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Design and Modelling of a Variable Stiffness Manipulator for Surgical Robots

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Abstract

In Natural Orifice Transluminal Endoscopic Surgery (NOTES), a surgical robot that can access the human colon or stomach via natural orifices should have sufficient flexibility to pass through tortuous paths and to be operated in a confined space. In addition, the robot should possess an acceptable stiffness level to hold payloads during the surgery. This paper presents a new design concept for variable stiffness manipulators using thermoplastic material Polyethylene Terephthalate (PET) and a flexible stainless steel sheath as a heating media. The stiffness phases of PET can be actively adjusted through temperature. Experiments at different conditions showed that the proposed design was at least as flexible as a typical commercial endoscope in compliant mode and at least 9 times stiffer than the endoscope in stiff mode. In addition, flexural modulus of the proposed manipulator with respect to temperature, current, and time was modeled and validated through both simulation and experiments. A tendon-driven flexible robotic arm integrated with a variable stiffness spine was also developed, and ex vivo tests on fresh porcine tissue were conducted. The manipulator in compliant mode can be easily controlled through the tendons, and it is able to hold its shape against considerably large loads in stiff mode. The results demonstrate not only the high potential of the design concept for the future medical application but also the first steps toward building a complete surgical robotic system with fully controlled variable stiffness.

Key Terms: *Minimally Invasive Surgery (MIS); NOTES; surgical robot; variable stiffness robot; variable stiffness material.*

1. Introduction

Surgical robots are gaining popularity in the field of Natural Orifice Transluminal Endoscopic Surgery (NOTES), an endoscopic surgical intervention technique for treatment within the intraperitoneal cavity through natural orifices such as mouth, vagina, and anus [1, 2]. In NOTES, 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. In this case, the endoscope serves as a working platform for these end-effectors. Therefore, the endoscope in NOTES, on one hand, needs to be flexible to transverse through tortuous paths in human bodies without damaging human tissues; on the other hand, it has to be stiff enough to be pushed forward and to hold its shape against external forces when the end-effectors are working with the target. The stiffness variation in these two cases is significant, but existing endoscopes fail to fully meet these two conflicting requirements, which limits the performance and the use of surgical robots.

To advance endoscope design for these requirements, this paper proposes a new variable stiffness manipulator whose stiffness can be actively adjusted significantly. Apart from being crucial for endoscopes, variable stiffness also plays the same important roles in other surgical applications such as flexible robotic arms, catheters, surgical tools [3-5]. These medical devices have to be flexible to follow the tiny, non-linear paths inside human body and guarantee safety, and they also need to be stiff enough to transmit force during biopsies or grasping tasks, support other tools or increase the positioning and surgical accuracy [5].

In the literature, the working principles of variable stiffness can be categorized into two different domains: structures and materials. In the former category, stiffness is adjusted by reorganizing and/or reconnecting parts via attachable/detachable links inside the structure. Compliance is obtained when parts are detached, and high stiffness is obtained when parts are reattached [6]. For example, Yagi et al. [7] developed an outer sheath with a pneumatic driven slider linkage lock for endoscopic surgery. In this design, multiple cylindrical pieces consisting of the sheath, link, sliders, and channels are connected serially. If the inner channel is empty, all the parts stay in compliant mode. In contrast, rigid state is achieved if the air is applied into fluid channel, resulting in higher friction force between the piece and the slider. Although being advanced, the design is quite complex due to many small parts involved and the use of air pressure. In addition, the structures are too bulky to be used in the confined spaces in human. In other studies, cable tension was employed to stiffen cable-driven surgical robots [8-11]. The main disadvantage of using cable tension is the need of highly durable cables and links. Recently, particle jamming technology based on granular materials, e.g., dry sand or coffee beans, is gaining wide attentions [12-16]. The benefits of this approach include fast response and dramatic stiffness change between two states. However, high stiffness requires substantial volume of granular materials, resulting in bulky structures. In a recent study [17], a variable stiffness robotic link consisting of a cylindrical silicon outer tube and an inner plastic embedded mesh was developed. Stiffness is controlled by air pressure, resulting in large structure dimensions and therefore it is not suitable for surgical applications.

