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Robotics-Assisted Catheter Manipulation for Improving Cardiac Ablation Efficiency

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Abstract— **The quality of contact between the catheter tip and cardiac tissue has been identified as an important factor in the efficacy of the catheter-based cardiac ablation procedures. However, maintaining a constant tip/tissue contact force during the procedure is difficult due to cardiac and respiratory motions. Robotic manipulation of the catheter has the potential to overcome this difficulty and decrease the range of variations in the contact force during the ablation procedure. This paper investigates the possibility of performing motion compensation for conventional steerable ablation catheters using a robotic manipulator. The behavior of such catheters is analyzed in free space as well as in contact with static and moving targets and the limitations in the actuation mechanism are identified. Based on this analysis, a technique for synchronizing the motion of the catheter tip with cardiac motion is proposed. The suggested control system estimates the frequency of the moving target and reshapes the input trajectory accordingly. The performance of the resulting system is evaluated experimentally. The results show that in the experimental setting, the proposed technique reduces the variations in the contact force and noticeably improves the quality of tip/tissue contact.**

I. Introduction

During cardiac arrhythmia, the normal heart rhythm is affected by false electric signals originating from the myocardium. Minimally invasive treatment of arrhythmia is realized using steerable ablation catheters. The flexibility, length and small diameter of the catheter allow it to be steered to the appropriate heart chamber through the vasculature. The catheter tip is positioned on the cardiac tissue to ablate the source of false signals by delivering some type of energy (most commonly radiofrequency (RF)) to the tissue. The ablation procedure is usually performed under X-ray fluoroscopy guidance, which does not provide a good visualization of the cardiac tissue, but displays the position of catheters and guidewires.

The recent development of force sensing catheters has facilitated research on defining the relationship between the amount of contact force and the outcome of the ablation procedure [1]–[3]. Considering that insufficient force results in incomplete tissue ablation and possible recurrence of arrhythmia, and excessive force increases the risk of perforation, it is important to keep the contact force within a certain range at all times during the ablation procedure. Moreover, a recent study suggests that the quality of contact plays an important role in the efficacy of the procedure [4]. Using a bench model that simulates the beating heart, Shah *et al.* [4] define three contact patterns: 1) constant contact, 2) variable contact ($F_{min} = 10$ gf and $F_{max} = 20$ gf) and 3) intermittent contact ($F_{min} = 0$ gf and $F_{max} = 20$ gf) [4]. Establishing a constant contact between the catheter tip and the tissue results in complete ablation and reduces the maximum applied force [5]. However, due to heartbeat and respiration, there is an intermittent contact between the catheter tip and cardiac tissue [6]. If the procedure is performed under apnea, where the respiratory movements are minimal, a variable contact can be achieved and the contact quality is improved considerably [6]. Providing a constant contact force further improves the quality of contact. It is expected that a constant force can be attained if the catheter tip moves synchronously with the cardiac tissue and compensates for cardiac and respiratory motions.

Motion compensation of beating heart in robotics-assisted minimally invasive cardiac interventions is usually realized through tracking visual clues on the heart surface and/or using a motion compensation robotic tool (*e.g.*, [7]–[9]). However, these techniques are mainly developed for surgeries performed on the external surface of the heart using rigid tools. Yuen et al. designed a rigid motion compensation device for performing intracardiac procedures under ultrasound guidance [10], [11]. A similar approach was then adapted to design a flexible robotic motion compensation catheter for intracardiac surgeries, namely the mitral valve annuloplasty procedure [12], [13]. The designed catheter is translated in and out of the sheath to compensate for the beating heart motion. The contact force is measured by a prototype force sensor integrated within the catheter tip [14].

