 \\ r = r_2: k \frac{\partial T}{\partial r} \Big|_{r=r_2} + h(T_{r=r_2} - T_{\text{env}}) + \varepsilon \sigma (T_{r=r_2}^4 - T_{\text{env}}^4) = 0 \end{cases} \quad (3.7)$$

$$t = 0: T_0 = T_{\text{env}}, T = T_{\text{env}} \quad (3.8)$$

Where c_0, k_0, ε_0 and c, k, ε, h - the temperature-dependent coefficients which correspond to the heat capacity, heat conductivity, the emissivity, and convection heat transfer of the stainless steel sheath and PET layer respectively; T_0 and T - temperature distribution along the sheath and PET tube; T_{env} and σ - environment temperature and Stefan-Boltzmann constant.

Because of high non-linearity with time-dependent coefficients, it is challenging to obtain the exact analytic transient solutions for the heat transfer problem (Eq. (3.4) and (3.5)). As a consequence, COMSOL Multiphysics software built by COMSOL, Inc. from US was used to calculate the solutions numerically. Since the VST design is symmetric, for the computational cost and time saving, a short tube is built and simulated in COMSOL environment. The simulated coefficients are as follows:

a) Heat conductivity coefficients k_0 and $k(W/mK)$

The coefficient k_0 is linearly interpolated with temperature variable based on the data from [130] (stainless steel grade SUS304 –Table 3.4).

Table 3. 4: Heat conductivity coefficient of SUS304 with different temperatures.

Temperature (K)	k_0 (W.m ⁻¹ .K ⁻¹)
293	14.76
300	14.89
350	15.79
400	16.61

The coefficient k of PET is described by Eq. (3. 9):

$$k = f(T[K]) = -0.14 + 0.003T - 8.96 \times 10^{-6} \times T^2 + 1.05 \times 10^{-8} \times T^3 \quad (3. 9)$$

b) Heat capacity coefficients $c_0 = 500(J/kgK)$ and c as suggested in [131, 132] (Table 3. 5)

Table 3. 5: Heat capacity coefficient of PET with different temperatures.

Temperature (K)	c (J / kg.K)
300	1172
400	1820

In the above section, the sheath coil is approximated as a tube for ease of modeling. A dummy resistivity of the tube is proposed based on the geometry difference between the coil and the tube.

However, with the same length L and the diameter of the wire of the coil (equal to the thickness of the tube) (Figure 3. 12, r_1 and r_2 are the inner and outer radius), the resistance of the coil is much higher than that of the tube. Therefore, we will use the resistance of the coil instead of the tube in the Eq. (3. 4) during the simulation.

The resistance of a coil and a tube as shown in Figure 3. 12 can be expressed by:

Figure 3. 12. The geometry of the coil and the tube with the same length and thickness.

$$\begin{cases} R_{coil} = \rho \frac{2\pi L}{r_2 - r_1} \frac{r_2 + r_1}{2} \frac{1}{\pi \left(\frac{r_2 - r_1}{2}\right)^2} = \rho \frac{4L(r_2 + r_1)}{(r_2 - r_1)^3} \\ R_{tube} = \rho \frac{L}{\pi(r_2^2 - r_1^2)} \end{cases} \quad (3. 10)$$

Eq. (3. 10) formulates the resistance of the coil and the tube with the related parameters. Based on that, we can derive the heat generated due to the applied electrical current (Eq. (3. 11)). The length L is absent in this Eq. (3. 11) because the proposed model is only 2D (per unit length) due to the symmetric design.

$$f_0(r, t) = I^2 \rho \frac{4(r_2 + r_1)}{(r_2 - r_1)^3} \quad (3. 11)$$

Where I is the direct current, ρ is the resistivity of SUS304 (using the first-order function regarding temperature and data from [130] – Table 3. 6), r_1 and r_2 are the inner and outer radius of the coil.

COMSOL simulation with the tube shape was conducted to verify Eq. (3. 10). In this simulation, we developed and used a dummy resistivity so that the tube will have the same resistance as the coil. The formula for this dummy resistivity is $\rho^* = k_{tube}\rho$, where $k_{tube} = R_{coil}/R_{tube}$. Figure 3. 13 and Figure 3. 14 show that the COMSOL simulation results are very close to the coil and the tube design although there is only a slight difference because the stainless steel tube has better contact with the PET tube.

c) The convection heat transfer to the air of the horizontal cylindrical surface is obtained from [128] (Eq. (3. 12)).

$$h = [0.1254(1/d)^{1/2} + 1.0932(T - T_{env})^{1/6}]^2 \quad (3. 12)$$

Where d is the diameter of the cylinder.

d) The resistivity of SUS304

Table 3. 6: The resistivity of SUS304 with different temperatures.

Temperature (K)	$\rho(10^{-8}\Omega m)$
293	71.3
300	71.9
350	76.0
400	79.8

e) Other constants:

$$\sigma = 5.67 \times 10^{-8}(W/m^2K^4), \quad r_0 = 0.615(mm), \quad r_1 = 0.735(mm), \quad r_2 = 0.925(mm),$$

The transient temperature results have been shown in Figure 3. 13 in the case of direct input current of 0.4 A.

Figure 3. 13. The transient temperature at different times with 0.4 A direct current applied and the coil shape (a) and the tube shape (b).

Figure 3. 8 depicts the flexural modulus versus temperature and Figure 3. 9 presents the flexural modulus versus time. Based on these two sets of data, we selected two points from each curve (linear interpolation) to obtain the curve of temperature versus time

and compared to that from COMSOL simulation and that measured from thermal couple (see Figure 3. 14). As shown, the data extracted from DMA and 3-point bending experiments is close to the simulation result, which means that the model and solutions are relatively accurate.

Due to the fast rate of temperature changing (from 23 °C to 90 °C in only 10 s) and the delay of the thermal couples, the temperature error is about 10 °C. In terms of time, there is 1 s delay with real-time temperature measurement from thermal couples. The time constant, namely the time required to reach 63.2 % of an instantaneous temperature change [133], of the thermal couples is 1 s, which matches with the measurement here.

