Towards a MR Image-Guided SMA-Actuated Neurosurgical Robot

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Abstract—We present our work towards the development of a MR image-guided SMA-actuated neurosurgical robot. We used two antagonistic SMA wires as actuators for each joint in the robot, so that each joint can be actuated independently. We also modeled and tested the force behaviour of SMA wires in the bent configuration, which can be used as a guideline for SMA actuator selection. Due to the size scale of the robot, it is impossible to have individual position sensors at each joint and hence we rely primarily on vision feedback to control the joint motion of the robot. The images used to control the robot in this paper were obtained from a camera with the goal of eventually using MR images to control the end-effector motion. We then developed a control strategy and a switching circuit to control multiple links simultaneously and independently using only one power supply. Experimental results from our current prototype of a 3-DOF robot showed that we can actuate the robot and hence observe joint motion in a gelatin slab. We also did a series of experiments inside the MRI to show that the robot is fully MRI-compatible and creates no significant image distortion in the MR images.

I. INTRODUCTION

Shape Memory Alloy (SMA) has a special ability to memorize its shape at a lower temperature and recover large deformation on thermal activation. SMA can recover up to a maximum strain of about 8% and generate extremely large forces when it encounters external loads during phase transformation. This phenomenon is advantageous for use of SMA as an actuator. SMA actuator possess interesting properties such as large force density, biocompatibility, light weight, and noiseless operation when compared to conventional actuators, which makes SMA actuator a good candidate for a range of applications in surgical robots, such as active endoscope [1], active forceps for laparoscopic surgery [2], gastro-intestinal intervention system [3], and active catheter [4].

The space in MRI bore is usually limited, hence the actuator should be as small as possible, while at the same time, it should be able to generate enough force for the robot to move inside a tightly enclosed environment. Therefore, it is important to develop a theoretical model, which can be used to characterize the force behaviour of the SMA actuator, as a guidance to specify the actuator size (SMA wire diameter). A widely used constitutive model by Tanaka [5], Liang and Rogers [6], and Brinson [7] has been developed to describe the uniaxial tensile thermomechanical behaviour of the one-way shape memory effect while Thier [8], Plietsch [9] and Rahman [10] worked on the bending vs. external loading behaviour of SMA. For our application, the generated force behaviour in the bending configuration (to enable revolute joint motion) is the most important parameter which we are interested in characterizing and hence it is one of the primary focus of this paper.

In our previous work [11], we have developed a preliminary prototype of a Minimally Invasive Neurosurgical Intracranial Robot (MINIR) using SMA wires as actuators. In [12] and [13], a new design of MINIR which improves several limitations of the previous prototype is proposed. The new MINIR has individual SMA actuators for each joint, all joints are on the outside surface and all wirings are inside the robot. The design makes the robot more dextrous, more compact and easier to shield. In this paper, we demonstrate using a vision-based controller to control the various joints of the robot. Furthermore, we test the MRI compatibility of the robot so that continuous MR images can be used to provide visualization of the target as well as the tip position of the robot. We also developed electrosurgery probes at the tip of MINIR to electrocauterize the tumor.

In this paper, we discuss the expected functions and the design considerations of the robot in section II and present the bending force modeling and the simulation results in section III. Section IV describes the design of the vision-based feedback control system. In section V, we show the results of our SMA force characterization experiment as well as a series of the MRI compatibility tests of MINIR. We also show that the 3-DOF in able to move in a gelatin medium towards the target and the bending displacement of each joint can be reliably controlled by using vision feedback. Finally, in section VI, we make some concluding remarks.

II. ROBOT DESIGN

The design of the current prototype of MINIR has been detailed in our previous work [13]. The hollow core design (Fig. 1) enables all joints to be on the outside surface of the robot while keeping the robot center hollow. This hollow core allows for the passage of electrical wiring and makes the robot more compact and easier to shield. Furthermore, we
used two antagonistic SMA wires for each joint (Fig. 2), so that each joint could be moved back and forth and operated individually. This design greatly increased the motion range of the robot with independent joint actuation capability.

Due to the special operation environment, there were two major requirements that we have to address in this paper, namely, small actuator size and MRI compatibility of the robot. We will discuss these two requirements in detail in the following two subsections.

