Design and Control of Motion Compensation Cardiac Catheters

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Samuel B. Kesner, *Student Member, IEEE,* and Robert D. Howe, *Senior Member, IEEE*

***Abstract*—Robotic cardiac catheters have the potential to revolutionize heart surgery by extending minimally invasive techniques to complex surgical repairs inside the heart. However, catheter technologies are currently unable to track fast tissue motion, which is required to perform delicate procedures inside a beating heart. This paper proposes an actuated catheter tool that compensates for the motion of heart structures like the mitral valve apparatus by servoing a catheter guidewire inside a flexible sheath. We examine design and operation parameters that affect performance and establish that friction and backlash limit the tracking performance of the catheter system. Based on the results of these experiments and a model of the backlash behavior, we propose and implement compensation methods to improve trajectory tracking performance. The catheter system is evaluated with 3D ultrasound guidance in simulate *in vivo* conditions. The results demonstrate that with mechanical and control system design improvements, a robotic catheter system can accurately track the fast motion of the human mitral valve.**

I. Introduction

Innovations in catheter technology have greatly expanded the range of procedures that interventional cardiologists can perform inside the heart using minimally invasive techniques. Procedures that are now performed using catheters include measuring cardiac physiological function, dilating vessels and valves, and implanting prosthetics and devices [1]. While catheters can perform many functions inside the heart, they do not yet allow clinicians to interact with heart tissue with the same level of skill as in open heart surgery. A primary reason for this is that current catheters do not have the dexterity, speed, and force capabilities to perform complex tissue modifications.

Cardiac catheters are long and thin flexible tubes that are inserted into the human vascular system and passed into the heart. Current robotic cardiac catheters, such as the commercially available Artisan Control Catheter (Hansen Medical, Mountain View CA), allow for teleoperated guidance of a catheter tool inside the heart [2]-[4]. These devices permit a human operator to control the positioning of a catheter *in vivo*. However, these technologies do not provide sufficient speeds to compensate for the motion of the heart, which is required for many beating heart procedures to prevent the catheter from colliding with internal cardiac structures [5],[6].

Researchers have developed robotic approaches to compensating for the motion of the beating heart [7]-[9], but these techniques are directed at procedures that repair arteries on the external surface of the heart. In previous work, we have developed robotic devices that compensate for the motion of internal heart structures *in vivo* with a handheld robotic instrument inserted through incisions in the heart wall [6], [10]-[12]. The motion of the tissue target is tracked in real time using 3D ultrasound guidance as described in [11]. This work shows that single degree of freedom (DOF) servoing is sufficient to accurately track the motion of certain cardiac structures, including the human mitral valve annulus [5],[12]. While this approach enables beating heart procedures to alleviate the risks associated with stopped heart techniques [13], the necessity of creating incisions in the heart wall means that this approach is not minimally invasive.

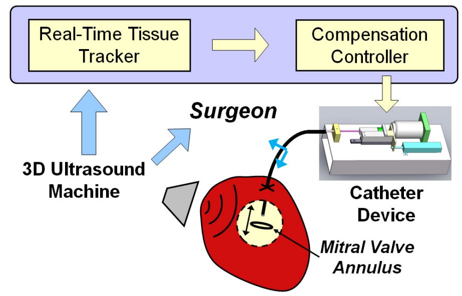

Fig. 1. The actuated catheter system.

We propose to apply our successful robotic cardiac motion compensation techniques to catheters in order to minimize the invasiveness. In the envisioned clinical system, an actuator at the base end of the catheter system will drive a catheter guidewire inside a flexible sheath. The sheath is manually advanced into the heart and positioned by a clinician. At the distal end of the catheter inside the heart, the guidewire tip translates in and out of the sheath under ultrasound guidance to compensate for the movement of the cardiac structures and perform repair. Fig. 1 presents a diagram of the catheter system.

This paper examines the friction and backlash behavior exhibited by the robotic catheter system and evaluates design and control methods to improve the system's motion compensation performance.

II. System Design

The design parameters for the actuated catheter system were selected from the human mitral valve physiology values

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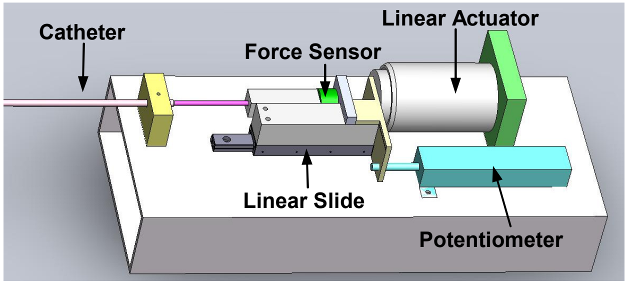

Fig. 2. Actuated catheter system prototype.