Figure 3. 14. Transient temperatures from COMSOL simulation and experiments

3.3.4: Mathematical interpolation for flexural modulus

In this section, mathematical formulas of the flexural modulus with temperature variable T are introduced based on two different interpolation functions, namely hyperbolic and polynomial functions. The measurement data is from DMA testing (run

1) with 370 data points. The proposed general functions with related parameters are shown in Eqs. (3. 13) and (3. 14). Although many other functions were considered, these two types of functions were selected eventually because of the small errors they have with the experimental data. Note that dividing the temperature range does not give better results with the hyperbolic equation. As a result, the interpolation is conducted on the whole temperature range in this case.

$$E_1 = f_1(T) = a_0 \tanh(a_1 T + a_2) + a_3 \quad T \in [30; 100] \quad (3. 13)$$

$$\begin{cases} f_2(T) = b_0 T_1^6 + b_1 T_1^5 + b_2 T_1^4 + b_3 T_1^3 + b_4 T_1^2 + b_5 T_1 + b_6 \\ f_3(T) = c_0 T_2^8 + c_1 T_2^7 + c_2 T_2^6 + c_3 T_2^5 + c_4 T_2^4 + c_5 T_2^3 + c_6 T_2^2 + c_7 T_2 + c_8 \end{cases} \quad (3. 14)$$

Where $T_1 (= T/10) \in [3; 7.5]$ and $T_2 (= T/10) \in [7.5; 10]$.

Genetic Algorithm is an optimization approach that allows global space searching based on genetics and natural selections including several operators such as crossover, mutation, and reproduction [134-136]. This method exploits the natural evolution to generates a new generation where unfit elements are eliminated from the original one using operators and fitness function evaluation. For our proposed models, the fitness function is defined as in the Eq. (3. 15). The goal is to minimize the value of fitness function to get the best parameters.

$$Fitness = \frac{1}{N} \sum_{i=1}^N (E(i) - E_{1,2}(i))^2 \quad (3. 15)$$

Here, N is the number of sample points collected from the DMA experiments; the i - sampling index; $E(i)$ and $E_{1,2}(i)$ are the data from the experiment and the proposed models (Eqs. (3. 13) and (3. 14)). The fitness function is defined as the mean of squared error between the proposed model and real experimental data.

With the above-developed function and algorithm, MATLAB Optimization Toolbox (using *optimtool* command) is employed for mathematical calculations of Eq. (3. 13) and the polynomial curve fitting (*polyfit* command) for Eq. (3. 14). The identified results are summarized in Table 3. 7.

Table 3. 7. The identified coefficients for the functions given by Eqs. (3. 13) and (3. 14).

a_0	1.222
a_1	-0.109
a_2	8.126
a_3	1.172
b_0	-0.002601864627564
b_1	0.072979477716920
b_2	-0.846862742927548
b_3	5.201612215813454
b_4	-17.830808616908705
b_5	32.286137528800232
b_6	-21.651085667368704
c_0	-0.000000115137420*10 ⁶
c_1	0.000008240173302*10 ⁶
c_2	-0.000257670384674*10 ⁶
c_3	0.004598012513484*10 ⁶
c_4	-0.051210101538069*10 ⁶
c_5	0.364505485627160*10 ⁶
c_6	-1.619187967251756*10 ⁶
c_7	4.103905326934227*10 ⁶
c_8	-4.543584462187940*10 ⁶

With polynomial functions, there are many digits presented here because the value of the high order polynomial functions are sensitive regarding both coefficients and variables, which means that only small changes in coefficients or variables can lead to huge changes in function value.

Figure 3. 15. The interpolation results and the error with two different proposed functions.

Figure 3. 15 depicts and compares the values of hyperbolic and polynomial functions to the DMA experiment results. The errors here are simply calculated by the subtractions between the experiment and computational results. It is shown that polynomial functions can accurately represent DMA data, while hyperbolic function still has some significant errors. To evaluate the performance of the interpolation, the mean square error (MSE) is used and expressed by (Eq. (3. 16)):

$$MSE = \frac{1}{N} \sum_{i=1}^N (f_{\text{experiment}}(i) - f_{\text{interpolation}}(i))^2 \quad (3. 16)$$

Where i is the sampling index and N is the total number of samples from experimental results, $f_{\text{experiment}}$ is the DMA data, $f_{\text{interpolation}}$ is the interpolation function values. After the calculation, the MSE of the hyperbolic function is 0.005121, while the MSE of the polynomial function is 8.6233×10^{-6} which is much smaller and better.

3.4: Discussion and conclusion

In this chapter, a new and promising design concept for variable stiffness is proposed using a PET tube and stainless steel sheath. Multiple tests, namely DMA tests and three-point bending tests, on the proposed design and a typical commercialized endoscope show that when flexibility is desired, the proposed VST is at least as flexible as the most flexible part of the current commercialized endoscope; when stiffness is desired, the proposed VST is nine times stiffer than the stiffest part of the endoscope (Figure 3.7). These outcomes prove the design's high potential toward variable stiffness applications for surgery.

Characteristic evaluation tests and modeling were conducted to investigate the relationship between stiffness and temperature as well as the heat transfer from coiled sheath to the PET tube; Based on DMA tests, the flexural modulus with respect to temperature was accurately interpolated with polynomial functions. The highly non-linear heat transfer model was built and numerically solved by COMSOL simulations, followed by the comparison with the transient temperature measurement using thermal couples. Although the simulated solutions in COMSOL are highly close to the extracted data from DMA and three-point bending tests, there is still a nearly constant delay with real time temperature measurement from thermal couples, which is due to the time constant (1 s in this case) of the sensors. However, this delay, which is nearly constant, can be compensated when more accurate measurement or control is applied.

Chapter 4: Temperature-dependent, Variable-stiffness Endoscopic Manipulator with Active Heating and Cooling

This chapter presents a novel active variable-stiffness robotic arm for surgical applications. By using multiple VSTs which were described in chapter 3, an air-cooling mechanism was developed to cool down the heated robotic arm by utilizing the inner channels of the PET tubes for air circulation, leading to a compact and safe design. The robotic arm has two bending segments and can be bend into “S” shapes by the 8 driving tendons. The arm is 15 mm in diameter and 140 mm long. Tests have been conducted to characterize the stiffness, heating, and cooling characteristics of the robotic arm. Models have also been developed to describe the temperature with respect to time, applied current, and the velocity of the cooling air.

4.1: Variable-stiffness robotic arm design

A schematic view of the robotic arm is given in Figure 4. 1. The robotic arm consists of a single continuum backbone made of a superelastic nitinol tube, 12 spacer disks, 8 stainless steel cables, and 4 variable-stiffness PET tubes. The spacer disks are fixed on the backbone and the spacing between them is 10mm. The robotic arm has two independently bending segments (Figure 4. 1a). Each segment has 6 spacer disks and is actuated by four cables routed through the guiding pinholes on the spacer disks as shown in Figure 4. 1b. To completely cancel the motion coupling effect of the two segments, the tendons for the second segment are run through four flexible sheaths and converge close to the center of the first segment (Figure 4. 1b). The sheaths apply counterbalanced forces (equal to the tension on the tendons) to the first spacer disk of the second bending segment so that the first bending segment would not bend when the second segment is actuated [137, 138]. Guided by the spacers, tendons are attached to the distal end of each segment by the knots and fixed to the actuators from the other end. The prototype is shown in Figure 4. 1d.