A. Actuator Selection

MINIR was designed especially for neurosurgical applications and it will be used in a human brain. Human brain is a tightly enclosed environment, therefore, MINIR has to have enough force to move inside a brain. MINIR will be introduced at the tumor site by passing through a narrow surgical corridor dissected by the neurosurgeon. The neurosurgeon will put MINIR inside a flexible cannula and guide the cannula through the corridor till it reaches the target location. Once at the target location, MINIR will slowly move back and forth to electrocauterize the tumor. It is obvious that larger SMA actuator can generate larger force, however, the space inside the robot is very limited. Besides, larger SMA actuator could also increase the rigidity of the robot and thus decrease the motion range. As a result, to find the suitable SMA actuator which can generate enough force while minimizing the size of it is also a major challenge when designing the robot. To achieve this goal, we developed a theoretical model to characterize the force behaviour of SMA wires in bending configuration. We also developed an experimental setup to perform the temperature vs. generated force test to compare the theoretical model with the experimental data.

B. MR compatibility

MINIR was designed to be operated under continuous MRI to eliminate radiation (such as in CT) and provide excellent soft-tissue contrast. Thus all materials of MINIR have to be MRI compatible. Our previous work [13] was shown that brass and SMA are MRI compatible, however, MRI compatibility is not the only requirement for MINIR. MINIR was also required to not interfere with the magnetic field inside the MRI bore and thus provide clear images as visual feedback for surgeons. Based on the success of this approach, we performed a series of MR imagining experiments when MINIR is actuated inside the MRI bore to see if the images are clear enough to be used for visual feedback.

III. SMA FORCE CHARACTERIZATION

The space inside the robot is very limited, thus, the size of actuators is extremely crucial. To size SMA actuators for MINIR, it is important to investigate the constrained recovery (blocked force) behavior of the material. In other words, a theoretical model to estimate the force capability of the actuators is necessary.

A. Modeling

This section describes the details of the model and outlines the necessary parameters for this model. We used a widely used Tanaka’s model [5], to describe the material behavior of the one-way shape memory effect. Most of the constitutive models were developed for uniaxial loading and assumed that the material at each instant is in thermodynamic equilibrium. According to [5], [6], [7], the strain ($\varepsilon$), temperature ($T$), and martensite phase ratio ($\xi$) are assumed to be the only and independent state variables in the constitutive equation of SMA. The constitutive equation can be written as:

$$\sigma - \sigma_0 = E(\xi)(\varepsilon - \varepsilon_0) + \Omega(\xi)(\varepsilon - \varepsilon_0) + \Theta(T - T_0)$$  \hspace{1cm} (1)

where $E(\xi)$ represents the effective Young’s modulus, $\Omega(\xi)$ is the phase transformation coefficient, $\Theta$ is the thermal expansion coefficient, $\xi$ is the martensite phase ratio, $\sigma$ is the stress of SMA and $(\varepsilon_0, \sigma_0, \varepsilon_0, T_0)$ are the material initial conditions. $\Theta$ is small and can be neglected. $\Omega$ is a constant for a specific material and can be correlated with Young’s modulus. If we define the maximum recoverable strain as $\varepsilon_L$, $\Omega$ can be defined as:

$$\Omega(\xi) = -\varepsilon_L E(\xi)$$  \hspace{1cm} (2)

The initial strain, $\varepsilon_0$, applied to each SMA wire for MINIR is 2.5%, the maximum recoverable strain, $\varepsilon_L$, of the SMA wire is about 8%. Since ($\varepsilon_0 < \varepsilon_L$), $\varepsilon_0$ can be fully recovered when the SMA wire is actuated. As a result, the $\varepsilon_L$ in (2) can be replaced by $\varepsilon_0$. Moreover, constrained recovery implies that the strain in the wire is constant ($\varepsilon - \varepsilon_0 = 0$), and the testing involves measurement of stress-temperature characteristics at a given pre-strain. MINIR is initially in a stress free condition ($\sigma_0 = 0$) and at room temperature ($\xi_0 = 1$). Based on the above discussion, (1) can be rewritten as:

$$\sigma = \varepsilon_0 E(\xi)(1 - \xi)$$  \hspace{1cm} (3)
The above equation is only valid for uniaxial loading case. However, in the operation of MINIR, bending behaviour is necessary to characterize. Thus, a new model based on (3) has to be developed to describe the bending behaviour of SMA. In this derivation, only pure bending is taken into consideration, ignoring the effect of shear stresses and assuming that the neutral plane remains at the center of the cross-section of the SMA wire. Considering that the SMA wire is pre-strained to an arc shape, the strain distribution \( \varepsilon \) of the wire can be defined as:

\[
\varepsilon(y) = \frac{y}{r} \quad (4)
\]