Phase-change materials are the other candidates for stiffness control, including electrorheological fluids, magnetorheological fluids [18-20], low melting point alloy (LMPA), phase changing alloy (PCA) [21, 22], and thermoplastic polymers [23-26]. 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. Although this approach is able to provide short activation time, high voltages and currents are required that is risky in surgical applications. Furthermore, these materials in rigid state are not stiff enough for use in some applications [6]. Recently, a variable stiffness manually operating platform using a mixture of indium, gallium and stannum was developed for stiffness adjustment in laparo-endoscopic single site surgery (LESS) [22]. There still exist several limitations in this design. For example, 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, 22 s with 89 °C hot water and 15 s with 18 °C cold water respectively. Field's metal alloy (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

was used to develop a continuum manipulator [21]. However, a high current of around 4 A is required for this design and therefore it is not safe for use in human. Although LMPA have relatively low transition temperatures, but they are not stable (rubidium and Gallium) and even toxic (Cerrolow 117). In addition, they do not possess high stiffness in rigid mode [5, 27, 28]. As a consequence, they are not ideal candidates for variable stiffness designs for surgical applications. In another study, Huan et al. [29] 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 similar method but different materials, researchers in [24, 25] have employed thermoplastic polymers polylactic acid (PLA) and acrylonitrile butadiene styrene (ABS) as variable stiffness solutions and shape memory alloy as heating method to change the stiffness of fabrics. However, PLA is too brittle in stiff mode [30], and ABS' glass transition temperature (105 °C) is too high for human body. In addition, although encouraging preliminary results have been obtained in these studies, there is a lack of studies on the comparison between the proposed designs and existing devices such as commercialized endoscopes, the modeling and control of these designs, etc.

In this paper, a new variable stiffness method using Polyethylene Terephthalate (PET) is proposed. The stiffness of the proposed structure can be significantly decreased upon heating using a flexible stainless steel sheath as an electrical resistor. Apart from immense stiffness change, PET was selected among other thermoplastic materials due to its biocompatibility, high strength, relatively low glass transition temperature (around 67 °C), high chemical resistance, and low cost [31]. The variable stiffness tube (VST) made of PET tube and flexible sheath was constructed and tested and compared to a commercial endoscope. The proposed design is at least as flexible as the commercial endoscope when flexibility is desired and at least 9 times stiffer than the endoscope when stiffness is desired. The flexural modulus of the proposed manipulator with respect to temperature, current, and time was modeled and validated through both simulation and experiments. To further demonstrate the effectiveness of the design, the VST was also validated in ex-vivo experiment (fresh pig tissue) with a flexible manipulator. It is worth to mention that this paper is the extended version of the authors' conference paper [32].

The detailed design, working principle, and preliminary testing results are given in Section 2, followed by Section 3 with stiffness modeling and related validation experiments. Section 4 describes the variable stiffness robotic arm, and Section 5 presents the conclusions and future work.

2. Conceptual design and preliminary experiment results

2.1. Materials

Thermoplastic materials are potential candidates for variable stiffness structures due to their flexibility upon heating and rigidity upon cooling. To provide variable stiffness manipulators for surgical applications, the materials should satisfy the following criteria: (1) high stiffness variation ratio; (2) glass transition temperature, i.e., the temperature at which the stiffness of the

material changes dramatically; (3) biocompatibility; (4) high strength. Table.1 shows these properties of common thermoplastic materials [33-36]. Among the current thermoplastic materials, Nylon and PET (Polyethylene terephthalate) have outstanding features to fulfill the requirements. Due to the humidity-dependent transition temperature of Nylon that can result in challenges for future control problem, PET was selected eventually.

Table 1: Properties of common thermoplastics [33-36]

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

Developed in the 1940's, PET 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 °C – as provided by Vention Medical Inc., USA). 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. In this study, we utilize the advantage of this thermoplastic feature to design a new type of variable stiffness tube. Furthermore, PET is also a great candidate for medical applications due to biocompatibility, clearness, lightweight, high strength, stiffness, favorable creep characteristics, low flavor absorption, high chemical resistance, barrier properties, low cost, and relatively low transition temperature [31]. The PET tubes used in this study are from Vention Medical Inc., USA.