By pulling a tendon at the proximal end, the backbone will bend toward the contracted tendon, leading to the bending of that segment. The stiffness of the structure mostly relies on the flexible backbone and the four PET tubes running along the robot's body. Inside each PET tube, there is a flexible stainless steel sheath working as the heating element (Figure 3. 1). By applying electric current into the sheaths, resistive heat will be generated due to the electrical resistance of the coiled sheath (essentially, long and thin stainless steel wire) and transferred to the PET tubes. If the temperature is below the transition threshold (around 67°C-70°C), PET will stay in a stiff glass state. Once the temperature reaches the threshold, the tubes will transform from glassy to rubbery state, which makes them and the whole robot soft. The flexural modulus of PET material is 2.41-3.1 (GPa) in the glass state and 82-124 (MPa) in rubber state and [139]. After being heated up, the robot needs to be cooled down when high stiffness is needed. Active cooling can help to remove the heat from PET tube faster than ambient cooling.

A number of active cooling approaches that have been studied for shape memory alloy are also applicable for proposed system [140]. While liquids (e.g., water) are fast for cooling, air is more secure for setups where electricity is involved. For that reason, cool air is chosen as the cooling media in the current design. To have a compact design, the inner channels of the four PET tubes are also utilized as the cooling channels for the air. A pair of tubes are connected at the distal end to form the air circulation vessels; one is for the inlet and the other the outlet (Figure 4. 2). The cooling fluids come in and out all the way from the proximal end of the arm to the distal end, taking thermal energy out of the robotic arm.

Previous studies indicate that the external diameter of the endoscopic devices should be within 15 mm [4]. For that reason, we chose the outer diameter of the disk to be 14 mm, the width of the disk equal 2mm, and a thermal insulation layer (made of Aerogel) with a thickness of 0.5 mm, which results in a robotic arm with the diameter of 15 mm. The detailed dimensions are given in Table 4. 1. Although these dimensions were rationally selected based on space constraints, available commercial materials, and functional requirements, they are not fully optimized because the focus of the study is to test the

concepts of changing the stiffness of the robotic arm using PET material and the heating and cooling mechanisms.

Figure 4. 1. Schematic view of the robot's design. (a) Overview of the robot. (b) Detailed inside components. (c) Cross-section of the disk. (d) Robot's prototype.

Table 4. 1: The dimensions of major components

Length	140 mm
Overall diameter	15 mm
Disk's OD	14 mm
Disk's width	2 mm
Backbone diameter	0.9 mm
Aerogel thickness	0.5 mm
Pinholes diameter	0.5 mm
Tendon diameter	0.42 mm
PET tubes' OD/ID	2.26 mm/1.35 mm
Sheath's OD	1.34 mm/0.9 mm

Figure 4. 2. The cooling system including inlet, connecting, and outlet channels.

4.2: Variable stiffness characterization and validation

In this section, three-point bending experiments using the Instron machine are carried on to characterize the variable stiffness of the given design. The manipulator was tested in both stiff and flexible modes. After that, the validation experiment of the design under transverse force loading (mimicking external challenging loading in real surgical scenarios-1N, 3.5N, and 16N) is also presented.

4.2.1: Variable stiffness characterization

The experiment set up is described in Figure 4. 3. The support span was set at 80 mm for all the tests. The samples were loaded at the speed of 5 mm/min until 5 mm deflection was reached. Note that the support points and the loading point are at the disks of the robotic arm. The force and displacements were recorded. Three trials for each state were conducted. Figure 4. 4 shows the force-displacement relationships of the trials for both the glass state (rigid) and rubber state (flexible).

Figure 4. 3. The 3-point bending experiment set up.

Figure 4. 4. Force and displacement data from the 3-point bending experiment of the robotic arm.

Based on the collected force and deflection, the stiffness, which is the ratio of the force to the deflection of the manipulator in glass state and rubber state, is calculated and equal to 4.4 N/mm and 0.2 N/mm, respectively. The stiffness ratio between the two states is 22.0. This stiffness variation ratio is much higher than that of the CF-240AL endoscope (Olympus, Japan) whose stiffness ratio is only about 1.59. In terms of flexural rigidity, the mentioned endoscope (the bending section) in the flexible mode has the smallest flexural rigidity of 103.74 Ncm²; in the rigid state, it holds the maximum flexural rigidity of 164.84 Ncm² [4, 114]. The flexural rigidity of the robotic manipulator is also estimated using the formula $EI = \frac{FL^3}{48\delta}$. In which, the production of EI represents the flexural rigidity, F -the bending force, L -the support span distance, δ -the displacement. By substituting the bending data into the equation, the manipulator's rigidities are obtained as 469.33 Ncm² ($EI = \frac{11 \times 8^3}{48 \times 2.5 \times 10^{-1}}$) and 21.33 Ncm² ($EI = \frac{0.5 \times 8^3}{48 \times 2.5 \times 10^{-1}}$) in the stiff and compliant modes respectively. It can be seen that the proposed scope is much stiffer than the Olympus endoscope in rigidified mode, while it's also much more flexible than the Olympus endoscope in compliant mode.

Table 4. 2 below summarizes the performances of several variable-stiffness devices available on the literature. In quantitative comparison with structure-based devices, the proposed design outperforms several ones in terms of stiffness [141] or stiffness ratio [38]. In particular, the slider linkage lock mechanism [141] can bear only 0.1N for the third joint, while the experiments show that the our robot has very small displacement with 1N force applied. Moreover, the proposed manipulator yields high stiffness ratio (22) between two states, while that of STIFF-FLOP module [38] is only 1.36. When comparing with other material-based devices, the studied one has better activation time and stiffness ratio. For example, the activation time of the proposed robot is 30s, while that of the continuum manipulator [107] is up to 63s. Furthermore, the platform for single site surgery [108] has the stiffness variation of 4, while ours is up to 22.

Table 4. 2: The performance comparisons of several variable-stiffness manipulators.