In our previous work, we studied the static behavior of the catheter tip under applied forces [15], [16]. We also showed that controlling the contact force at the tip of an ablation catheter can be achieved through adjusting the position of the prismatic knob on the catheter's proximal handle [17]. In this paper, we propose a technique for compensating for heart motion to reduce the variations in the contact force between the tip of **a conventional unidirectional steerable ablation catheter** and the environment (cardiac tissue). A thorough study is carried out to determine the characteristics of a conventional ablation catheter when the catheter handle is actuated in different frequencies. The results of this study

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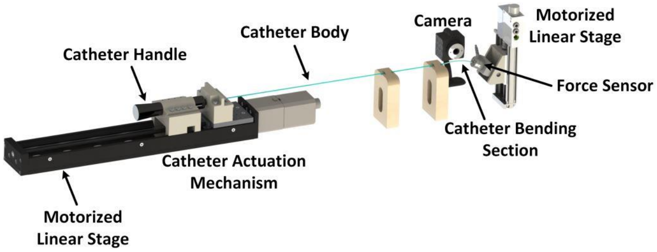


Fig. 1: Schematic of the experimental setup.

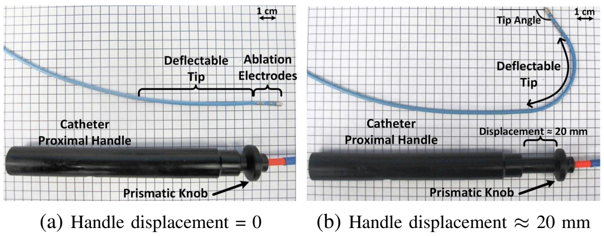


Fig. 2: A close view of the proximal handle and the bending tip of a steerable ablation catheter

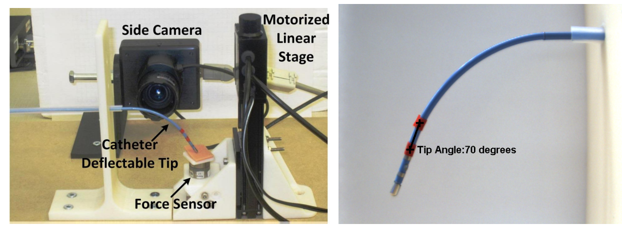
are then used in designing a control system that estimates the frequency of the environmental motion, calculates the actuation delay and generates the proper control signal. The experimental results show that the proposed technique is capable of reducing the variations in the contact force caused by the moving tissue and noticeably improves the performance of the uncompensated system.

II. Experimental Setup

The setup is shown schematically in Figure 1. The proximal catheter handle is fixed to the base of a motorized linear stage (T-LSR300B, Zaber Technologies Inc., Vancouver, BC, Canada) by an adapter piece. The prismatic knob on the catheter handle is attached to the sliding platform on the linear stage and the catheter actuation is realized by moving the sliding platform back and forth. In this setup, the catheter proximal handle can be actuated with a maximum speed of 20 mm/s. A conventional unidirectional 7-Fr *RF* ablation catheter (Biosense Webster, Diamond Bar, CA) is passed through a sheath which is fixed in place along its length.

The distal section of the catheter is the deflectable shaft that can be actuated by moving the proximal prismatic knob. The prismatic knob on the adapted catheter can move from 0 to about 21 mm (Figure 2). When the knob is at its initial position (handle displacement = 0), the catheter tip is completely straight. For a displacement of about 21 mm, the catheter tip bends to over 90° (Figure 2).

To measure the contact force without altering the structure of the catheter, a force sensor (Nano17, ATI Industrial Automation, Apex, NC, USA) is used as the environment with which the catheter interacts (Figure 3a). A small piece of artificial tissue is fixed to the top plate of the sensor. To adjust the angle between the catheter tip and the force sensor, the sensor is mounted on a small adapter piece. A motorized linear stage (T-LSM050A, Zaber Technologies Inc., Vancouver, BC, Canada) follows the commands to move the force sensor up and down and thus provides a moving



(a) (b)

Fig. 3: (a) A view of the catheter tip, camera and force sensor, (b) a sample frame

target for the experiments. A camera (Dragonfly®, Point Grey Research Inc., Richmond, BC, Canada) is positioned to show a planar view of the catheter tip and provides images with a resolution of 640×480 pixels from the deflectable tip. The images are streamed at 30 frames per second (fps).

In order to have real-time feedback from the position and orientation of the catheter tip, two markers are fixed at the end of the deflectable section behind the rigid part reserved for ablation electrodes and are tracked in the streamed video (Figure 3b).