For the heat transmission, Thomas et al. [24] employed a shape memory alloy (SMA) cable with applied current as heating source. The disadvantage of this idea is that the design will have undesirable shape changes due to SMA's shape memory effect and the contact area for heat transfer is very limited. In the proposed design reported here, we used a flexible stainless steel coiled sheath from Asahi Intecc Japan, Inc. Compared to the SMA cables used in [24], stainless

steel does not have shape memory effect and thus does not result in undesired shape changes. In addition, the coiled stainless steel sheath result in larger contact area for more efficient heat transfer. Stainless steel is also biocompatible and has a high electrical resistance which results in low applied current. This is because the applied current is fixed during the heating process. As a result, the generated heat by resistive heating is RI^2 . So, for the same amount of needed heat, higher resistance will result in lower required current. Moreover, the selected stainless steel sheath is highly flexible and thus has little effect on the stiffness of the manipulator in the rubbery state.

2.2. Design

The design and working principle are depicted in Fig. 1. The variable stiffness tube (VST) consists of an outer PET tube and an inner flexible stainless steel sheath (coiled tube). In terms of heating mechanism, using sheath does occupy the space from the whole structure. However, there are several advantages of using it in the proposed variable stiffness design. Firstly, using sheath as a heating mechanism speeds up the stiffness variation because, with the same overall length, coiled sheath can generate more heat and better heat distribution than single wire. Secondly, with the same tubular configuration as the PET tube, the inner channel of coiled sheath can be utilized for other purposes such as wiring, cooling, or instrument insertion in future applications. At room temperature, the PET tube is in its glass state and thus is stiff. At higher temperature (around 67 °C- glass transition threshold), PET tube will change from the glass state to the rubbery state, making the VST flexible. External current is applied directly to the flexible stainless steel sheath to generate heat. The VST can be employed as catheter outer tube Fig. 1(b) or robot's backbone as shown in Fig. 1(c).

Fig. 1. (a) Structure and working principle of the VST. (b) Catheter with variable stiffness overtube. (c) Flexible robotic arm backbone with its integrated VST

2.3. Preliminary testing

To measure the flexible modulus of the proposed VST, three-point bending tests are performed using the Instron 5569 material testing machine with 500 N and 50 N load cells (Instron Singapore ITW Pte Ltd.). The same tests were also performed for a commercial endoscope (GIF-2T240 from Olympus, Japan) for comparisons.

2.3.1. VST bending tests

The three-point bending diagram and the experimental setup are shown in Fig. 2. Three different tubes were used: a single PET tube, a VST in the glass state, and a VST in rubbery state. It is worth to notice that dimensions are very important when it comes to applications. However, the scope of this paper is on the conceptual design, modeling, and experimental validations rather than on the designs for specific applications. As a result, all the components are off-the-shelf items with the sizes specified in Table 2. In the future, the dimensions of the tube and the sheath will be optimized to fulfill specific application requirements on the stiffness, flexibility, and the

heating time.

Table 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 rubbery state. The bending test for glassy VST was conducted at room temperature, while that of rubbery VST started after supplying 0.3 A into the stainless steel sheath for 30 s. This was to ensure enough time for the VST change from the glass state to the rubbery state.

Fig. 2. VST bending tests. (a) The diagram of three-point bending test. (b) Instron 5569 material testing machine. (c) Bending test with single PET tube. (d) Bending test with VST in glass state. (e) Bending test with VST in rubbery state.

2.3.2. Bending test with commercial flexible endoscope

The bending test was also carried out for a commercial endoscope (Model GIF-2T240 from Olympus Co., Japan) to compare with the proposed VST in terms of stiffness and flexibility. Bending test was carried out at four different positions in the endoscope: at the middle of the flexible tip, and at positions of 20 cm, 40 cm, and 60 cm away from the end of flexible tip. For each position, the endoscope bending was measured with locked and unlocked modes. The endoscope is expected to be stiff in locked mode and compliant in unlocked mode. The experiment setup is described in Fig. 3. The endoscope was loaded with the speed of 2 mm/min, and the maximum deflection was set as 4 mm.