Approach	Activation Time	Stiffness in rigid state	Stiffness in compliant state	Stiffness ratio	Reference
Structure-based	No data (short in principle)	The third joint can bear only 10gf (0.1N) with 400kPa pressure	No data	No data	[141]
	No data (short in principle)	0.529 N/mm with 59.9N cable tension	No data	No data	[94]
	No data (short in principle)	1489 Ncm ²	No data	No data	[85]
	No data (short in principle)	No data	No data	36% increase	[38]
Material-based	37s	0.23N/mm	0.056N/mm	4	[108]
	63s	No data	No data	No data	[107]
	No data (long in principle)	No data	No data	No data	[109]
Proposed Design	30s	469.33 Ncm ²	21.33 Ncm ²	22	

4.2.2: Variable stiffness validation

For navigation in the human body, the endoscope needs to be flexible to adapt to the tortuous path of the gastrointestinal tract. When the endoscopic instruments (e.g., graspers, endoloops) are working on the target tissue, the endoscope needs to be stiff enough to provide a strong platform. In this case, the most critical loading scenario is when a transverse force is applied to the tip (e.g., the grasper is lifting a heavy tissue). This transverse force can significantly bend the tip and thus deviate the endoscope from the desired orientation. This can be even worse when the force keeps changing during operation because the endoscope tip would dynamically move or wander around and thus make the operation difficult. In this section, experiments were conducted to see how the robot reacts to external transverse loading at the rigid or flexible states.

The experimental set up is shown in Figure 4. 5. The robot was placed on top of a black panel, and an endoscopic grasper was inside one of the endoscope tool channels with its end-effector reaching out of the endoscope tip. A tendon was fixed to the grasper so that a transverse force can be applied to the grasper. To track the robot's shape, a set of white markers was painted on the black insulation sheet. The shape of the robot can be captured by a camera through these white markers. A checkerboard on the bottom was used for the calibration of the camera.

Figure 4. 5. The validation experiment setup. (a) Robot in the original shape. (b) Robot in the changed shape after applying force F .

The procedure has four main steps, described as bellow:

Step 1: Bend the robot to make an S-shape (the original shape).

Step 2: Take a photo of the original shape.

Step 3: Apply a transverse force to the distal end.

Step 4: Take a photo of the changed shape.

The taken photos are handled using Matlab software (*extrinsics* and *pointsToWorld* functions) to get the markers' coordinates. The process is divided into four phases. Starting from color image (Figure 4. 6a) which is then filtered to produce a binary image

(Figure 4. 6b). The noise was then removed (Figure 4. 6c) to get the final refined points (Figure 4. 6d).

Figure 4. 6. The image processing scheme. (a) Color image. (b) Binary image. (c) Noise removed the image. (d) Marker points reconstruction.

Three different scenarios, a robot without PET tubes, robot with PET tubes in the glass state, and a robot with PET tubes in the rubbery state, were tested. In addition, three various force magnitudes (1 N, 3.5 N, and 16 N), which are in the force range of surgical tasks [4, 142], were selected. In order to get both qualitative and quantitative comparisons, the average distance error is measured based on the markers' coordinates (Eq. (4. 1)).

$$e = \frac{1}{N} \sum_{n=1}^N \sqrt{(x_c - x_i)^2 + (y_c - y_i)^2} \quad (4.1)$$

where N is the number of markers ($N = 14$ in this case); x_c , y_c and x_i , y_i denote the coordinates in the changed and initial shape respectively. With different combinations of robot configurations and forces, totally 5 cases were investigated.

In the first two cases, which are the robot without PET tubes and the robot with PET tubes in rubbery state, the distal end was pulled by a weight with the force equal to 1 N while the actuation cable tensions were kept at 5 N. After reconstruction, the marker points are plotted in Figure 4. 7. Furthermore, the average distance errors are estimated ($e_1 = 33.61 \text{ mm}$, $e_2 = 33.31 \text{ mm}$).

Figure 4. 7. The data points in the event of 1 N applied force. (a) The robot without PET tubes. (b) The robot with PET tubes in the rubbery state.

The robot with embedded PET tubes in the glass state was exerted 1 N, 3.5 N, and 16 N in the other three cases (no tension on other cables). The results are shown in Figure 4. 8. Similar to the previous cases, the average distance errors are computed with the formula in the Eq. (4. 1) ($e_3 = 3.63 \text{ mm}$, $e_4 = 8.36 \text{ mm}$, $e_5 = 23.61 \text{ mm}$).

Figure 4. 8. The reconstructed dots along the robot body. (a) With 1 N force applied. (b) With 3.5 N force applied. (c) With 16 N force applied.

With the consideration of the above graphs and measured average distance errors, several conclusions can be made. Firstly, the cable tension has less impact on holding the robot's shape (huge e_1) due to the under-actuation feature (the number of motors is less than the degrees of freedom of the robot arm). Secondly, while PET tubes are compliant in rubbery state ($e_1 \approx e_2$), the robot is significantly stiffened due to the PET tubes in glass state (much smaller e_3 compared to e_1). Ultimately, having a great performance in term of holding shape when applying force within surgical range (small e_4), the design can bear the ultimate force.

4.3: Heating and cooling experiments

In the current design, resistive heating is achieved by applying an electrical current to the flexible stainless steel sheathes embedded inside the PET tube (Figure 4. 2). Heat is generated and transferred into the PET tube, followed by the phase change of PET once the temperature reaches the glass transition temperature. For cooling, cool air is circulated inside the circulation channels (the inner channels of the PET tubes). Experiments were conducted to test the heating and cooling processes on the robotic arm.

The robotic arm with thermal sensors (model TH100PT, Thorlabs Inc., United State) is shown in Figure 4. 9(a), and the diagram of the experimental setup in Figure 4. 9(b). Air from the pump passes through a tube that is dipped into iced water to create cold air. An airflow gauge was used at the inlet of the PET channels to measure the flow rate of the air, followed by a thermal sensor 1 which measures the temperature of the cooling air. Thermal sensor 2 is put on the PET tube surface (see Figure 4. 1), and the thermal sensor 3 is to measure the temperature outside the insulation layer. Sensor 2 is important because the state of the PET tubes is judged based on sensor 2's reading. All data from sensors is sent back to Arduino circuit for processing.

Figure 4. 9. The manipulator with temperature sensors (a) and the diagram of the heating and cooling experiment set up (b). Sensors 1, 2, 3 are to measure the temperatures of the cooling air at the inlet, PET tubes inside the manipulator, and the outer surface of the insulation layer, respectively.