[5], [12]. The system's principal functional requirements are that it has a single actuated linear degree of freedom with at least 20 mm of travel that can provide a maximum velocity and acceleration of at least 210 mm/s and 3800 mm/s^2, respectively. These values have been shown to be sufficient to compensate for the human mitral valve motion [5], [12]. The catheter components in the device should have the same dimensions and material properties as current clinical cardiac catheters. Finally, the system should be able to apply a sufficient force to modify cardiac tissue, approximately 4 N.

The experimental system used in this study (Fig. 2) is composed of a linear voice coil actuator with 50.8 mm of travel and a peak force of 26.7 N (NCC20-18-02-1X, H2W Technologies Inc, Valencia CA), a linear ball bearing slide (BX3-3, Tusk Direct, Inc., Bethel CT), and a linear potentiometer position sensor (A-MAC-B62, Midori America Corp, Fullerton CA). The catheter sheaths are 85 cm long sections of Teflon tubing, and the guidewires are stainless steel close-wound springs. A geometric description of the various combinations of sheaths and guidewires is detailed below. The system can be divided into two modules: the base module, which includes the linear actuator, slide, and potentiometer, and the catheter module, which includes the catheter sheath and guidewire. The catheter sheath can be flexed into different configurations as required by the vascular geometry (bent, twisted, etc.) while the guidewire is servoed by the base module.

A PID control system running at 1 kHz is used to control the position of the linear actuator at the base of the catheter. Custom C++ code is used to control the system and make measurements via a data acquisition card (DAQCard-6024E, National Instruments Corp., Austin TX). Commands to the linear actuator are amplified by a bipolar voltage-to-current power supply (BOP 36-12M, Kepco Inc., Flushing NY). The friction reaction forces between the guidewire and actuation mechanism are measured with a miniature force sensor (LCFD-1KG, Omega Engineering, Stamford CT, range: 10 N, accuracy: +/-0.015 N) connected to a differential amplifier (AM502, Tektronix, Beaverton OR). The catheter tip position is measured with an ultra-low friction rotary potentiometer (CP-2UTX, Midori America Corp, Fullerton CA). The linear motion of the tip is converted into rotation of the potentiometer through a lightweight lever arm that connects the tip of the catheter to the sensor. A long lever arm is used to ensure that the

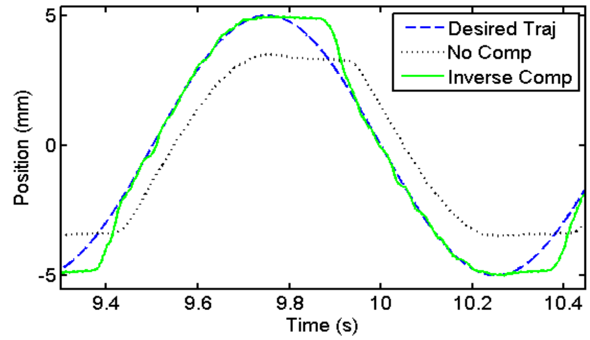

Fig. 3. The desired sinusoidal trajectory, the tip trajectory without compensation, and the improved trajectory with compensation.

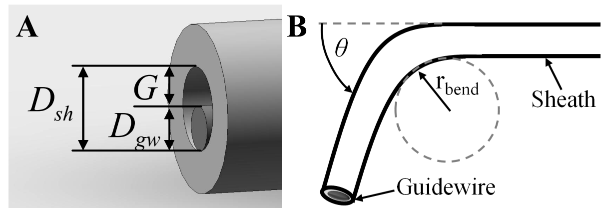

Fig. 4. An illustration of a sheath and a guidewire. (A) The gap size G is the difference between the sheath inner diameter (D_{sh}) and the guidewire diameter (D_{gw}). (B) The bending configuration is specified by the bend angle (θ) and bend radius (r_{bend}).

measurements are linear. In a clinical setting, tip position can be measured with an electromagnetic tracker or ultrasound imaging.

III. Performance Issues

Operation of this prototype system reveals two principal performance limitations**:** the friction forces experienced by the guidewire and the backlash behavior exhibited by the guidewire-sheath interaction. These two issues degrade the trajectory tracking accuracy and response time of the actuated catheter end effector. Fig. 3 illustrates an example of the uncompensated catheter tip failing to accurately track a desired trajectory. A large number of factors are involved in determining the friction and backlash properties of the catheter system. The experimental variables examined in this study include the gap size between the sheath and guidewire and the bending configuration of the catheter, characterized by the bend radii and bend angles of the sheath (Fig. 4). The catheter material properties and the external forces were held constant.