In this study, the largest motion frequency in the experiments is lower than the frequency of cardiac motion ($\approx$ 1 Hz). The reason for this is explained in detail in Section V. It is reported in [18] that the mean amplitude of the motion of the aortic root is 11.6 mm and that of the atrial wall is about 8.5 mm [19].

III. System Characteristics

Catheters have applications in a variety of vascular and cardiac interventions, *e.g.,* cardiac ablation and coronary angioplasty. Cardiac ablation catheters are usually 5-Fr or 7-Fr catheters and can be either unidirectional or bidirectional. It is shown that even forces as low as 77 gf ($\approx$ 0.75 N) can result in cardiac perforation [3] and force levels for achieving efficient ablation are generally lower than 35 gf ($\approx$ 0.34 N) [2]. Cardiac ablation catheters are designed to apply forces in the required range.

In this section, we study how the catheter tip responds to actuating the proximal handle at different frequencies. Hereinafter, the tip position refers to the position of the red marker closest to the ablation tip (Figure 3b) and the tip angle is defined as shown in Figure 2b.

A. Frequency Response in Free Space

From the results presented in [17], the contact force can be controlled through adjusting the position of the handle knob. Since the target, *i.e.,* the heart tissue, is moving very fast, it is expected that for maintaining a constant contact, the catheter handle should be actuated with a high frequency ($\approx$ 1 Hz) as well. In this section, we study how the catheter tip responds to actuation at the proximal knob.

The catheter handle is actuated with a 6 mm sinusoidal command centered at $d = 14$ mm, taking into account that for $d < 9$ mm, the deflection of the catheter tip is insignificant. The handle is actuated at 20 different frequencies, ranging from 0.02 Hz to 0.65 Hz for 20 s. At

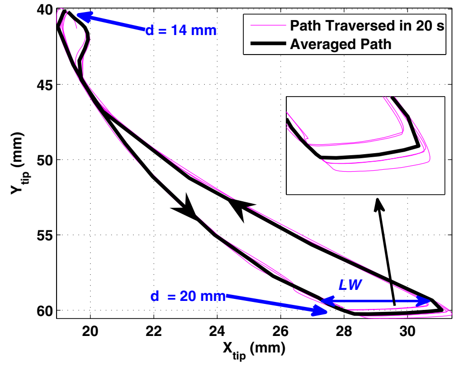


Fig. 4: The path that the tip follows when F_a = 0.27 Hz.

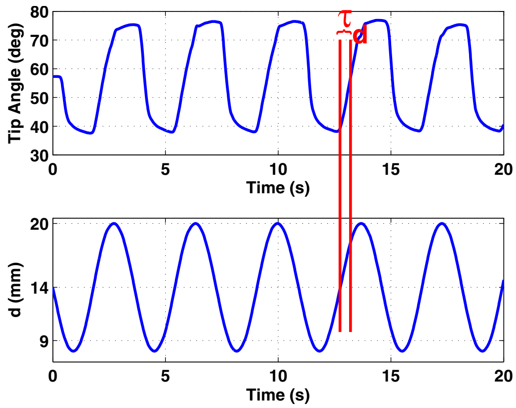


Fig. 5: Variations in the tip angle when F_a = 0.27 Hz: (top) tip angle, (bottom) handle displacement

each actuation frequency, F_a, the experiment is repeated 3 times. The corresponding signals are recorded and for each repetition, the following features are extracted (Figures 4-5):

- F_a (Hz) is the frequency of actuating the proximal knob.
- F_t (Hz) is the frequency at which the tip moves.
- LW (mm) is the maximum width of the hysteresis loop corresponding to the tip position.
- PL (mm) is the average length of the path that the catheter tip traverses when the knob is actuated from $d = 0$ to $d = 22$ mm and back to $d = 0$.
- Φ_{min} (deg) is the minimum angle that the catheter tip returns to when $d = 0$.
- Φ_{max} (deg) is the maximum angle that the catheter tip reaches for $d = 22$ mm.
- τ_d (s) is the time delay, *i.e.*, the time elapsed between when $d = 14$ mm and the time at which the tip angle reaches its mean value.