The first study is to investigate the effect of different currents on heating time and ambient cooling time. Three different currents (0.2A, 0.3A, and 0.4A) are chosen. The main reason for selecting these currents is through trials. Authors experienced that currents below 0.2A could result in too long heating process, while currents above 0.4A would lead to too fast heating speed which is hard to control and possibly the overheat of the PET tubes. Considering the transition temperature of PET material is around 67°C-70°C and the overshoot (i.e., the temperature keeps rising for a while after turning

off the current), the current was turned off when the temperature of the PET tubes raised to 60°C in the heating phase, and the cooling was stopped once the PET temperature drops to 60°C in the cooling phase. Figure 4. 10, which plots the temperature versus time, indicates that higher current produces faster heating speed, while ambient cooling takes quite a long time. For example, it costs 15s and 7.5s to reach 60°C with 0.3A and 0.4A respectively. It is also obvious that the ambient cooling time is longer in the case of higher applied current because the maximally reached temperature is higher. For instance, the ambient cooling times for 0.2A, 0.3A, and 0.4A are 36.1s, 100.3s, and 184.1s respectively.

Figure 4. 10. The temperature outside the PET tube in the heating and cooling phases. Different electric currents were tested.

In order to accelerate the cooling speed, ambient cooling was replaced by forced convective cooling using cold air at 13°C (created by pumping ambient air through an iced water tray Figure 4. 9). As seen in Figure 4. 11 which depicts the PET tube's temperature versus time, overshoot and cooling time are vastly shortened in comparison

with ambient cooling. In particular, for the case of 0.3A electric current, the overshoot duration was reduced from 12.3s to 4.8s (60% less) and cooling was shortened from 100.3s to 11.9s (almost 10 times faster). Also, faster airspeed resulted in shorter overshoot and cooling duration. For example, overshoot and cooling times are 6.91s and 16.6s when using 0.04 L/s air flow rate, while those of 0.05 L/s are 6.39s and 14.54s and those of 0.06 L/s are 4.85s and 11.96s. In short, the higher the cooling airspeed, the shorter the overshoot and cooling time. Although there is no standard to regulate the activation time, however, authors believe that the achieved cycle of the proposed device is acceptable considering the fact that the frequency of switching between two modes is not high during the surgical scenario.

Figure 4. 11. The temperature outside the PET tube when heating with 0.3A current, air, and ambient cooling.

Figure 4. 12. The temperature outside the Aerogel surface in 3 different cases.

It is also important to notice that the temperatures on the thermal insulation surface stay below 35 °C during the experiments (Figure 4. 12) which is in the safety range for the human body (≤ 41 °C [4]). With the active cooling system, the device's performance was improved greatly by reducing the overshoot and cooling time. In addition, the thermal insulation layer makes the design more applicable for human body in terms of safety.

4.4: Heating and cooling models

In the current design, the heat generated from the stainless steel sheath is transferred to the internal parts inside the robot (PET tubes, disks, backbone, and thermal insulation layer), which is the conduction phenomenon. Apart from that, heat is partially dissipated to the surrounding environment and taken away by the air cooling, producing convection process. The combination of the conduction and convection forms the general heat transfer model of the system. Due to the complexity of the system and the high nonlinearity of the heat transfer equations, it is impossible to acquire accurate analytical solutions. For that reason, an approach that integrates the simplified heat

transfer model and the experimental data is proposed. Two different scenarios are investigated, one is with only ambient cooling and the other with active air cooling. Each scenario consists of three separate phases: heating, overshoot, and cooling.

4.4.1: Heating and ambient cooling models

To simplify the model, the robot was considered as a whole entity. Figure 4. 13 represents the energy flows in and out of the robot during the heating and ambient cooling.

Figure 4. 13. The energy flows during the heating and ambient cooling process
According to the conservation of energy principle, the Eq. (4. 2) defines the heating stage:

$$dQ = dQ_{in} - dQ_{out} \leftrightarrow RI^2 dt - h_{am} A_{am} (T - T_{\infty}) dt = mcdT \quad (4. 2)$$

Where R - the sheath resistance; I - the applied current; h_{am} and A_{am} - the convective heat transfer coefficient and the surface area; T and T_{∞} - the temperature of the robot and environment respectively; m and c - the weight and specific heat of the robot.

Eq. (4. 2) yields the heating solution as Eq. (4. 3); the Eq. (4. 4) is be used instead of Eq. (4. 4) for calculation and presentation purposes.

$$T = T_{\infty} + \frac{\beta I^2}{\alpha} (1 - e^{-\alpha t}) \quad t \in [0; t_h] \quad (4. 3)$$

$$T = T_{\infty} + k_{11} I^2 (1 - e^{-k_{12} t}) \quad t \in [0; t_h] \quad (4. 4)$$

Where $\alpha = \frac{h_{am}A_{am}}{mc}$ and $\beta = \frac{R}{mc}$; t_h -the heating time (the moment of turning off the current); k_{11} and k_{12} are both constant.

During the experiments, authors experienced the overshoot phenomenon when the temperature of the PET tubes still keep raising for a while after turning off the power supply. This was due to an induced transient current caused by a sudden change of state [143]. The Eq. (4. 5) describes the mentioned current in the overshoot phase.

$$I = I_0 e^{-\gamma t} \quad (4. 5)$$

Where I_0 – the magnitude of the electric current; γ - a constant of transient current.

Substituting (4. 5) into (4. 2), we obtain the overshoot as Eq. (4. 6). Solving the Eq. (4. 6) gives us the solution (4. 7).

$$RI_0^2 e^{-2\gamma t} dt - h_{am1} A_{am1} (T - T_\infty) dt = mcdT \quad (4. 6)$$

$$T = T_\infty + e^{-\alpha_1 t} \left(\frac{\beta_1 I_0^2}{\alpha_1 - \gamma_1} e^{-(\gamma_1 - \alpha_1)t} + c \right) \quad (4. 7)$$

Where $\beta_1 = \frac{R}{mc}$; $\alpha_1 = \frac{h_{am1}A_{am1}}{mc}$; $\gamma_1 = 2\gamma$; c is the constant defined by the initial conditions.

For the ease of calculation and presentation, Eq. (4. 8) will be used.

$$T = T_i + k_{21} I^2 e^{-k_{22}(t-t_h)} (1 - e^{-k_{23}(t-t_h)}) \quad t \in [t_h; t_o] \quad (4. 8)$$

Here k_{21} , k_{22} , and k_{23} are all constant; t_h -the heating time; t_o -the end of overshoot (the moment temperature reaches the maximum).

