Control of Shape Memory Alloy Actuated Flexible Needle Using Multimodal Sensory Feedbacks

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***Abstract*—In this paper, closed loop nonlinear control of a shape memory alloy (SMA) actuated flexible needle is presented. Merits of different modalities of sensory feedbacks are studied. In order to track the given desired trajectory, the actual position of the needle tip is controlled using three different feedback signals (i) Vision, (ii) Electromagnetic (EM), and (iii) Ultrasound (US) imaging feedback signals are considered independently. From the experimental results it appeared that the EM sensor feedback control of the flexible needle has the best tracking performance compared to the other two feedbacks.**

***Index Terms*—first term, second term, third term, fourth term, fifth term, sixth term**

I. Introduction

Percutaneous intervention has recently peaked significant interest. Such procedures, performed in vivo, show promise for improvement in regards to efficiency, accuracy/precision, as well as minimizing the degree of surgical invasiveness, if automated. Tumor biopsy, surgical ablation, brachytherapy, deep brain stimulation, and localized drug delivery, among other procedures, can all benefit from an improved operative technique in order to reduce tissue trauma, scarring, and ultimately hospitalization/healing times. In the recent years, considerable interest in developing SMA actuators has stemmed from their inherent advantages in producing large elastic deformations, having high power-to-weight ratio and requiring low driving voltages. These advantages coupled with the biocompatibility that the alloys exhibit makes them good candidates for medical applications where the surgical workspace is limited.

Applications of SMA actuators in the biomedical engineering field include: smart needle [1]-[5], active catheters [6], and artificial muscles for prosthesis. In addition, SMA actuators have found applications in robotics such as in the areas of vibration control and active control of space structures because of their highpower to weight ratio. SMAs demonstrate a hysteresis characteristic as they are coerced from heating to cooling phases and vice versa, calling for more intricate control strategies that take this phenomenon into consideration. Hence, control accuracy is challenging. In percutaneous interventions, accurate placement of the needle to the desired target is very important. This accuracy is affected by many factors such as tissue-needle interaction and sensory feedback of the needle. In this study we have considered the sensory feedback of the needle affecting the accuracy of the needle placements intended for percutaneous interventions. So far we know, a comparison study with different feedback signals in achieving the closed loop control of a SMA actuated flexible needling device is not performed. Therefore, we

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are motivated to do this comparison study to assess the influence of multimodal sensing feedbacks.

The organization of the paper is as follows: First, the literature survey discussing the related work in the field of flexible needle control in percutaneous interventions is discussed in Section II. Then, Section III describes the Materials and Methods involved in this work. Section IV explains the experimental setup and Section V elaborates the results with discussion. Finally, conclusions are made in Section VI.

II. RELATED WORK

In the past two decades, several attempts have been made in the control of flexible needles. A detailed literature survey pushing the state of art in the needle control field is given in this section. Abolhassani *et al.* [7] have proposed a needle deflection model for controlling the needle along the desired tracks using US imaging feedback. Maghsoudi and Jahed [8] have presented a model based control strategy by estimating the force applied to the needle in percutaneous intervention. Ko *et al.* [9] have proposed a closed loop kinematic control scheme of a steerable probe using EM sensor data for a 2D trajectory tracking task. They have proposed the concept of "programmable bevel" for steering the probe in a defined direction using an approximate linearization technique.