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

2.4. Results and comparisons

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

From Fig. 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 formula (Eq. 1) of the middle point deflection of an elastic beam of length L loaded by a central force F in three-point bending test [37]:

$$E = \frac{FL^3}{48\delta I} \quad (0)$$

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

Based on Eq. (1), the flexural modulus of glassy and rubbery VST are 2.141 GPa and 38.88 MPa, respectively. It can be seen that the stiffness of VST drops 55 times when shifting from the glass state to the rubbery state. Compared to the liquid metal [22] whose stiffness is only four times different between rigid and flexible states, the stiffness of the proposed PET-based VST can change more significantly.

Fig. 4. Bending test results for VST.

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

Fig. 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.

The bending test results of unlocked and locked endoscope and the comparison at each position in two different modes are shown in Fig. 5 and Fig. 6, respectively. 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 [38]. The principle 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.

Table 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 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 current commercialized endoscope; when stiffness is desired, the proposed VST is 9 times stiffer than the stiffest part of the endoscope (Fig. 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.

Fig. 7. Stiffness comparison between VST and the endoscope.

3. Stiffness modeling and experiments

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. 2 depicts this relationship:

$$\begin{cases} E = f_1(T) \\ T = f_2(d, t, I) \end{cases} \text{ or } E = f_1(f_2(d, t, I)) \quad (1)$$

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 modelling 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.

3.1. Dynamic Mechanical Analysis (DMA)

Dynamic Mechanical Analysis (DMA) is a widely employed approach to characterizing mechanical properties of polymers or composites upon temperature effect [39-41]. In a DMA test, an oscillating force which plays the similar role as the bending force is applied to a sample, and the material's responses to that force was 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 which includes the real part (E') as the storage modulus and the imaginary part (E'') as the loss modulus with its magnitude [42].

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

In this study, the tests were conducted in DMA Q800 from TA Instrument, 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 (Fig. 8) from DMA machines with three different runs illustrates 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.

Fig. 8. PET flexural testing results from DMA machine.

3.2. Three-point bending result

The dynamic flexural modulus varied with time was also measured and computed from 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. 1 and depicted in Fig. 9 with three different trials and 0.4 A current applied. Furthermore, the reverse direction from rubbery to glass state is also examined by 3-point bending and plotted in Fig. 10. The tests show that with only ambience cooling, it takes up to 80 s–100 s to change the state from flexible to stiff.

Fig. 9. Instron bending result with 0.4 A current applied

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

3.3. Heat transfer modeling

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 mathematic heat transfer model is based on the approach for electrical cables [43, 44] with different boundary conditions. Fig. 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. 4) and the homogeneous non-linear heat conduction equation for the PET layer (Eq. 5), which relate to each other by the conjugation conditions at $r = r_1$ (Eq. 6), the boundary conditions at $r = r_0$ and $r = r_2$ (Eq. 7), and the initial conditions (Eq. 8).

Fig. 11. The cross section geometry of VST

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

$$c \frac{\partial T}{\partial t} = \frac{1}{r} \frac{\partial}{\partial r} \left(r k \frac{\partial T}{\partial r} \right), \quad r \in (r_1, r_2) \quad (4)$$

$$\begin{cases} T_0|_{r=r_1} = T|_{r=r_1} \\ k_0 \frac{\partial T_0}{\partial r} \Big|_{r=r_1-0} = k_0 \frac{\partial T_0}{\partial r} \Big|_{r=r_1+0} \end{cases} \quad (5)$$

$$\begin{cases} r = r_0 : & k_0 \frac{\partial T_0}{\partial r} \Big|_{r=r_0} + \varepsilon_0 \sigma (T_{r=r_0}^4 - T_{env}^4) = 0 \\ r = r_2 : & k \frac{\partial T}{\partial r} \Big|_{r=r_2} + h (T_{r=r_2} - T_{env}) + \varepsilon \sigma (T_{r=r_2}^4 - T_{env}^4) = 0 \end{cases} \quad (6)$$

$$t = 0 : T_0 = T_{env}, T = T_{env} \quad (7)$$

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. 4 and 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 [45] (stainless steel grade SUS304 – Table 4).

Table 4: Heat conductivity coefficient of SUS304 with different temperature.