Finally, the ambient convective cooling process takes place. Canceling out the generated heat term in the Eq. (4. 2) gives us the following ambient cooling Eq. (4. 9) and the solution (4. 10).

$$-h_{am2} A_{am2} (T - T_\infty) dt = mcdT \quad (4. 9)$$

$$T = T_{\infty} + (T_{i2} - T_{\infty})e^{-\alpha_2 t} \quad (4. 10)$$

For the ease of calculation and presentation, Eq. in the form of (4. 11) will be used.

$$T = T_{\infty} + (T_{i2} - T_{\infty})e^{-k_{31}(t-t_0)} \quad t \in [t_o; t_e] \quad (4. 11)$$

Here t_o -the end of overshoot (the start of the cooling process); t_e -the end of cooling stage; k_{31} is a constant.

4.4.2: Heating and active air cooling models

The same procedure is carried out in order to build the model in the case of active air cooling. Although the term “active air” is being used, ambient cooling is always happening. The heating solution is same as Eq. (4. 3). For the purpose of differentiation and presentation, other notations are employed as Eq. (4. 12) and (4. 13).

$$T = T_{\infty} + \frac{\beta_a I^2}{\alpha_a} (1 - e^{-\alpha_a t}) \quad (4. 12)$$

$$T = T_{\infty} + k_{41} I^2 (1 - e^{-k_{42} t}) \quad t \in [0; t_h] \quad (4. 13)$$

Where $\alpha_a = \frac{h_{ama} A_{ama}}{mc}$ and $\beta_a = \frac{R}{mc}$; k_{41} and k_{42} are constant.

Cold air is pumped in simultaneously with turning off the current. For this reason, the overshoot Eq. (4. 14) is unlike the ambient case.

$$RI_0^2 e^{-2\gamma_a t} dt - h_{ama1} A_{ama1} (T - T_{\infty}) dt - h_{a1} A_{a1} (T - T_a) dt = mcdT \quad (4. 14)$$

Where T_a - the temperature of cold air; h_{a1} and A_{a1} - the convective heat transfer coefficient and the contact surface area.

Solving the Eq. (4. 14) provides us the solution (4. 15) or (4. 16).

$$T = \frac{\tau_{a1}}{\alpha_{a1}} + e^{-\alpha_{a1}t} \left(\frac{\beta_{a1}I^2}{\alpha_{a1} - \gamma_{a1}} e^{-(\gamma_{a1}-\alpha_{a1})t} + c \right) \quad (4.15)$$

$$T = T_{ai1} + \frac{k_{52}}{k_{51}}(1 - e^{-k_{51}(t-t_h)}) + k_{54}I^2 e^{-k_{51}(t-t_h)}(1 - e^{-k_{53}(t-t_h)}) \quad t \in [t_h; t_o] \quad (4.16)$$

Where $\alpha_{a1} = \frac{h_{ama1}A_{ama1}+h_{a1}A_{a1}}{mc}$; $\beta_{a1} = \frac{R}{mc}$; $\tau_{a1} = \frac{h_{ama1}A_{ama1}T_{\infty}+h_{a1}A_{a1}T_a}{mc}$; $\gamma_{a1} = 2\gamma_a$; T_{ai1} - the initial temperature when the overshoot starts; c-constant defined by initial conditions. Here k_{51}, k_{52}, k_{53} , and k_{54} are all constant. Because the air convective heat transfer coefficient h_{a1} is proportional to the velocity v [144], the terms α_{a1} and τ_{a1} can be rewritten as (4. 17) for the calculation purpose.

$$\begin{cases} k_{51} = \alpha_{a1} = x_{a1} + y_{a1}v \\ k_{52} = \tau_{a1} = x_{\tau1} + y_{\tau1}v \end{cases} \quad (4.17)$$

Lastly, the active convective cooling process takes place. Canceling out the generated heat term in the Eq. (4. 14) gives us the following active cooling Eq. (4. 18) and the solution (4. 19) or (4. 20).

$$-h_{ama2}A_{ama2}(T - T_{\infty})dt - h_{a2}A_{a2}(T - T_a)dt = mcdT \quad (4.18)$$

$$T = \frac{\tau_{a2}}{\alpha_{a2}} + \left(T_{ai2} - \frac{\tau_{a2}}{\alpha_{a2}} \right) e^{-\alpha_{a2}t} \quad (4.19)$$

$$T = k_{61} + (T_{ai2} - k_{61})e^{-k_{62}(t-t_o)} \quad t \in [t_o; t_e] \quad (4.20)$$

Where $\alpha_{a2} = \frac{h_{ama2}A_{ama2}+h_{a2}A_{a2}}{mc}$; $\tau_{a2} = \frac{h_{ama2}A_{ama2}T_{\infty}+h_{a2}A_{a2}T_a}{mc}$; T_{ai2} - the initial temperature of the cooling step (the maximum value). Here t_o -the end of overshoot (the start of the cooling process); t_e -the end of cooling stage; k_{61} and k_{62} is a constant.

Similarly, $\alpha_{a2}, \tau_{a2}, k_{61}$, and k_{62} can be simplified as (4. 21).

$$\begin{cases} k_{61} = \alpha_{a2} = x_{\alpha2} + y_{\alpha2}v \\ k_{62} = \tau_{a2} = x_{\tau2} + y_{\tau2}v \end{cases} \quad (4.21)$$

In order to tune the coefficients in the abovementioned models, the Curve Fitting Toolbox and the command *lsqcurvefit* inside the MATLAB are utilized.

There are totally six models (Eqs. (4.4) (4.8) (4.11) (4.13) (4.16) (4.20)), which are equally divided into two scenarios, with three major variables including time- t , current- I , and the air velocity- v . Two of them require two data sets to define coefficients (Eqs. (4.16) (4.20)), while the rest need only one data set. The reason for that is because of the relationships between the velocity variable (v) and the coefficients k_{51}, k_{52}, k_{61} , and k_{62} can only be solved with two pairs of known coefficients (Eqs. (4.17) (4.21)).

Table 4. 3: The values of coefficients of the above-derived models.

Model	Data sets used for parameter identification	Data sets used for model verification	Results
(4.4)	Am ($I = 0.2 A$)	Am ($I = 0.3A, 0.4A$)	$T_{\infty} = 23.5; k_{12} = 0.006; k_{11} = 6338.975$
(4.8)	Am ($I = 0.4A$)	Am ($I = 0.2A, 0.3A$)	$k_{21} = 1622.047; k_{22} = 0.071; k_{23} = 0.027$
(4.11)	Am ($I = 0.2A$)	Am ($I = 0.3A, 0.4A$)	$T_{\infty} = 23.5; k_{31} = 0.045$
(4.13)	Am ($I = 0.2 A$)	Ac ($I = 0.3 A$)	$T_{\infty} = 23.5; k_{42} = 0.006; k_{41} = 6338.975$
(4.16)	Ac ($f = \frac{0.04l}{s}, \frac{0.06l}{s}$)	Ac ($f = \frac{0.05l}{s}$)	$x_{\alpha1} = 0.026; y_{\alpha1} = 0.215; x_{\tau1} = 0.066; y_{\tau1} = 0.012; k_{53} = 0.1734; k_{54} = 183.16$
(4.20)	Ac ($f = \frac{0.04l}{s}, \frac{0.06l}{s}$)	Ac ($f = \frac{0.05l}{s}$)	$x_{\alpha2} = 0.4; y_{\alpha2} = 9.98; x_{\tau2} = 0.215; y_{\tau2} = 0.0128$

Here, “Am” refers to the case of heating and ambient cooling; “Ac” is considered as the instance of heating and active air cooling.