In order to minimize the off-plane error, Kallem and Cowan proposed a novel plane-alignment control algorithm for needle steering along planar trajectories [10]. Using a stereo camera in order to measure the needle tip coordinates of a flexible needle they implemented a full-state observer to approximate missing states (such as the rotational degrees of freedom of the needle, which have been immeasurable). Additionally, Shoham and Glozman utilized fluoroscopically obtained images to measure the deflection of a rigid non-bevel-tipped needle when inserted into tissue [11]. In [12], a position approximate that is based on stereo camera images has been used to steer a thin and flexible bevel-tipped needle into gelatin. An "ON–OFF" controller, switching between the bevel left and right directionshas complemented by a path planning module, torsioncompensation, and an off-plane error minimization algorithm. While these approaches have been successful, they rely upon external sensing elements and complex image processing that limit the range of applications for this technology. Ayvali *et al.* [13] created a pulse width modulation (PWM) based temperature and vision feedback control of a discrete steerable cannula for diagnostic and therapeutic procedures. A PWM-based control system utilizing vision and temperature feedbacks has been successfully implemented to demonstrate local actuation of the cannula and a vision-based feedback controller has also been used to regulate and preserve the joint angle inside gelatin. In vitro experimental results demonstrate that both kinematic and control methods performed as expected. These early experimental results are promising and research to date offers significant scope for future work. Specifically, further studies are planned on the miniaturization of the current prototype down to a clinically acceptable size. Recently, Ruiz *et al.* [14] have performed a set point tracking control of a flexible needle actuated by SMA using EM sensory feedback in air and in-water environments. They have analyzed the performance of PID and PID-P3 controllers in the set point tracking task.

For minimally invasive medical procedure, Dupond *et al.* [15] proposed the design and control of preloaded concentric robotic tubes thereby having curvilinear configuration. The application of SMA in active catheters is first included by Haga *et al.* [16]. In [17], dynamic control of flexible needles for percutaneous intervention is performed based on Partial feedback linearization method. Hauser *et al.* proposed an algorithm facilitated by variable helical paths, based on the principle that the flexible needle generates a helical path during simultaneous insertion and rotation in tissue [18]. Lencioni *et al.* [19] presented a percutaneous image-guided radiofrequency ablation of liver tumors. Rucker *et al.* [20] have proposed a novel sliding mode controller for steerable needles using image guided interventions for both set point and trajectory tracking tasks within tissue environment. Recently, Okazawa *et al.* [21] have presented a steerable needle device for percutaneous surgeries that allows the surgeon to steer the needle tip during insertion which in turn lessens the difficulty linked with reaching the target using 3D US data. In the next section, the materials and methods involved in our current work are elaborated.

III. MATERIALS AND METHODS

A. Closed Loop Control

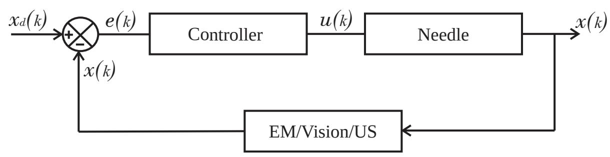

Figure 1. Closed loop control block diagram.

The closed loop control block diagram of the needle control system using the EM sensor, vision and US imaging feedback for trajectory tracking task is shown in Fig. 1. Two control algorithms discrete-time Proportional, Integral, Derivative (PID) and a nonlinear Proportional, Integral, Derivative, Cubic Error (PID-P^3) whose output equations given in (1) and (2) are employed for the experimental verification of closed loop control of the Nitinol SMA actuated flexible needle shown in Fig. 2 with three different real-time feedback signals namely EM, Vision and US imaging feedbacks.

Figure 2. Needle prototype used in control experiment.

The discrete-time PID controller is given by,

$$u(k) = u(k-1) + K_P[e(k) - e(k-1)] + \frac{K_P h}{T_I} e(k)$$

$$+K_P T_D[e_f(k) - 2e_f(k-1) + e_f(k-2)] \quad (1)$$

where h is the time step, $e(k) = x_d(k) - x(k)$ is the error at k^{th} instant. $e_f(k)$ is the low pass filtered error given by $e_f(k) = \left(\frac{1}{0.1T_D}\right)e(k)exp^{-t/0.1T_D}u(k)$ is the controller output at k^{th} instant. K_P , $\frac{K_P h}{T_I}$, $K_P T_D$ are the proportional, integral and derivative gains respectively. $h$is the time step. $e(k) = x_d(k) - x(k)$ is the error at k^{th} instant. The detailed derivation of the discrete-time PID controller is given in Appendix A.