During the actuation period of 20 s, the handle is displaced between $d = 0$ and $d = 22$ mm repeatedly and each cycle takes $1/F_a$ seconds. As a result, the catheter tip traverses the same path recurrently. The defined parameters are derived for each actuation cycle and the final value for each parameter would then be the average of the corresponding value in all cycles. The values obtained for different repetitions of the same actuation frequency are found to be very close. Samples of the obtained results are shown in Figures 4 and 5.

The experimental results suggest that the frequency of the tip motion is the same as the actuation frequency as expected, *i.e.*, $F_t \approx F_a$, and the maximum width of the position hysteresis loop, LW, does not depend on F_a. However, increasing the actuation frequency causes:

- the catheter tip to traverse a shorter path. The catheter tip returns to almost the same position for $d = 0$, but as F_a is increased, the catheter tip does not reach its maximum bending configuration and the furthest point that it can reach gets closer to its starting point.
- the range of Φ decreases, *i.e.*, Φ_{max}-Φ_{min} has a smaller value for higher F_a.

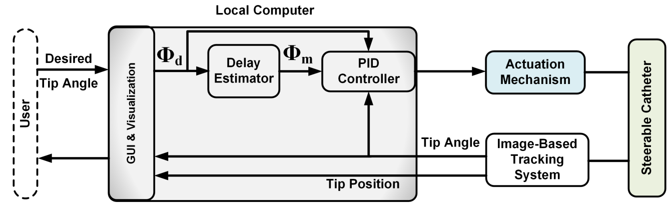


Fig. 6: The control system for achieving a desired angle at the catheter tip.

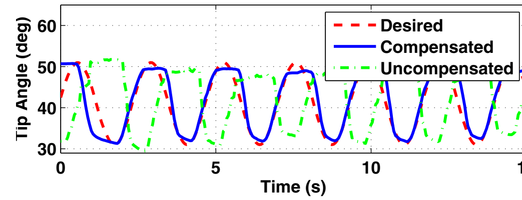


Fig. 7: Performance of the control system in following a desired tip angle.

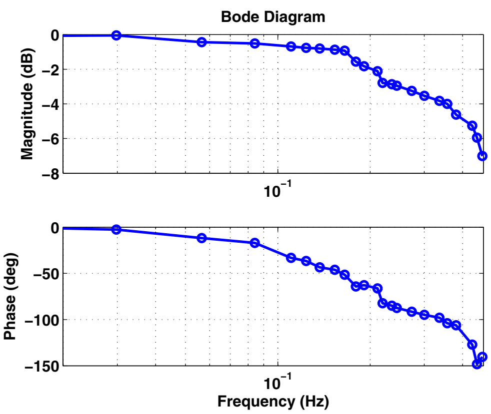


Fig. 8: Bode diagram for the system with the desired tip angle as input and the actual tip angle as output.

B. Achieving a Desired Tip Angle

In this section, we investigate the problem of controlling the catheter tip in free space. A PID controller is used for this purpose (Figure 6) and it is commanded to follow a time-varying tip angle in the shape of a sinusoidal signal centered at 42° with an amplitude of 10°. The experiments show that the system output is in the shape of a sinusoidal and that the time delay has an undesirable but noticeable effect on the response (Figure 7). Hence, the commanded trajectory is modified to compensate for the delay:

$$\Phi_m(t) = \Phi_{des}(t + \tau_d) \qquad (1)$$

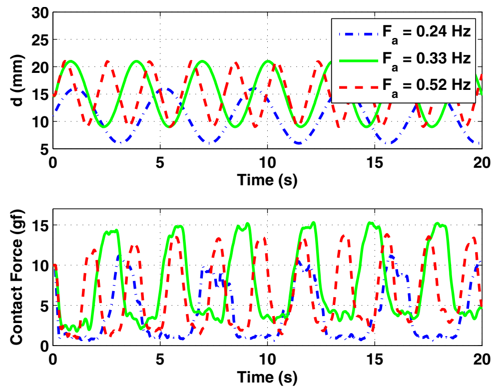


Fig. 9: Catheter interacting with a static environment: (top) commanded handle displacement, (b) tip/tissue contact force.

where Φ_{des} is the desired trajectory. τ_d is the time delay and can be obtained from Figure 8. Φ_m is the modified trajectory that is fed into the control system. Since the main source of error in tracking the desired tip angle is the time delay, the modified trajectory is the desired trajectory with a phase shift. Figure 7 shows that reshaping the desired trajectory and compensating for the time delay improves the tracking performance noticeably, reducing the root mean square of tracking error from 12.71° to 2.83°.