Figure 4. 14 and Figure 4. 15 depict the results of the developed models and the experiment data for the qualitative comparison, while *Table 4. 4* summarizes the Root Mean Square Errors (RMSEs) for the quantitative comparison. Overall, the models can accurately predict the heating and cooling temperatures with small RMSEs.

Figure 4. 14. Experiment data and results from models in the case of ambient cooling

Figure 4. 15. Experiment data and results from models in the case of active cooling

Table 4. 4: Root mean square errors from the models.

Case	Current/Air velocity	Root Mean square error (RMSE)
Heating and Ambient Cooling	$I = 0.2\text{A}$	0.7504
	$I = 0.3\text{A}$	0.7984
	$I = 0.4\text{A}$	1.3066
Heating and Active Cooling	$f=0.04\text{l/s}$	0.8465
	$f=0.05\text{l/s}$	0.5790
	$f=0.06\text{l/s}$	0.6799

4.5: Discussion and conclusion

In this chapter, a novel variable-stiffness manipulator whose stiffness can be adjusted on-command through joule heating and active air-cooling is developed. The proposed

manipulator has improved stiffness and flexibility in the two modes (rigid and flexible modes) when compared with the commercial variable-stiffness endoscope, and the former exhibits a significantly higher stiffness-changing ratio (22) than the latter (1.59). The validation study confirms that the robot has smaller unwanted deflection (8.36mm) under typical transverse force in NOTE (3.5N) while still be highly flexible in the flexible mode. With active cooling, the cooling time (11.9s) is almost 10 times shorter than the ambient cooling time (100.3s). Based on our experience on flexible endoscopy [142, 145], the activation time, in particular, 12 s for cooling, is sufficient. This can be understood from the detailed operation procedures of the robotic flexible endoscopic procedure: (1), the endoscope is inserted to the surgical site, and this insertion process is normally slow and takes more than 10 minutes. In this process, the endoscope should be in the flexible mode which can be achieved within 7.5-15 seconds with the proposed heating mechanism; (2) once the endoscope arrives at the surgical cite, robotic/manual instruments are inserted through the tool channels of the endoscope, which takes around 2~3 minutes or longer; during this time period, the endoscope can be cooled down to become stiff in around 12 seconds. Note that, during robotic endoscopic surgery, the robotic instruments inside the endoscope tool channels move frequently but the endoscope itself does not move much because it needs to serves as the stable platform of the robotic instruments; the surgeon may need to move the endoscope from one site to the other (not frequent), and this normally involves further manoeuvres of the endoscope which can take several minutes; thus, it is quite affordable to allow 12 seconds to adjust the stiffness. In addition, a mathematical model that describes the heating and cooling processes was derived and verified with multiple experiments. The RSME of the developed models ranges from 0.5790 °C to 1.3066 °C. The results confirm the feasibility of building a compact, variable-stiffness endoscopic manipulator with significantly high stiffness-changing ratio (low stiffness in the flexile mode and high stiffness in the rigid mode), shortened time for switching the modes, and safe thermal insulation.

Chapter 5: Conclusions and Recommendations for Future Work

This final chapter will summarize the main results and recommend several future research directions for this dissertation. The first section will present the main findings as the contributions of my work. The proposed variable-stiffness structure and its performance will be also outlined in this part. The limitations of the research will be given in the next section. Several recommendations for future work such as optimal design problem, variable stiffness control scheme will be discussed.

5.1: Summary of contributions

The major contributions of the PhD work are summarized as below

- Introducing a novel variable stiffness concept that has potential in the surgical robot using the combination of thermoplastic material (PET) and joule heat generated from applying current to the stainless steel coiled sheath.
- Developing a novel variable-stiffness tendon-driven robotic manipulator with embedded active heating and cooling. The stiffness of the robot can be significantly adjusted with varying temperature. The design has realistic features such as fast mode switching (short activation time) using active cooling, thermal insulation for human safety, and comparable size with the endoscopes, for real surgical applications.
- Tests and modeling of the multi-physics properties such as stiffness (flexural modulus), temperature, current, and air velocity. A series of tests like DMA, three-point bending, heating and cooling temperature measurements are combined with the heat transfer phenomenon to derive and verify the models.

5.1.1: A novel variable stiffness approach using PET

The first novelty contributed in this thesis is the development of a variable stiffness concept that makes use of a phase-changing material (PET). The selected material can exist in both flexible (rubbery) and stiff (glass) states upon the temperature that is adjusted by applying current into a stainless steel sheath embedded inside the PET tube.

The variable-stiffness tube (VST) is simple in design due to the combination of the PET tube and the flexible coiled sheath. Based on the three-point bending tests, the flexural modulus (stiffness) of the VST shows the immense difference between the two states. In particular, the flexural moduli are 2.141 GPa and 38.88 MPa in glass and rubber states respectively. After normalization, the Olympus commercial endoscope has only the flexural modulus of about 190 MPa which is 9 times smaller than the VST. In short, while maintaining the flexibility in the rubbery state, the VST is much more stable in the glass state. The flexural modulus of PET is also modeled by using the DMA tests, three-point bending tests, and the heat transfer analysis, which is also the novel knowledge of this study. This mathematical model can be used to calculate the flexural modulus at the given time during the heating process.