A. Friction

1) Experimental Methods

The first set of experiments examined the friction experienced by the catheter system as a function of four different sheath-guidewire gap sizes (Table 1 and Fig. 4), three bending angles (90°,180°, and 360°), and two bend radii (25 and 50 mm). The friction was calculated by commanding a series of constant velocities from the actuator in both the positive and negative directions. Force sensor readings during the constant velocity portion of the trajectory were averaged and plotted against the velocities. The friction data was summarized for each configuration by

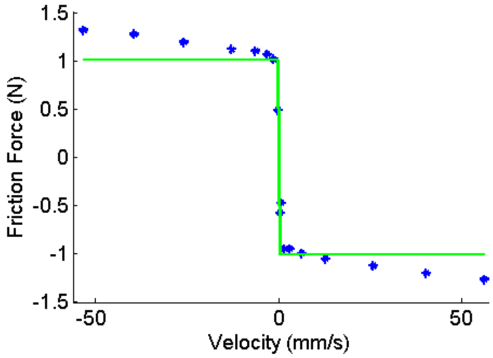

Fig. 5. Typical friction force versus velocity for the catheter system. The points are experimental values and the line is a Coulombic model approximation of the data.

taking the average of the determined friction values.

2) Friction Results

Fig. 5 presents a typical friction-velocity curve for this system. The observed behavior can be approximated as constant dynamic friction plus a component that varies linearly with velocity. For this case, the Coulomb term can be approximated as 1.0 N of friction, and the velocity dependent term as 0.006 N/(mm/s). Configurations with less than 0.05 N of friction were assumed to be frictionless because the friction was on the order of the sensor drift for the length of the experiment.

The results of the friction experiments, summarized in Fig. 6, illustrate a number of trends. The data was analyzed with a three-way analysis of variance (ANOVA). The most significant result is that the gap size has the strongest influence on guidewire friction ($p < 0.0001$). The gap size, the interior space between the guidewire and the sheath (Fig. 4), directly affects the normal forces applied to the guidewire by the sheath. The normal force is created by any sections of the sheath that might be pinch or kinked, locations where the guidewire is constrained to conform to the inner wall of the bending sheath, and guidewire or sheath discontinuities that cause the two components to come into contact. The small gap size amplifies these issues because smaller deformations in the catheter system cause the sheath and guidewire to interact. Large gap sizes, on the other hand, allow more space for misalignments. Therefore, increasing the gap size decreases the friction experienced by the guidewire.

The results also show that bend angle has an effect on the friction forces (p=0.004). One reason for this trend is that bending causes the sheaths' cross sections to deform slightly. This deformation can pinch the guidewire, thus increasing the applied normal forces. Also, the bending of the sheath forces the inner guidewire to also bend in order to conform to the outer sheath. The reaction forces generated by the conforming guidewire increase the normal force and therefore the friction on the guidewire. The bending radii used in this study, which span the typical range for cardiac catheters, do not appear to have an impact on the friction measurements (p=0.64).

These results indicate that for certain conditions, only the

TABLE I
EXPERIMENTAL CATHETER DIMENSIONS

Symbol	Sheath Inner Diameter	Guidewire Diameter	Gap Size (*G*)
▽	1.59 mm	0.76 mm	0.83 mm
*	1.59 mm	1.50 mm	0.09 mm
○	2.38 mm	1.50 mm	0.88 mm
□	2.38 mm	2.23 mm	0.15 mm

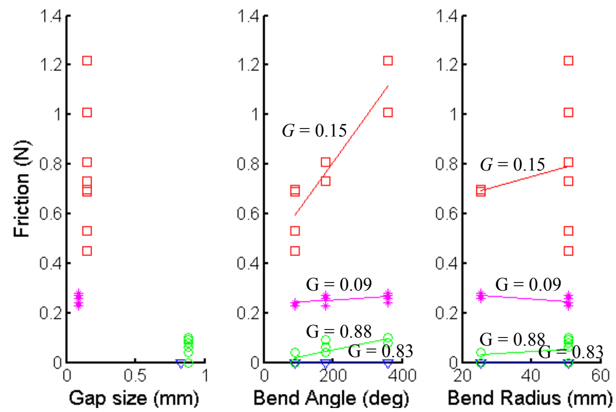

Fig. 6. Summary of the friction results versus tested parameters. See Table I for an explanation of the symbols in the plots.

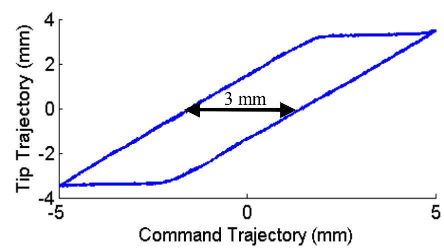

Fig. 6. Position hysteresis curve for a sheath-guidewire system with a 3 mm backlash width following a sinusoidal trajectory.

gap size and catheter bending are required to estimate the friction in the system. However, the large number of additional factors that impact the friction experienced by the guidewire, including the sheath and guidewire materials and dimensions, the catheter seals and connectors, and the external forces applied to the system, make developing a general model of the system friction challenging.