Where \( r \) is the radius of curvature of the SMA wire and \( y \) is the distance from the neutral plane. Since the SMA wire is at a pre-strain state, \( \varepsilon \) can be considered as the initial strain in the SMA wire \((\varepsilon_0 = \varepsilon)\). When the SMA wire is heated up, it intends to recover its original straight configuration and thus tensile stress is generated. The generated stresses then cause a bending moment \( (M) \) which can be expressed as a function of \( y \) and is given by:

\[
M = \int_A \sigma(y) \, da \quad (5)
\]

where \( a \) represents the cross-sectional area of the SMA wire. To obtain the stress distribution, \( \sigma(y) \), (3) has to be solved. However, Tanaka’s model assumes an exponential function for the martensite phase ratio. During heating transformation (martensite phase \( (M) \) to austenite phase \( (A) \)), \( \xi \) is given by:

\[
\xi_{M \rightarrow A}(\sigma, T) = e^{\sigma/(A_s - T)} + b_A \sigma \quad (6)
\]

Since the martensite phase ratio, \( \xi \), is dependent on the stress, \( \sigma \), in (6), the stress distribution has to be solved numerically.

**B. Numerical resolution**

Finite difference method is used to solve the bending behaviour of the SMA wire. We model the SMA wire to consist of \( i \) discrete layers along the \( y \) direction with \( t \) as the thickness of each layer and \( y_i \) as the distance from the neutral plane to layer \( i \) (see Fig.3). We assume that the thickness, \( t \), is small and the initial strain of each layer can be written as:

\[
\varepsilon_i = \frac{y_i}{r} \quad (7)
\]

Since the radius of curvature, \( r \), is known, the stress distribution can be obtained using Newton-Raphson’s method (8) to solve (3) iteratively after substituting the expression of \( \xi \) from (6).

\[
\sigma_{new} = \sigma_{old} - \frac{f(\sigma_{old})}{f'(\sigma_{old})} \quad (8)
\]

Since the effective Young’s modulus of SMA can be expressed as:

\[
E(\xi) = \xi E_M + (1 - \xi)E_A \quad (9)
\]

The function \( f(\sigma_{old}) \) and \( f'(\sigma_{old}) \) are obtained from the following expressions:

\[
\begin{align*}
 f(\sigma) &= \sigma - \varepsilon_0 E(\xi)(1 - \xi) \\
 f'(\sigma) &= 1 - \varepsilon_0 (E_M - E_A)b_A \xi + \varepsilon_0 (E_M - E_A)b_A \xi^2 + \varepsilon_0 E(\xi)b_A \xi 
\end{align*}
\]

After obtaining the stress distribution, the generated moment can be computed by:

\[
M = 2 \sum_{i=0}^{n} \sigma_i y_i da \quad (11)
\]

Since the moment arm is known, the tip force of each link of MINIR can be obtained.

To summarize, we first used the geometry of MINIR to derive the strain distribution in (4) of the pre-bent SMA actuator. We then used Tanaka’s model in (3) and Newton-Raphson’s method in (8) to find out the generated stress distribution of the SMA wire. Finally, we used the concept of finite difference to derive the generated moment given by (11) and thus the generated force at the tip of each link is obtained. Fig. 4 shows the simulated generated forces of a 0.508 mm diameter SMA wire as a function of temperature with different pre-strain. The parameters used in the simulation were experimentally determined based on our previous work [13]. It is apparent that the generated forces in the motion range of MINIR \((0^\circ \text{ to } \pm 30^\circ)\) are about the same. In other words, the generated forces are relatively independent to pre-strain. This significant observation implies that each joint of MINIR is able to generate the same force at different positions.