IV. Tip/Tissue Motion Synchronization

In the previous section, we studied how the catheter tip responds to actuating the proximal handle in free space and identified the delay inherent in the system. In this section, we study the catheter behavior in contact with the environment.

A. Static Environment

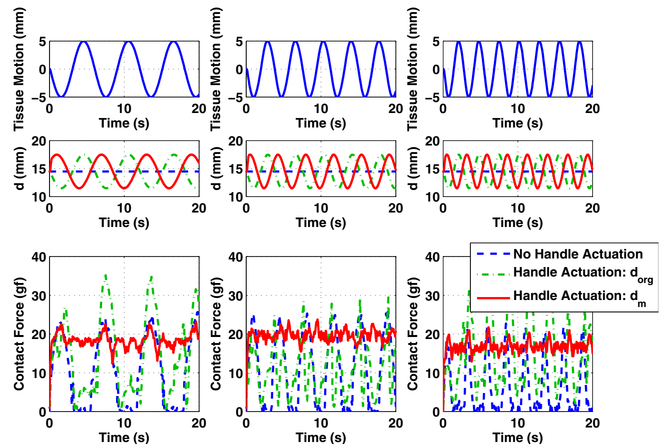
During the ablation procedure, the catheter tip is in contact with a relatively fast-moving environment, *i.e.*, cardiac tissue. When the tissue moves further from the tip, its contact with the catheter tip might be lost. The question that arises is if it is possible to maintain the contact by controlling the flexing of the tip. To investigate the answer, we first study if flexing the tip will change the force applied on a static environment.

In the experimental setup, the force sensor is placed such that when the tip flexes, it comes in contact with the artificial tissue. The proximal knob is then commanded with a sinusoidal signal at different frequencies. The results are given in Figure 9 and show how the contact force changes when the handle is actuated: when the handle is displaced between 5 mm and 20 mm, the variations in the contact force are about 15 gf.

These results confirm that the catheter tip/tissue contact force changes when the catheter proximal handle is actuated.

B. Moving Environment

In the experimental setup, the force sensor is commanded to follow a 5 mm sinusoidal signal as a simplified version of the motion of the atrial wall. The proximal handle is also commanded to follow a sinusoidal signal to provide the necessary flexing of the catheter tip. It is intuitive that in order to synchronize the motion of the tip with that of



(a) F_m = 0.17 Hz (b) F_m = 0.27 Hz (c) F_m = 0.37 Hz

Fig. 10: Catheter tip/tissue contact force when the tissue has a sinusoidal motion: (top) motion of the tissue. (middle) different scenarios for actuating the proximal handle, (bottom) contact force; F_m: Frequency of motion

the environment, the frequency of handle actuation and the frequency of motion should be the same.

Based on the results presented in Section III, the input command to the proximal knob is reshaped to compensate for the time delay between the handle displacement and the resulting change in the shape of the tip:

$$d_m(t) = d_{org}(t + \tau_d) \quad (2)$$

where d_{org} is the sinusoidal signal with the same phase as that of the motion signal. τ_d is the time delay obtained from Figure 8. d_m is the modified command to the proximal handle which compensates for the time delay.

Figure 10 shows the tissue motion, the handle displacement and the measured contact force at three different frequencies. The force sensor is placed such that the pattern of the contact force matches the intermittent contact [4]. It is observed that actuating the catheter handle without compensating for the system delay causes larger variations in the contact force. Obtaining the frequency response of the system allows the time delay to be estimated at each frequency and to modify the command to the proximal handle accordingly. Figure 10 shows that synchronizing the motion of the catheter tip with that of the tissue results in a noticeable improvement in the quality of contact.