Similarly the discrete-time PID-P[3] controller is given by,

$$\begin{aligned} u(k) = u(k-1) + K_P[e(k) - e(k-1)] + \frac{K_P h}{T_I}e(k) \\ +K_P T_D[e_f(k) - 2e_f(k-1) + e_f(k-2)] \\ +K_T[e(k) - e(k-1)]^3 \end{aligned} \quad (2)$$

B. *Experiment Design and Setup*

To track the needle tip using the EM sensor, the sensor coil is attached to the needle tip by aligning along the length of the needle as shown in Fig. 3. The SMA actuator wire is mounted at the proximal end of the needle thereby the interruption between the SMA wire and the EM sensor is avoided. The EM sensor is employed to track the coordinates of the needle tip's position. The sensor is an Aurora 5DOF sensor made by Northern Digital Inc. (Waterloo, Ontario, Canada) [22]. It is interfaced to the control computer through serial port. The respective lengths of the needle and the SMA actuator are 110mm and 70 mm. The discrete-time controller output $u(k)$ to the power amplifier circuit is limited in the range of (0 to 10V).

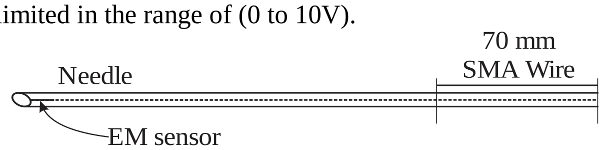

Figure 3. Installation of SMA actuator and EM sensor on the needle.

The power amplifier circuit is enabled to have the current passing through the SMA wire in the range of 0 to 1A. The control output is voltage and the process variable is the position of the needle tip.

The experimental setup of the SMA actuated flexible needling device mounted with EM sensor to perform the given desired trajectory tracking task is shown in Fig. 4.

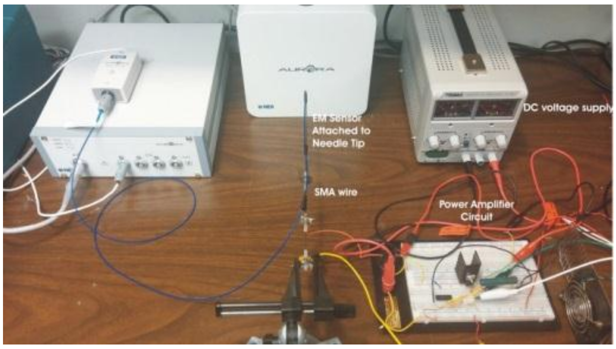

Figure 4. Experimental setup of the EM sensor based needle control system.

The purpose of this experimental work is to track the given trajectory. The dynamics of the system is not considered hence, it is purely a kinematic control. The experimental setups of the needle with Vision and US feedbacks are shown in Fig. 5 and Fig. 6, respectively. Asshown in Fig. 5, the vision system (USB 2.0 Logitec C920 HD WebCam, Newark, CA) is mounted vertically on top to capture the needle tip movement real-time at the rate of 30 frames per second.

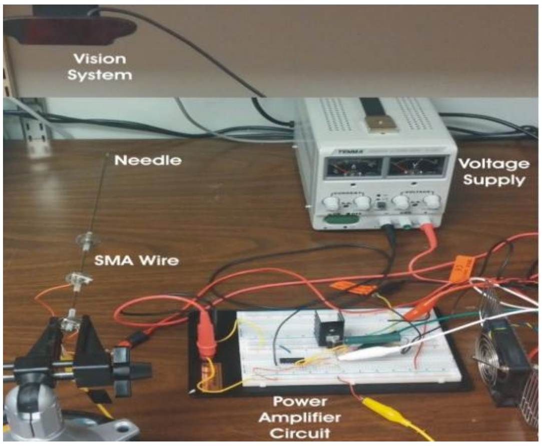

Figure 5. Experimental set up of the vision sensor based needle control system.