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. 9:

$$k = f(T[K]) = -0.14 + 0.003T - 9.84 \times 10^{-6} \times T^2 + 1.05 \times 10^{-8} \times T^3 \quad (8)$$

b) Heat capacity coefficients $c_0 = 500(J/kg.K)$ and $c(J/kg.K)$ as suggested in [46, 47] (Table 5)

Table 5: Heat capacity coefficient of PET with different temperature.

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

In the above section, the sheath coil is approximated as a tube for the 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) (Fig. 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. 4 during the simulation.

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

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

$$\begin{cases} R_{coil} = \rho \frac{2\pi L}{r_2 - r_1} \frac{r_2 + r_1}{2} \frac{1}{\pi \left(\frac{r_2 - r_1}{2} \right)^2} = \rho \frac{4L(r_2 + r_1)}{(r_2 - r_1)^3} \\ R_{tube} = \rho \frac{L}{\pi(r_2^2 - r_1^2)} \end{cases} \quad (9)$$

Eq. 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. 11). The length L is absent in this Eq. 11 because the proposed model is only 2D (per unit length) due to the symmetric design.

$$f_0(T) = I^2 \rho \frac{4(r_2 + r_1)}{(r_2 - r_1)^3}, \quad (10)$$

Where I is the direct current, ρ is the resistivity of SUS304 (using first order function regarding to temperature and data from [45] – Table 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. 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 $k_{tube} = R_{coil} / R_{tube}$. Figs. 13, and 14 show that the COMSOL simulation results are very close for the coil and the tube design although there is only a slight difference because the stainless steel tube has a better contact with the PET tube.

c) The convection heat transfer to air of the horizontal cylindrical surface is obtained from [43] (Eq. 12).

$$h = \left[0.1254(1/d)^{1/2} + 1.0932(T - T_{env})^{1/6} \right]^2 \quad (11)$$

Where d is the diameter of the cylinder.

d) Table 6: The resistivity of SUS304 with different temperature.

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

e) Other constants:

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

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

Fig. 13. Transient temperature at different time with 0.4 A direct current applied and the coil shape (a) and the tube shape (b).

The Fig. 8 depicts the flexural modulus versus temperature and Fig. 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 Fig. 14). As shown, the data extracted from DMA and 3-point bending experiments is really close to the simulation result, which means that the model and solutions are relatively accurate.

Fig. 14. Transient temperatures from COMSOL simulation and experiments

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 [48], of the thermal couples is 1 s, which matches with the measurement here.

3.4. Mathematical interpolation for flexural modulus experiment results

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. 13 and 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 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.

$$E_1 = f_1(T) = a_0 \tanh(a_1 T + a_2) + a_3 \quad T \in [30; 100] \quad (12)$$

$$E_2 = \begin{cases} f_2(T_1) = b_0 T_1^6 + b_1 T_1^5 + b_2 T_1^4 + b_3 T_1^3 + b_4 T_1^2 + b_5 T_1 + b_6 \\ f_3(T_2) = c_0 T_2^8 + c_1 T_2^7 + c_2 T_2^6 + c_3 T_2^5 + c_4 T_2^4 + c_5 T_2^3 + c_6 T_2^2 + c_7 T_2 + c_8 \end{cases} \quad (13)$$

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

Genetic Algorithm is an optimization approach which allows global space searching based on genetics and natural selections including several operators such as crossover, mutation, and reproduction [49-51]. 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. 15. The goal is to minimize the value of fitness function to get the best parameters.

$$Fitness = \frac{1}{N} \sum_{i=1}^N (E(i) - E_{1,2}(i))^2 \quad (14)$$

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

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

Table 7: The identified coefficients for the functions given by Eq. 13 and 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 to both coefficients and variables, which means that only small changes in coefficients or variables can lead to huge changes in function value.

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

Fig. 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. 16):

$$MSE = \frac{1}{N} \sum_{i=1}^N (f_{\text{experiment}}(i) - f_{\text{interpolation}}(i))^2 \quad (15)$$

Where i is the sampling index and N is the total number of samples from experimental results, $f_{\text{experiment}}$ is the actual 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 polynomial function is 8.6233×10^{-6} which is much smaller and better.