B. Backlash

1) Experimental Methods

The backlash properties of the sheath-guidewire system were investigated with the same experimental variables (gap size, bend angle, bend radius) as the friction experiments above. The backlash was examined by commanding the base of the catheter system to follow a 5 mm, 1 Hz sinusoidal trajectory (Fig. 3). This trajectory is a highly simplified version of a mitral valve annulus motion of a heart beating at 60 beats per minute (BPM). The hysteresis curve for the system plots the commanded versus measured tip position trajectory.

The amount of backlash was quantified for each experiment by the width of the backlash hysteresis curve. For example, the hysteresis curve in Fig. 7 has a width of approximately 3 mm. The width of the hysteresis is the amount of displacement commanded at the base of the catheter that does not result in any movement at the tip.

2) Backlash Results

The experimental data presented in Fig. 8 summarizes the effect of the three experimental parameters on the amount of backlash. The backlash data was analyzed with a three-way ANOVA. Bend angle has the clearest effect on backlash ($p < 0.0001$). The backlash width was found to be approximately proportional to the bend angle. The other parameter that was found to affect the backlash was the gap size ($p < 0.0001$). While the gap size value did not proportionally relate to backlash, the data does suggest that the larger the gap size, the larger the possible amount of backlash. Bend radius was not found to have a significant effect on the backlash width (p=0.53).

3) Backlash Model

We developed a model to explain the backlash width values in these experimental results. The catheter guidewires utilized in this system are different from tendon transmission mechanisms because unlike tendons, the guidewires are used both in tension and compression, which can result in buckling [14]-[16]. Unlike backlash models that describe the effects of backlash on displacement and force transmission, our model predicts the size of the backlash deadzone [17]. If there is a bend in the catheter sheath, the guidewire must conform to the bend inside of the sheath. Under tension, the guidewire uses the inside of the curve as a bearing surface and slides along this inner surface of the sheath. When the guidewire force changes directions to compression, the guidewire is forced to switch positions and conform to the outside of the sheath (Fig. 9). As the force F switches from pulling the guidewire in tension to pushing it in compression, the guidewire tip does not initially move despite the translation of the base. The backlash width is a function of the change in the length of guidewire required to conform to the curvature inside the sheath. This change depends on the physical configuration and dimensions of the system. The backlash width B can be predicted as the change in curve length

$$B = \theta(r_{bend} + D_{sh} - \tfrac{1}{2} D_{gw}) - \theta(r_{bend} + \tfrac{1}{2} D_{gw}) \qquad (1)$$

where θ is the bend angle of the sheath, r_{bend} is the bend radius of the sheath, D_{sh} is the inner diameter of the sheath, and D_{gw} is the diameter of the guidewire. This equation can be simplified to a function of the bend angle and gap size

$$B = \theta(D_{sh} - D_{gw}) \qquad (2)$$

The backlash model (2) was evaluated with the backlash data presented in Fig. 8. The model predicted values are plotted against the experimental data in Fig 10. The root mean square (RMS) error for the model is 0.4 mm and the coefficient of determination r^2 is 0.93.

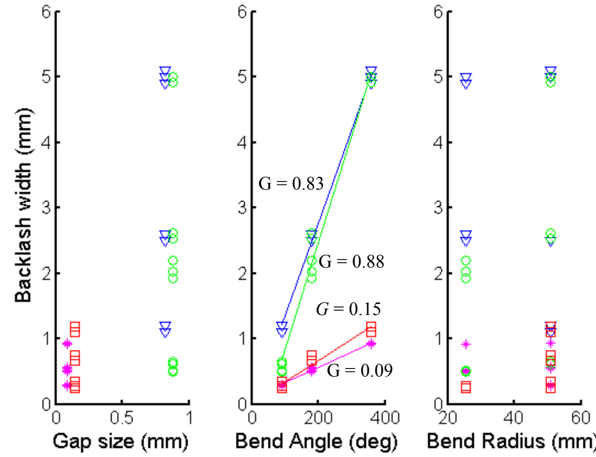

Fig. 8. Summary of the backlash results versus tested parameters. See Table I for an explanation of the symbols in the plots.