Based on the simulation results of this section, we can say that the 0.508 mm diameter SMA wire can generate up to 1.5N force, which is sufficient for our application. In section V-B, we show that the theoretical model matches the experimentally measured force quite well. Later in the same section, we show that MINIR is able to move in a gelatin slab. Gelatin is commonly used in the ballistic test to simulate human muscle and human muscle is a lot stiffer than brain tissue. Hence, we hypothesize that if MINIR is able to move in the gelatin, then we do not anticipate any problem in moving it inside the brain tissue.
IV. VISION-BASED CONTROL OF MINIR

In our previous work [13], we have successfully developed a thermomechanical model to control the motion of MINIR by monitoring the temperatures of the SMA wires through thermocouples. Though the approach was successful, there were some technical limitations to make it work reliably in the intended application. First, it is very difficult to attach a thermocouple to the SMA wire. When the joints are moving, the thermocouples sometimes detach from SMA wires, which makes the temperature measurement unreliable. Besides, since the diameter of the thermocouple is large (0.8 mm), it significantly increases the rigidity of the robot. As a result, the motion range and the generated force of the robot were reduced dramatically. To overcome this limitation, we decided not to use temperature as the feedback signal but instead rely on vision feedback to control the joint angle.

The goal of this section was to implement a vision-based feedback control system for MINIR. The vision feedback could be used to track specific features on the robot and obtain the position of those features. With the position of those features, we can then compute the bending displacement of each link. We plan to use this approach to control the motion of MINIR in MRI based on sequential MR images obtained during its motion. To track the features on the robot, we chose to use Lucas-Kanade Optical Flow Method [14]. Optical flow is widely used to track the motion of objects, surfaces, and edges in an image sequence. It assumes that the flow is essentially constant in a local neighbourhood of the pixel under consideration and solves the basic optical flow equations for all the pixels in that neighbourhood by the least squares criterion.

Since the bending displacement of each link of MINIR can be measured, we then implemented a controller to control the motion of each link. The current position \( P_{\text{cur}} \) of each link can be obtained through the vision-based feedback system. The position commands \( P_{\text{set}} \) of each joint can be sent either in real time or pre-programmed through the user interface. If the difference between \( P_{\text{set}} \) and \( P_{\text{cur}} \) for a specific link is larger than a threshold \( \delta \), the corresponding SMA actuator will be activated for a period of time \( t \). The activation time was computed using the following equation.

\[
t = k(P_{\text{cur}} - P_{\text{set}})
\]  

(12)

The proportional gain, \( k \), can be adjusted to change system performance. Note that the payload of each link is different and thus the gain, \( k \), is different for each link.

The control command, \( t \), is the width of the “ON” time of the discrete ON/OFF signal and it was sent to a switching circuit (see Fig. 5) to control the state of the switches. Note that only one power supply is required to control six SMA wires using this circuit. The DC power supply is used to provide a constant current to SMA wires and heat them up. When the control command is sent to the circuit, the corresponding switch would close for \( t \) seconds, enabling the assigned SMA wire to be heated for \( t \) seconds. This time period should be limited to an upper value, \( t_{\text{max}} \), to prevent overheat and the temperature drop of other SMA wires. When the link is actuated to a desired position, \( P_{\text{set}} \), or activated for \( t_{\text{max}} \) seconds, the control system switches to monitor the next SMA wire and therefore keeps all wires under control. If \( (P_{\text{cur}} - P_{\text{set}}) < \delta \), the system will immediately switch to monitor the next SMA wire. By switching current to the SMA wires at a high frequency, multiple SMA wires can be heated and maintained at a constant temperature by using one power supply.

V. RESULTS AND DISCUSSIONS

A. Robot Design

The electrocautery system has been successfully integrated to the current 3-DOF prototype of MINIR (see Fig. 6), and all wirings are routed inside the robot. The tip link of the robot was made by plastic because a lot of debris could be attached to the electrocautery tips after electrocauterizing tissues. Thus, the tip link is disposable and plastic can also provide better MR image quality than brass. Since the tip position is the most important area that we are interested in monitoring, using plastic is also advantageous for vision-based feedback control.

B. Generated force test

For the experimental apparatus design, we used similar design as what we used to test SMA parameters in [13]. We extended the device to perform the tests of force-temperature relation by adding a force sensor, as shown in Fig. 7. Two 0.508 mm diameter SMA wires were distributed antagonistically in a link of MINIR. The link was pre-bent to a desired position and attached to the device. The SMA wire was heated to a specific temperature and the corresponding force reading was recorded. The experimental results are shown in Fig. 8 which were consistent with the theoretical values that we derived in section III. The maximum generated force at
Fig. 7. Experimental setup for force testing

Fig. 8. Generated force at the tip of a link (experiment)

Fig. 9. Experimental testing of the 3-DOF robot in gelatin. (a) Home position, (b) tip joint actuated, (c) middle joint and (d) base joint actuated.