C. Online Estimation of Motion Frequency

In the previous section, it was assumed that the frequency of tissue motion is known. The input signal d_m was then designed to compensate for the system delay. Knowing that in order to have effective tip/tissue motion synchronization, the proximal knob should move with the same frequency as that of the tissue, online estimation of the frequency of tissue motion is required. To this end, the catheter is brought in contact with the tissue and the handle actuation is started at a pre-defined frequency. The position of the tissue is tracked in real-time and the instant when the direction of motion changes is also detected and recorded.

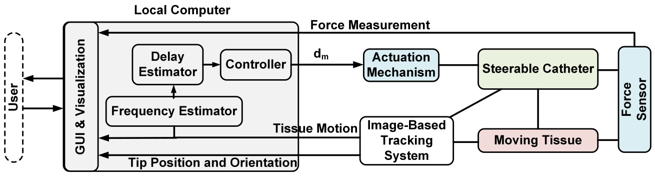


Fig. 11: Schematic of the control system with online frequency estimator.

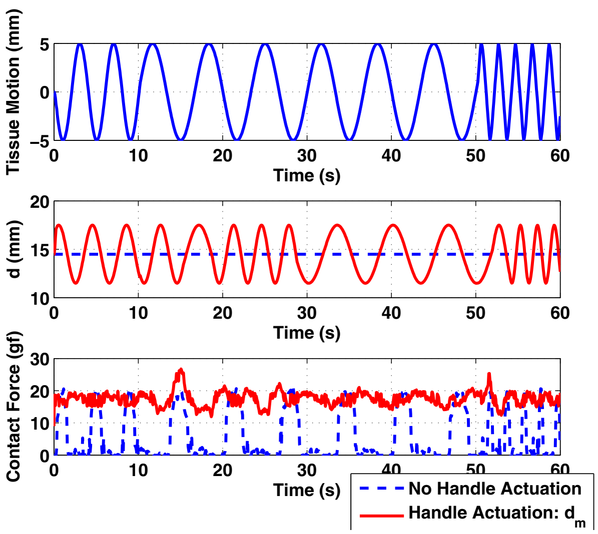


Fig. 12: Catheter tip/tissue contact force when the frequency of the tissue motion changes: (top) motion of the tissue. (middle) proximal handle actuation, (bottom) contact force.

TABLE I: FTI calculated for the cases in Figures 10 and 12.

Case	FTI (gf.s) with $d = 0$	FTI (gf.s) with $d = d_m$	improvement%
Figure 10a	212.77	367.25	173%
Figure 10b	227.8247	391.5554	172%
Figure 10c	122.8314	337.8225	275%
Figure 12	341.24	1050	307%

The frequency of motion is calculated for each motion cycle and the actuation frequency is regulated accordingly. Since the frequency of the control loop is much higher than that of motion, the actuation frequency adjusts to the calculated motion frequency quite rapidly.

Figure 12 shows the performance of the control system shown in Figure 11: the synchronization between actuation and motion is lost for the first cycles after the change in the frequency occurs, but after that the frequency change is detected, the corresponding time delay is calculated and the tip/tissue motion synchronization is restored.

D. *Evaluating Force-Time Integral*

Shah *et al.* [4] showed that force-time integral (FTI) can predict the size of lesion formed during tissue ablation. Therefore, the FTI can be used as a measure to quantify the quality of contact between the catheter tip and the tissue. To study how synchronizing the motion of the catheter tip and that of the tissue affects the quality of tip/tissue contact in the experimental setting, the FTI, *i.e.*, the area under the contact force curve, is calculated for the cases shown in Figures 10 and 12. Table I summarizes the results. For the first three entries, the FTI is calculated over 20 s and the last entry is calculated over 60 s. It is observed that the proposed technique significantly improves the quality of contact.

V. Discussion

A. *Performance Limitations*

The experiments carried out during this study indicated certain limitations in integrating current ablation catheters within motion compensated robotic catheter manipulation. The observed limitations are listed below:

1) The actuation mechanism of the catheter breaks when it is actuated in frequencies above 0.65 Hz. The pull-wires that connect the prismatic knob to the flexing tip could not stand the 60 s of actuation which was required for doing this study. At frequencies of about 0.65 Hz, the tendon connections inside the proximal handle snapped and the catheter needed to be replaced.