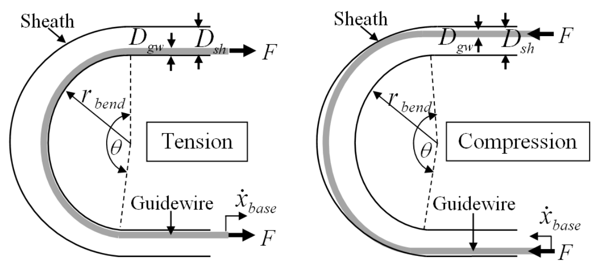

Fig. 9. Guidewire position in the sheath under tension and compression.

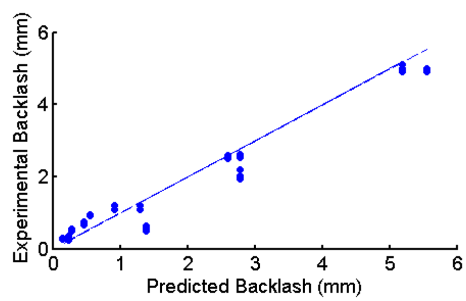

Fig. 10. Model predictions of backlash versus experimental results.

The results in Fig. 10 show that the model accurately predicts the backlash width. The model slightly underestimates the backlash for lower backlash values and overestimates for larger values. This trend is probably caused by the level of friction experienced by the guidewire. Systems with smaller gap sizes have greater friction, which causes the guidewire to buckle in compression during operation and deforms the outer flexible sheath, thus increasing the backlash width. Systems with larger gaps experience decreased friction forces, which in turn reduce the forces that drive the guidewire to conform to the inner wall of the sheath. A detailed analysis of compliant guidewires buckling inside rigid sheaths was examined in [18], which could be extended to account for the sheath deformation seen here.

4) Guidewire Friction Effects on Backlash

To evaluate the effects of guidewire friction forces on the accuracy of the backlash prediction model, a 0.76 mm

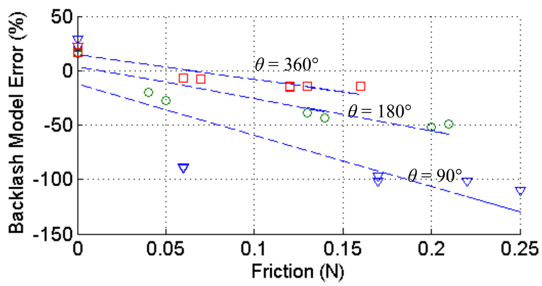

Fig. 11. Backlash model prediction error versus friction for three bend angles. Increased friction decreases the accuracy of the model.

diameter guidewire and a 1.59 mm inner diameter sheath were commanded to follow a 1 Hz trajectory in various bend angle configurations. The amount of friction experienced by the guidewire was varied by applying a normal force to the sheath at the tip end of the guidewire.

The experimental results, presented in Fig. 11, support the hypothesis that the amount of catheter friction affects the accuracy of the backlash model. The trends show that the model percent error decreases as the friction increases for each configuration. This understanding of how the amount of backlash is affected by friction resistance can be used to improve backlash compensation.

IV. Compensation Techniques

The above results demonstrate that we have identified the major factors that affect catheter system trajectory tracking performance. This understanding can be used to improve performance through both mechanical design and control system modifications. Friction in the catheter system can be reduced through material selection, material coatings, and lubrication. The backlash can be decreased by reducing the gap between the guidewire and the sheath. However, reducing the gap will also increase the amount of friction experienced by the guidewire. This design tradeoff will need to be explored further based on the performance requirements and design constraints of future applications.

The system backlash and friction can also be reduced through improvements to the control system. For example, feedforward Coulomb friction compensation can be used to reduce the friction force effects in the base module [19]. However, friction compensation primarily improves the trajectory tracking of the base module. It is not able to reduce the main source of trajectory tracking error at the tip, the backlash behavior of the guidewire. While the backlash is related to the friction resistance in the catheter, compensating for the friction at the base actuator does not reduce the backlash effects on the guidewire.

An enhanced control system can reduce the backlash behavior by modifying the trajectory commanded at the base of the catheter. For example, the trajectory can be extended to ensure that the tip of the catheter overcomes the backlash deadzone and reaches the desired location. An example of a standard backlash deadzone compensating method is to solve for the inverse of the backlash [17]. This method, known as

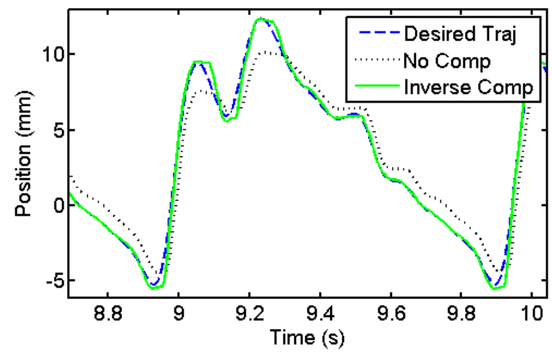

Fig. 12. The mitral valve annulus trajectory, the tip trajectory, and the improved tip trajectory with inverse compensations.

inverse compensation, consists of commanding the system to follow a new trajectory created by adding the tracking error to the original trajectory. Limitations of this method are that it assumes the system can traverse the deadzone region instantaneously and that the system backlash width is constant and not velocity-dependent [17].