The tip of the link is about 1.4N at 75°C and the maximum forces at different positions were about the same. Since 75°C is too high for neurosurgical applications, it is well-known that the transformation temperatures of SMA can be tuned through heat treatment [15]. Hence in our future prototype for tissue experiments, we plan to have SMA with a 42°C austenite finish temperature to prevent damaging the healthy tissue.

We then tested a 3-DOF MINIR inside a gelatin slab to make sure that the 1.4N maximum force was enough for MINIR to move inside a tightly enclosed environment such as a human brain. We actuated each link independently and the test results are shown in Fig. 9. The results clearly demonstrate that the 3-DOF robot is able to move in a tightly enclosed environment and push the gelatin away. The horizontal displacement of the robot tip is about 20 mm, which is of the order of the size of a brain tumor.

C. MR comparability test

For MRI compatibility test, we evaluated the image quality of a 3-link MINIR with all wirings and electrocautery system. We put the robot in the gelatin and it was surrounded by gelatin to simulate the tightly enclosed environment in a brain. As seen in Figs. 10(a) and 10(b), the tip of the robot can be clearly identified in the image. The clarity of the tips in the MR images will enable us to control the motion of MINIR using MR image feedback. The bright white spots in the images were caused by the sharp edges of the link profiles, which can be fixed by rounding the edges. The blurred areas around the links are caused by air cavities which are created when we inserted the robot into the gelatin.

D. Vision-based control

We used the Lucas-Kanade optical flow method to track the motion of MINIR. By tracking the fiducial at each joint, we can compute the joint angle of each joint. The current joint angle can be compared with the desired joint angle and a straightforward proportional control scheme was implemented. The experimental setup is shown in Fig. 11.

In the experiment, we first moved the robot to a specific point. We then move the tip link back and forth to simulate the motion of removing the tumor using electrocautery and maintain the base and middle link at the same position. Note that after reaching the set point, the link cools down for 20 seconds and then the new command is sent. The positioning result is shown in Fig. 12, where it clearly shows that the controller can be used to control the motion of MINIR reliably. It also demonstrates that the controller is capable of controlling multiple links simultaneously and independently while using only one power supply. Figure 13 shows three different positions of MINIR when controlled using a vision-based control scheme. Note that the configuration of the robot here is defined by the angles of each link with respect to the robot base frame.

E. Motion of MINIR in MRI

In this section, we tested the 3-DOF MINIR under continuous MRI and used MR images as feedback to move the robot towards a target. We used omentum from rats to simulate brain tumor and embedded it into the gelatin as the target. The major composition of omentum is fat,
which provides good contrast in MR images. We then took MR images continuously (the frame rate was 4 fps) and actuated the robot manually towards the omentum. The experimental results are shown as Fig. 14. The image is clear and the electrocautery tips can be easily identified in the images. Hence, it should be possible for the surgeon to track and control the tip position of the robot visually under continuous MRI when performing electrocauterization. To semi-automate the process and provide the surgeon with possible trajectory plans, we will work towards implementing a image-based tracking scheme whereby the electrocautery tip of MINIR can be controlled to carry out the procedure.

VI. CONCLUSIONS

In this paper, we investigated the force behaviour of SMA wire in the bent configuration and used the result as the guideline to correctly select the size the SMA actuator. We also evaluated the motion of the robot in gelatin and observed that the generated force was enough for the robot to move inside the gelatin and the motion range was sufficient for neurosurgical applications. We also did a series of tests to demonstrate that MINIR is MRI compatible by actuating it in the MRI bore. We also demonstrated that a vision-based control scheme can be used to precisely control the motion of MINIR. Furthermore, the MR images also showed no significant artifacts which enabled us to use continuous MR images as feedback to control the robot to move towards the target with continuous user input. Although the quality of MR images is good enough for the surgeon to control the robot manually, it is not good enough for automating or semi-automating the procedure. The next step of this project will focus on improving the MR image quality using plastic material rather than brass to make the robot and finding suitable shield materials to cover the entire robot. We will then develop image-based control algorithms to enable the surgeon to do trajectory planning for optimally electrocauterizing the tumor.

REFERENCES