The needle is made sure that it is not mounted with any sensor during the vision feedback based control experiment. A small marker is affixed on the tip of the needle and using the conventional grey scale value pyramid pattern matching algorithm in LabView (NI Ireland), the marker is tracked in real-time. The US transrectal transducer (BK Medical, MA, USA) is employed to detect the needle using a DVI2USB 2.0 frame grabber (Epipham, CA, USA). After, detecting the needle in water, pattern matching algorithm is used to track the needle tip online.

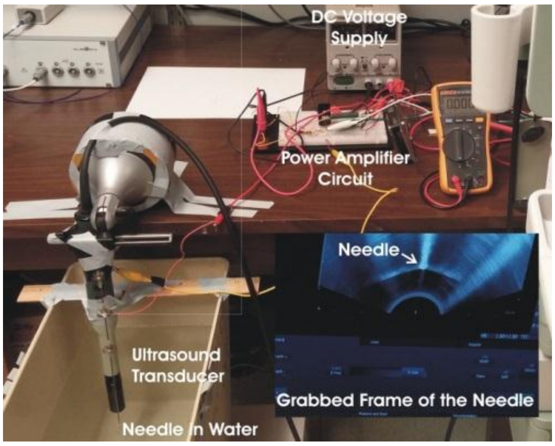

Figure 6. Experimental setup of the US imaging based needle control system.

Thus, we could able prepare the experimental setup to perform both the in-air and in-water experiments with the needle prototype. In the next section, both the in-air and in-water experimental results are presented.

IV. RESULTS

As, the PID-P^3 controller outperforms PID controller [9] we have shown only the results of the PID-P^3 controller alone in this paper. First, the closed loop control of the needle with EM sensor feedback is performed in-air. The trajectory tracking of the needle tip with the desired trajectory is shown in Fig. 7. The given desired trajectory consists of linearly incrementing the displacement to ten millimeters over twenty seconds, sustaining the ten millimeter displacement for five seconds, then decrementing the displacement linearly back to the starting point over another five seconds. The EM sensor data are collected at the rate of 50ms sampling time. The error plot of the trajectory tracking is shown in Fig. 8. The error plot is obtained by plotting the Euclidean distance between each desired way point and the actual way point of the needle tip.

Then, the trajectory tracking experimental task with the vision system feedback is performed for the needle in-air. Here, the sampling time is also 50ms. The trajectory tracking and the error plots related to the vision based feedback system are shown in Figs. 9 and 10, respectively. Then, finally, the in-water trajectory tracking closed loop control experiment of the needle is performed with the US transducer. The sampling period in extracting the US feedback signal is same as in the cases of EM and Vision sensors (50 ms). The trajectory tracking and the error plots for the US based needle control experiment are shown in Fig. 11 and Fig. 12, respectively.

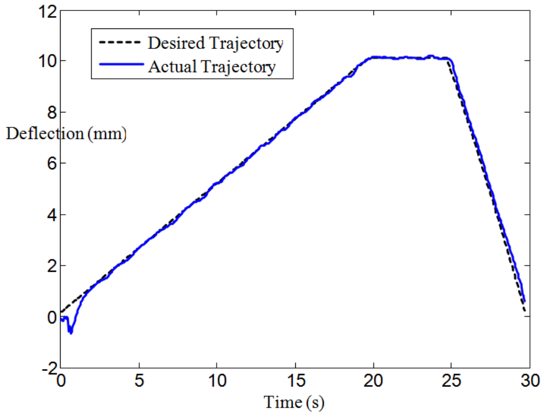

Figure 7. Trajectory matching between the desired trajectory and the actual needle tip trajectory with EM sensor data.