2) Performance of the catheters deteriorates significantly with continuous actuation of the handle. Figure 13 shows an example of how the catheter behavior changes as a result of continuous actuation. In this experiment, the catheter handle is actuated with a sinusoidal signal, while the environment is static (see Section IV-A). The experiment is done first with a less used catheter and then with a catheter worn out during the experiments. The handle actuation is similar in both cases, nevertheless, the measured contact force is very different. The reason is that the actuation mechanism of the catheter loses its functionality over time, especially with repeated actuation. Initially, the catheter tip bends more than 90°, but after repeated usage, the maximum bending angle reduces to less than 60°.

3) The orientation of the catheter tip with respect to the myocardium affects the lesion formation (*e.g.,* [20]). During the experiments, the angle of the catheter tip related to the artificial tissue is between 70° and 90°. It was found out that if the catheter is flexed to a smaller angle, the performance of the catheter decreases rapidly mainly due to the effect of hysteresis, and has a negative impact on the repeatability of the results.

Due to these limitations, the catheter had to be replaced several times during the experiments. The catheters used were of the same type; nevertheless, discrepancies existed in some of their features, such as the maximum displacement of the proximal knob and the maximum angle that the tip can reach in its ultimate bending configuration.

The observations on the limitations of the actuation mechanism suggest that this mechanism should be redesigned for using the ablation catheters with robotic manipulators.

Remark 1: In the experimental setting, the catheter is free to move back and forth in the sheath. However, the catheter does not slide in the sheath. For this study, the position of the base of the proximal handle is fixed, *i.e.*, the catheter body is not inserted/retracted. The actuation is applied on the prismatic knob of the handle to flex the catheter tip.

The average contact force that the catheter tip can apply on the tissue depends on the relative position between the

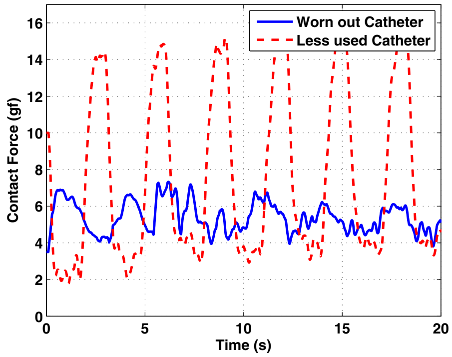


Fig. 13: The effect of wear of the catheter on the experimental results.

catheter tip and the tissue. If higher contact forces are desired, the catheter body should be inserted further into the sheath to bring the tip closer to the tissue. Arrangements for inserting/retracting the catheter in/from the sheath are not included in the preliminary experimental setting used for this study, but will be considered in future work. In such a setting, the effects of backlash and friction on motion compensation should be considered as well [12], [13].

Remark 2: The simulated motion for the atrial wall is simplified. To better evaluate the performance of the proposed technique, a study should be carried out on how different points inside the atria move during each heartbeat and the pattern as well as the amplitude of wall motion should be determined. The motion compensation system can then benefit from motion estimators to predict the wall motion and adjust the handle actuation accordingly to maintain constant contact with the tissue.

VI. Conclusions

In this paper, the initial steps towards designing a motion compensated robotics-assisted catheter manipulation system were presented. In this regard, the behavior of **conventional steerable ablation catheters** in free space when the proximal handle is actuated at different frequencies was investigated. The frequency analysis revealed that the main factor that must be considered in designing robotics-assisted catheter control systems is the time delay that exists between the handle actuation and the flexing of the catheter tip. By actuating the catheter handle at the same frequency as that of the moving target and reshaping the input handle displacement to compensate for the time delay, the motion of the catheter tip was synchronized with the target motion. Evaluating the force-time integral shows that the proposed technique improves the quality of tip/tissue contact noticeably. The experiments also pointed out the limitations in the design of actuation mechanisms for current ablation catheters. This issue would need to be addressed in order to actuate the catheters at frequencies close to those of cardiac motion.

VII. Acknowledgment

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