4. Validation experiments

To further validate the proposed VST, a multi-section, snake-like, and tendon-driven (the tendon is 0.27 mm in diameter bought from Asahi Intecc Co.) manipulator was designed, manufactured (3D printed), and then performs several tasks. Each single cylindrical link (12 mm in diameter) consists of one spherical joint and two channels of 2.4 mm and 2.0 mm intended for surgical

instrument and VST respectively (Fig. 16 (a, b)). After joining 10 similar links together, the final arm has a total length of 45 mm (Fig. 16 (c)).

Fig. 16. The details of multi-section robotic arm. (a) The front view of a single link. (b) The top view of a single link. (c) The complete system with 10 assembled links.

The first experiment is to verify the flexibility and bending capability of the robotic arm with embedded VST in rubbery state. The VST with the same dimension used in the heat transfer section was employed. 0.4 A current was supplied for 15 s before the test was started to make sure that the VST was in the rubbery state. For the visualization of robot's movements, readers are recommended to refer to the attached videos. Fig. 17 shows that the manipulator with VST in rubbery state can be significantly bent, the same as the one without VST.

Fig. 17. The bending capability of the robotic arm in two different scenarios. (a) Robot before bending. (b) Bending robot without VST. (c) Bending robot with VST in rubbery state.

The next test was conducted to compare the arm's capability of bearing weight in two different cases, with VST (in glass state) backbone and without it. With VST included, initially it is heated up to be flexible, then the robot bends 90° and stops there. After that, the current is removed to change the VST state from rubbery to glass. Ultimately, the weight is hang up to the robot's tip. Note that there is only a low cable tension (0.3 N) in both cases.

Fig. 18. The load holding ability of the robot (low tension (0.3 N) applied to the tendons). (a) Without VST and 10 g weight. (b) With VST in glass state and 10 g. (c) With VST in glass state and 20 g. (d) With VST in glass state and 50 g

The performances prove that with developed VST as backbone, the robotic arm is still able to bend easily but hold much more weight compared to non-backbone mechanism (Fig. 18).

Finally, an experiment with porcine stomach tissue from a local supermarket (ShengSiong Group Ltd., Singapore) was performed in order to observe the robot's movements in a surgery-like situation. In this section, one surgical instrument (biopsy forceps manufactured by Olympus) is inserted through the designed channel to approach and grasp the tissue. Then the tissue was lifted

up by cable tension force while the embedded VST was in compliant mode. Ultimately, the current was cut off to change the VST to the stiff mode before releasing the cable tension.

Fig. 19. Manipulator performance with pig stomach when cables tension is released. (a) Manipulator with glass VST. (b) Manipulator without VST

The experiment demonstrates that the manipulator is able to hold the big tissue without cable tension when the VST is inserted (Fig. 19). Refer to Video 1 and Video 2 in the supplementary materials for details.

5. Conclusions and Future Work

We have developed a new and promising design concept for variable stiffness manipulators using a PET tube and stainless steel sheath for surgical applications. 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 (Fig. 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.

Finally, a flexible snake-like multi-channel manipulator was designed and fabricated to test the performance with the pig tissue. Both surgical instrument and the developed VST can be inserted into the manipulator. It is encouraging to see that the manipulator with the VST can hold the tissue firmly, which would be difficult without the VST.

In conclusion, we proposed a new concept for the design of variable stiffness manipulators with various advantages such as the simplicity, the biocompatibility, and significant stiffness change between two states. Yet, several limitations still need to be solved. For instance, the glass transition temperature is relatively high for human body so that heat isolation needed to be considered in the design. In addition, currently, the working cycle is long because of passive ambient cooling. It can be observed from Fig. 9 that with only the current of 0.4 A, only 7 s–8 s is required to soften the structure. However, it takes up to 80 s–100 s to harden it by ambience cooling (Fig. 10). In the future, active cooling methods [52] will be investigated to shorten the cooling time as well as enhance the overall performance. Models and characterization data obtained in this study will also be utilized for the optimal design of variable stiffness manipulators. Designing a robust control algorithm is also our next goal to achieve, followed by the development of practical variable stiffness surgical tools such as endoscopes, robotic end-effectors, or other applications such as wearable devices, rehabilitation systems, and human-machine interfaces. Finally, related in-vivo experiments will be conducted to verify the performances of the developed devices.

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Supplementary materials

Supplementary material associated with this article can be found, in the online version, at doi:10.1016/j.mechatronics.2018.05.012.

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