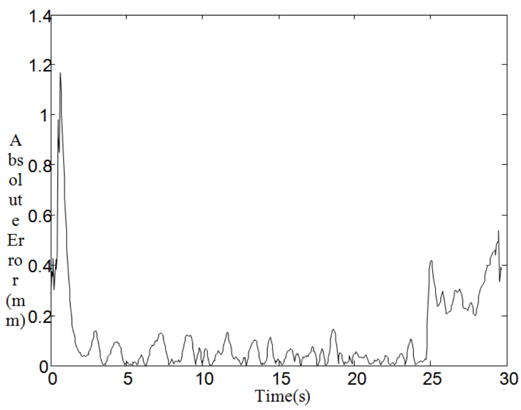

Figure 8. Error plot in the EM sensory feedback control task.

The Root Mean Square Error (RMSE) values of trajectory matching with the corresponding EM, Vision and US feedbacks are tabulated in Table I. From, Table I it is observed that the tracking with the EM sensory feedback has smaller RMSE value compared to that of the Vision and US imaging feedbacks.

TABLE I. COMPARISON OF THE RMSE VALUES WITHIN THE THREE EXPERIMENTAL CASES

Feedback type	RMSE (mm)
EM	0.1128
Vision	0.1502
US	0.1621

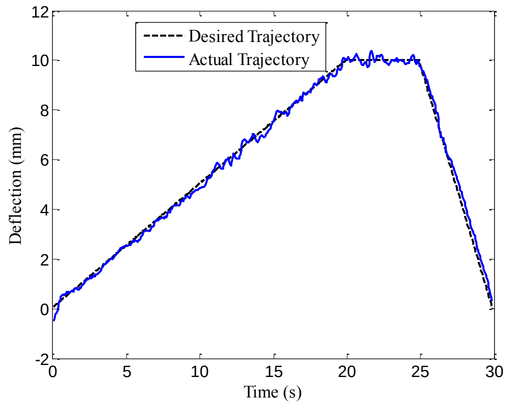

Figure 9. Trajectory matching between the desired trajectory and the actual needle tip trajectory with vision sensor data.

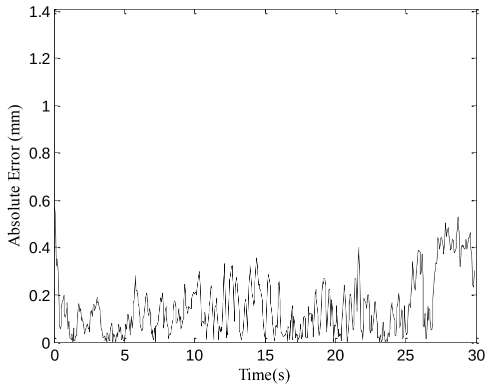

Figure 10. Error plot in the Vision sensory feedback control task.

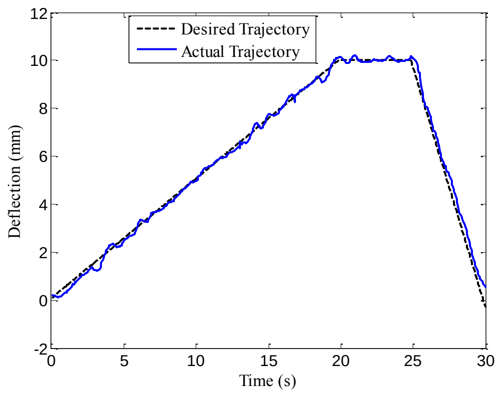

Figure 11. Trajectory matching between the desired trajectory and the actual needle tip trajectory with US imaging data.