5.1.2: A novel variable-stiffness manipulator with active heating and cooling

The other novelty contributed in this thesis is the development of a variable-stiffness manipulator with active heating and cooling. Although various research on variable stiffness has been done and published, the majority of them still focused on the concept of working principle with the application designs overlooked. The proper application designs play important roles in extensively proving the initialized concept. For that reason, a continuum tendon-driven variable-stiffness endoscopic robotic manipulator was designed and developed. Embedding multiple VSTs, the robot's stiffness can actively be modified through joule heat and an active air-cooling mechanism. The robotic manipulator (Ø15 mm) has two bending sections and can be bend into "S" shapes by 8 driving tendons. Tests were conducted to characterize the stiffness, heating, and cooling characteristics of the robotic manipulator, and models were developed to describe the temperature with respect to time, applied current, and the velocity of the cooling air. The experiment results show that the proposed robotic manipulator not only has good flexibility, but also a high stiffness-changing ratio (22) between rigid and flexible states which is 13.84 times as large as that (1.59) of a commercial variable-stiffness endoscope. The thermal insulation keeps the temperature of the outer surface in the safe range (below 35°C) when the temperature on the PET tube was as high as 90°C. With the active air-cooling mechanism, the cooling time was shortened from

100.3s (ambient cooling) to 11.9s. The mathematical models for the heating and cooling processes have root mean square errors ranging from 0.5790 °C to 1.3066 °C compared with the experiment data. The results confirm the feasibility of a compact variable-stiffness endoscopic manipulator with significantly high stiffness-changing ratio (low stiffness in the flexible mode and high stiffness in the rigid mode), shortened time for switching the modes, and safe thermal insulation. The proposed variable-stiffness manipulator may be applied for flexible endoscopes in general flexible endoscopy and key-hole surgery.

5.2: Advantages and disadvantages

The PhD work provides an innovative solution for variable stiffness with various advantages such as the simplicity in design, the relatively short activation time, and the high stiffness ratio between two states. However, there still exist several limitations that need to be studied in future research.

5.2.1: Design

The VST consists of only a PET tube and a flexible stainless steel sheath making the structure simple. While PET is a great material that has relatively low transition temperature and high stiffness variation ratio, the sheath working as the heating element also brings several benefits. Firstly, its high resistance is beneficial for Joule heating process. In addition, it does not affect the stiffness of the design due to its high flexibility. Lastly, the sheath's tubular shape provides not only the sufficient contacting area with the PET tube for the heat transfer during the heating stage but also the active cooling channel, which helps shorten the activation period. However, some drawbacks still exist. For instance, the embedded electrical cables inside the PET tubes prevent us from exploiting the faster cooling media (e.g., water) due to the short circuit issue. Besides, in this study, the robot's dimensions are decided in such a way that it has a reasonable performance (stiffness, heating and cooling time), comparable size with the commercial endoscope, and fits the available off-the-shelf items (sheath, PET tube), but not fully optimized. This issue can be addressed with further study of the optimization design problem.

5.2.2: Modeling

The research provides a framework for modeling the heating and cooling process by combining the simplified heat transfer model and the experimental data. While this method greatly reduces the computational cost by avoiding the full complex heat transfer problem, it still accurately represents the experiment data with small root mean square errors. Nonetheless, the derived models with identified coefficients are only applicable for the given design. The reason is because the coefficients are determined by a certain set of experimental data that is design-specific. While the framework is reusable, the process of tuning the coefficients has to be redone if the design's dimensions change. For that reason, a generic model should be taken into consideration in the future.

5.2.3: Control

This work concentrates on the development, modeling, and validation of a variable stiffness approach. But, the design and implementation of a stiffness control scheme are still unaccomplished, which is another limitation of the research.

5.2.4: Experiment

Throughout the study, thorough experiments have been conducted to characterize and validate the design concept. However, they haven't been done in a realistic environment (e.g., inside a living pig) in which other factors such as temperature, humidity, and surrounding organs can affect the performance of the robot. In addition, more repetitive cycles of heating and cooling should be investigated to see whether it downgrades the PET thermal and mechanical properties.

5.3: Recommendations for Future Work

In this thesis, a novel variable stiffness design concept has been proposed and its good performance has been demonstrated via extensive experiments. However, there are still some limitations in the current study, which will be elaborated along with the possible future research directions.

5.3.1: The design optimization

The optimal design is the solution of a multi-objective optimization problem with the presence of trade-offs between conflicting objectives such as dimension constraints, overall stiffness and flexibility, heating and cooling time. For example, providing 2.3 mm channels, if high stiffness is required, PET tubes with considerable thickness are preferred. Nonetheless, enlarging PET tube thickness would lead to longer heating because of increasing the amount of heat needed. In addition, the cooling time is also prolonged due to the reduced diameter of cooling channels. To date, there is still a lack of related criteria such as the required stiffness or activation time. Thus, to have an optimal design, further data collection is necessary.

In the future, PET can be customized to different shapes (not limited by the tubular shape) using 3D printing or injection molding. By doing that, the designs can be expanded into other robotic applications such as variable-stiffness joints, links, and graspers.

5.3.2: Implementation of a real-time stiffness control scheme

In the current design, the stiffness of the robot mainly relies on the PET tubes whose rigidity is alterable with temperatures. Therefore, the stiffness control problem turns into the temperature control question. Temperature is adjusted through joule heat generated by the electrical current supplied to the stainless steel coils and an active air-cooling mechanism. A strategy that regulates the current and air cooling on/off in a collaborative manner needs to be designed and implemented to actively control the stiffness. In addition, the combination of more complex control algorithms such as PID or adaptive control with the temperature feedback from the thermal sensor 2 will also provide more accurate results.

5.3.3: Evaluation of the design in a more realistic surgical scenario

The environment (temperature, humidity, and surrounding organs) inside the human body are vastly different from benchtop setups, which will have effects on the efficiency of the proposed design. To have a deeper assessment of the robot performance, further

evaluation tests in terms of the stiffness, activation time, unwanted deflection, and the improvement it brings to the operators need to be conducted in more realistic surgical scenario (ex vivo or in vivo endoscopic procedures).

List of Publications

International Journals

- [1] **H.M. Le**, T.N. Do, and S.J. Phee, “A survey on actuators-driven surgical robots”. Sensors and Actuators A: Physical, 2016. 247: p. 323–354. (**Q1, IF=2.739**).
- [2] **H. M. Le**, L. Cao, T. N. Do, and S. J. Phee, “Design and modelling of a variable stiffness manipulator for surgical robots”, Mechatronics, vol. 53, pp. 109-123, 2018/08/01/ 2018. (**Q1, IF=2.978**).
- [3] **H. M. Le**, P. T. Phan, L. Cao, J. Liu, and S. J. Phee, “A Temperature-dependent, Variable-stiffness Endoscopic Robotic Manipulator with Active Heating and Cooling” to Annals of Biomedical Engineering, (**Q1, IF=3.474**), **Minor Revision**.

International Conferences

- [1] **H.M. Le**, T. N. Do, L. Cao, and S. J. Phee, "Towards active variable stiffness manipulators for surgical robots," in 2017 IEEE International Conference on Robotics and Automation (ICRA), 2017, pp. 1766-1771.

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