1) *Backlash Compensation Evaluation*

The inverse compensation method was evaluated on the actuated catheter system in conditions that simulated a cardiac intervention. In this experiment, a 0.8 mm diameter guidewire and a 1.6 mm inner diameter sheath were constrained to a configuration with two 90° bends that simulated a realistic anatomical approach of passing the catheter from the inferior vena cava into the right atrium with a 50 mm bending radius, crossing the atrial septum, and then turning towards the mitral valve with a 25 mm bend radius. A rubber seal attached to the end of the sheath simulated a valve used to prevent the gap between the sheath and guidewire from filling with blood.

Inverse compensation was first applied to the 1 Hz sinusoidal trajectory. Initially, the tip position trajectory tracking mean absolute error (MAE) for the sinusoidal trajectory was 1.28 mm. The inverse compensation trajectory improved the tip position trajectory tracking by 80%, to a MAE error of 0.26 mm. Fig. 3 shows the goal sinusoidal trajectory, the original catheter tip trajectory, and the inverse compensation improved tip trajectory. The main reason the inverse trajectory does not perfectly remove all of the backlash behavior is because the guidewire does not move instantaneously through the deadzone region (e.g. near 9.8 sec in Fig. 3).

The ultimate goal for the catheter system is to perform repairs inside the heart. An example of a challenging cardiac structure to track is the mitral valve annulus, the outer rim of the heart valve that sits between the left atrium and ventricle. Modification of this structure (i.e. mitral annuloplasty) is a common aspect of mitral repair surgery. The inverse compensation method was applied to a typical mitral valve annulus trajectory taken from human ultrasound data [5] (Fig. 12). Initially, the catheter tip failed to track the extremes of the mitral valve trajectory. However, the tip trajectory tracking greatly improved when the inverse

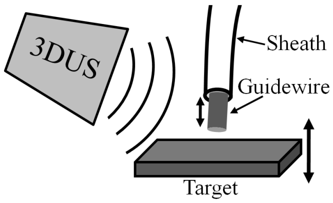

Fig. 13. The water tank *in vivo* simulator setup. The target motion replicates the 1D motion of the human mitral valve.

compensation trajectory was applied to the system. The inverse method decreased the MAE error from 1.19 mm to 0.24 mm, an improvement of almost 80%.

While this compensation method greatly improves the catheter tip tracking, it requires the system to first follow the commanded trajectory inaccurately and then calculate how to alter the trajectory to improve tracking. This approach is impractical for the real time control of a system *in vivo* because it assumes that the environmental conditions are constant during operation. Other methods that would improve trajectory tracking *in vivo* include an adaptive compensator that updates a model of the system backlash based on the tracking performance or a repetitive control system that takes advantage of the periodicity of the cardiac motion [17], [20], [21]. Another strategy is to provide close-looped control for the catheter tip position. These potential methods will be explored in future research.

V. IMAGE BASED CATHETER CONTROL

To investigate the feasibility of using the catheter motion control system *in vivo*, we integrated the catheter system described above with the ultrasound visual servoing system we developed in previous work [5],[6],[10]-[12] and evaluated the system in a water tank *in vivo* simulator (Fig. 13). Controlling a catheter to follow the motion of internal cardiac structures requires real-time sensing of both the catheter tip and tissue target positions. 3D ultrasound must be used for guidance because it is currently the only real-time volumetric imaging technique that can image tissue through blood.

In the ultrasound servoing system, 3D image volumes are streamed via ethernet to an image processing computer (Fig. 1). A GPU-based Radon transform algorithm finds the catheter axis in real-time. The target tissue is then located by projecting the axis forward through the image volume until tissue is encountered; this allows the clinician to designate the target to be tracked by simply pointing at it with the catheter. To compensate for the 50-100 ms delay in image acquisition and processing, an extended Kalman filter estimates the current tissue location based on a Fourier decomposition of the cardiac cycle. *In vivo* experiments using this servoing system showed that a rigid instrument system was capable of accurate tracking within the heart, with an RMS error of 1.0 mm. See [10]-[12] for a detailed description of the system.