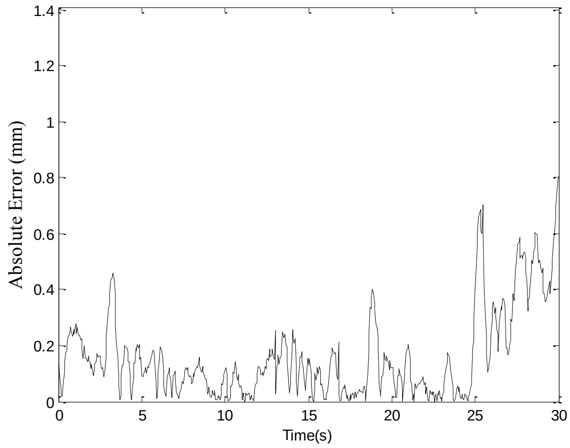

Figure 12. Error plot in the US sensory feedback control task.

V. Discussion and Conclusion

Experiments with a closed loop nonlinear control of a flexible needle actuated by SMA wire are performed using three different feedbacks which include EM, Vision and US. As per the quantitative measure shown with the RMSE values, the performance of the EM sensory feedback based closed loop controller is better with acceptable accuracy in the trajectory tracking task. Second best is through webcam vision feedback, and the worst performance is seen in the US imaging feedback. The most probable reason for this discrepancy is the fact that the EM sensor relays near-instantaneous coordinate feedback while the two image based feedbacks need to perform the extraneous step of pattern matching a template to the current image frame in order to extract coordinate data. Additionally, the poor image quality of the US when compared to the webcam vision requires more computation for the pattern matching step, making it the least efficacious method. The limitations in our current study include (i) non-involvement of the dynamics of the system (both thermodynamic characteristics of the SMA actuator and the tissue-needle interaction dynamics), (ii) controller adaptability in case of varying environment conditions including disturbances. In the near future, we will use dynamics of a discretely actuated steerable cannula in an imaging environment and also we plan to develop a high-level controller to switch between image-based and EM sensor based feedbacks for position control, especially when there is poor image quality from the imaging system. We are also looking into incorporating tissue models into the control framework in order to enable better trajectory planning. The work presented in this paper is the first step towards realizing these aforementioned goals.

AppendixA Appendix Title

The discrete-time PID controller is derived originally from the continuous-time PID controller:

$$u(t) = K_p e(t) + \frac{K_p}{T_I}\int_0^t e(t)dt + K_p T_D\, \dot{e}_f(t) \qquad \text{(A1.1)}$$

Here, $e_f(t)$ is the filtered output obtained by removing the noise through a low-pass filter:

$$e_f(s) = \frac{1}{\tau_f s+1} e(s) \qquad \text{(A1.2)}$$

where, τ_f is the time constant of the low-pass filter. It is selected as,

$$\tau_f = cT_D \qquad \text{(A1.3)}$$

Usually, c = 0.1;

By, differentiating both sides of (A1.1) we get,

$$\dot{u}(t) = K_p \dot{e}(t) + \frac{K_p}{T_I} e(t) + K_p T_D \ddot{e}_f(t) \qquad \text{(A1.4)}$$

By Backward differentiation method, we can substitute$\dot{u}$, $\dot{e}$ and $\ddot{e}_f$ by,

$$\dot{u}(t) = \frac{u(k) - u(k-1)}{h}$$

$$\dot{e}(t) = \frac{e(k) - e(k-1)}{h}$$

$$\ddot{e}_f(t) = \frac{e_f(k) - e_f(k-1)}{h} - \frac{e_f(k-1) - e_f(k-2)}{h}$$

Thus, (A1.4) becomes,

$$\frac{u(k) - u(k-1)}{h} = K_p \frac{e(k) - e(k-1)}{h} + \frac{K_p}{T_I} e(k) + K_p T_D \frac{\frac{e_f(k)-e_f(k-1)}{h} - \frac{e_f(k-1)-e_f(k-2)}{h}}{h}$$

Eventually, by solving for $u(k)$ from the above equation, we get the discrete-time PID controller:

$$u(k) = u(k-1) + K_p[e(k) - e(k-1)] + \frac{K_p h}{T_I} e(k) + \frac{K_p T_D}{h}[e_f(k) - 2e_f(k-1) + e_f(k-2)] \qquad \text{(A1.5)}$$

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