Valve motion was simulated by a cam follower mechanism that replicated the dominant 1D motion component of the human mitral valve as determined in [5]. The cam mechanism simulated a heart rate of 60 beats per minute. The moving target attached to the cam mechanism was extended into a water tank, to permit imaging by a 3DUS machine (SONOS 7500, Philips Healthcare, Andover, MA, USA). A potentiometer measured target position for off-line assessment of system accuracy.

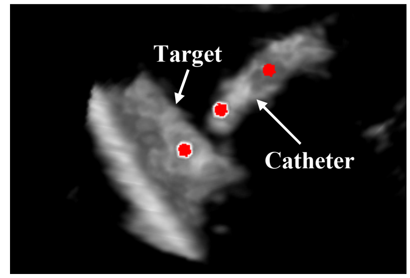

Fig. 14. Ultrasound image showing catheter and target surface. The red dots are placed by the tracking algorithm.

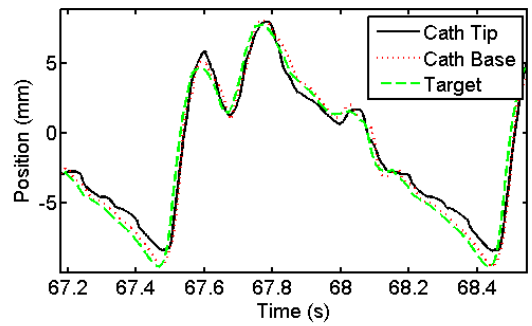

Fig. 15. Trajectory of target surface, catheter base, and catheter tip.

The catheter system consisted of a sheath with a 1.59 mm inner diameter and a guidewire with a 1.50 mm outer diameter. An electromagnetic tracker (trakSTAR 1.5 mm sensor, Ascension Technology Corp., Burlington, VT, USA, measured RMS error of 0.3 mm) was affixed to the guidewire tip to assess control accuracy.

During the experiment, the catheter was fixed in a shape that roughly corresponds to the path from the femoral vein into the left atrium as in Section IV.1 so that the tip was 1-2 cm from the target. The catheter controller performs a simple calibration routine that estimates the magnitude of the friction force and backlash deadzone. Next, the image processing routines locate the catheter using the Radon transform algorithm and then project forward to find the target. The catheter is then servoed to maintain a constant distance between the catheter and tip and the target.

A. Tracking Results

The catheter system was successful in tracking the target. Fig. 14 shows a cross section through a typical ultrasound image volume, showing the catheter and target surface. The image includes two red dots along the catheter axis and one dot at the target, which are placed by the image processing algorithm to permit visual confirmation of tracking during experiments.

Fig. 15 shows a typical plot of catheter trajectory at both the base and tip, together with the commanded trajectory, demonstrating that the system achieves good accuracy. The RMS error between the catheter tip (measured by the electromagnetic tracker) and the actual target location was 0.86 mm. The calibration routines estimated the friction levels in the catheter to be 0.5-0.8 N and the amount of backlash to be less than 0.2 mm. In this experiment, the main source of tracking error was friction.

VI. Conclusion

Robotic catheters have the potential to revolutionize cardiology by allowing clinicians to minimally invasively perform complicated surgical tasks inside the beating heart. One of the major technological challenges to realizing this concept is the tracking of the fast moving cardiac tissue with the catheter tool. In this paper, we identified the mechanical and control challenges of servoing a guidewire inside a catheter sheath. Friction was found to increase as a function of bending angle but decrease as a function of the guidewire-sheath gap size. The backlash width was found to be a function of the catheter gap size and the bending angles. Compensation improved the catheter tip trajectory tracking by 80%. To the authors' knowledge, the system described in this paper is the first robotic catheter that can compensate for the fast motion of structures inside the heart.

These results also demonstrate the feasibility of using image-based servoing to match catheter trajectories to the motion of cardiac structures. This required integrating the catheter system with ultrasound imaging and an image processing system. Water tank studies showed that excellent tracking can be obtained, with RMS errors of less than one mm. The ultrasound servo system has previously been used successfully *in vivo* with rigid instruments with equivalent accuracy; this is encouraging for the upcoming animal trials of the catheter system.

The next step for this project is to design adaptive controllers to improve the catheter trajectory tracking *in vivo* and control the forces applied by the catheter tip to the cardiac tissue. This work suggests that a single DOF robotic catheters can achieve the performance required for specific intracardiac repair applications such as mitral valve annuloplasty. This approach is complementary to and could be readily combined with current commercial catheter robot systems like the Hansen Medical Artisan Catheter to replace manual aiming with fast actuation of the servoing direction.

Future work will be required to extend this motion compensation technology to cardiac surgery applications that require additional DOF for complex tissue trajectory tracking. In contrast to existing robotic catheter systems that move at roughly manual speeds, fast motion in the lateral directions will be required for these applications.

Acknowledgment

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References

[1] D. S. Baim, *Grossman's Cardiac Catheterization, Angiography, and Intervention.* Lippincott Williams & Wilkins, 2005, pp. 992.

[2] T. Fukuda, *et al.*, "Micro active catheter system with multi degrees of freedom," in *Proc. IEEE Int. Conf. Robotics & Automation,* 1994, pp. 2290-95.

[3] J. Jayender, R. V. Patel and S. Nikumb, "Robot-assisted catheter insertion using hybrid impedance control," in *Proc. of IEEE Int. Conf. Robotics and Automation,* 2006, pp. 607-612.

[4] D. B. Camarillo, C. F. Milne, C. R. Carlson, M. R. Zinn and J. K. Salisbury, "Mechanics Modeling of Tendon-Driven Continuum Manipulators," *IEEE Trans. Robotics,* vol. 24, pp. 1262-73, 2008.

[5] D.T. Kettler, *et al.,* "An active motion compensation instrument for beating heart mitral valve surgery," in *Proc. of IEEE/RSJ Int. Conf. Intelligent Robots and Systems,* 2007, pp. 1290-95.

[6] S. G. Yuen, S. B. Kesner, N. V. Vasilyev, P. J. del Nido and R. D. Howe, "3D ultrasound-guided motion compensation system for beating heart mitral valve repair," in *Medical Image Computing and Computer-Assisted Intervention,* LCNS vol. 5241, 2008, pp. 711-719.

[7] O. Bebek and M. Cavusoglu, "Intelligent control algorithms for robotic assisted beating heart surgery," *IEEE Trans. Robotics,* vol. 23, pp. 468–480, 2007.

[8] R. Ginhoux, *et al.*, "Active filtering of physiological motion in robotized surgery using predictive control," *IEEE Trans. Robotics,* vol. 21, pp. 67–79, 2005.

[9] Y. Nakamura, K. Kishi and H. Kawakami, "Heartbeat synchronization for robotic cardiac surgery," in *Proc. IEEE Int. Conf. Robotics and Automation,* 2001, pp. 2014–19.

[10] S. G. Yuen, P. M. Novotny and R. D. Howe, "Quasiperiodic predictive filtering for robot-assisted beating heart surgery," in *Proc. IEEE Int. Conf. Robotics and Automation,* 2008, pp. 3875-80.

[11] P. M. Novotny, J. A. Stoll, P. E. Dupont and R. D. Howe, "Real-time visual servoing of a robot using three-dimensional ultrasound," in *Proc. IEEE Int. Conf. Robotics and Automation,* 2007, pp. 2655-60.

[12] S. G. Yuen, D. T. Kettler, P. M. Novotny, R. D. Plowes, and R. D. Howe, Robotic "Robotic Motion Compensation for Beating Heart Intracardiac Surgery," *Int. J. Robotics Research*, vol. 28, no. 10, pp. 1355-72, 2009.

[13] M. F. Newman, *et al.,* "Longitudinal Assessment of Neurocognitive Function after Coronary-Artery Bypass Surgery," *New England J Med*, vol. 344, no. 6, pp. 395-402, 2001.

[14] M. Kaneko, T. Yamashita and K. Tanie, "Basic considerations on transmission characteristics for tendon drive robots," in *Proc. Fifth Int. Conf. Advanced Robotics,* 1991, pp. 827-832.

[15] A. Nahvi, J. M. Hollerbach, Y. Xu and I. W. Hunter, "An investigation of the transmission system of a tendon driven robot hand," in *Proc. IEEE/RSJ Int. Conf. Intelligent Robots and Systems,* 1994, pp. 202-208.

[16] G. Palli and C. Melchiorri, "Model and control of tendon-sheath transmission systems," in *Proc. IEEE Int. Conf. Robotics and Automation,* 2006, pp. 988-993.

[17] M. Nordin and P. Gutman, "Controlling mechanical systems with backlash - a survey," *Automatica,* vol. 38, pp. 1633-1649, 2002.

[18] E. K. Bassett, *et al.*, "Design of a mechanical clutch-based needle-insertion device," *PNAS,* vol. 106, pp. 5540-5545, 2009.

[19] B. Armstrong-Helouvry, P. E. Dupont and C. Canudas De Wit, "A survey of analysis tools and compensation methods for control of machines with friction," *Automatica,* vol. 30, pp. 1083-1138, 1994.

[20] D. A. Recker, P. V. Kokotovic, D. Rhode and J. Winkelman, "Adaptive nonlinear control of systems containing a deadzone," in *Proc. IEEE Conf. Decision & Control,* 1991, pp. 2111-15.

[21] R. Horowitz, "Learning Control of Robot Manipulators," *Trans. ASME,* vol. 115, pp. 402-411, 1993.